Mixed and Boundary Lubrication in Natural Synovial Joints

by

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The candidate confirms that the work submitted is his own and that appropriate credit has been given where reference has been made to the work of others.

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Dedicated to

My Sisters

Sally and Lindsey

Good Luck & Best Wishes for the Future

to all Four of You!

Abstract

Human synovial joints have to withstand complex, varied and often harsh loading regimes. For approximately 90% of the time synovial joints and articular cartilage undergo very little motion, and these periods are commonly associated with significant amounts of prolonged loading. During these periods the lubricating films are too thin to separate the opposing articular cartilage surfaces and direct contact occurs within a mixed or boundary regime. Under these conditions friction must remain low to prevent damage to the articular cartilage. The lubrication mechanisms which continue to provide low friction and minimum wear in synovial joints are far from clearly understood and few adequate studies have been undertaken to address this important issue. The specific objective of this study was to enable an enhanced comprehension of how synovial joints perform under conditions of breakdown of the fluid film by investigating and clarifying the lubrication mechanism(s) which occur while mixed and boundary regimes predominate for articular cartilage specimens.

Friction testing of cartilage/metal, cartilage/cartilage and cartilage/hydrogel contacts were conducted when boundary and mixed lubrication was predicted. Both Ringer's solution and synovial fluid were used throughout as the lubricants in order to assess whether or not the constituents of the synovial fluid, not present in the Ringer's solution, served to reduce friction levels. The key variables for the friction tests were load duration and loading history of the cartilage specimens, choice of lubricant, contact configuration and loading regime (i.e. stationary, reciprocating or cyclic).

Loading time, and subsequent load removal, has throughout the course of this project been shown to be a major determinant of friction for articular cartilage. Articular cartilage consists of a solid matrix phase and interstitial fluid phase. The interaction of these two phases is largely responsible for the biomechanical behaviour of the 'biphasic' cartilage. It has been reasonably argued that there exists a strong link between the fluid phase and friction coefficient of articular cartilage. The biphasic lubrication mechanism has been proposed as a rational explanation of the relationship between the fluid content and tribological behaviour of cartilage. This mechanism is in keeping with the broadly held opinion that fluid flow occurs away from the contact zone and loaded areas of the tissue. In a mixed regime, it is believed that the proportion of load carried by the fluid phase contributes little to the total or aggregate friction force. As the loading period is increased the load carried by the fluid phase decreases and that carried by the solid phase increases. Hence, the overall friction and friction coefficient increases. In respect of these conclusions, the conventional view of mixed lubrication in diarthrodial joints has been reappraised.

Well hydrated cartilage surfaces are considered to be of prime importance in maintaining low friction coefficients within a mixed lubrication regime. The fluid phase load carriage mechanism in reducing friction could be specific to a certain region of the cartilage, e.g. the superficial tangential zone. It is also quite possible that the cartilage's surface or boundary layer acts independently from the cartilage matrix in affording biphasic lubrication. In view of the normal physiological

kinematics of healthy diarthrodial joints, during day-to-day activities, which permit the cartilage layers to remain well hydrated, the occurrence of high, sustainable coefficients of friction, leading to cartilage wear and the potential onset of arthritis is thought to be unlikely.

Synovial fluid produced systematically reduced friction coefficients, in comparison to Ringer's solution, for the cartilage/metal contacts under constant load reciprocating motion and the cartilage/cartilage stationary loaded contacts; which were statistically significant ($p<0.05$). The synovial fluid is thought to replenish the cartilage's phospholipid/glycoprotein boundary layer and so aid boundary lubrication, and also provide a boosted lubrication mechanism whereby the large spherical hyaluronic acid-protein macromolecular complexes (0.2-0.5 μm in diameter, in their unloaded state) minimise further asperity contact between the opposing surfaces in a mixed lubrication regime.

In addition, surface analysis of articular cartilage was undertaken. The techniques utilised for this investigation were stylus and laser profilometry and scanning, environmental scanning and transmission electron microscopy. The main aim of this particular work was to reliably quantify the surface roughness of articular cartilage, and to identify and characterise the features of any boundary layer that might exist. It was concluded that for general analyses of synovial joint lubrication, e.g. for the prediction of full fluid film or mixed regime lubrication, it was appropriate to ascribe a quantitative value of roughness of 1-2 μm. This however was largely associated with the waviness or form of the cartilage surface. A more realistic value of intrinsic cartilage surface roughness of ~0.1-0.3 μm, due to surface features such as collagen fibres, was put forward. Cartilage lubrication theories involving microscopic analysis of surface roughness within a mixed lubrication regime would be better suited to adopt this range of values. A distinct acellular, noncollagenous surface/boundary layer was observed (<1 μm thick) and, in consideration of the friction test results, was deemed to be an inherent feature of the articular cartilage surface. This layer has potential implications for cartilage nutrition and permeability, aside from its role in synovial joint lubrication.

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Notation and Abbreviations

SF	Synovial fluid
HA	Hyaluronic acid
DPPC	dipalmitoyl phosphatidylcholine
μ	Coefficient of friction (μ=F/W)
F	Friction force
W	Applied normal force
PG	Proteoglycan
EHL	Elastohydrodynamic lubrication
OA	Osteoarthritis
SEM	scanning electron microscopy
ESEM	Environmental scanning electron microscopy
CSEM	Conventional scanning electron microscopy
TEM	Transmission electron microscopy
$p_{applied}$	externally applied pressure
$p_{swelling}$	inherent swelling pressure of cartilage
$p_{elastic}$	Solid matrix elastic pressure
Δp	net pressure differential

1. Introduction

Over the last 20-30 years detailed theoretical and experimental studies have been directed towards the understanding of lubrication mechanisms that occur in synovial joints, particularly in the lower limb during periods of walking and running. In order to provide the low friction and minimum wear characteristics associated with healthy synovial joints it is generally regarded that during walking and running a lubricating fluid film is present between the two opposing articular cartilage joint surfaces. The fluid film is supported by mechanisms such as squeeze film, elastohydrodynamic and micro-elastohydrodynamic lubrication, which are discussed in section 1.1 below. While the possibility of some direct cartilage-cartilage contact was not totally ruled out, its occurrence and subsequent effect was deemed to be negligible. However, the lubrication of lower limb synovial joints during walking and running, while being considerably important, is only a part of the story.

All synovial joints will, also, invariably be required to operate under conditions when the lubricating film is insufficiently thick to separate the cartilage surfaces and a mixed lubrication regime is said to predominate. A mixed lubrication regime will prevail when joints are less mobile or stationary and subjected to load. The lubrication mechanisms which continue to provide low friction and minimum wear in synovial joints are far from clearly understood and few adequate studies have been undertaken to address this important issue.

The specific objective of this study was to enable an enhanced comprehension of how synovial joints perform under conditions of breakdown of the fluid film by investigating and clarifying the lubrication mechanism(s) which occur while mixed and boundary regimes predominated for articular cartilage specimens.

1.1 Literature Review

1.1.1 Introduction

1.1.1.1 Joints

The three types of joints that exist in the human body are the *fibrous* (synarthroses), *cartilaginous* (amphiarthroses) and *synovial* (diarthroses). Fibrous joints are those in which the bony surfaces have very little relative movement (e.g. junctions of bones in the skull). Cartilaginous joints are those in which the bony surfaces may have some relative movement (e.g. the joints between the two pubic bones and those between the vertebrae of the spine). Synovial or diarthrodial joints allow a large degree of relative motion between the opposing bones. Some examples of this type of joint are the shoulder, elbow, hip, knee and ankle.

Diarthrodial joints have some common features (Figure 1-1):

i. Enclosed by a strong *fibrous capsule* .

ii. Inner surfaces of the joint capsules are lined with a metabolically active tissue, the *synovium* (or synovial membrane/lining). The synovium secretes *synovial fluid* and the nutrients required for the tissues within the joint and adsorbs cellular waste products.

iii. Bone ends are lined with a thin layer of *articular cartilage*. These two linings, i.e. the synovium and the articular cartilage, form the joint cavity that contains the synovial fluid. The synovial fluid (lubricant), articular cartilage (bearing material) and supporting bone form the smooth, low friction and wear, bearing system of the body.

Joint Loading and Motion

The loads transmitted by synovial joints in the lower limb in walking vary from almost zero to values which are several times greater than body weight (Table 1-1).

Joint	Peak load (body weight)
Hip	3-6
Knee	3-7
Ankle	2-5

Table 1-1 Typical values of peak load (in units of body weight) experienced by lower limb joints during walking.

Peak loads occur just after heel strike and just before toe off (Paul, 1967) and it should be noted that the average loading during the swing phase is a small fraction of body weight.

The sliding speeds encountered in synovial joints in the lower limb seem to be in the range 0-0.1 m/s. Furthermore, the speeds tend to be lowest when the loading is most severe (but only for relatively short periods); a situation which discourages hydrodynamic film formation due to entraining action (Dowson, 1981a).

The motion is often a combination of sliding and rolling, particularly in the knee, such that the region of load transmission moves across both surfaces of the articular cartilage.

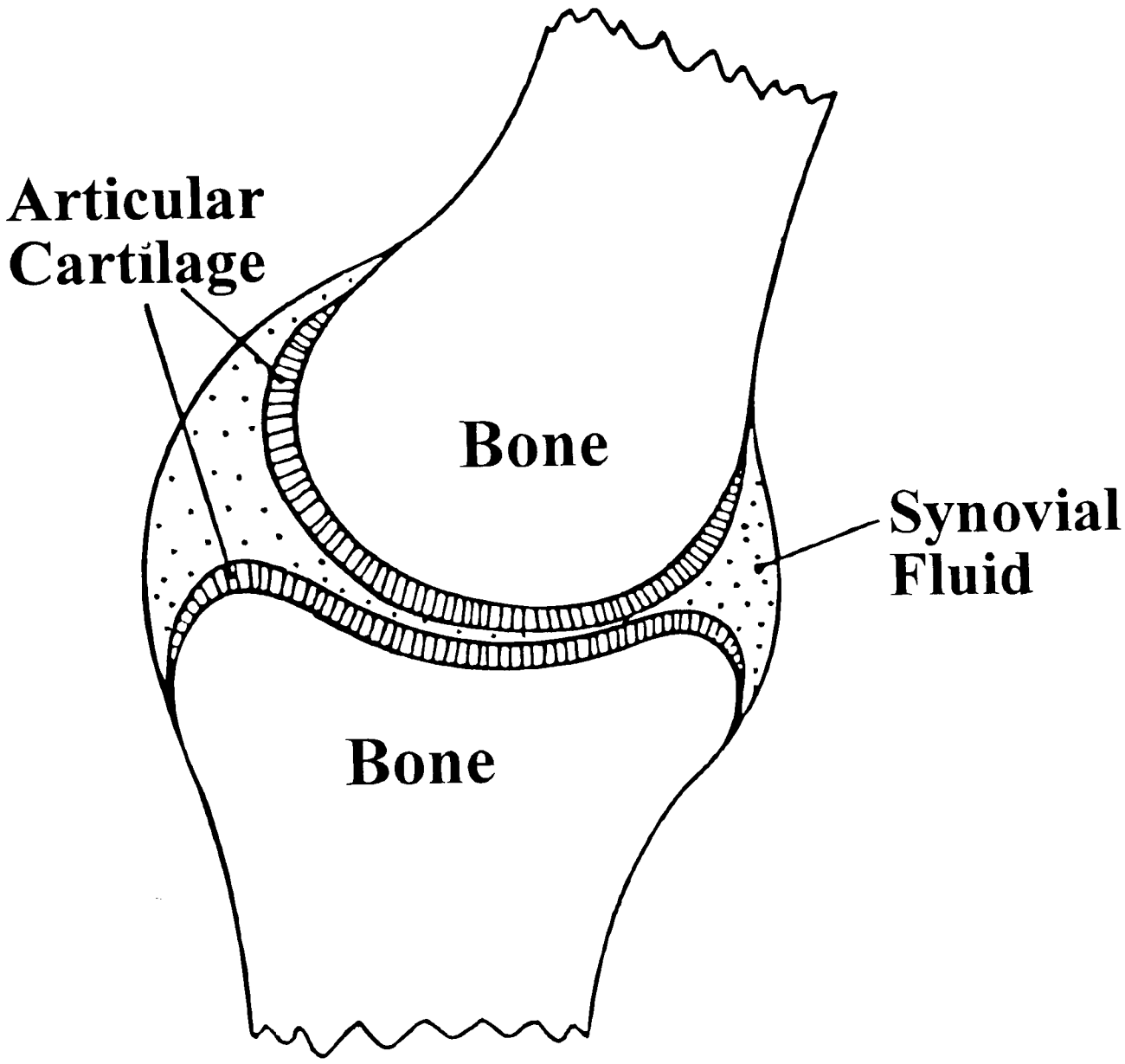

Figure 1-1 Schematic diagram of a synovial joint.

1.1.1.2 Synovial Fluid

Synovial fluid is a clear or slightly pale yellow, viscous liquid. Aspiration of a normal knee joint rarely yields more than 0.2 ml of fluid (Barnett et al, 1961). Thus the majority of the literature is devoted to studies of normal synovial fluid of larger animals, particularly bovine, and pathological human synovial fluids which are available in much larger quantities. The quantity and properties of synovial fluid vary, not only from species to species but also from joint to joint in the same animal and in the corresponding joint from animal to animal of the same species (Barnett et al, 1961). In cattle the average volume of synovial fluid in the knee joint is 10 ml and in the ankle joint 25 ml.

Synovial fluid is a dialysate of blood plasma with the addition of a protein - polysaccharide, hyaluronic acid (HA), complex which is responsible for its high viscosity. At low shear rates (0.1 s^{-1}) the viscosity of synovial fluid is of the order of a few tens of Pa s, whereas at much higher shear rates (1000 s^{-1}) it is only one thousandth of this value, ~0.02 Pa s (Cooke et al., 1978), clearly portraying its non-Newtonian behaviour. Under physiological shear rates, being ~10^5-10^6 s^{-1}, extrapolation of the data available shows that the viscosity of synovial fluid can be little more than twice that of water (Dowson, 1990).

1.1.1.3 Articular Cartilage

Articular cartilage covers the end of the articulating bones in the synovial joint. Its thickness varies between species, particular joints and with location within a specific joint (Simon, 1970). Typically it ranges in thickness from 0.1 to 0.5 mm in rabbit knee joints to 1.0 to 5.0 mm in human joints, the latter occurring beneath the human patella. The main functions of this compliant tissue layer are to spread the applied load over a large area of the joint (Brown and Shaw, 1983) and to provide minimal friction and wear.

The structure of articular cartilage can be described of in terms of zones (Davies et al., 1962). Chondrocytes, the sparsely distributed cells in articular cartilage, account for less than 10% of the tissues volume. The zonal arrangement of these cells is shown schematically in Figure 1-2. Chondrocytes manufacture and maintain the organic component of the extracellular cartilage matrix. The organic matrix is composed of a dense network of fine collagen fibrils (Figure 1-3) enmeshed in a concentrated solution of proteoglycans (PGs). The collagen content of cartilage tissue ranges from 10 to 30 % by weight and 3-10 % by weight for the proteoglycans; the remaining 60 to 87 % is water, inorganic salts, and small amounts of other matrix proteins, glycoproteins, and lipids (Armstrong and Mow, 1982), collectively known as the interstitial fluid. Collagen fibrils and PGs are the structural components transmitting the internal mechanical stresses that result from loads being applied to the cartilage. The structural components (the solid matrix) together with water determine the biomechanical behaviour of this tissue (Mow et al., 1991). For this reason it is often modelled as a biphasic

material (Spilker et al., 1992; Goldsmith et al., 1996) or even triphasic with an additional ionic component (Mow et al., 1990). Under load the tissue expresses interstitial fluid into its unloaded areas and into the joint capsule. This is referred to as exudation. During the exudation process, the pressure differences produced in the fluid phase, that cause the flow, also carry varying amounts of the load, effectively reducing the load carriage and stresses in the solid matrix (Macirowski et al., 1994; Ateshian et al., 1994; Ateshian and Wang, 1995). When the load is removed the tissue recovers the lost water, due to its inherent swelling pressure. This is referred to as imbibition, rehydration or swelling. It is particularly interesting to acknowledge that the fluid phase of the cartilage tissue has been accredited with this ability to support up to 90% of the applied load (Ateshian and Wang, 1995).

Articular cartilage is a porous, permeable material with an effective pore size of 2.0 to 6.5 nm (Dowson, 1990). It is difficult to represent the mechanical properties of such a complex biphasic material since they are dependent on the nature and magnitude of the loading, time, age and condition. A representative range of the effective Young's Modulus for physiological loading frequencies is 10-20 MNm^{-2} (Dowson, 1990).

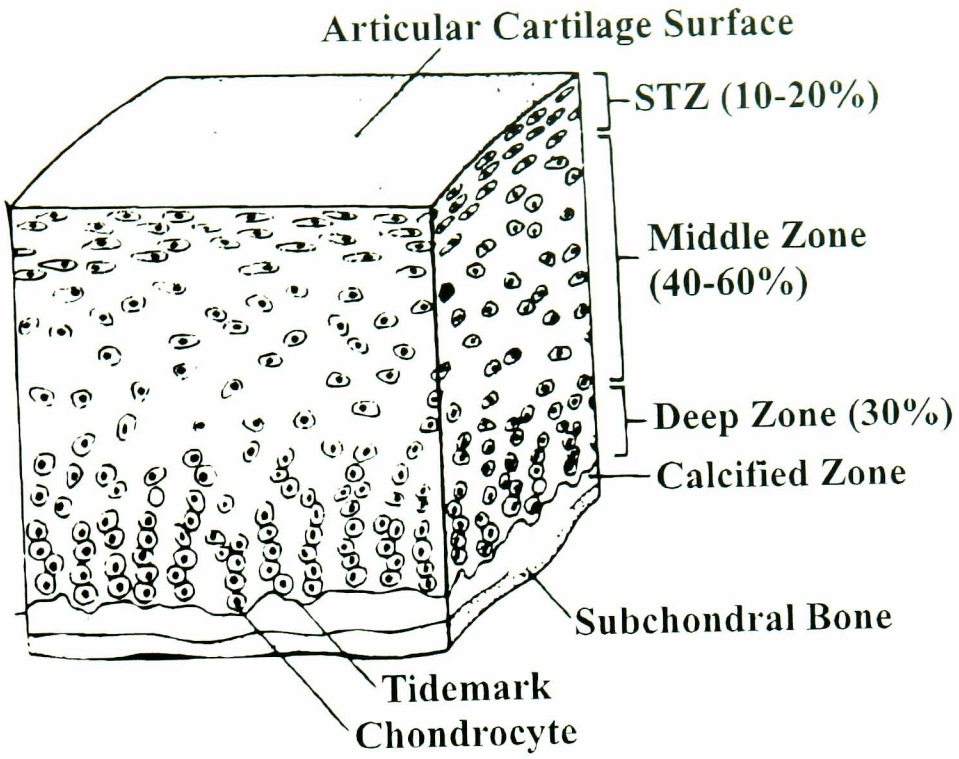

Figure 1-2 Schematic representation of the chondrocyte arrangement in articular cartilage. In the superficial tangential zone (STZ), chondrocytes are ellipsoid with their long axes aligned parallel to the surface. In the middle zone, the chondrocytes are round and randomly distributed. Chondrocytes in the deep zone are arranged in columnar fashion, perpendicular to the tidemark, being the demarcation between the calcified and noncalcified tissue. Adapted from Mow et al. (1989).

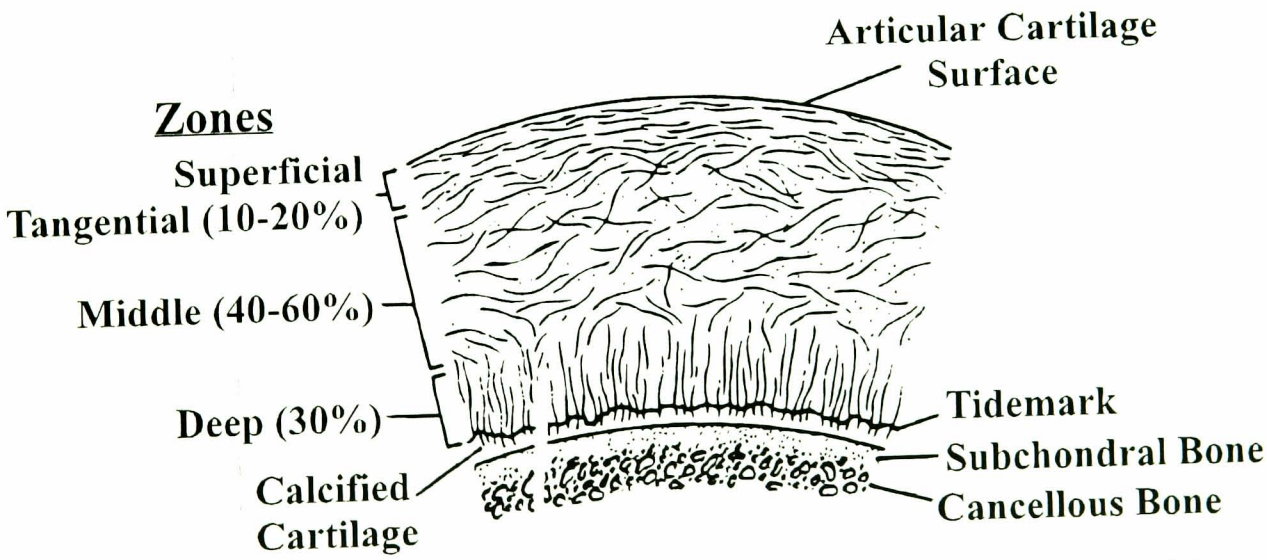

Figure 1-3 Schematic representation of the arrangement of the collagen network throughout the depth of the articular cartilage. In the superficial tangential zone collagen fibrils are tightly woven into sheets arranged parallel to the surface. In the middle zone randomly arranged fibrils are less densely packed to accommodate the high concentration of PGs and water. The collagen fibrils of the deep zone form larger, radially orientated fibre bundles that cross the tidemark, enter the calcified zone, and anchor the tissue to the underlying bone. Note the correspondence between this collagen fibre architecture and the spatial arrangement of the chondrocytes shown in Figure 1-2. Adapted from Mow et al. (1989).

1.1.1.4 *Articular Cartilage Loading and Interstitial Fluid Flow*

When a cartilage specimen which is laterally unconfined is subjected to a compressive load, two effects are observed. First there is an instantaneous elastic deformation involving a change in the shape but not in the volume of the specimen. Secondly, fluid begins to be expressed from the loaded region, which results in cartilage creep. This creep is entirely due to fluid loss from cartilage. When cartilage is not under an external load, the difference in the osmotic pressure between the inside of the tissue and the outside medium, i.e. the swelling pressure $p_{swelling}$, is balanced by the hydrostatic pressure due to the elastic forces in the collagen network $p_{elastic}$, such that

$$p_{elastic} = p_{swelling} \quad \text{(Maroudas, 1980)}$$

When an external compressive load is locally applied to cartilage, the sum of the applied hydrostatic pressure $p_{applied}$ and the hydrostatic pressure produced by the elastic stresses $p_{elastic}$ exceeds the swelling pressure $p_{swelling}$, with the result that fluid flow takes place away from the loaded region under the effect of a net pressure differential given by

$$\Delta p = p_{applied} + p_{elastic} - p_{swelling} \quad \text{(Maroudas, 1980)}$$

It is important to note that flow can only occur from loaded to unloaded regions (Maroudas, 1980). As liquid is gradually squeezed out of the matrix, $p_{swelling}$ increases, since the concentration of the hydrophilic proteoglycans in the tissue becomes greater while $p_{elastic}$ decreases. Eventually, a stage is reached when

$$p_{applied} = p_{swelling} - p_{elastic} \quad \text{(Maroudas, 1980)}$$

and a new state of equilibrium is achieved, with $\Delta p = 0$; hence no further fluid flow occurs.

During the loading process, the rate of fluid expression Q_1 is proportional to the net pressure differential, thus

$$Q_1 = B(\Delta p) \quad \text{(Maroudas, 1980)}$$

where B is a proportionality factor dependent on the permeability of the specimen and on its geometry. The term $p_{swelling} - p_{elastic}$ represents the internal cartilage pressure. If the load is removed fluid will be imbibed at a rate given by

$$Q_2 = B(p_{swelling} - p_{elastic}).$$

It is important to consider that for a cartilage specimen subjected to an external load, the parameters B, $p_{elastic}$, and $p_{swelling}$ vary during load carriage as a function of the water content. Thus, in consequence the permeability of the cartilage matrix changes with time during loading and load removal, explaining the non-linear relationship between loading time and creep deformation and recovery (Edwards, 1967). Furthermore, the stress-strain and stress-liquid content relationships *at equilibrium* are grossly non-linear (Edwards, 1967).

This particular section has covered the fundamental basics of the biphasic principles of articular cartilage used to describe the tissue's biomechanical behaviour (Mow et al., 1984). From this basic understanding far more elaborate finite element modelling of articular cartilage has been undertaken over the last 10-15 years (Goldsmith et al., 1996).

1.1.2 Joint Lubrication Studies

1.1.2.1 Introduction

Two basic methods have been adopted to experimentally analyse joint lubrication and friction. The first method uses the fact that frictional forces cause a decay in the amplitude of the oscillations of a freely vibrating system. One of the simplest vibrating systems is a pendulum oscillating in the atmosphere, pivoted to a fixed frame. By recording the oscillations over a sufficient number of cycles and performing the necessary calculations the frictional forces in the pivot (animal or human joint) may be found. The effects of air resistance, if necessary, and more importantly any forces exerted due to muscles, tendons, ligaments and any other remaining extraneous tissues must be considered in these calculations.

The following procedures have been previously adopted to calculate friction coefficients. The motion of a pendulum swinging through an angle θ is governed by the inertial, gravitational, and frictional forces acting on it. By resolving the moment of these forces, it has been shown (Clarke et al., 1975) that when there is a linear decay in amplitude of the pendulum swing, the coefficient of friction is given by

$$\mu = \frac{\text{amplitude decay per oscillation}}{4 \text{ x radius of the bearing (i.e. the synovial joint)}}$$

Thus upon observing a constant amplitude decrease per cycle the above equation can be applied to entire joint pendulum experiments (Clarke et al., 1975). As the decay per oscillation and the bearing radius are both constant, the friction coefficient is a constant and boundary lubrication is therefore indicated.

For a nonlinear amplitude decay rate the situation is more complex; fluid film / mixed lubrication is indicated. Velocity-dependent and velocity-independent frictional forces must be calculated separately (Clarke et al., 1975).

The presence of boundary lubrication will result in a linear decay of the amplitude of the pendulum's oscillations, whilst full fluid film lubrication will cause the amplitude to decay exponentially. This simple difference can be obscured if a system is lubricated with a non-Newtonian fluid (e.g. synovial fluid), having a viscosity which varies with shear rate. The apparent behaviour of such a system will tend to approach that of boundary lubrication (Swanson, 1979).

The second to be considered consists of small cartilage specimens being rubbed against a longer flat counterface, generally another material (usually glass). The cartilage surface is placed flat against the counterface and a known force, W, is applied normal to the contact surfaces. The force (in the direction of motion), F, needed to sustain relative motion is then measured such that the coefficient of friction, μ, being F/W, can be calculated.

1.1.2.2 Entire Joint Pendulum Experiments

The first work of this kind to be conducted was performed by Jones (1934), measuring the static friction of horse stifle joints. Joint pendulum studies were also carried out by Charnley (1959), Barnett and Cobbold (1962) and Little et al. (1969). Unsworth et al. (1975) and O'Kelly et al. (1978) used a different type of pendulum in which the frictional turning moment was measured directly instead of being calculated from the observed decay of free oscillations.

Typical values obtained by these workers are shown in Table 2. All the workers had taken steps to exclude forces exerted by capsules and ligaments, which may provide erroneous indications of the frictional forces. However, when considering studies which removed the joint capsule, which was usually the case, it was often unclear as to how irrigation of the joint surfaces (to prevent dehydration) was maintained and what lubricant, if any, was used.

Reference	Joint	μ
Jones (1934)	Horse stifle	0.02
Charnley (1959)	Human ankle	0.014 , 0.024
Barnett & Cobbold (1962)	Canine ankle	0.018-0.03
Linn (1968)	Canine ankle	0.0044
Little et al. (1969)	Human hip	0.003-0.015
Unsworth et al. (1975)	Human hip	0.02-0.042
Clarke et al. (1975)	Human hip	0.001-0.030
O'Kelly et al. (1978)	Human hip	0.01-0.08
*Roberts et al. (1982)	Human hip	0.04
Mabuchi et al. (1994)	Canine hip	0.007 (±0.004)
Higaki et al. (1995)	Porcine shoulder joint	0.003-0.023

Table 1-2 Coefficient of friction, μ, measured in entire synovial joints. *Friction measured by hip function simulator machine. Static or start-up friction was recorded by Jones (1934); the other μ values refer to dynamic or steady state friction measurements.

Experiments on living joints would face the problem that ligaments would exert forces probably greater than the frictional forces and muscles will always possess passive elastic and viscous forces. Thus giving an overall measurement of 'joint stiffness'. Nothing is known of any possible differences between the frictional characteristics of living and dead joints.

Experiments on entire joints have the obvious advantage that the cartilage surfaces are preserved in, as nearly as possible, the same state as *in vivo*. However, removal of muscles, tendons and intracapsular and extracapsular ligaments must cause some degree of instability and alteration of joint kinematics. While if they remain intact how accurately are they accounted for when calculating μ? Furthermore, unless the joint surfaces are nearly flat, mechanical considerations arising from the shape of the loaded areas can make it difficult to calculate μ. In the calculations only sliding can be accommodated for and therefore rolling motion must be inhibited during the experiment, necessarily restricting joint motion to generally no more than 10° (Charnley, 1959). In addition Edwards (1959) made this comment "Unfortunately whilst the pendulum method shows fairly conclusively when boundary lubrication prevails, the results are much less informative when the type of lubrication varies during the cycle".

It is interesting to note that despite the low μ values obtained in Table 1-2 some of the workers stated that a boundary lubrication regime was in operation (Charnley, 1959; Little et al., 1969; Higaki et al., 1995), as opposed to full fluid film lubrication.

1.1.2.3 Small Cartilage Specimens

The disadvantage of involved calculations to ascertain μ values using whole joints in pendulums is avoided by employing specimens small enough to be effectively flat. However the cartilage specimens are necessarily restricted to dimensions of a few millimetres, making the recording of μ for cartilage/cartilage contacts very difficult (Swanson, 1979). Thus a longer section of another material is employed as a counterface. Another disadvantage is that the flow of fluid in such a system, with the shorter flow path from the centre of the loaded area to its edges, as well as having exposed areas of the cartilage matrix, may differ from that in an intact joint.

An important advantage is that the use of small cartilage specimens on a flat counterface allows for a better understanding and control of the tribological conditions of the experiment. Table 3 gives a general indication of coefficients of friction measured by various workers. For example the water content of the cartilage specimen, by load application and removal, can be controlled and the subsequent influence on friction coefficients examined. This type of analysis would be impossible for moving joint pendulum friction studies with loaded regions of both opposing cartilage surfaces constantly moving.

Reference	Contacts	Lubricant	μ
McCutchen (1962)	cartilage/glass	SF	0.003-0.4
Dowson et al. (1968)	cartilage/glass	SF	0.1-0.9
Walker et al. (1970)	cartilage/glass	SF	0.0014-0.07
Chappuis et al. (1983)	cartilage/glass	SF	0.01-0.1
Clift et al. (1989)	cartilage/glass	None	0.1-0.2
Ikeuchi et al. (1994)	cartilage/PMMA cartilage/cartilage	saline saline	~0-0.28 0.016-0.028
Stachowiak et al. (1994)	cartilage/metal	SF	0.02-0.20
Forster et al. (1995)	cartilage/metal	SF	0.003-0.35

Table 1-3 Coefficient of friction, μ, measured for small cartilage specimens. SF - Synovial fluid.

1.1.2.4 Conclusions

The results in Table 1-2 and Table 1-3 show that the friction in synovial joints compares favourably with engineering bearings. The only engineering systems which can maintain such low μ values over a similar range of operating conditions (speed, load) are hydrostatically pressurised systems. The differing μ values in Table 1-2 and Table 1-3 are attributable to the varying tribological conditions, resulting in varied lubrication regimes, and differences in measurement methodologies.

Thus, considering not only the coefficient of friction but also the range of conditions in which it can be maintained, synovial joints can be seen to perform well by comparison with other systems at present known to engineers. This is exemplified by the consideration that synovial joints generally require little to no 'servicing' for up to 80 years.

Close inspection of the friction coefficients in Table 1-2 and Table 1-3 reveals that while a lower range of values (~0.005-0.030) are obtained for both types of friction studies, only the small cartilage specimen studies attained values greater than 0.050. This occurrence has never previously been pointed out. A possible reason, for this anomaly between the two types of experiment configurations, lies in the fact that for the small cartilage specimen tests the loaded region of the cartilage remains constant, whereas for the entire joint studies the loaded regions

are constantly moving across the opposing surfaces of the joint. The constant loading upon a relatively small area of cartilage will, of course, considerably deplete fluid content of the tissue. In the course of this particular study depletion of the cartilage's water content has been cited as a key factor in increasing friction levels.

1.1.3 Fluid Film Lubrication

1.1.3.1 Introduction

Many modes of lubrication have been proposed to explain the minimal friction and low wear characteristics of diarthrodial joints. To be acceptable, each proposed mode(s) of lubrication must be able to account for the friction and wear characteristics of these joints under a variety of loading and motion conditions. For fluid film lubrication, the minimum fluid film thickness predicted by a specific lubrication theory must exceed the combined roughness of the opposing cartilage surfaces. If the calculated fluid film gap is too thin to separate the two cartilage surfaces then solid-to-solid contact is predicted, with the possibility of boundary lubrication in action. Both lubrication types appear to occur in articular cartilage (Mow et al., 1989).

Fluid film lubrication provides low friction and a high resistance to wear; frictional resistance arises purely from the shearing of the viscous fluid. Coefficients of friction are generally in the range of 0.001-0.02 in bearings exhibiting full fluid film lubrication (Dowson, 1981a). The fluid film thickness associated with engineering bearings is usually less than 20 μm. The following terms all relate to fluid film lubrication.

Hydrodynamic lubrication occurs when non-parallel rigid bearing surfaces lubricated by a fluid film move tangentially with respect to each other (i.e. slide on each other), forming a converging wedge of fluid. A lifting pressure is generated in this wedge by the fluid viscosity as the bearing motion drags the fluid into the gap between the surfaces.

Squeeze film lubrication affords protection to bearing surfaces separated by a film of lubricant during normal approach. Since a viscous lubricant cannot be instantaneously squeezed out a pressure is built up. The pressure field in the fluid film is capable of supporting large loads. Typically, with rigid bearings, a 20 μm thick film can resist several megapascals of pressure for many minutes before the film becomes depleted.

Hydrostatic lubrication relies on a pressurised lubricant to be pumped in between the opposing surfaces. The system is often used when the velocities of the bearing surfaces are low and the loads extremely high.

Elastohydrodynamic lubrication (EHL) occurs when the pressure generated in the fluid film substantially deforms the solid surfaces (Dowson, 1967). These elastic deformations tend to increase the surface area, beneficially altering film geometry. Because the bearing contact area is increased, the lubricants escapes less readily from between the bearing surfaces, a longer lasting film is generated, and the stress of articulation is lower and more sustainable.

Many bearings operate under a combination of fluid film and boundary lubrication known as *mixed lubrication*. Here, part of the load is carried by the fluid film and part by the solid-to-solid contact. The coefficients of friction tend to be between 0.01 to 0.5 (Unsworth, 1993).

1.1.3.2 Experimental Work

This section will discuss the various lubrication regimes and mechanisms suggested to provide full fluid films in synovial joints.

Hydrodynamic

In 1932 an anatomist called MacConaill was the first to propose hydrodynamic lubrication for synovial joints. MacConaill hypothesised that the high viscosity of synovial fluid and the relative motion of the joint surfaces and structures (including the menisci) could create the wedge-shaped fluid film to operate sufficiently in the knee. However this would need continuous high speed sliding

and even so Tanner (1959) demonstrated that the viscosity of synovial fluid would be too low to form hydrodynamic films at physiological shear rates.

Elastohydrodynamic (EHL)

Dintenfass (1963) also found that the equations for hydrodynamic lubrication could not be satisfied by any credible combination of film thickness and viscosity. He suggested that EHL was possible due to the compliant nature of articular cartilage.

Tanner (1966) calculated EHL films in human hip joints to be approximately 0.1 μm thick with a coefficient of friction of 0.003, which might be increased by boundary contacts of the cartilage surfaces. More recently film thicknesses provided by this mechanism have been calculated to be as high as 1.25 and 1.3 μm (Hamrock and Dowson, 1978; Dowson, 1981b), respectively. These calculations have subsequently been shown to be too large as they were based on semi infinite elastic solids, with smooth surfaces. Film thicknesses calculated by Jin (1988) and Yao (1990) for thin layer contacts, with realistic contact areas, show film thicknesses of approximately 0.5 μm, less than three times the roughness of cartilage. So for fluid film to exist either micro-EHL (below) is necessary or the cartilage surface is much smoother than previously thought.

To date EHL remains a strong contender as one of the lubrication regimes in synovial joints.

Micro-EHL

Quite recently Dowson and Jin (1986, 1987) proposed a theory that was an extension of EHL, called micro-EHL. By the mid-1980's it had been generally accepted that the unloaded surfaces of articular cartilage were rough (Ra ≅ 1-6 μm) compared with film thicknesses of 0.5-1 μm predicted by EHL analyses for thin layers of smooth, low elastic modulus material on a rigid substrate (Dowson, 1990). However, there was a reasonably good indication of fluid film lubrication

in synovial joints from experimental studies. Thus an anomaly between the experimental and theoretical evidence existed.

Micro-EHL theory described locally generated pressures on the cartilage surface that have the ability to smooth the articular cartilage asperities as they pass through the loaded region. This work demonstrated predictions of film thickness several times greater than the effective roughness of the opposing cartilage surfaces and seems to be a contributing mechanism to fluid film joint lubrication.

Weeping and Interstitial Fluid Flow

The possibility of 'weeping' or 'self-pressurised hydrostatic' lubrication was introduced by McCutchen (McCutchen, 1959, 1962; Lewis and McCutchen, 1959). From experiments on cartilage/glass contacts McCutchen concluded that fluid, squeezed out of the cartilage under load, was maintaining the film between the contacts, as fluid leaked away at the edges.

There is no doubt that liquid is expressed from cartilage by the application of pressure (Edwards, 1967; Unsworth et al., 1975). However there is no satisfactory evidence that this exudation occurs in the most highly loaded area necessary for weeping lubrication (Swanson, 1979). Furthermore, since articular cartilage is impermeable to hyaluronate (Ogston and Stanier, 1953), any liquid expressed must be of a saline solution nature having a lower viscosity, and fluid flow rates have been found to be minute (Maroudas, 1980).

In work conducted by Mow and co-workers, the articular cartilage was modelled as a thin layer of biphasic medium supported by a hard bony substrate. Cartilage was subjected to a constant width, parabolically distributed normal load sliding over its surface at physiological speeds. This formulation yielded a 'self-generating mechanism' whereby fluid is exuded under the leading half of a moving parabolic load and resorbed under the trailing half (Mow et al., 1989). Fluid exudation was believed to begin ahead of the oncoming load and increase as the load approached. After the load passed the expelled fluid was resorbed as the

solid matrix recovered its lost fluid. This self-generating mechanism was apparently confirmed by using an optical sliding-contact analytical rheometer (Mow and Lai, 1979).

This theory, however, is again compromised by uncertainties regarding times required for sufficient exudation and subsequent resorption to occur (also note that for repeated cycles over any given period the amount of fluid exuded from the cartilage must equal that resorbed for this mechanism to work continuously); and by the probability that fluid from the cartilage matrix is not as viscous as synovial fluid.

Squeeze-Film, Boosted and Ultrafiltration Mechanisms

Tanner (1959) while disclaiming hydrodynamic lubrication as a possible mechanism in diarthroses made this statement "However, the high viscosity at low shear rates may well serve the important purpose of preventing the joint from being squeezed dry when standing still". Following on from this, Fein (1967) investigated squeeze film lubrication using a lens and an optical flat, with oil as the lubricant, to measure interference fringes as a function of time after loading. Fein calculated, for a human hip joint, that the known compliance of articular cartilage and viscosity of synovial fluid would enable a squeeze film to persist for physiologically useful times, and that hydrodynamic lubrication (although more probably EHL) replenished the squeeze film during the swing phase when the joint was lightly loaded. Unsworth et al. (1975), in joint pendulum studies, having observed increasing μ after the first few seconds of sudden loading concluded squeeze lubrication to be evident. Further analytical work by Higginson and Unsworth (1981) has also supported Fein's conclusions.

In a detailed mathematical analysis of squeeze film lubrication Hou et al. (1992) reached the following conclusions.

i. The cartilage layer deforms to enlarge the load-supporting area.
ii. Cartilage deformation acts to reduce the lateral fluid speed in the lubricant; thus prolonging the squeeze film time which ranges from 1 to 10 seconds.

iii. Fluid in the gap is forced from the central high-pressure region into the cartilage and expelled from the tissue at the low-pressure periphery of the load bearing region.

Another analysis (Jin et al., 1992) considered the effect of porosity of the articular cartilage on squeeze film lubrication in the normal human hip joint. They concluded that porosity of articular cartilage depletes the lubricant film thickness, particularly when film thickness becomes small (< 0.5 μm). Thus, if the lubricant film thickness is sufficiently large the porosity effects can be ignored. Fluid was deemed to be forced into the articular cartilage within the centre of the contact, in agreement with Hou et al. (1992), suggesting a boosted (see below) rather than a weeping mechanism.

The earlier work (Fein, 1967) ignored two important aspects involving the time dependence of cartilage deformation during squeeze film action and the permeability of cartilage. The effect of permeability in a squeeze film bearing would usually be to increase the rate of leakage and reduce the squeeze film times. However, in synovial joints the size of the HA-protein complex would cause it to be trapped between the cartilage surfaces under load, instead of passing into the cartilage matrix with the water and small solutes (Ogston and Stanier, 1951). The question of the concentration of synovial fluid in the loaded region has been experimentally evaluated by Maroudas (1967) and Walker et al. (1968), and theoretically investigated by Dowson et al. (1970) and Hlavácek and Novák (1995).

Maroudas (1967) calculated that the film thickness at which the escape of water via the cartilage supersedes the lateral flow in the gap to be ~ 0.5-1.0 μm (suggesting that porosity effects could be ignored for film thicknesses above 1 μm). A final equilibrium film thickness of ~ 0.02-0.03 μm was predicted, such that cartilage-cartilage contact would not occur, (Maroudas, 1979). Thus an HA-water gel apparently is formed by an 'ultrafiltration mechanism'. These values are, however, too low to separate cartilage surfaces when considering their

accepted roughness, this mechanism cannot therefore be assumed to provide fluid film lubrication by itself.

Walker et al. (1968) adopted perhaps a more realistic approach. Considering the roughness of articular cartilage they described trapped pools of concentrated, HA-protein complex rich, synovial fluid, with boundary contacts occurring at the pool edges. This was, therefore, presumed to occur within a mixed lubrication regime. This mechanism was termed 'boosted lubrication'.

Remaining questions are: Can the concentrate be produced within physiologically useful time scales of ~ 0.01 seconds during walking and running? Can trapped pools be formed under sliding and rolling motions? For it would appear that at least one surface requires a constant loading, i.e. loading region (on one surface at least) must be stationary. At what film thicknesses does the concentration of synovial fluid begin to occur? Does the concentration of synovial fluid (and the boundary layer!) affect cartilage permeability?

Perhaps current squeeze film theory (Jin et al., 1992; Hou et al., 1992), alongside EHL, micro-EHL and occasional boundary action, is sufficient to explain the low friction under normal ambulatory loading conditions. It seems, at present, quite likely that any ultrafiltration mechanism may only come into affect after extended periods of stationary loading.

1.1.3.3 Summary

It is unlikely that the varied demands on diarthrodial joints can be satisfied by a single mode of lubrication (Dowson, 1967). As yet, it is still impossible to state definitely under which conditions a particular lubrication mechanism may operate. Nevertheless, using the human hip during walking as an example, some general statements are possible.

Both EHL and squeeze-film lubrication, with the support of micro-EHL, probably play an important role in reducing the friction and wear in the joint. During the lightly loaded swing phase, a relatively thick film of fluid can be entrained between the cartilage surfaces. When the heel strikes the ground, the load is suddenly very high while entraining velocity is very low. Now the thick fluid film begins to squeeze out and the film thickness reduces. Since the load is applied for only a short time, this film does not reduce substantially. As the cycle progresses to the stance phase, the load on the hip reduces and the entraining velocity increases. This is the phase in the walking cycle in which (micro-)EHL is predominant. Finally, at toe-off, the load is maximum and the entrainment velocity is very low, but under normal conditions squeeze film can maintain a fluid film and prevent surface-to-surface contact (Unsworth, 1993).

After extended periods of stationary loading (e.g. standing still) and other such extreme loading conditions involving high loads and low entraining velocities for a sufficient period, any full fluid film would become depleted. For these conditions it is envisaged that the cartilage surfaces are protected by a boundary layer and possibly a concentrated synovial fluid gel, as well.

It is apparent that some workers generally describe boundary contacts as only occurring under extreme loading conditions, e.g. Unsworth (1993) and Mow et al. (1989). Although this could be the case, it is far from being conclusively proven.

1.1.4 Boundary Lubrication

1.1.4.1 Introduction

In engineering bearings, if the two surfaces involved are deliberately 'contaminated' with a substance chosen for its ability to be adsorbed by the surfaces and to remain adsorbed at a range of loads and rubbing speeds, whilst having a molecular structure offering little resistance to relative motion of the two adsorbed layers, boundary lubrication is said to exist (Swanson, 1979). Thus, the boundary layer upon the two opposing surfaces serves to reduce both friction and wear.

1.1.4.2 Experimental Work

Charnley (1959) was the first to investigate the possibility of boundary lubrication in joints. Charnley measured μ in a human knee joint (all extraneous tissue removed) via a 'see-saw' type of arrangement. The joint was weighted (W) at the centre while a spring-balance was used to apply the force (F) alternatively at either end. The force (F) necessary to start movement both in flexion and extension could then be recorded. Values of μ varied from 0.005 to 0.023 with an average, from seven readings, of 0.013. No difference in μ was observed when all visible traces of synovial fluid were removed by wiping the cartilage surfaces with a dry cloth or when the joint was freely lubricated with an excess of synovial fluid.

For a pendulum experiment Charnley choose a human ankle joint as it offered a high degree of congruity, enabling sliding rather than rolling to predominate, with a stable location, over angles of 5° each way. From this experiment two values of μ were quoted for two different specimens, being 0.024 and 0.014. Plots of decay of amplitude with the number of cycles were found to be linear indicating boundary lubrication. It was concluded that the cartilage surface was "intrinsically the antifriction element" and that synovial fluid was "not an essential feature", although fluid in the cartilage could play a role. Thus, despite the very low μ values, full fluid film lubrication was not deemed to occur.

Tanner (1959) constructed a metal human ankle joint and used both synovial fluid and industrial lubricants. Synovial fluid showed little superiority and it was again decided that the cartilage surface was the most important factor.

In another pendulum friction study, using human hip joints (Little et al., 1969), the coefficient of friction values, for synovial fluid and Ringer's solution respectively, were found to increase from approximately 0.003 and 0.006 up to 0.006 and 0.011, after 30 minutes of loading. A 2:1 solution of chloroform/methanol was then used as a lipid solvent upon the cartilage surfaces. Upon repeating the tests μ values, for synovial fluid and Ringer's solution

respectively, were found to be higher increasing from approximately 0.011 and 0.016 up to 0.025 and 0.029, after 30 minutes of loading. It was concluded that even in the presence of synovial fluid the lubrication regime was boundary and low friction was maintained by a lipid boundary layer. It is important to note that no other characteristics (gross appearance, histology, permeability, indentation stiffness and wet weight) of the articular cartilage were reported to have altered due to the lipid extraction procedure.

More recent pendulum studies, performed upon porcine shoulder joints (Higaki et al., 1995), revealed similar increases in μ after the cartilage surfaces were treated with a highly surface active surfactant (polyoxyethylene p-t-octylphenyl ether), to act as a boundary layer 'detergent'. The recorded friction coefficient values, using hyaluronic acid (HA) solutions as control lubricants, were ~ 0.008 and 0.020 before and after detergent treatment respectively. Further testing with synovial fluid and two other bio-synthetic boundary lubricants (protein and phospholipid based) did reduce friction levels but not quite back to their original pre- 'detergent' levels. This study was perhaps the first and only one to objectively assess the lubricating ability of potential macromolecular boundary lubricants. The results were however ambiguous with both the γ-globulin (glycoprotein) and phospholipid (DPPC), both in HA solutions, reducing friction levels after 'detergent' treatment by much the same amounts.

For both of these studies (Little et al., 1969; Higaki et al., 1995) even the higher μ values, (which were recorded after boundary layer removal), were still very low at < 0.03. These low levels of μ could not be regarded as catastrophic failure of synovial joint lubrication, following boundary layer removal. There is therefore a good possibility that another lubrication mechanism was still in operation to maintain friction levels at < 0.03.

Since the mid-Sixties a series of studies have investigated the ability of synovial fluid to act as a boundary lubricant for rubber/glass contacts (McCutchen, 1966;

Wilkins, 1968; Reimann et al., 1975; Reimann, 1976; Davis et al., 1978; Davis et al., 1979; Jay, 1992). The apparatus essentially always consisted of an oscillatory rotating device with a metal ring covered with a rubber layer loaded against a glass disc. Immersion of the contacts in the lubricants prior to testing seemed to be preferred to rather than using a 'lubricant bath' during testing.

McCutchen (1966) found that μ decreased with increased synovial fluid immersion periods ($\mu \sim 0.01$-0.2). Filtration at between 0.22-0.65 μm, of the synovial fluid, as well as treatment with hyaluronidase and then recording μ led McCutchen to believe that boundary lubrication was evident and that the HA-protein complex was responsible. Walker et al. (1970), from their work applying SEM and friction testing upon cartilage/glass contacts, proposed that the HA-protein rich concentrated synovial fluid "collapses", under severe loading, to form a protective boundary layer.

It had previously been found that hyaluronidase depolymerises the HA molecule, whilst trypsin destroyed the protein component of the complex (Ogston and Stanier, 1953). Subsequently, Wilkins (1968) analysed the effects of both hyaluronidase and trypsin on synovial fluid's lubricating ability. After 5 minutes digestion with hyaluronidase μ remained the same (~ 0.01), while 5 minutes digestion with trypsin increased μ to over 0.2. He concluded that the integrity of the protein component was essential to boundary lubrication and that while HA seemed necessary its natural length can be considerably shortened (reducing the viscosity) without much reduction of lubricating ability.

Reimann et al. (1975), during standardisation, found that the tension and cleanliness of the rubber could substantially affect μ values. After soaking periods of 15 minutes and longer, bovine synovial fluid was shown to lubricate better than 0.9% NaCl, 0.10 compared to 0.05. Trypsin added to synovial fluid gave a consistent μ of 0.275, again stressing the importance of the protein component in boundary lubrication.

Using the same apparatus the boundary lubricating properties of pathological synovial fluid from the knee joints of 80 patients was assessed (Reimann, 1976). The equilibrium μ values were as follows: rheumatoid arthritis 0.181 ±0.014 (19 tests); osteoarthritis 0.110 ±0.023 (10 tests); torn meniscus 0.071 ±0.005 (38 tests). The differences between the torn meniscus values and the rheumatoid arthritis values were statistically significant. In half of the knee joints with torn menisci the articular cartilage and synovial membrane were considered to be normal. From these 19 cases the mean μ value was 0.052 which was statistically lower than the mean value from the remaining cases with torn menisci, being 0.090. Variations in the boundary lubricating (and viscous) properties of synovial fluid, with different diseases, were thought to relate to the degree of cartilage degeneration (and synovitis) respectively. In contrast, a very similar study (Davis et al., 1978) found that during quiescent periods of degenerative joint disease, synovial fluid retains its normal lubricating ability. These results might suggest that defective lubrication by synovial fluid is not a major pathogenic mechanism in osteoarthritis.

In further studies (Davis et al., 1979) cartilage/cartilage and cartilage/glass contacts were employed. Bovine synovial fluid was found to provide lower μ values (0.025-0.1) than Gey's balanced salt solution, over varying conditions of load, speed and cartilage thickness. However, the cartilage was sectioned bovine nasal septum and not articular cartilage. Also both these contacts were abandoned because they were too time consuming and lacked reproducibility. Returning to the rubber/glass bearing, the results confirmed previous findings namely: ultrafiltration (pore size not specified) showed that the lubricating ability of synovial fluid resided in the retentate; trypsin digestion did not alter the viscosity of the retentate but destroyed its lubricating ability; and hyaluronidase digestion did not result in a loss of lubricating ability (results concerning latter two conclusions were not detailed).

A boundary lubrication model for synovial joints was proposed (Davis et al., 1979), based on the structuring of water within the boundary lubricant multi-

macromolecular layers, said to be glycoprotein (Swann et al., 1974). A point of note concerning this study is that synovial fluid was reported to actually increase the friction between most synthetic surfaces (highlighting its selective nature), including those commonly used for contemporary joint prostheses.

Swann and Radin (1972) and Swann et al. (1974, 1981, 1985) were the first workers to look in detail at the possible macromolecules responsible for boundary lubrication. The friction apparatus was constructed from a gramophone turntable with rabbit cartilage specimens loaded against a rotating glass disc (Swann et al., 1981). Complex biochemical preparations were involved. Fractionation studies supposedly revealed that the boundary lubricant component was present in a glycoprotein fraction and indicated that it was a high molecular weight "lubricating glycoprotein 1 (LGP-1)" also referred to as lubricin (Figure 1-4).

Using bovine synovial fluid and salt solution as lubricant controls the purified lubricin samples, in solution at varying concentrations, were shown to provide an approximate 40-60% reduction in friction, especially at speeds less than 1 cm/s (Swann et al., 1985). Coefficients of friction were never referred to in any of these studies, either in the written text or the figures, instead a detector response was quoted as a friction measurement. The mechanism of boundary lubrication was thought to possibly involve two processes: an interaction with the cartilage surface (naturally!) and the ability to maintain stable layers of water molecules at the interface (Swann et al., 1981).

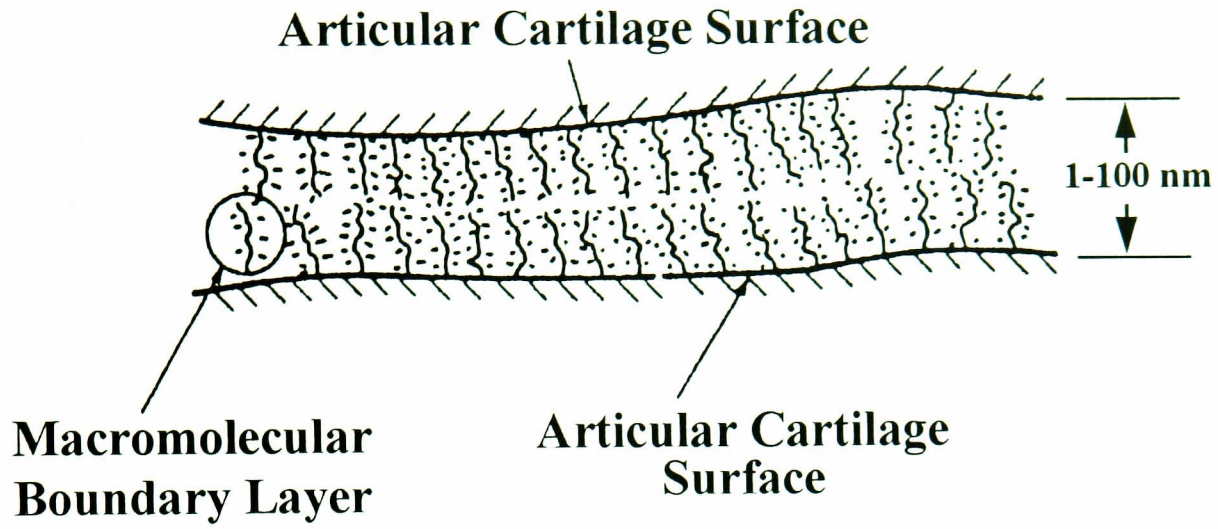

Figure 1-4 Boundary lubrication of articular cartilage. The boundary layer is currently believed to consist of macromolecular layer(s) of either phospholipids or glycoproteins. The boundary layer acts to reduce friction and helps to prevent cartilage wear when the two opposing cartilage surfaces are in contact. Adapted from Mow et al. (1989).

It was further found that lubricin isolated from three different bovine joints and from human degenerative or traumatised joints apparently all had a similar chemical structure and specific lubricating ability *in vitro* (Swann et al., 1985).

Another study (Jay, 1992) using the same experimental set-up as Davis et al. (1979), seemed to confirm that a glycoprotein was the boundary lubricant. It was referred to as purified synovial lubricating factor, PLSF, and described as being very similar to Swann's lubricin or LGP-1. Using 0.9% NaCl as a control lubricant, the control coefficient of friction value for the rubber/glass contact was established, μ=0.110($\pm$0.006). The lubricating ability of bovine synovial fluid, purified hyaluronic acid and purified glycoprotein was then investigated; μ varied by $\pm$0.06 about the 0.110 control value for these lubricants. The boundary lubricating ability of PSLF was maximal at a concentration of 260 μg/ml reducing μ by approximately 0.05. At this concentration a boundary monolayer of PSLF was deemed to have formed.

It was proposed that repulsive hydration forces, rather than electrostatic (Roberts, 1971; Stanescu and Leibovich, 1982), were the responsible mechanism in the boundary lubrication of the rubber/glass contact, with separation of the boundary surfaces having to be below 3 nm.

In further work (Jay et al., 1992) an interaction between PSLF and hyaluronate was demonstrated by three methods: enhancement of lubrication in rubber/glass contacts; viscometry; and electron microscopy. It was thought that while hyaluronate was not intrinsically a boundary lubricant, it beneficially bound with PSLF in the boundary lubricating activity.

Earlier work whereupon joint friction was increased by 130% when the articular surfaces were rinsed with a lipid solvent (Little et al., 1969) indicated the potential importance of lipids in joint lubrication. Recent work has proposed that phospholipids are the macromolecules responsible for boundary lubrication (Hills, 1989, 1990; Williams et al., 1993). This work apparently dismisses the glycoproteins as being responsible for boundary lubrication.

Having observed phospholipids to be active surfactants in other physiological environments (Hills et al., 1982), synovial joint lubrication was investigated (Hills and Butler, 1984; Hills, 1989, 1990). Optically flat quartz/quartz contacts were used and lubricated with dried on layers of purified phospholipid and commercially available phospholipid (DPPC - dipalmitoyl phosphatidylcholine, the major lipidic component of synovial fluid {45%}). Cartilage/quartz contacts were initially used but abandoned in favour of a better defined system, even if the quartz/quartz contacts were far less physiological. "Synthetic synovial fluid", being commercial HA dissolved in distilled water and pH adjusted to 6-7 with the addition of DPPC (3 mg/ml), was also used as a lubricant.

Values of μ ranged from 0.002-0.006 for both dried on layers and the synthetic synovial fluid. If HA was not added to the synthetic synovial fluid μ values were of the order of 0.01-0.1. This finding seems to question the ability of the DPPC in

acting as an independent boundary lubricant. Considering the fact that some of these layers were 'dried on', these μ values do seem incredibly low but without any fluid present must have been recorded within a boundary regime.

Electron microscopy analysis (Hills, 1990) revealed what was considered to be oligolamellar phospholipid lining the articular cartilage surface. The mechanism of boundary lubrication was proposed to be imparted by the shearing between multi-molecular layers (6-8 layers evident) of phospholipid on the articular cartilage surface.

Similar work (Williams et al., 1993) involving glass/glass contacts of known surface roughness used solvent cast DPPC layers of estimated physiological thickness, 70 nm. In general the μ values for clean glass surfaces were approximately 5 times higher than for the lubricated surfaces, 0.70 and 0.14 respectively. The test set-up employed appeared substantially removed from physiological conditions and was not thoroughly validated. Even so they inferred that their work supported the previously mentioned hypothesis (Hills, 1989), which regarded DPPC to be the active boundary lubricant in synovial joints.

It would have been interesting to have investigated the effects of similar coatings of HA and glycoproteins for this study. This would have demonstrated the effectiveness of DPPC as a boundary lubricant far more objectively.

1.1.4.3 Conclusions

Synthetic/synthetic contacts with synovial fluid as the lubricant (or a bio-synthetic boundary lubricant) has commonly been adopted to investigate boundary lubrication. These studies are, however, far removed from physiological conditions. The problem of such studies is the interpretation of the results in relation to the biotribology of synovial joints *in vivo*. Their scope is limited to the lubricating ability of synovial fluid and bio-synthetic boundary lubricants afforded to synthetic contacts, with possible test variables of viscosity, lubricant filtration and selective enzyme degradation. The unsubstantiated assumption of these

studies is that boundary lubrication is solely imparted by the synovial fluid. Even if this were so the articular cartilage surface should remain a key feature in these investigations, as boundary layer bonding to synthetic surfaces and its subsequent lubricating abilities may be quite different. Thus synthetic/synthetic contacts using synovial fluid as the lubricant would seem inappropriate for investigating boundary lubrication in diarthroses.

Furthermore the variability in the applied methodology is wide. Different studies have used varying speeds and loads; reciprocating or continuous motion; differing contacts (counterface roughnesses are often disregarded), lubricants, and type of friction apparatus; as well as varying conditions of enzyme degradation, specimen storage and preparation. Results cannot, therefore, be directly correlated and useful comparisons and conclusions from data of this kind can only be made within a carefully controlled study (Clift et al., 1989). All these tests suggest that boundary lubrication is the lubrication regime without detailing the surface roughnesses or film thicknesses. Whether or not boundary lubrication is actually occurring for a given experimental set-up has been previously questioned (Jay (1992) referring to Swann et al.(1981)). A final comment is that none of these synthetic/synthetic studies seem to have adequately demonstrated , in an objective manner, the lubricating advantages of any of the boundary lubricants mentioned.

1.1.5 Overview of Mixed and Boundary Lubrication Regime

The conventional view of synovial joint lubrication during episodes of full fluid film breakdown involves two components, Figure 1-5. The first is the presence of a boundary layer[1], generally quoted at being ~1-100 nm thick which maintains low friction and prevents wear when direct cartilage-cartilage asperity contacts occur. Secondly, in between the asperity contacts there are pools of fluid existing between the two opposing joint surfaces. The favoured lubrication mechanism operating in these pools is a boosted or ultrafiltration mechanism, whereby for a sufficiently thin film of < 1 μm, the solute of the synovial fluid is forced into the

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[1] Discussed in detail within Chapter 3.

cartilage leaving behind a highly concentrated HA-protein macromolecular rich synovial fluid gel which is better able to support the load and thus minimise further asperity contact. Another, more controversial, weeping lubricating mechanism has been proposed such that interstitial fluid from the cartilage flows *into* these fluid pockets, and in doing so again maintains the fluid pockets, and minimises further asperity contact. However it is widely thought that upon load application the net flow of fluid will be directed away from the loaded contacting cartilage surfaces; thus, supporting boosted lubrication and not the weeping mechanism.

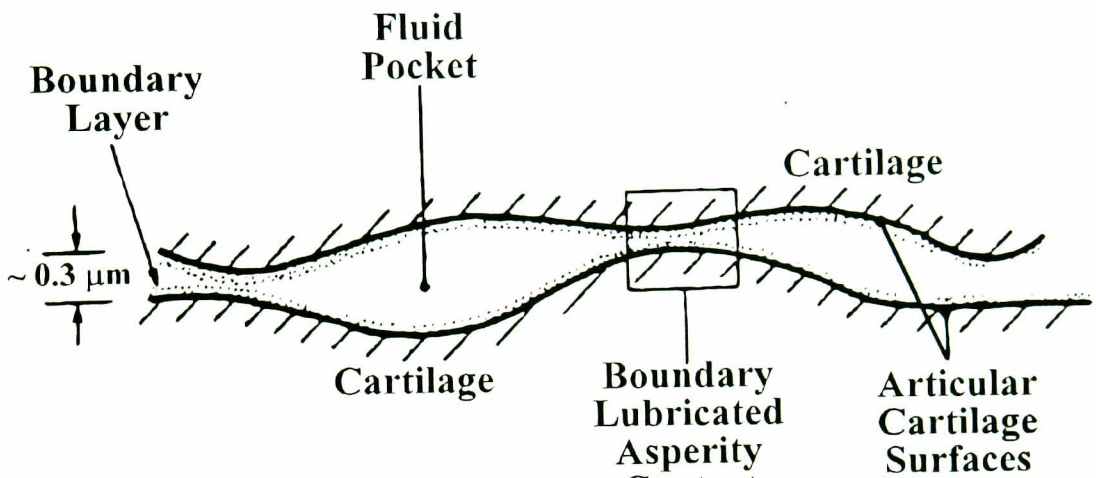

Figure 1-5 In the mixed lubrication regime low friction and minimum wear is believed to be maintained by a boundary layer protecting the asperity contacts and by pressurised fluid pockets helping to minimise further asperity contact. Adapted from Mow et al. (1989).

1.2 Introduction to this Study

Human synovial joints have to withstand complex, varied and often harsh loading regimes, being subjected to both dynamic and static loading cycles under conditions of sliding and rolling; often after considerable periods of stationary loading with little motion. In order to explain the low friction and minimum wear of diarthrodial joints, throughout most people's lifetime, it is very probable that more than one lubrication mode is in operation (Dowson, 1967). It appears generally accepted that some combination of fluid-film and boundary/mixed lubrication must coexist (Mow and Soslowsky, 1991). Although not necessarily working simultaneously together, the operation of the various lubrication mechanisms are dependent upon the loads and loading time, kinematics, anatomy, and chemical and physical properties of the tissues and fluids within a specific joint at any given moment.

At the start of this project it was clear that a detailed understanding of how synovial joints retained their low friction and wear characteristics under conditions of breakdown of fluid film lubrication was lacking. Furthermore, the few *in vitro* studies that have attempted to address this issue were far removed from physiological conditions (Reimann et al., 1975, 1976; Davis et al., 1978, 1979; Hills, 1989, 1990; Jay, 1992; Williams et al., 1993), with their synthetic/synthetic contacts, usually rubber on glass, appearing inappropriate for investigating boundary and mixed lubrication in diathroses. More generally, the variability in the applied methodology was also wide. Different studies have used varying speeds and loads; reciprocating or continuous motion; differing contacts (counterface roughness often being disregarded), lubricants, and type of friction apparatus; as well as varying conditions of enzyme degradation, specimen storage and preparation. Results, therefore, cannot be correlated and useful comparisons and conclusions from data of this kind can only be made within a carefully controlled study (Clift et al., 1989). Another important factor to consider is that all these studies suggest that they were measuring friction under conditions of full fluid film breakdown (mixed or boundary lubrication regime) without having

conducted any prior tribological analyses to confirm this. Furthermore only one or two of the boundary lubrication studies seem to have objectively shown the lubricating advantages of any of the boundary lubricants mentioned in section 1.1, as compared to a control lubricant such as Ringer's solution for example.

1.2.1 Hypothesis

Under conditions of direct cartilage-cartilage contact in synovial joints it is proposed that two lubrication mechanisms are in operation. Firstly, some form of boundary lubrication attributed to the synovial fluid, and secondly a lubrication mechanism related to the high water content, biphasic articular cartilage. By careful control of the tribological conditions, and in particular the choice of lubricant (Ringer's solution / synovial fluid) and loading regime (controlling the water content of the cartilage) the importance of each mechanism in reducing friction can be determined.

The purpose of this study was:

To design and identify experimental conditions when boundary or mixed lubrication was prevalent in cartilage/metal and cartilage/cartilage contacts in order to then analyse the effects of the tribological variables upon the coefficient of friction. Both Ringer's solution and synovial fluid were used throughout[2] as the lubricants in order to assess whether or not the constituents of the synovial fluid, not present in the Ringer's solution, served to reduce friction levels.

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[2] Except for the cyclic loading friction tests - Ringer's solution only.

1.2.2 Summary of Friction Experiments & Surface Analysis Work Conducted

The key variables for the friction tests were load duration and loading history of the cartilage specimens, choose of lubricant, contact configuration and loading regime (i.e. stationary, reciprocating or cyclic). These variables are described below, in addition to surface analysis work which was undertaken to characterise the cartilage surface.

- Stationary loading time - this was the amount of time that specimens were subject to load, prior to sliding and measuring friction. By varying the stationary loading times the water content and fluid flow biphasic properties of the cartilage were changed due to exudation while under load. This was found to have dramatic influences upon the coefficient of friction values. The effects of wear did not have to be considered due to the lack of continuous motion in the test procedure.
- The effect of stationary loading upon cartilage friction coefficients was investigated using the following contact configurations.
  i. 3 mm and 9 mm diameter cartilage plugs sliding against a metal counterface (Chapter 4).
  ii. Metal plugs sliding against cartilage counterfaces (Chapter 5). This was a reversal of the contact configuration presented in Chapter 4, being cartilage plugs sliding against a metal counterface.
  iii. 9 mm diameter cartilage plugs sliding against cartilage counterfaces (Chapter 6).

  For the contact configurations using a metal counterface both *steady state* (during sliding) and *start-up* (at the instigation of sliding) coefficients of friction could be recorded. Steady state μ values for the cartilage counterfaces were however found to be unreliable due to the inherent unevenness of the relatively large synovial joint cartilage surface areas required to be used as cartilage counterfaces.
- The 9 mm diameter cartilage plug on metal counterface configuration was examined further under constant load, reciprocating motion conditions (Chapter 7). The effects of continuous reciprocating motion were investigated using the same friction apparatus as for the stationary loading tests and thereby

allowing comparison of the two data sets to be undertaken. Hence the effects of wear and cartilage exudation upon the coefficient of friction were analysed both in isolation and simultaneously.

- Tests were also carried out whereby continuous cycles of varying short duration loading and varying short duration load removal were applied to cartilage specimens while the friction was constantly monitored (Chapter 8). These cyclic loading friction tests were conducted using the 9 mm diameter cartilage plug on metal counterface configuration, with only Ringer's solution as the lubricant. These, more physiological, loading regimes provided a good illustration of the ranges of load-on phase / load-off phase cycles that cartilage requires in order to maintain low friction in conditions of full fluid film breakdown.
- An interesting experimental variation was also briefly investigated; namely a water swollen polymer (hydrogel) material, possessing biphasic properties, was used as a counterface upon which cartilage specimens were tested. This cartilage on hydrogel contact configuration allowed both start-up and steady state friction to be recorded in order to further understand the tribology of biphasic on biphasic (i.e. cartilage/hydrogel) contacts.
- In addition, surface analysis of articular cartilage was undertaken. The techniques utilised for this investigation were stylus and laser profilometry and scanning, environmental scanning and transmission electron microscopy (SEM, ESEM & TEM). The main aim of this particular work was to reliably quantify the surface roughness of articular cartilage and characterise the features of the boundary layer, if indeed it could be identified. A full account of the findings, coupled with a review of other similar studies, is rendered in Chapter 3, prior to the presentation of the cartilage friction experiments (Chapters 4-8).
- In the Materials and Methods (Chapter 2) collection, handling and storage procedures for the cartilage and synovial fluid samples are explained, along with details of the metal plugs and counterfaces. The arrangement of the friction apparatus and the validation of recording reliable data is put forward. The basic working principles of the stylus and laser profilometry, and electron microscopy techniques are also deliberated.

Aims and Objectives

While much work has been directed at understanding synovial joint lubrication during such actions as when walking or running it is estimated that for approximately 90% of the time synovial joints and articular cartilage undergo very little motion. These periods are commonly associated with significant amounts of prolonged loading. During these periods the EHL lubricating films produced by entraining or squeeze film action are too thin to separate the opposing articular cartilage surfaces and direct contact occurs within a mixed or boundary regime. Relatively little is known about the lubrication mechanisms which persist under these conditions to prevent damage to the articular cartilage. It has been previously proposed that both boundary macromolecules from synovial fluid and the biphasic nature and high water content of the cartilage help to sustain low friction. However there are conflicting views of the relative contributions and importance of each mechanism.

The specific objective of this study was to enable an enhanced comprehension of how synovial joints perform under conditions of full fluid film breakdown by isolating and clarifying the lubrication mechanism(s) which occur while mixed and boundary regimes predominated for articular cartilage specimens. To achieve this objective, coefficient of friction measurements were to be undertaken under a variety of tribological conditions and contact configurations.

2. Materials and Methods

2.1 Introduction

This chapters describes the collection, production, handling and storage procedures of 3 mm diameter disc cartilage on bone plugs (3 mm cartilage plugs), 9 mm diameter cartilage on bone plugs (9 mm cartilage plugs), cartilage counterfaces and synovial fluid, (section 2.2). All these materials were required for the friction studies. Cartilage plugs were also required for surface analysis studies. The specifications of metal plugs, metal counterfaces and hydrogel counterfaces, also used in the friction studies, are detailed as well. In section 2.3 the basic principles of the stylus and laser profilometry, and electron microscopy techniques, used to examine the surface of articular cartilage, are explained. The adopted methods for recording coefficient of friction values for all the contact configurations studied is presented in section 2.4. In section 2.4 details of the friction rig are provided along with a validation of the test protocol in terms of establishment of full fluid film breakdown and consistency of results.

Further details about the experiments carried out, such as exact number of specimens used and the friction sampling times, has been provided in the particular chapter concerned.

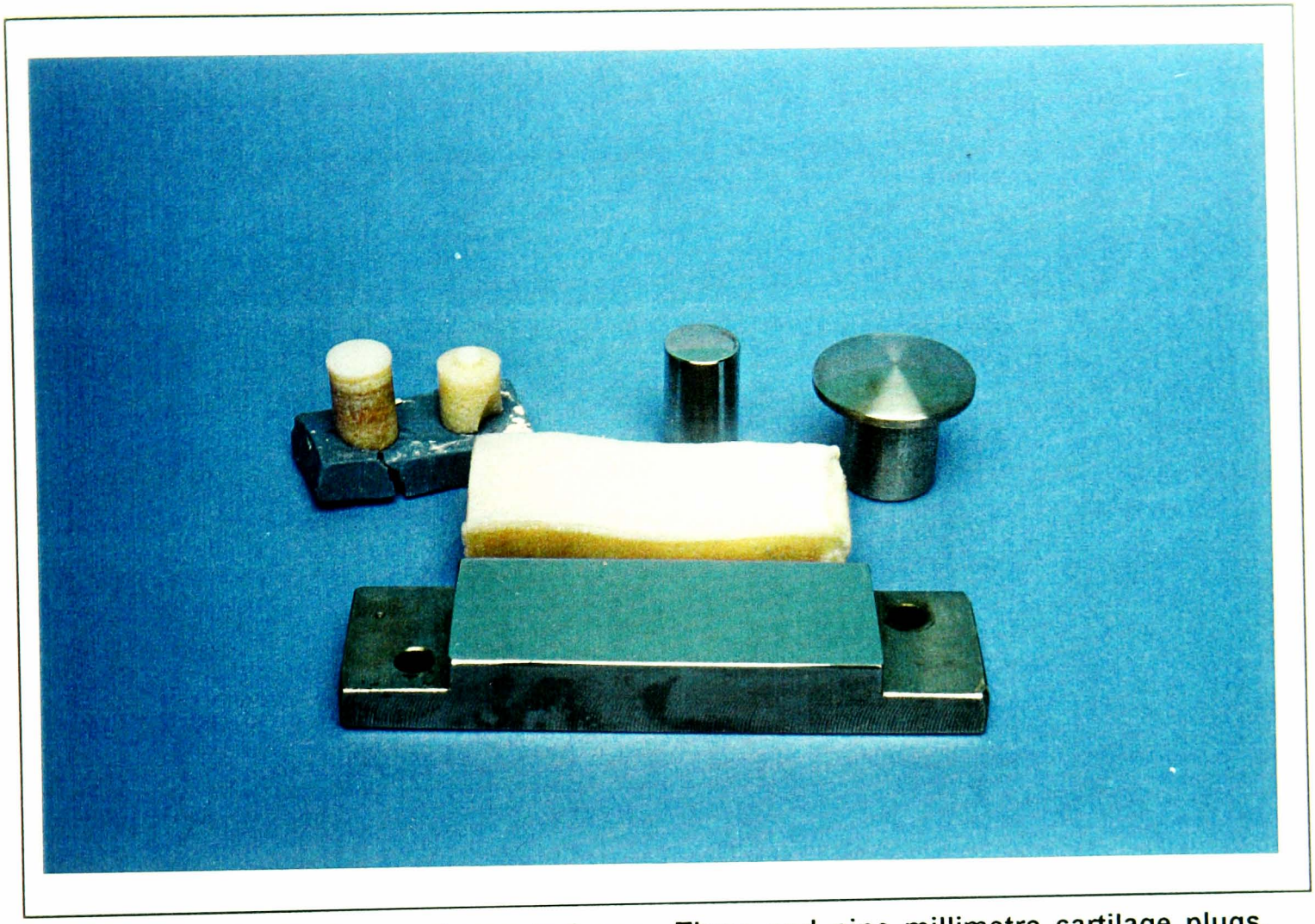

Figure 2-1 Sliders and counterfaces. Three and nine millimetre cartilage plugs, metal plugs and cartilage and metal counterfaces used in this study.

2.2 Articular Cartilage and Synovial Fluid Samples

Articular cartilage was collected from the distal end of bovine femurs. The cartilage samples were collected from an abattoir as the bovine carcasses were 'deboned' in a cold room, within 36 hours following commercial slaughter. Prior to 'deboning' the carcasses were kept in a cold storage area with the knee (cartilage specimens) and ankle joints (synovial fluid specimens) skinned but still fully intact. The collected cartilage samples were obtained from skeletally mature animals and were of a healthy and normal appearance. The distal end of the femurs were sectioned using a hack saw, placed in large plastic specimen bags and then soaked fully by spraying with Ringer's solution. The bags were kept in a cooler box for approximately 1-2 hours until they were removed and placed into laboratory fridges. The bovine femurs were often kept in the fridges overnight until cartilage specimen preparation began as outlined below.

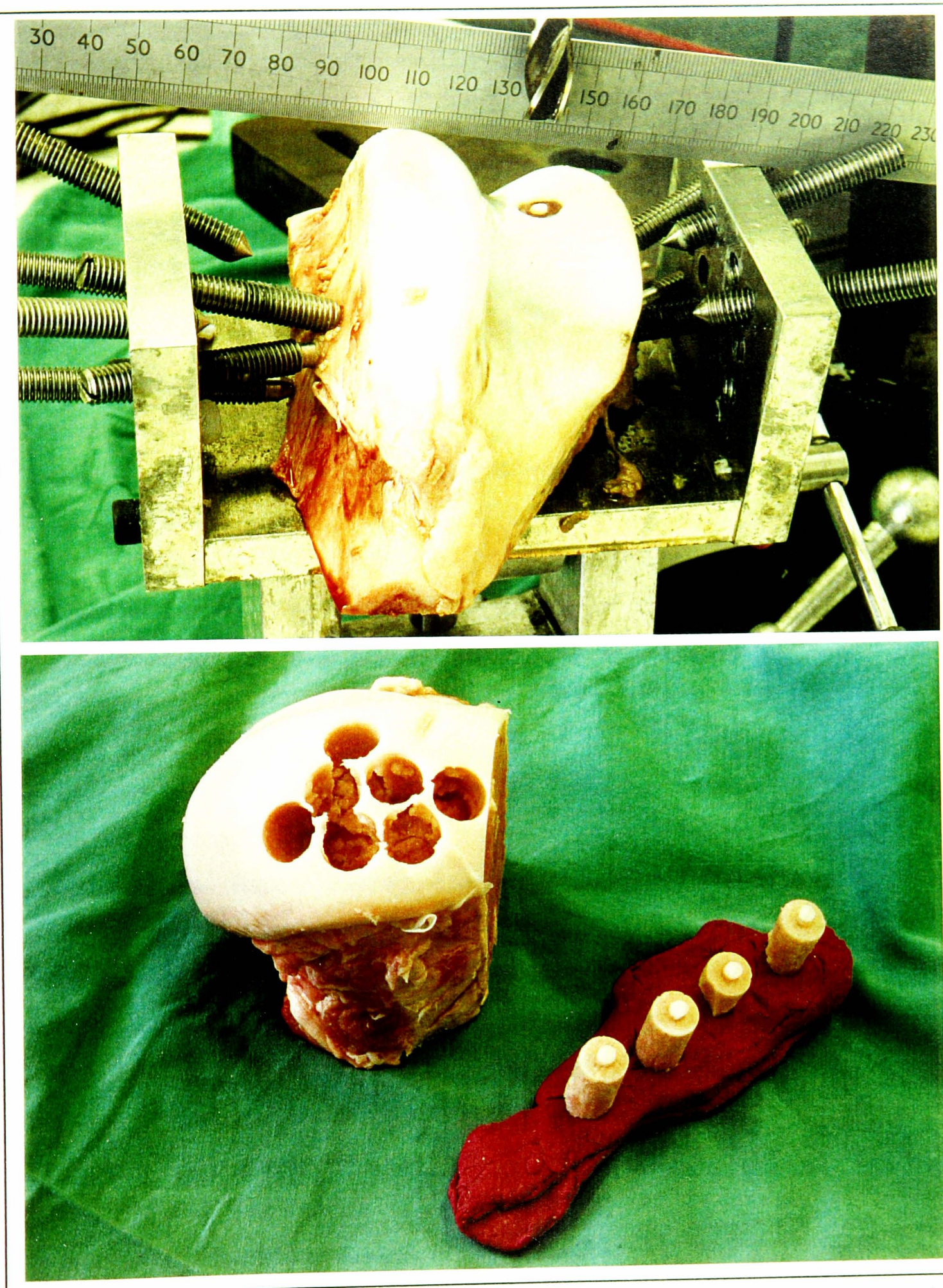

Figure 2-2 The top photograph shows the parapatellar portion of the distal end of an adult bovine femur. The cartilage disc drill is shown, having just produced a 3 mm diameter disc of cartilage on the joint surface. A ruler marked in millimetres was provided in the background to denote scale. The bottom photograph displays 3 mm diameter cartilage disc specimens extracted from a bovine femoral condyle which is also shown.

The technique used to produce the cartilage plugs was designed in a manner similar to that adopted by other workers (Lipshitz and Glimcher, 1974). By the use of this technique the following necessary requirements were achieved.

I. The surface of the articular cartilage was both flat (or the radius of curvature was centrally located) and perpendicular to the cylindrical shaft of the underlying subchondral and cancellous bone.

II. Full hydration of cartilage surfaces maintained.

III. The cartilage tissue of the extracted plugs remained untouched and undisrupted.

IV. Production of reproducible cartilage contact faces, both for the 3 and 9 mm cartilage plugs.

The distal end of the femurs were sectioned into three distinct components, using a band saw, namely the lateral femoral condyle, medial femoral condyle and the femoropatella groove. These sections were then of a suitable size to be placed within a screw vice, (Figure 2-2), as described below. A special rig was developed to carefully extract the specimens from the femoral components. This comprised of a universal drilling machine with customised drill bits to allow ease of production and eliminate damage of the extracted cartilage plugs. During drilling operations, the femoral components had to be secured very firmly in order to maintain a permanent, lasting perpendicular alignment between the drill bit and cartilage surface (for each cartilage plug). This was achieved by a specially designed screw vice, Figure 2-2, which consisted of a pivoted ball joint clamp for realigning so that several cartilage specimens could be taken per femoral component. The screw vice was, therefore, fully adjustable in three dimensions such that drilling could always be undertaken normal to the joint surface. Throughout these operations the cartilage was kept hydrated by frequent spraying application of Ringer's solution; at least every couple of minutes was deemed necessary (Speer et al., 1990; Kirk et al., 1993b). The Ringer's solution was also used as a coolant fluid during drilling. Excessive heat build up, which could have

degraded the cartilage plugs, was further eliminated by using slow revolution drilling speeds and periodic removal of the rotating drill bit.

Once the joint surface had been fixed in a position perpendicular to the cutting tool direction a hole punch of 3 mm diameter was attached to the chuck of the drilling rig. This was used to cut 3 mm diameter rings into the articular cartilage. While maintaining correct alignment, the hole punch was exchanged by a cartilage disc drill. This drill cut 10 mm diameter rings, through the full thickness of the cartilage, without touching the initial 3 mm diameter cartilage discs, Figure 2-2. The initial use of the hole punch ensured that the cartilage disc drill did not transfer any shear or propagate tears across the surface and bulk of the 3 mm cartilage discs. A 9 mm surgical bone plug drill bit, having been machined down to fit the chuck, was used to cut a bone plug with the 3 mm cartilage disc positioned in the centre. A complimentary surgical bone plug extractor was then able to carefully remove the specimen from the joint without contacting or disturbing the cartilage disc surface. The 9 millimetre cartilage plugs were manufactured in much the same way, only requiring use of the 9 mm surgical bone plug drill bit followed by the bone plug extractor.

In synovial joints the cartilage surfaces tend to have an associated curvature which poses problems in the production of reproducible cartilage plugs for friction testing. The smaller 3 mm cartilage plugs produced flat, reproducible contact areas which remained approximately constant during the period of loading. Also, higher contact stresses for a given load could be generated compared to the 9 mm cartilage plugs. The friction rig used had a load limit of 50 N. The 3 mm cartilage disc plugs allowed the contact area and stress to be controlled more precisely, as they better approximated to a flat contact. However the sectioned areas of the cartilage matrix, surrounding the 3 mm cartilage disc, would be directly exposed to the applied loads which may affect the exudation and fluid flow properties of the cartilage tissue under load (especially for such a relatively small volume of cartilage). The advantage of a 9 mm cartilage plug was that the sectioned region of the cartilage matrix was generally at a distance from the loading region and the larger volume of the cartilage made this concern less

significant anyway. The 9 mm cartilage plug was also considered to be more stable under constant load reciprocating motion conditions.

The subchondral bone and underlying cancellous bone was retained, in the form of 9 millimetre diameter cylindrical bone plugs approximately 15 mm in length with a 1-2 mm thick cartilage layer, Figure 2-2. This retention of the underlying bone facilitated handling and mounting of the cartilage for the friction tests, and ensured cartilage integrity (Ghadially et al., 1982).

Cartilage counterfaces were sectioned from bovine humeral and femoral heads using a band saw, in such a manner as to provide as even a cartilage surface area as possible. The dimensions of approximately 30 x 70 mm^2 and 10-20 mm thick (largely consisting of the underlying subchondral and cancellous bone) were similar to those of the metal counterfaces, Figure 2-1.

Synovial fluid was collected from bovine ankle joints within a period of 36 hours following commercial slaughter. Synovial fluid was exposed by sectioning across intact, cleaned and skinned, ankle joints and extracted by a syringe through a 2 mm diameter flexible plastic tube. This technique was found to provide easy and rapid collection; contamination by blood and hair etc. was also eliminated. The ankle joints provided a plentiful and fresh supply of synovial fluid. The fluid was filtered through a 500 μm sieve to remove cartilage debris.

Articular cartilage and synovial fluid samples were stored frozen at -20°C. The cartilage specimens were kept frozen in Ringer's solution. Ringer's solution is widely used as an irrigation fluid during arthroscopy (Yang et al., 1993; Jurvelin et al., 1994). Ringer's solution is, therefore, regarded as a good medium for storage of articular cartilage and has also often been used as a lubricant during friction testing (Little et al., 1969); as well as an immersion fluid in many other biomechanical studies, (Jurvelin et al., 1994). Prior to the tests the cartilage specimens were defrosted and immersed in the appropriate lubricant for at least 1 hour before use. Additional tests were also carried out on fresh articular cartilage

and synovial fluid samples that had not been frozen, in order to assess the effect of freezing the specimens. By comparing results from previously frozen and fresh samples, the freezing was found not to have an influence upon friction readings.

2.3 Surface Analysis Techniques

In this section the techniques used to characterise the surface of articular cartilage are described. The results of this study, as well as an appraisal of each of these techniques in analysing the cartilage surface, are presented in Chapter 3.

It was necessary for the stylus profilometry measurements to be conducted by trained technical staff. The data files of the line profiles were saved to an on-line computer and consequently processed by the author. All other analytical equipment mentioned in this section was solely operated by the author, following initial instruction where necessary.

2.3.1 Stylus Profilometry

This is a very familiar tool used widely by engineers and tribologists to quantitatively assess the surface texture of engineering materials. It involves a stylus traversing across the surface of a material under investigation. During the traverse a two dimensional trace or line profile of the surface is recorded, based upon vertical deflections of the stylus. Highly accurate reproduction of the specimen surfaces is possible with vertical resolution limits of ~0.01-0.02 μm and a fine spherical diamond tipped stylus of ~1.5 μm radius. Modern stylus profilometers (as employed in this study) are accompanied with microcomputers which can store and process the data from the line profiles to produce quantitative expressions of surface texture.

The stylus profilometer used in this study was a Form Talysurf Series 6 model (Rank Taylor Hobson). Important model specifications are provided below:

Straightness of Traverse	Within 2 parallel lines spaced 0.5 μm over 120 mm.
Measuring traverse speed	0.5 mm/s ±5%.
Vertical Resolution	20 nm at 12 mm range.
Standard Conisphere Diamond Tip	
Tip Radius	1.5-2.5 μm.

Applied Force of Stylus Tip 0.7-1 mN.

2.3.2 Laser Profilometry

Laser profilometry is a relatively new analytical technique used to provide quantitative information of surface texture in a similar format to that of stylus profilometry. To date no reference to its use on articular cartilage can be found in the literature. One key advantage of this system is that contact with the surface under examination does not occur, i.e. it is a non-contacting profilometry technique. The software package provided with the laser profilometer, (Microfocus Measurement System, UBM), could also display 3 dimensional quantitative images of the surface, using data collated from surface areas scans. The rudimentary working principle of the Microfocus Measurement System UBM laser profilometer is detailed below.

Infrared light from a semiconductor laser is focused to a spot (1 μm in diameter) by an objective lens. The light reflected by the object surface is directed by a beam splitter, through a prism, and is imaged as a pair of spots onto an arrangement of photodiodes. This is done in such a manner that when the objective lens is exactly its focal distance from the surface both diodes are illuminated equally. If the distance between object surface and objective lens then changes by some amount, the imaged focus point is shifted and the illumination of the photodiodes becomes unequal. This unequal illumination generates a focus error signal by means of a differential amplifier. To ensure exact measurements both the spot diameter and its light distribution across this diameter must be kept constant. With the UBM sensors, this is achieved by moving the objective lens as the lens/surface distance changes. A control circuit monitors the focus error signal and moves the objective lens accordingly. It is from these highly precise vertical movements of the objective lens, maintaining focus as the lens traverses across the object surface, that accurate line profiles and area scans of the surface are recorded.

Important specifications are provided below:

Vertical Measurement Ranges ± 50 μm & ± 500 μm

Corresponding Vertical Resolution	10 nm & 100 nm
Accuracy	dependent on surface structure
Light spot diameter	1 μm
Minimum surface reflection	1%

All the laser profilometry results were recorded within the ± 50 μm measurement range, except for the relatively large 1x1 mm^2 area scans and 6 mm long line profiles when the ± 500 μm measurement range was required.

2.3.3 Electron Microscopy

Both scanning and transmission electron microscopy, (SEM & TEM), have over the last 25-30 years become an indispensable tool in many areas of science and engineering research. Their operating principles are widely documented and understood, and will not therefore be discussed in depth here. Environmental scanning electron microscopy (ESEM) is a relatively new technique, developed and refined over the last ten years. The difference between the ESEM and conventional SEM (CSEM) is described below.

All SEM's consist of an electron gun, that creates a beam of electrons; a sample chamber, where the electron beam interacts with the sample; detectors, that monitor a variety of signals resulting from the beam-sample interaction; and a viewing system, that constructs an image from the signal. CSEM's are designed such that both the sample chamber and column share a common high vacuum environment. The column requires a high vacuum beam in order to generate and focus the electron beam. The sample chamber requires a high vacuum to permit the use of secondary electron detectors. CSEM's, therefore, require that samples be *vacuum tolerant* (not altered by high vacuum environment), *vacuum friendly* (i.e. do not degrade vacuum enough to damage detector or electron gun; no deposits on electron gun apertures degrading imaging performance; and no remnant contamination of sample chamber interfering with subsequent observations) and *electrically conductive* (sufficient charge flow to eliminate charging artefacts).

There are two key technological aspects that differentiate the ESEM from all other SEM's. The first is its multiple aperture, graduated vacuum system. This system maintains a high vacuum in most of the electron column while permitting relatively high pressures in the sample chamber. The gas in the sample chamber does scatter some electrons from the beam. However, within a scattering regime known as Partial Scattering, corresponding to a certain range of operating conditions (pressure, gas path length, temperature, accelerating voltage, and gas type), beam scattering does not degrade image resolution. The second key feature is the Environmental or Gaseous Secondary Electron Detector, using gas ionisation to detect and amplify the secondary electron signal. Gas ionisation also suppresses charging artefacts on insulating samples.

Analysis of biological materials with CSEM requires meticulous specimen fixing, dehydration, drying and gold coating in order to become vacuum tolerant, vacuum friendly and electrically conductive. For articular cartilage constituted of 60-80% water there are obvious concerns regarding the authenticity of CSEM images following dehydration of the cartilage. The use of an ESEM, requiring no specimen preparation whatsoever, other than being of a suitable size to fit into the specimen chamber, has obvious advantages. On the other hand the very fact that cartilage does have a very high water content may indeed 'fog' the image or make features appear indistinctive and difficult to focus upon.

The Electroscan ESEM model used in this project was located in the Biological Sciences Electron Microscopy Unit at the University of Manchester.

2.4 Friction Tests

The advantages and disadvantages of joint lubrication studies involving small cartilage specimens and whole joints have been discussed extensively in Chapter 1. In this study friction was determined using extracted cartilage plugs. This was done in order to have clear control upon the loading time and subsequent fluid content of the loaded cartilage under investigation, to permit direct measurements of friction to be recorded and to allow a better understanding of the tribological conditions of the contact.

It should be noted that further details relating to the friction apparatus, mentioned in the preceding pages, can be found in the appendix.

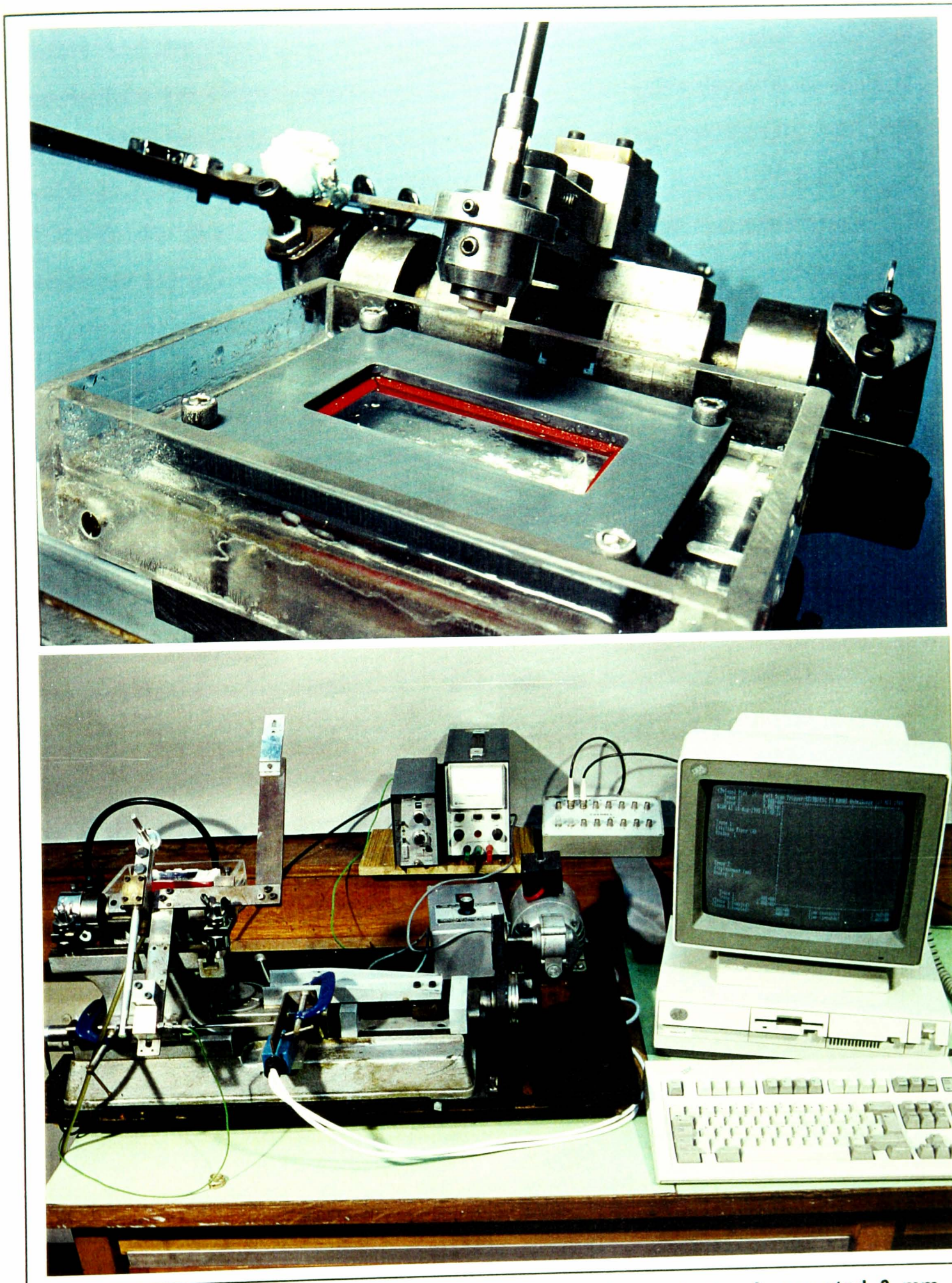

Figure 2-3 Photographs of the friction apparatus set-up. A mounted 3 mm cartilage plug is shown, poised above a red bordered metal counterface (top). The metal counterface is housed within a grey plastic lubricant bath which required only a small amount of lubricant to cover the surface of the counterface (~5 ml). The transducers, motor and on-line data acquisition computer, alongside the friction rig, are shown in the bottom photograph.

Friction was measured on a sliding-friction machine (Figure 2-3), using both synovial fluid and Ringer's solution as a lubricant. A flat metal (surgical grade 316L stainless steel) counterface, with a surface roughness of Ra≅0.02 μm, was fixed level in a bath of lubricant which was driven by a motor to slide a distance of 50 mm in one direction at a constant speed, or alternatively to continuously reciprocate ±25 mm about a mean position. The cartilage specimen was loaded onto the counterface while secured to one end of a balanced loading arm, pivoted at a fulcrum on an air bearing. The other end of the loading arm was restrained by a piezoelectric force transducer and thereby served to record the frictional force (N). Although piezoelectric force transducers are essentially used for measuring instantaneous dynamic loads, during these experiments and the calibration of the transducer, friction forces were measured for periods of approximately 10 seconds without any adversity to the data obtained.[1] Displacement of the metal counterface was monitored by a linear variable differential transducer. The signals from both transducers were transferred to a computer via an analogue-to-digital converter. The sliding speed and the coefficient of friction, μ, were recorded using the Unkelscope data acquisition software package. The software used previously determined calibration factors to convert the transducer signals into values of friction force and sliding distance, which were then plotted against time. The coefficient of friction readings had a precision of ±0.002. The calibration for friction was conducted by attaching a series of loads, from 0 to 1000 grammes (up to equivalent frictional forces of 10 N), to the specimen end of the loading arm, such as to mimic friction forces during testing. The Unkelscope software package, using an analogue-to-digital converter, recorded the voltages for each applied load and a linear least squares fitting procedure was used for the calibration graph for the transducer and data acquisition system. Prior to using the friction apparatus at the start of a day's testing a known load was always attached to the specimen end of the loading arm. The resultant 'friction force'

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[1] Any loss of charge for the piezoelectric cell was insignificant in relation to the general noise of the friction force traces, equivalent to μ variance of ±0.002.

was sampled in order to check that the friction rig and data acquisition procedures were fully operative.

With the specimen mounted the loading arm was balanced such that there was no resultant load acting at the contact between the cartilage specimen and metal counterface. The point of contact was always fixed such that the cylindrical shaft of the cartilage plug was perpendicular to the metal counterface thus ensuring that the opposing cartilage and metal surfaces were parallel to one another. A normal load of 30 N was applied to the specimens which produced contact stresses of 4 MNm^{-2} for the 3 mm cartilage plugs (reasonably assuming a flat 3 mm diameter cartilage surface) and 0.5-~2 MNm^{-2} for the 9 mm cartilage plugs (corresponding to increasing contact diameters with loading time from ~ 4 mm to 8 mm in diameter for the 9 mm cartilage plugs; the compliant 1-2 mm thick cartilage layers having given radii of curvature). These loading levels were considered to be appropriate physiological load ranges.

The period of loading was a main friction test variable. A 5 second loading period was the shortest time that could be practically adopted. At this time the predicted squeeze film thickness for both lubricants was less than 0.1 μm for both the 3 mm and 9 mm cartilage on metal counterface contact configurations. This predicted film thickness is approximately one tenth the roughness of the cartilage surface indicating that the contacts had entered the mixed lubrication regime. A low sliding speed of 4 mm/s was adopted throughout to ensure the contacts remained within a mixed or boundary lubrication regime (Section 2.4.3, page 60).

2.4.1.1 Metal Plugs, Metal Counterfaces and Hydrogel Counterface

The polished metal (stainless steel) plugs used for the metal plug on cartilage counterface contact configuration were referred to as metal plugs 1 and 2. Both metal plugs 1 and 2 had surface roughnesses, Ra values, of less than 0.15 μm, for a 0.25 mm sampling interval. Metal plug 1 had a radius of curvature of 66 mm. Metal plug 2 had a radius of curvature of 137 mm. The respective diameters of metal plugs 1 and 2 was 11 and 25 mm. The metal counterfaces were surgical grade 316L stainless steel, with a surface roughness of Ra≅0.02 μm. Their dimensions were approximately 30 x 70 mm^2 and 15 mm in depth, (Figure 2-1, page 38).

In addition to the polished metal and cartilage counterfaces a hydrogel layer, 0.8 mm thick, was also utilised as a counterface in some experiments. The water swollen hydrogel materials are designed to mimic the material properties of cartilage as closely as possible (Corkhill et al., 1990). The hydrogel material used in this study was N-vinyl pyrrolidone methyl methacrylate (PC110). This hydrogel possessed the following material properties; Tensile Elastic Modulus 9 MNm^{-2}; Tensile Strength 6.5 MNm^{-2}; Elongation to Fracture 252 %; and Equilibrium Water Content 43 % (Caravia, 1993).

2.4.1.2 Statistics

For each set of friction test conditions the specimens were generally placed into two groups, according to the chosen lubricant, being either Ringer's solution or synovial fluid. The mean and standard deviation of the friction coefficient was calculated for both of these groups. The statistical significance of the differences in the mean was assessed using the Student's t-Tests on a Microsoft Excel spreadsheet application software package. Statistical significance was tested at the 95% confidence level ($p<0.05$).

2.4.2 Repeatability and Reproducibility of Friction Results

The reliability of data from friction tests using cartilage specimens has in previous studies found to be problematic (Davis et al., 1979; Hills and Butler, 1984). At the outset of this work it was therefore important to establish a sufficient level of consistency in the friction results. The terms *repeatability* and *reproducibility* are often used in this study. Repeatability is defined as the consistency of friction coefficient values for any given specimen. Reproducibility is defined as the consistency of friction coefficient values between different cartilage specimens under identical conditions.

The first series of friction tests were designed to allow the assessment of both repeatability and reproducibility. Friction tests of 3 mm cartilage plugs loaded against a metal counterface were performed on four different groups of 7 specimens. The first two groups had friction measurements recorded after 5 seconds and 2 minutes of stationary loading, using either Ringer's solution or

synovial fluid as the lubricant (Figure 2-4 and Figure 2-5). The second two groups had friction measurements recorded after 5 seconds and 5 minutes of stationary loading, using either Ringer's solution or synovial fluid as the lubricant (Figure 2-6 and Figure 2-7). For all the specimens tested seven readings were taken for each stationary load time evaluated.

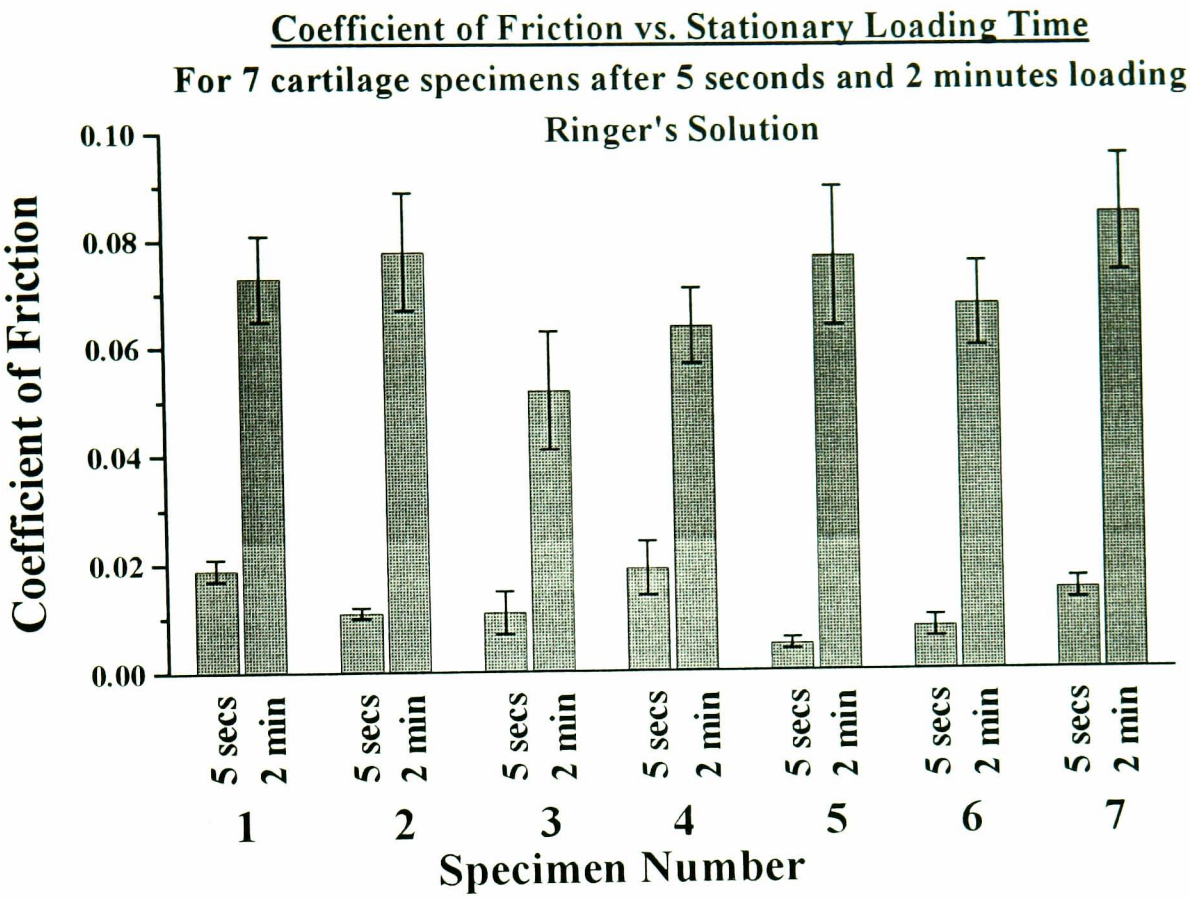

Figure 2-4 Means and standard deviations of 7 readings recorded after 5 seconds and 2 minutes stationary loading times for 7 specimens (labelled 1-7). Ringer's solution was the lubricant.

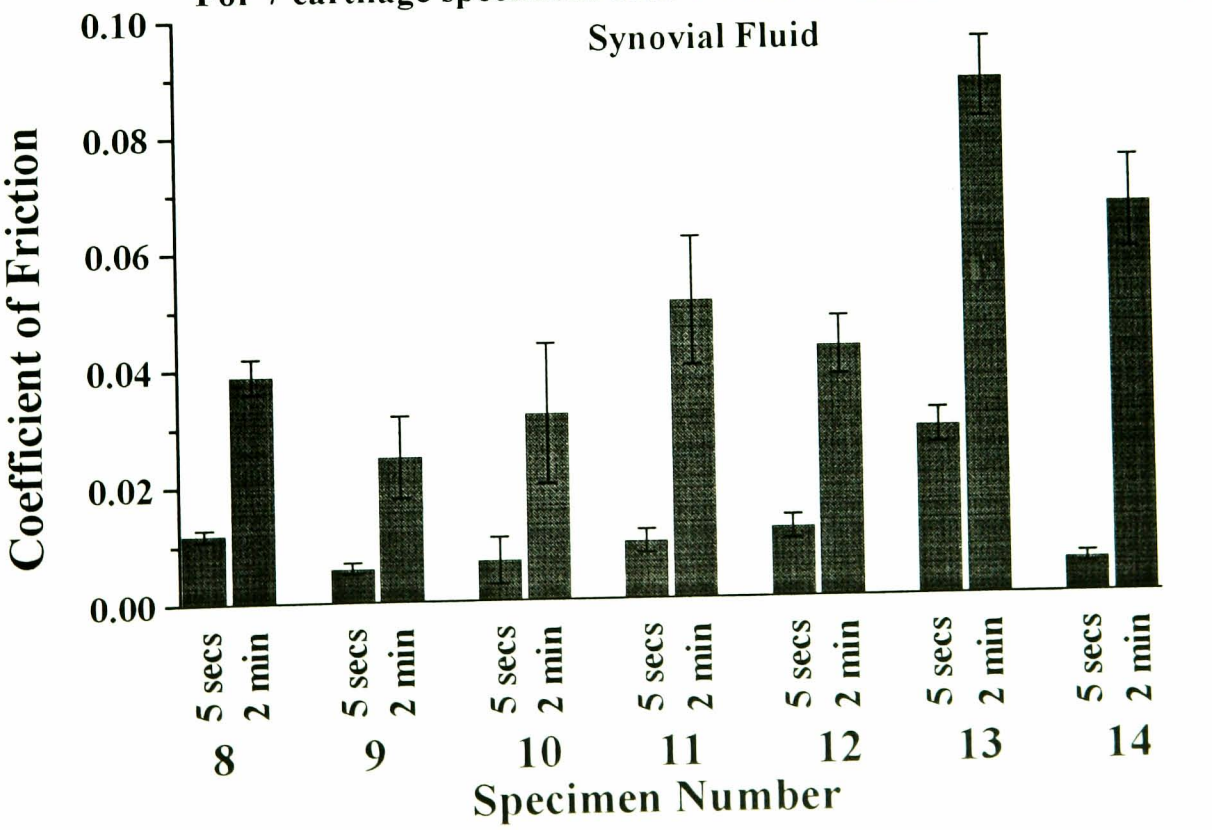

Figure 2-5 Means and standard deviations of 7 readings recorded after 5 seconds and 2 minutes stationary loading times for 7 specimens (labelled 8-14). Synovial fluid was the lubricant.

The friction tests carried out for 5 second/2 minute and 5 second/5 minute stationary loading periods for both Ringer's solution and synovial fluid, Figure 2-4, Figure 2-5, Figure 2-6 and Figure 2-7[2], provided a clear indication of the *repeatability* of the intra specimen friction readings, i.e. for a particular specimen, as well as the inter specimen *reproducibility* of the results, i.e. between the cartilage specimens. Note that the variance in the repeatability of the 5 second friction results can partially be accounted for by the assessed ±0.002 accuracy of the friction rig calibration for friction coefficient.

A quantitative evaluation of both the repeatability and reproducibility was possible by respectively calculating the coefficient of variance (standard deviation/mean) for each of the 7 readings taken per specimen at the evaluated 5 second, 2 minute or 5 minute stationary loading time, Figure 2-8; and then, using the mean of the 7 friction coefficient readings for each specimen as a single data value, calculate standard deviation/mean of the pooled specimen results for each of the groups, as displayed in Table 2-1.

The repeatability of the friction coefficients was very pleasing, with coefficient of variance generally ~0.2, Figure 2-8. The reproducibility of the friction coefficients was predictably higher, Table 2-1. However, aside from the 5 second results, the coefficient of variance for reproducibility was still more than adequate. The relatively high variance in friction results between cartilage specimens recorded after 5 seconds loading can be partially explained by the differing nature of the mixed lubrication regime, introduced after such a short duration of loading, between the various cartilage specimens/metal counterface contacts.

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[2] Note that specimen 13 (Figure 2-5) and specimen M (Figure 2-7) were considered to be outliers and excluded from any statistical analysis.

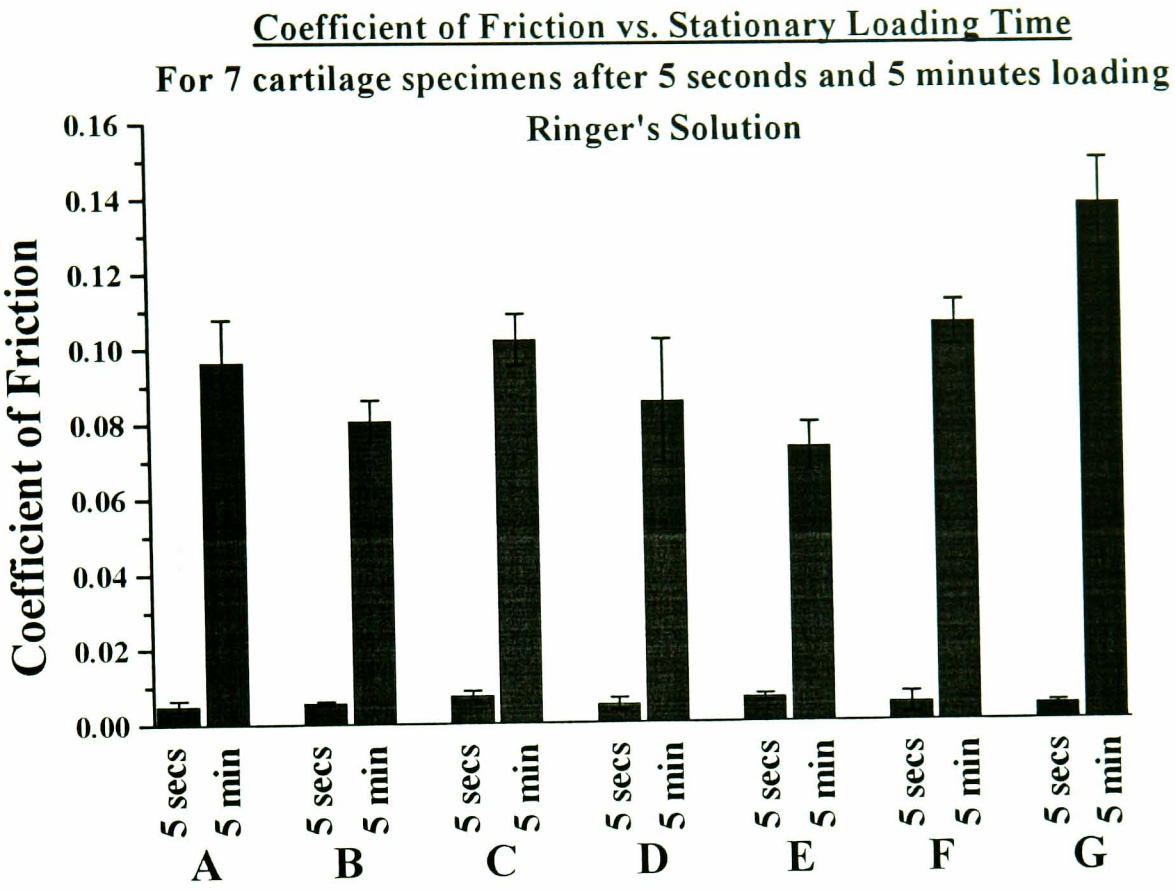


Figure 2-6 Means and standard deviations of 7 readings recorded after 5 seconds and 5 minutes stationary loading times for 7 specimens (labelled A-G). Ringer's solution was the lubricant.

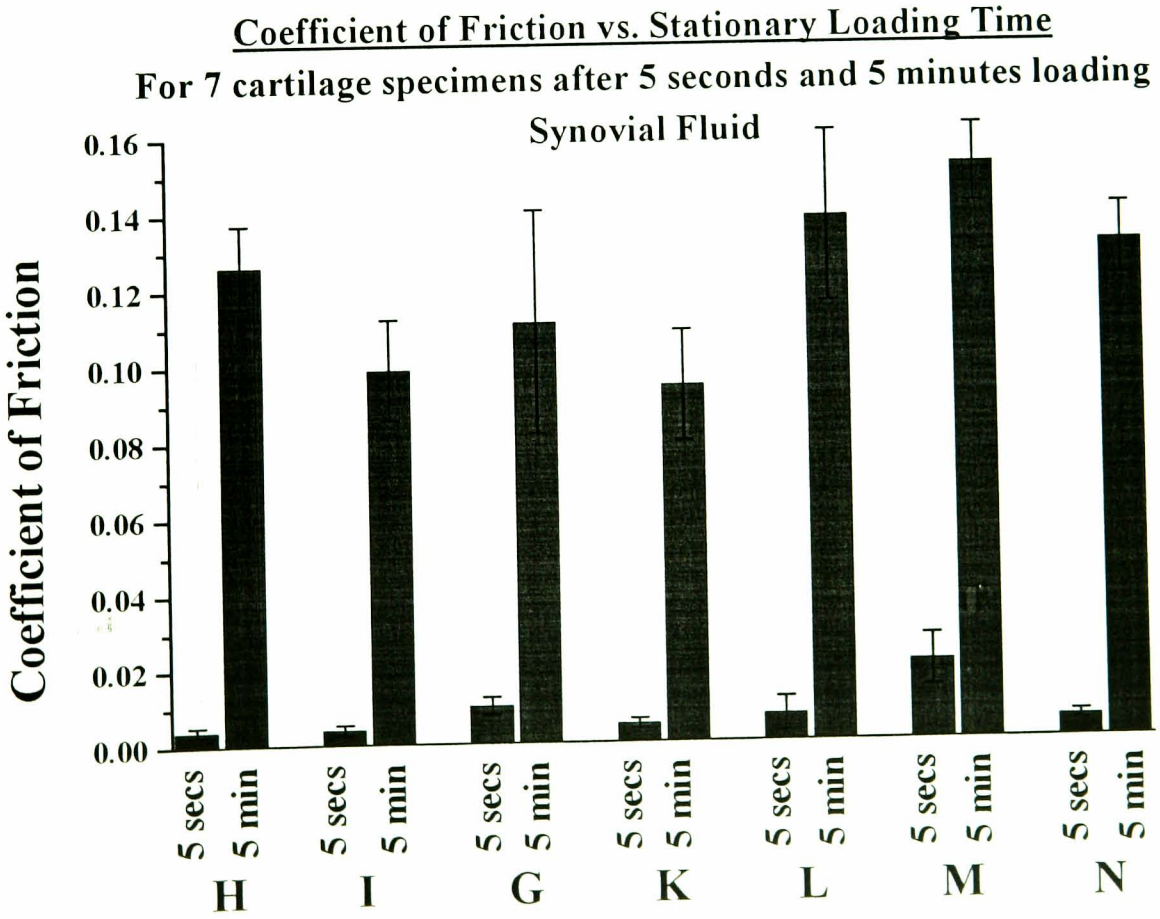
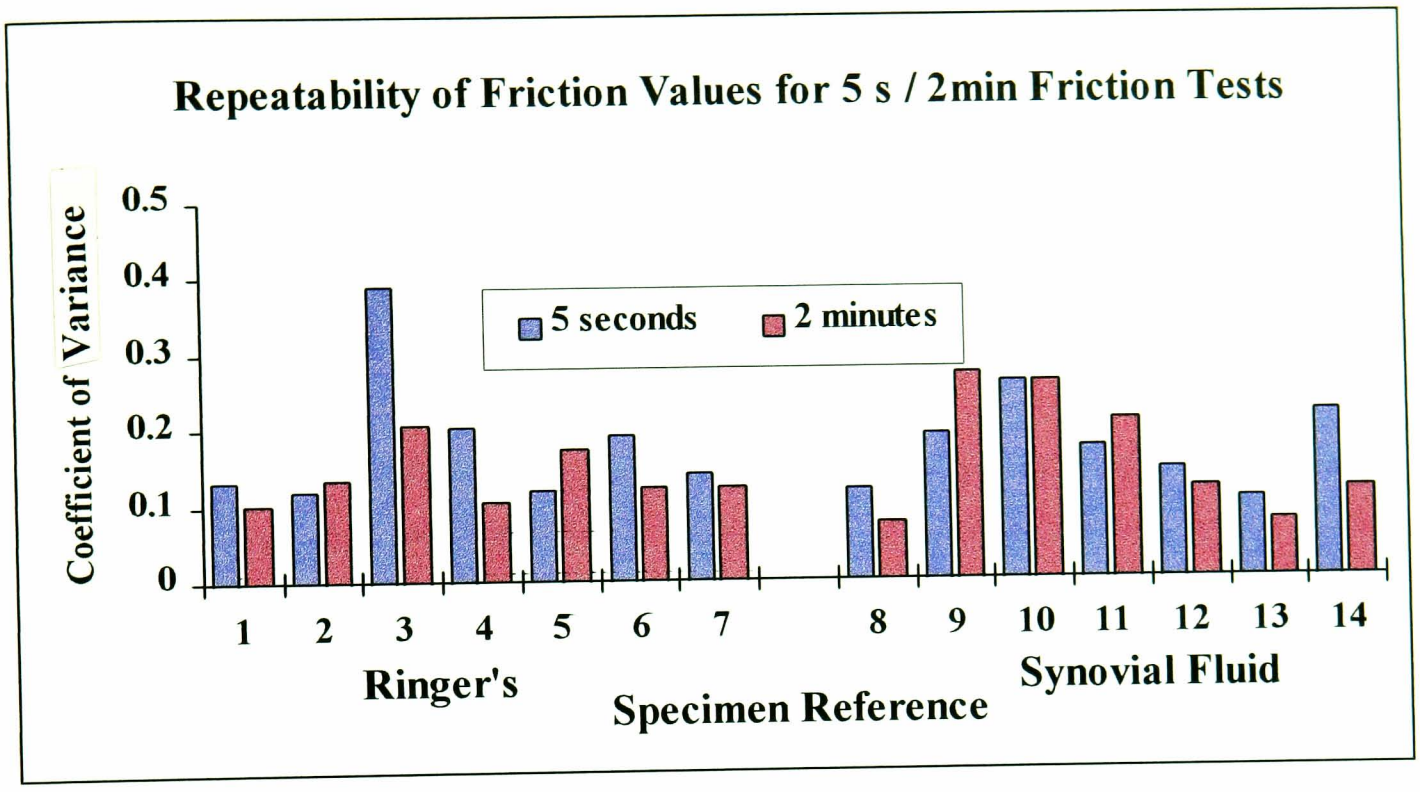


Figure 2-7 Means and standard deviations of 7 readings recorded after 5 seconds and 5 minutes stationary loading times for 7 specimens (labelled H-N). Synovial fluid was the lubricant.



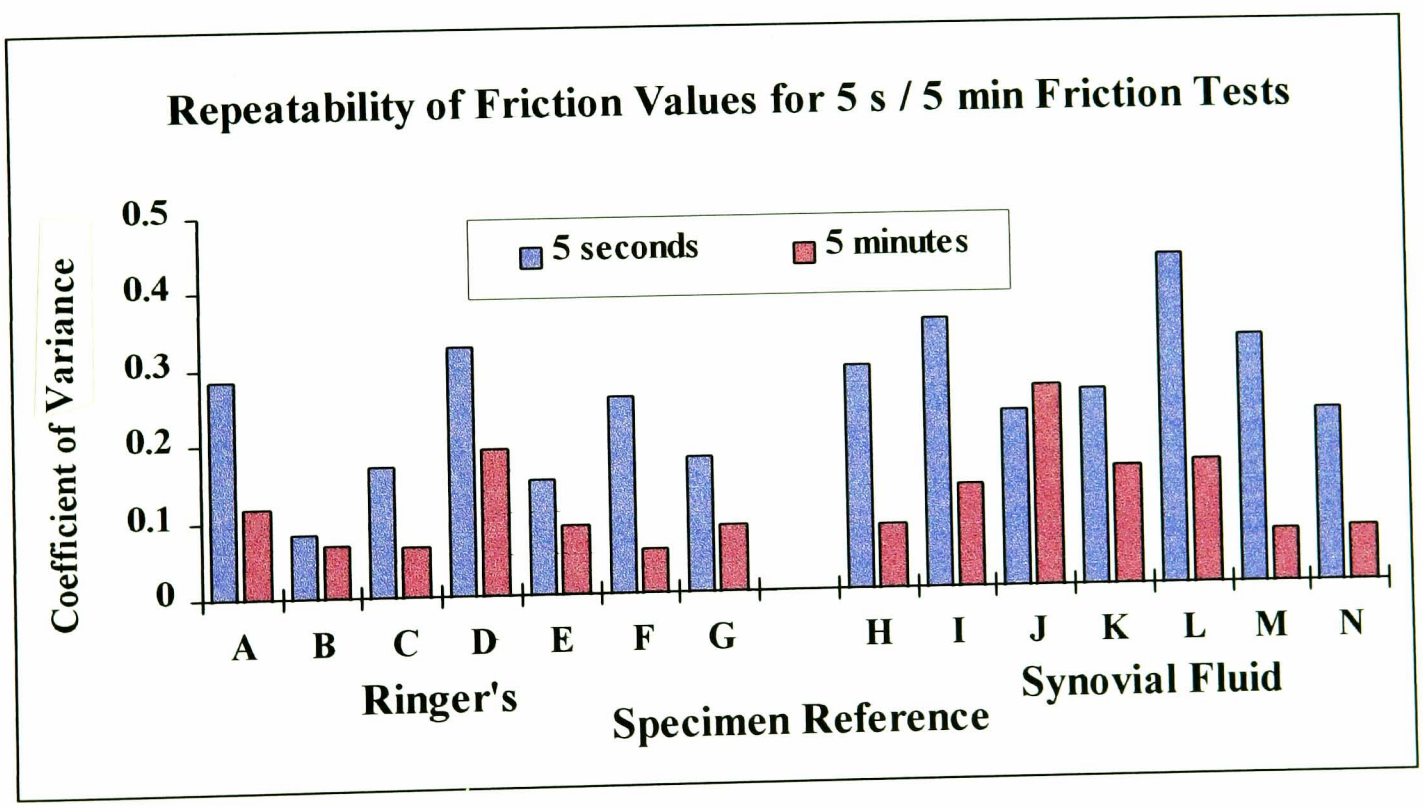


Figure 2-8 ***Repeatability*** **of friction values as demonstrated by the coefficient of variance (standard deviation/mean) for the respective 7 readings taken per specimen at the evaluated 5 second, 2 minute or 5 minute stationary loading time.**

Reproducibility Coefficients of Variance StDev/Mean					
Synovial Fluid			Ringer's solution		
5 s	2 min	5 min	5 s	2 min	5 min
0.69	0.35	0.15	0.56	0.15	0.22

Table 2-1 Friction data *reproducibility* for the results presented in Figure 2-4, Figure 2-5, Figure 2-6 and Figure 2-7.

Following these series of tests, as the reproducibility variance was greater than the repeatability variance, it was decided, henceforth, to take one measurement per specimen per loading time. Mean values and standard deviations were then calculated from testing groups of at least seven specimens tested under identical conditions. However, as explained in subsequent chapters, two readings per specimen per loading time were often recorded so as to ensure the continuous reliability of the friction data for the various contact configurations studied. This was sometimes conducted in such a manner that the possible occurrence of wear and biological degradation could also be monitored.

2.4.2.1 Indentation Analysis

During preliminary friction tests upon articular cartilage specimens it became evident that the amount of load removal allowed, in-between repeated friction measurements upon the same cartilage specimen, was clearly having an effect on the coefficient of friction results. The indentation test described below was used to study this phenomenon.

In order to investigate how the cartilage responded to loading and to obtain an idea of how the fluid was expressed out of the cartilage while under load and recovered upon load removal, an indentation test was performed. As the fluid is displaced away from the loaded biphasic cartilage the solid phase volume of the cartilage decreases, thereby making indentation tests a useful and popular way to characterise the biomechanical properties of this tissue and also quantitate fluid expression while under load (Edwards, 1967). A load of 6 N was applied via a

hemispherical indenter to a section of articular cartilage (femoral cartilage counterface) while immersed in Ringer's solution. The radius of the indenter was 3.2 mm. The indenter was lowered onto the cartilage surface while data sampling for both vertical displacement and load, with the Unkelscope data acquisition software package, was initiated at a frequency of 0.5 milliseconds so as to record accurately the point of contact between the indenter and the cartilage counterface (Swann and Seedhom, 1989). This point of contact was identified by a sharp rise in the measured load. Measurements of vertical displacement were then taken up to 60 minutes after the initial load was applied. This procedure was then repeated, upon exactly the same cartilage area, after the cartilage had been allowed to rehydrate while unloaded and immersed in Ringer's solution for another 60 minutes, Figure 2-9.

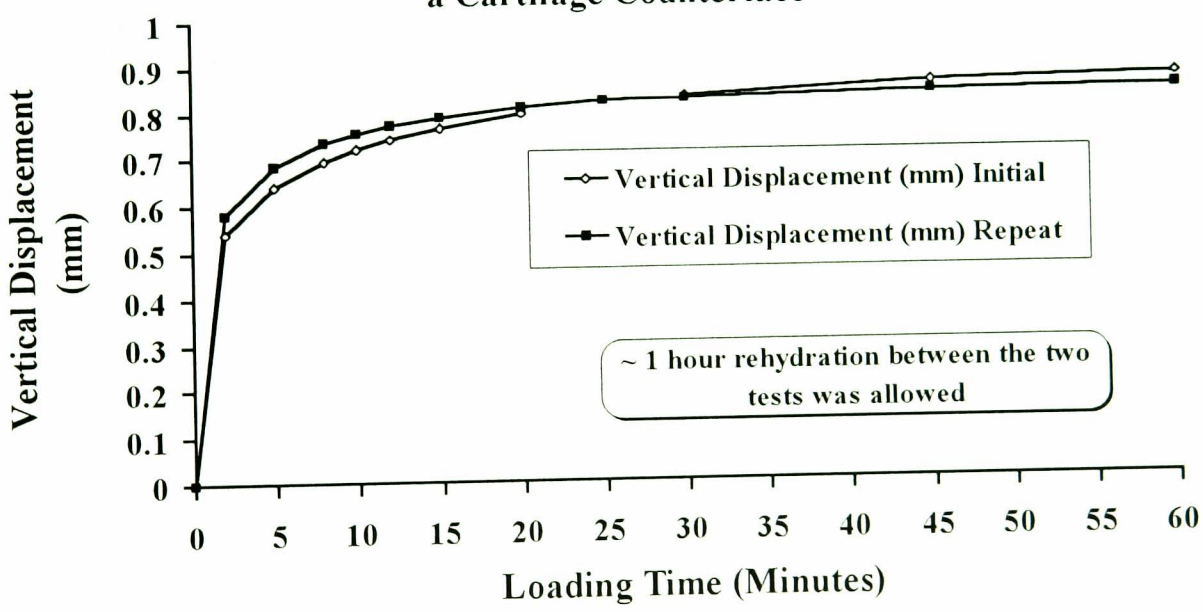


Figure 2-9 Indentation test conducted to assess recovery of fluid content for equal loading / load removal duration.

Following an equal period of load removal to that of previous loading, the indentation test produces a similar displacement-time curve when subsequently reloaded which was indicative of a full recovery of fluid content and this equal loading and load removal procedure was subsequently used in all friction tests. In

adopting this procedure repeat measurements on the same cartilage specimen produced repeatable values of friction coefficient. This procedure was therefore a major factor in ensuring good levels of both *repeatability* and *reproducibility* of friction coefficient results. If the amount of loading and load removal were not carefully monitored the friction coefficients were found to have far greater variability.

2.4.2.2 Different Anatomical Locations

As a particular point of interest a fresh 9 mm cartilage plug extracted from one of the bovine carpal synovial joints was friction tested at stationary loading times of 5 seconds and 5 minutes in Ringer's solution. The cartilage thickness from these regions was particularly thin, approximately ≤0.3 mm. The mean and standard deviation of seven steady state friction coefficient readings recorded at each loading time was 0.034 (±0.007) and 0.123 (±0.014) respectively. These values were far higher than the respective equivalent mean values for cartilage specimens taken from the bovine knee joint, being 0.012 (±0.006) and 0.073 (±0.017). Obtaining the cartilage specimens from the same anatomical sites was also therefore shown to be an important way to control the reproducibility of the friction results. No obvious disparity was seen between friction data collected from the various locations of the distal end of the bovine femur.

2.4.3 Film Thickness Predictions

As the purpose of this study was to investigate the frictional behaviour of articular cartilage under mixed and boundary lubrication regimes it was essential to design the experiments to avoid full fluid film lubrication. A mixed regime is predicted when the calculated film thicknesses are less than the combined roughness of the opposing contact surfaces. The methods for theoretically predicting the thickness of the lubricating film for the 3 & 9 mm cartilage plug on metal counterface contact configuration are outlined below.

Squeeze film - for stationary loading periods.

To model the normal approach of the cartilage discs towards the metal counterface and predict squeeze film thicknesses while under stationary load the

following equation, for perfectly smooth surfaces, was used after Dowson et al. (1991),

$$h = a^2 \sqrt{\frac{3\pi\eta}{4Wt}}$$

η - viscosity (Pa s)
W - load (N)
t - squeeze film time (s)
h - squeeze film thickness (m)
a - contact radius (m)

For this study a = 0.0015 m and W = 30 N. For Ringer's solution η was 0.001 Pa s and for synovial fluid η was assumed to be 0.1 Pa s.

For a stationary loading time of 5 seconds the predicted squeeze film thickness for both Ringer's solution and synovial fluid was calculated to be 0.01 and 0.09 μm respectively. As articular cartilage surface roughness is of the order of 1 μm, then clearly, at 5 seconds loading and beyond, a mixed lubrication regime was predicted for both lubricants.

Entraining action - during sliding.

The minimum film thickness, h_{min}, for entraining action during sliding of the cartilage specimens can be estimated by assuming that the flat-ended cylindrical cartilage disc could be approximated to a square pad such that the length of the square, L, was $L^2 = \pi r^2$, (Halling, 1978), where r is the radius of cartilage specimen (m) such that,

$$h_{min} = \sqrt{0.071\left(\frac{\eta u L}{W}\right)} \cdot L$$

where W - load (N)
η - viscosity (Pa s)
μ - sliding velocity (m/s)
$L = \sqrt{\pi . r}$ (m)

Accounting for some degree of shear thinning the viscosity of synovial fluid was taken to be 0.01 Pa s during sliding. For Ringer's solution and synovial fluid the predicted film thicknesses for entraining action were 0.01 μm and 0.04 μm respectively. Both of these predictions were equal to or less than the squeeze film calculations at 5 seconds and thereby still well within a mixed lubrication regime.

The values quoted above relate only to the 3 mm cartilage plugs but as previously mentioned for the 9 mm cartilage plugs, under both squeeze film and entraining action conditions, a mixed lubrication regime was again predicted.

2.5 Concluding Remarks

- Successful procedures for the collection and storage of articular cartilage and synovial fluid samples have been developed.
- Suitable techniques for the surface characterisation of articular cartilage have been identified; including non-contacting 3D laser profilometry which has never before been used on articular cartilage.
- Methods of measuring friction of articular cartilage in mixed lubrication regimes, with good repeatability and reproducibility, have been established.

3. Surface Characterisation of Articular Cartilage

3.1 Introduction

For any workers involved with the tribology of synovial joints a good understanding of the morphology and composition of the articular cartilage surface is required (Kirk et al., 1993a). From the literature however there are conflicting opinions regarding the characteristics of the cartilage surface. Using predominantly scanning and transmission electron microscopy, some studies have advocated the articular cartilage surface to be perfectly smooth (Ghadially, 1983; Kirk et al., 1993a; Bloebaum and Radley, 1995) while others attributed varying degrees of roughness to the surface (Walker et al., 1968; Sayles et al., 1979; Gardner et al., 1981). Also, the presence of a distinct surface layer lying above the superficial tangential zone has been noted, (MacConaill, 1951; Davies et al., 1962; Barnett et al., 1963; Balazs et al., 1966; Weiss et al., 1968; Meachim and Roy, 1969; Stanescu and Leibovich, 1982; Ghadially et al., 1982; Orford and Gardner, 1985; Jeffery et al., 1991; Kirk et al., 1993a; Shaw and Molyneux, 1994; Kobayashi et al., 1995). However the characteristics of the layer, (e.g. composition and thickness), as well as its function, still remain largely unestablished, with varying interpretations put forward within the literature.

The uncertainty regarding the nature of the cartilage surface made an independent investigation a necessary requirement for this project. This study involved the examination of bovine articular cartilage surfaces using a range of available techniques as follows,

I. Stylus Profilometry

II. Laser Profilometry

III. Scanning Electron Microscopy (SEM)

IV. Environmental Scanning Electron Microscopy (ESEM)

V. Transmission Electron Microscopy (TEM).

Several cartilage plugs were analysed for each of these methods.

None of the previous studies have scrutinised the cartilage surface with more than any two particular analytical techniques. Furthermore laser profilometry has never before been used to observe the surface of articular cartilage.

3.2 Analytical Techniques and Experimental Procedure

The basic principles of each of the five analytical techniques employed in this study have been referred to in Chapter 2. Unless otherwise stated the specimens examined for each technique had not been previously tested in the friction rig or investigated by any of the other analytical techniques.

3.2.1 Stylus Profilometry

Nine bovine cartilage plugs underwent a number of stylus traces using a Form Talysurf Series 6 stylus profilometer (Rank Taylor Hobson), Table 3-1. Each cartilage plug was taken from a different animal. Cartilage plugs were extracted from synovial joints of the carpal and metacarpal regions. These sections of the front hooves were routinely removed from the animals immediately after slaughter and therefore offered the freshest possible specimens for analysis. Femoral patella and condyle regions of the bovine knee joint were also used, as cartilage plugs from these areas were utilised for the friction tests. For each cartilage plug four line profiles, each one approximately 6 millimetres long, were recorded. These line profiles were sampled across the diameter of the cartilage surface. The first two line profiles were taken over the same assessment length (within $\pm$ 1 μm) and a further two across different assessment lengths.

Each line profile took about 30 seconds to record and none of the cartilage specimens were out of their Ringer's solution sample containers for more than 5 minutes during the surface analysis. Throughout these experiments the cartilage plugs remained fully hydrated. This was achieved by droplets of Ringer's solution being able to rest on and completely engulf the cartilage surface while the stylus traces were being performed without adversely affecting the results or impeding movement of the stylus (Sayles et al., 1979).

Specimen 1 (carpal/metacarpal)	Collected immediately after slaughter and stored in Ringer's solution for 3 days at 0-5°C prior to profilometry. Showed signs of *mild osteoarthritis*.
Specimen 2 (carpal/metacarpal)	Collected immediately after slaughter and stored in Ringer's solution for 3 days at 0-5°C prior to profilometry.
Specimen 3 (carpal/metacarpal)	Collected immediately after slaughter. Stored in Ringer's solution for 3 days at 0-5°C and examined by ESEM prior to profilometry.
Specimen 4 (carpal/metacarpal)	Collected immediately after slaughter. Stored in Ringer's solution for 3 days at 0-5°C and stationary load friction tested on a metal counterface prior to profilometry.
Specimen 5 (femoral patella)	Stored frozen in Ringer's solution for 3 months and stationary load friction tested on a metal counterface prior to profilometry. Defrozen 3 days before profilometry analysis.
Specimen 6 (femoral condyle)	Stored frozen in Ringer's solution for 3 weeks and stationary load friction tested on a metal counterface prior to profilometry. Defrozen 3 days before profilometry analysis.
Specimen 7 (femoral condyle)	Stored frozen in Ringer's solution for 3 weeks prior to profilometry. Defrozen on the same day as the profilometry analysis.
Specimen 8 (femoral condyle)	Stored frozen in Ringer's solution for 10 months prior to profilometry. Defrozen on the same day as the profilometry analysis.
Specimen 9 (femoral condyle)	Stored frozen in Ringer's solution for 10 months prior to profilometry. Defrozen on the same day as the profilometry analysis.

Table 3-1 Details of cartilage specimens used for stylus profilometry.

3.2.1.1 Surface Roughness Parameters

Numerous parameters can be calculated from a line profile. It should be remembered that these parameters were originally chiefly designed to describe engineering surfaces. These parameters describe the vertical characteristics of the surface deviations (amplitude/roughness parameters), the horizontal characteristics of the surface deviation (spacing/waviness parameters) or some combination of both (hybrid parameters). A profile is taken from the total surface for evaluation. This profile is fitted to a *mean line*, if sloped this is simply a line levelling process involving least square analysis, or for curved surfaces a curve fit

of given radius may provide a better outcome[1]. The mean line bisects the profile such that the area above it and below it are equal and a minimum. Roughness and waviness may be separated by means of the length of instrument *cut-off*. The profile is divided into equal *sample lengths* l (known as the cut-off), which are long enough to include a statistically reliable amount of roughness, yet short enough to exclude waviness from the roughness measurement. This procedure is termed filtering and the standard ISO filter was adopted, being available for both the stylus and laser profilometry software packages. The ISO filter provided results similar to the ISO PC and Gaussian filters but the two latter filters were only available for stylus profilometry. The *assessment* or *evaluation length* L is defined as the length of profile used for the measurement of surface roughness parameters (usually containing several sampling lengths; at least five consecutive sampling lengths are taken as standard).

For both the stylus and laser profilometry work the results of three roughness parameters were considered, namely R_a, R_{tm} and R_{3z}. These parameters are described below.

R_a R_a is the universally recognised, and most used, international parameter of roughness. It is the arithmetic mean of the departures of the roughness profile from the mean line.

R_{tm} R_{tm} (also known as $R_{z(DIN)}$) is the mean of all the maximum peak-to-valley heights (R_{ti} values) in each particular sampling length within the assessment length.

R_{3z} R_{3z} is the vertical mean from the third highest peak to the third lowest valley in each sample length over the assessment length.

R_{3z} is therefore similar to R_{tm} but ignores the first two highest peaks and deepest valleys in each cut-off length. The R_{3z} data can be a much more accurate representation of the surface. The highest peaks and deepest valleys can be untypical of the surface and susceptible to artefacts.

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[1] A polynomial fit option was also available for the software supplied with the laser profilometry equipment but not for the stylus profilometer.

All this data was processed by software packages, according to international standards, supplied with the particular stylus and laser profilometry tools. Further information relating to surface roughness parameters can be found in the profilometry appendix, page 319.

3.2.2 Laser Profilometry

A single cartilage plug was analysed using laser profilometry both before and after a reciprocating motion friction test. The cartilage plug was subject to a constant 30 N load and continuous sliding against a metal counterface for a total of 90 minutes, as detailed in chapter 7. The cartilage plug required no special preparation for laser profilometry. However to avoid reflection from the water the cartilage plug had to be devoid of any surface water. Therefore immediately prior to focusing the cartilage surface was gently blown dry by a clean air duster spray. The laser profilometry scans were intentionally limited to take no longer than 5 minutes to perform due to the possible detrimental effects of surface dehydration (Speer et al., 1990; Kirk et al., 1993b). This placed constraints on both the available scan area and point resolution of the acquired image. Directly after each scan the cartilage plug was resoaked in Ringer's solution for a period of approximately 5 minutes.

The cartilage plug was from a bovine femoral condyle collected and stored frozen at -20°C for 3 months prior to the profilometry analysis and reciprocating motion friction test. The cartilage plug was selected from a number of samples after examination by laser profilometry to confirm that it was a healthy, normal specimen with a typical surface appearance. Laser profilometry scans were undertaken both before and after a reciprocating motion friction test and are detailed in Table 3-2. The area scans provided a 3D visualisation of the cartilage surface while the line profiles were used to calculate the surface roughness parameters, R_a, R_{tm} and R_{3z} (see section 3.2.1.1).

Laser Profilometry Scans		
Number of Scans	Assessed Area or length (mm)	Resolution (points/mm)
1	1 x 1	250 x 250
2	0.5 x 0.5	250 x 250
2	0.1 x 0.1	1000 x 1000
5 Line Profiles	6	1000

Table 3-2 Laser profilometry scans recorded both before and after a reciprocating motion friction test. Each scan was sampled from a different area of the cartilage plug surface.

3.2.3 Scanning Electron Microscopy

Four femoral condyle cartilage plugs were analysed using JEOL T20 SEM. Three of the cartilage specimens were 9 millimetre diameter plugs, the remaining one was a three millimetre cartilage disc specimen and had previously been used for a stationary load friction test. All the cartilage plugs had been stored frozen in Ringer's solution at -20°C and processed for SEM examination within 3 months of their acquisition date. The fixation and dehydration procedure adopted was similar to one standardised method used by Bloebaum and Wilson (1980). Cartilage tissue preparation for both SEM and TEM was carried out by Mr Paul McPhie[2].

3.2.3.1 Cartilage fixation and dehydration for SEM and TEM examination

For the cartilage plugs fixation was by immersion in 2.5% aqueous glutaraldehyde in 0.1 M cacodylate buffer at pH 7.4 for at least 4 hours or left overnight at 4°C. The specimens were then washed for 2 hours at 4°C in three changes of 0.1 M cacodylate buffer at pH 7.4, post fixed with 1% osmium tetroxide in 0.1 M cacodylate buffer at pH 7.4 for 1-2 hours, and washed several times in 0.1 M cacodylate buffer at 4°C for 15 minutes each time. The specimens were dehydrated through ascending grades of ethanol (70% 2x10 minutes; 90% 2x10 minutes and 100% 2x10 minutes) and subjected to critical point drying in carbon

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[2] Senior Experimental Officer, Biochemistry and Molecular Biology Electron Microscopy Unit, University of Leeds.

dioxide. The specimens were finally gold coated (~20 nm in thickness) in a Nanotech SEM Prep. 2 vacuum coating apparatus.

Throughout these procedures the cartilage remained attached to the subchondral bone. The removal of the subchondral bone has previously been cited as a mechanism for the introduction of artefacts upon the cartilage surface (Ghadially et al., 1982) as the cartilage matrix contracts without the constraints of the underlying bone.

3.2.4 Environmental Scanning Electron Microscopy

The cartilage specimens were observed under conditions of full hydration; dehydration artefacts were thus eliminated. In the ESEM (Electroscan model) this is achieved by viewing hydrated specimens in a high water vapour pressure environment, separated from the electronic optic column by a series of isolating apertures and incorporating a unique electron detection system. A dew point is established by the operator by varying the specimen chamber pressure and temperature, whereby no condensation of water upon the specimen surface occurs and no dehydration of the specimen surface or bulk of the tissue occurs. For the cartilage specimens the dew point was set at a pressure of 986 Nm^{-2} and temperature of 8.0°C. If the specimen chamber temperature rose or the pressure fell the specimen would begin to dehydrate. Visa versa, if the specimen chamber temperature fell or the pressure rose the specimen surface would flood with water. The details of the cartilage specimens examined using ESEM are listed in Table 3-3. Prior to mounting the cartilage samples on the Electroscan specimen stage they were quickly rinsed in distilled water to avoid salt deposition from the Ringer's solution. The underlying bone of the cartilage plugs had to be sectioned down as much as possible so the cartilage surface was as close to the specimen stage base as possible. The temperature of the specimen stage base was regulated in order to control the dew point inside the specimen chamber. The pressure of the specimen chamber could also be regulated by the operator again to maintain the dew point of the hydrated imaging surface, as mentioned. During the

examination of the specimens video footage was recorded in conjunction with photographic and digital images.

Specimen 1 (femoral condyle)	Stored frozen in Ringer's solution for 10 months and defrozen 2 days before ESEM examination. This specimen had been analysed by stylus profilometry the day before the ESEM work.
Specimen 2 (carpal/metacarpal)	Collected immediately after slaughter and stored in Ringer's solution for 1 day at 0-5°C prior to ESEM.
Specimen 3 (femoral condyle)	Stored frozen in Ringer's solution for 3 weeks and defrozen 2 days before ESEM examination. This specimen had been analysed by stylus profilometry the day before the ESEM work. The cartilage was wiped firmly with moist tissues prior to ESEM imaging.
Specimen 4 (carpal/metacarpal)	Collected immediately after slaughter and stored in Ringer's solution for 2 days at 0-5°C prior to ESEM. Examined by stylus profilometry 2 days after ESEM work.
Specimen 5 (carpal/metacarpal)	Collected immediately after slaughter and stored in Ringer's solution for 2 days at 0-5°C prior to ESEM.. Showed signs of *mild osteoarthritis.*
Specimen 6 (femoral condyle)	Fixed and dehydrated cartilage specimen prepared for conventional SEM. Previously examined by SEM.

Table 3-3 Details of cartilage specimens used for environmental scanning electron microscopy. The specimens were analysed one at a time while the remaining specimens were stored in Ringer's solution at 0-5°C.

3.2.5 Transmission Electron Microscopy

Three femoral condyle cartilage plugs were prepared for TEM analysis. The specimens had been stored frozen at -20°C for 1-2 months prior to defrosting and subsequent fixation. The specimens were fixed and dehydrated using the same procedure as for the SEM specimens (see section 3.2.3.1). Further treatment was conducted in manners similar to those previously conducted by other workers (Ghadially et al., 1982; Morrison et al., 1993; Kirk et al., 1993a). The plugs were decalcified, infiltrated with propylene oxide and embedded in epoxy resin. Ultrathin sections, ~50-100 nm thick, were cut on a Riechert Ultracut using a diamond knife and stained with uranyl acetate and lead citrate. Sections of

cartilage ~0.5x0.5 mm in area, were viewed and photographed using a Phillips CM10 TEM.

3.3 Results and Discussion

3.3.1 Stylus Profilometry

For any particular specimen the repeatability of the roughness parameter values sampled by the four line profiles per specimen was very good. This was especially true for the first two recorded line profiles which traversed over the same assessment length to within an accuracy of ± 1 μm. The close matching data from these two 'repeat' line profiles indicated that permanent deformation caused by the stylus during its traverse was not occurring. This was backed up by observations made using an ESEM upon stylus profilometry cartilage specimens, see section 3.3.4. Plastic deformation due to the stylus trace has been previously, theoretically postulated not to occur (Sayles et al., 1979).

The mean and standard deviation values of the R_a, R_{tm} and R_{3z} roughness parameters were calculated from the average of the four line profiles values conducted on each specimen (the two 'repeat' line profiles were first averaged themselves and assessed as a single value). The R_a values were reasonably reproducible with the coefficient of variance being 0.6 and 0.3 respectively for the 0.8 and 0.08 mm sampling lengths, Figure 3-1. This level of reproducibility was pleasing in consideration of the differences between the specimens regarding their anatomical location, storage details etc., Table 3-1. The R_{tm} and R_{3z} values were also adequately reproducible with their coefficient of variance being 0.5 and 0.4 respectively. The finding that the R_{tm} values were not much higher than the R_{3z} values was indicative of the cartilage possessing few, if any, surface features which produced exceptionally high , one-off, peaks or low troughs within the assessed profiles. This finding was also the case for the specimen showing mild signs of osteoarthritis (OA), Figure 3-2. This finding also indicated that the stylus tip itself was not introducing any obvious anomalous data. For all three of the roughness parameters the OA specimen demonstrated markedly higher values than the mean values of the other 8 specimens examined. The mean R_a value for a 0.8 sampling length of 1.6 (±0.9) μm was in good agreement with previous stylus profilometry studies. For the same sampling length Sayles et al. (1979) quoted

values of 1.4 to 1.9 μm while Longfield et al. (1969), performing stylus traces upon replica castings of the cartilage surfaces, quoted R_a (termed centre line average in their study) values of 2-4 μm for healthy adult human cartilage.

In consideration of lubrication mechanisms, the R_a measurements calculated for sampling lengths of 0.8 mm may probably be the most appropriate for relatively large areas of opposing cartilage surfaces coming into contact. However these values were probably more a reflection upon the general waviness still existing over such 0.8 mm lengths. The 0.08 mm filtered line profiles exhibited no waviness as such. Their mean R_a values of 0.24 (±0.08) μm, Figure 3-1, were much more in keeping with the dimensions of surface features such as superficial collagen fibres observed by both SEM and TEM techniques. This quantitative value of roughness was also more akin to qualitative values quoted by previous TEM studies (Weiss et al., 1968; Ghadially et al., 1982). The order of magnitude, however, would appear to cause concern regarding the estimated magnitude of elastic deformation of the cartilage surface during traverse of the stylus tip, (see section 3.3.1.1).

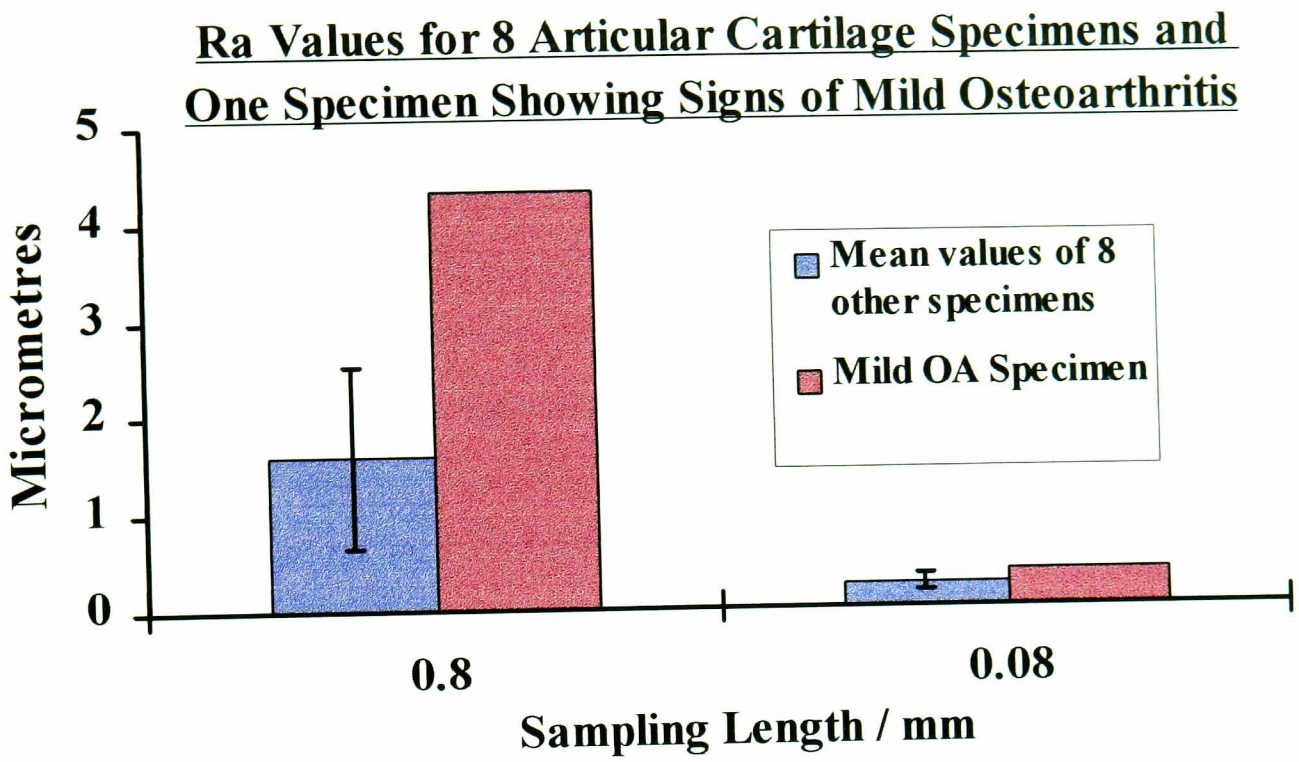


Figure 3-1 Mean and standard deviation Ra values taken from 8 different cartilage plugs and one cartilage plug showing signs of mild osteoarthritis, The results were calculated for two different sampling lengths as shown.

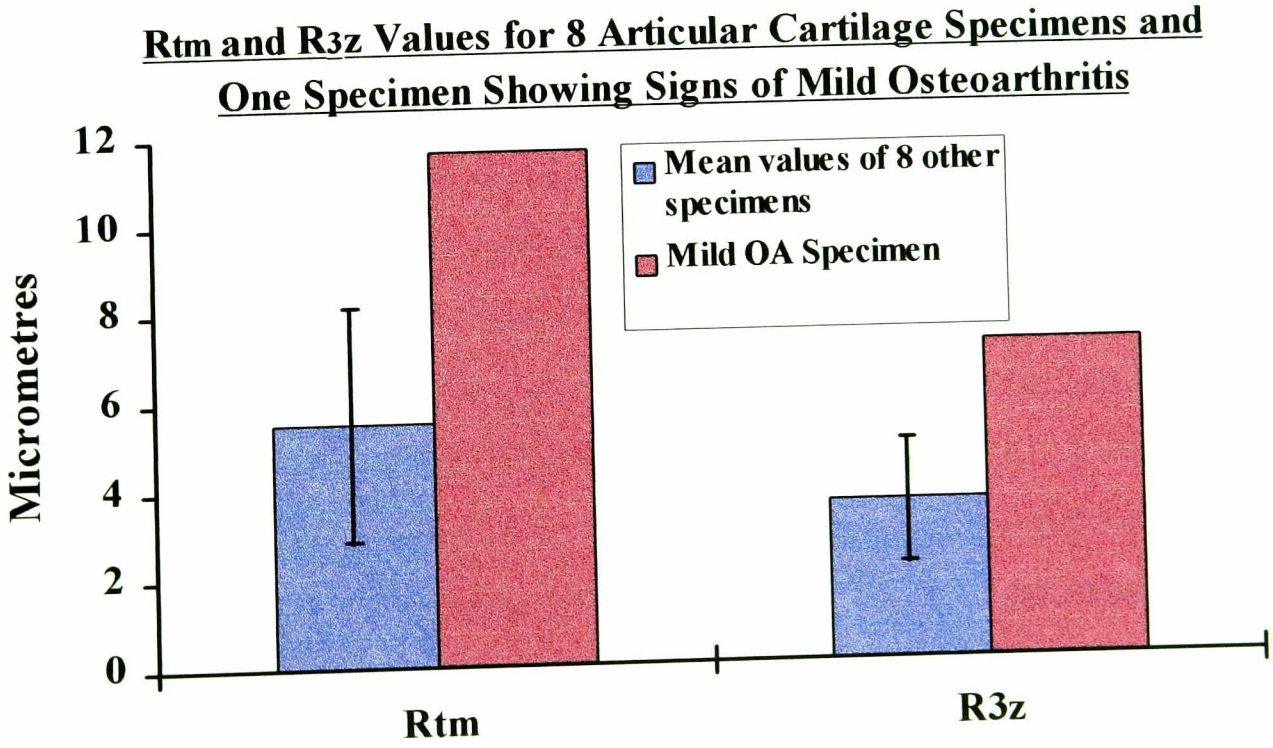


Figure 3-2 Mean and standard deviation R_{tm} and R_{3z} values taken from 8 different cartilage plugs and one cartilage plug showing signs of mild osteoarthritis. The results were calculated using a 0.8 mm sampling length.

3.3.1.1 Assessment of Stylus Profilometry Data

The use of acrylic replica casts of the cartilage surface has in the past been preferred to direct stylus traces of the cartilage surface due to concerns regarding elastic and plastic deformation of the cartilage surface under the stylus leading to erroneous data (Longfield et al., 1969). Although it was found that this replica technique produced reasonable results if the correct acrylic was chosen, the technique was considered unnecessary as direct stylus measurements did not damage the cartilage surface or introduce unacceptable errors (Sayles et al., 1979). A correct theoretical evaluation of elastic and plastic deformation caused by a diamond tip conisphere stylus used in this study, (1.5-2.5 μm in radius with a static resultant load of 0.7-1 mN traversing at 0.5 mm/s), upon a viscoelastic, anisotropic, inhomogeneous cartilage surface (while choosing the correct values for such required parameters as Young's modulus from the range of values stated) which may or may not possess a distinctive surface layer of unknown physio-chemical properties and thickness is a difficult, if not impossible, undertaking. Sayles et al. (1979) however performed both an experimental investigation and theoretical analysis of the interaction between a Talysurf (Rank Precision Industries) stylus (truncated pyramid, rectangular tip ~3x8 μm [3 μm in the direction of traverse] having a static resultant load of 1 mN traversing at 1 mm/s), and the cartilage surface. It should be noted that the stylus specifications were therefore similar to that of the present study. The indentation of the stylus tip was considered to be well within the elastic range of the cartilage, without therefore leading to permanent deformation. Elastic deformation of the cartilage solid matrix during traverse of the stylus tip was estimated at ~0.4 μm; deformation of a distinct surface or boundary layer was not considered. This value represented 20-30% of the quoted roughness measurements in this study, calculated for a 0.8 mm cut-off, and was deemed not to be sufficiently large as to invalidate them. It was further argued that if every element of the surface yielded by the same amount the resultant effect of elastic (and plastic) deformation would not have altered the calculation of the average roughness.

3.3.2 *Laser Profilometry*

This analysis tool demonstrated the occurrence of surface wear of a cartilage plug during reciprocation motion friction testing. This finding was established by comparing images scanned at different magnifications both before and after reciprocating motion, Figure 3-3 and Figure 3-4. For equal scale magnifications both of these comparisons show a rougher surface for the post friction test images. This has been statistically confirmed by the line profiles sampled before and after the friction test, Figure 3-7. A relatively high resolution image of the worn cartilage surface has been provided, Figure 3-5. In this image surface undulations and humps are evident. These features could be due to such phenomena as collapsed superficial chondrocytes (Clarke, 1971) or exposed collagen fibre bundles (Walker et al., 1969) but with a limiting resolution light spot diameter of 1 micrometer the nature of the surface features were difficult to ascertain with any amount of confidence. The necessity for the removal of water from the cartilage surface was evident after analysing samples with only the thinnest film of water on the surface, Figure 3-6. Generally the surface reflectivity of the cartilage was sufficiently uniform and high enough (>3%) to avoid artefacts. However sometimes steep peaks and troughs, clearly atypical of the cartilage surface, appeared and were deemed to be areas upon which focusing errors occurred. One such area where a deep trough ran adjacent to a relatively high peak can be seen in the forefront of the top image in Figure 3-3.

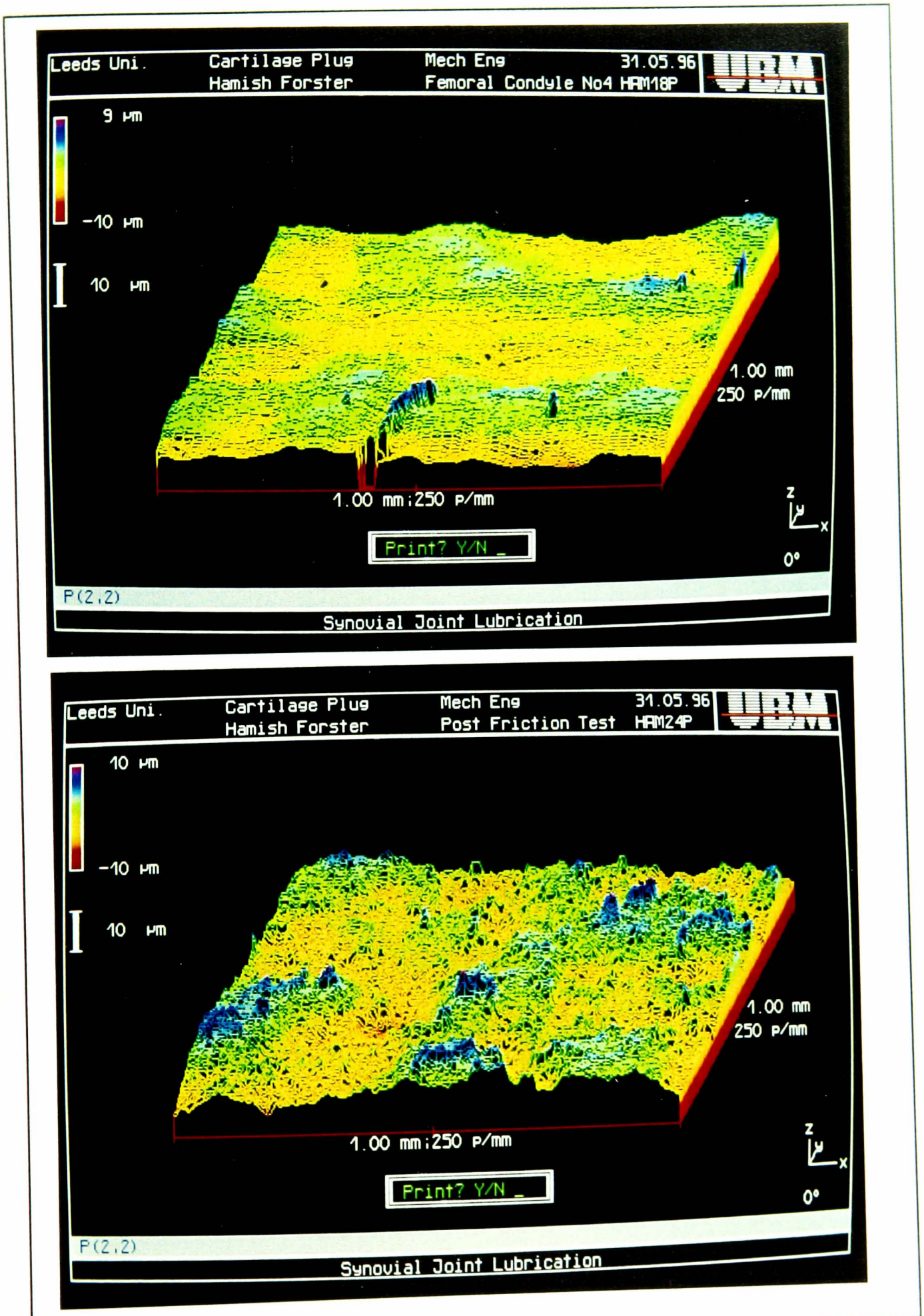


Figure 3-3 Laser profilometry scans of a cartilage plug taken before (top) and after (bottom) a reciprocating motion friction test. The x and y axes are the same scale of 1 mm x 1 mm with a 10 µm scale bar to represent the z axis for both scans. The z : x & y magnification ratio is ~8:1.

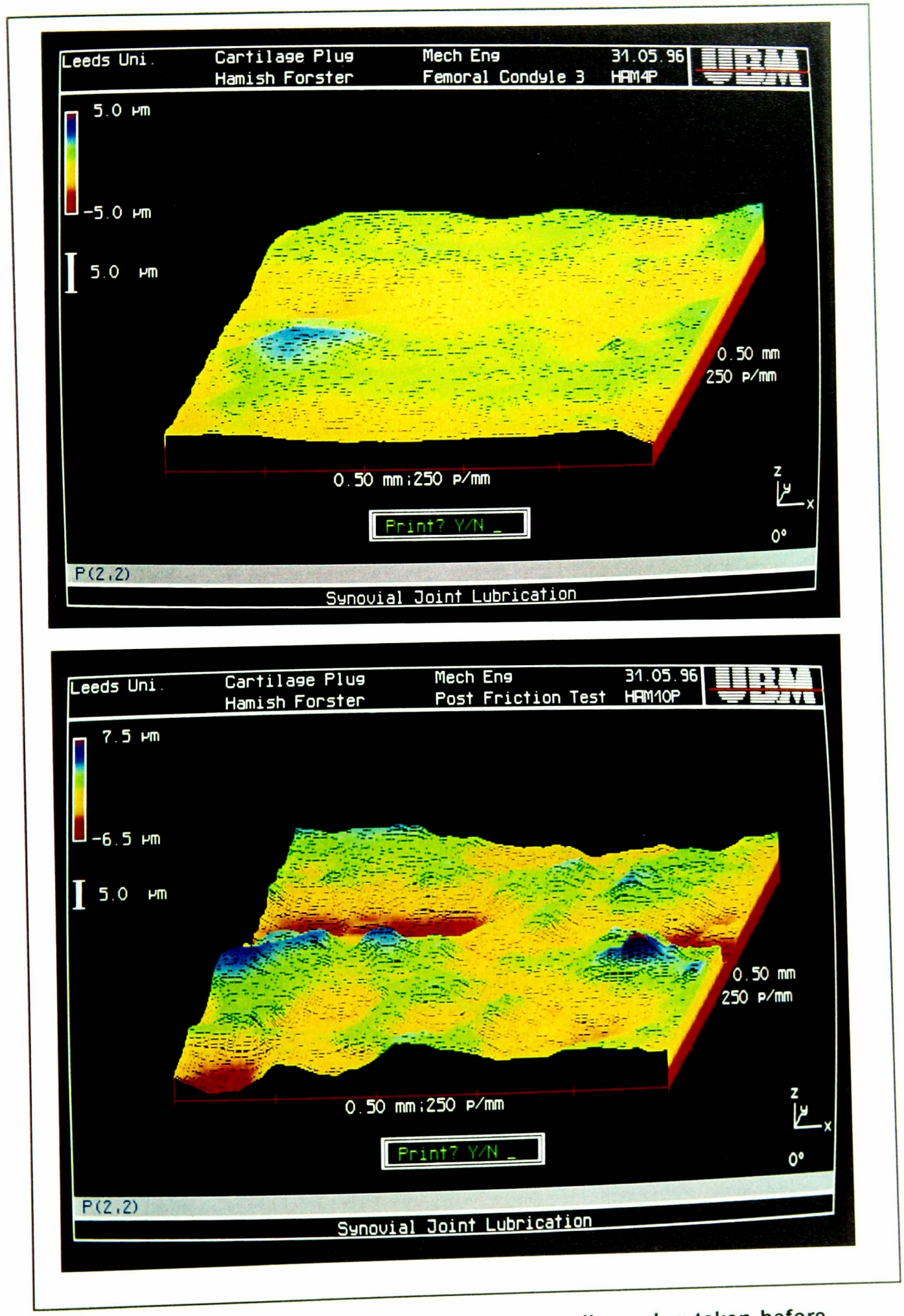


Figure 3-4 Laser profilometry scans of a cartilage plug taken before (top) and after (bottom) a reciprocating motion friction test. The x and y axes are the same scale of 0.5 mm x 0.5 mm with a 5 µm scale bar to represent the z axis for both scans. The z : x & y magnification ratio is ~6:1.

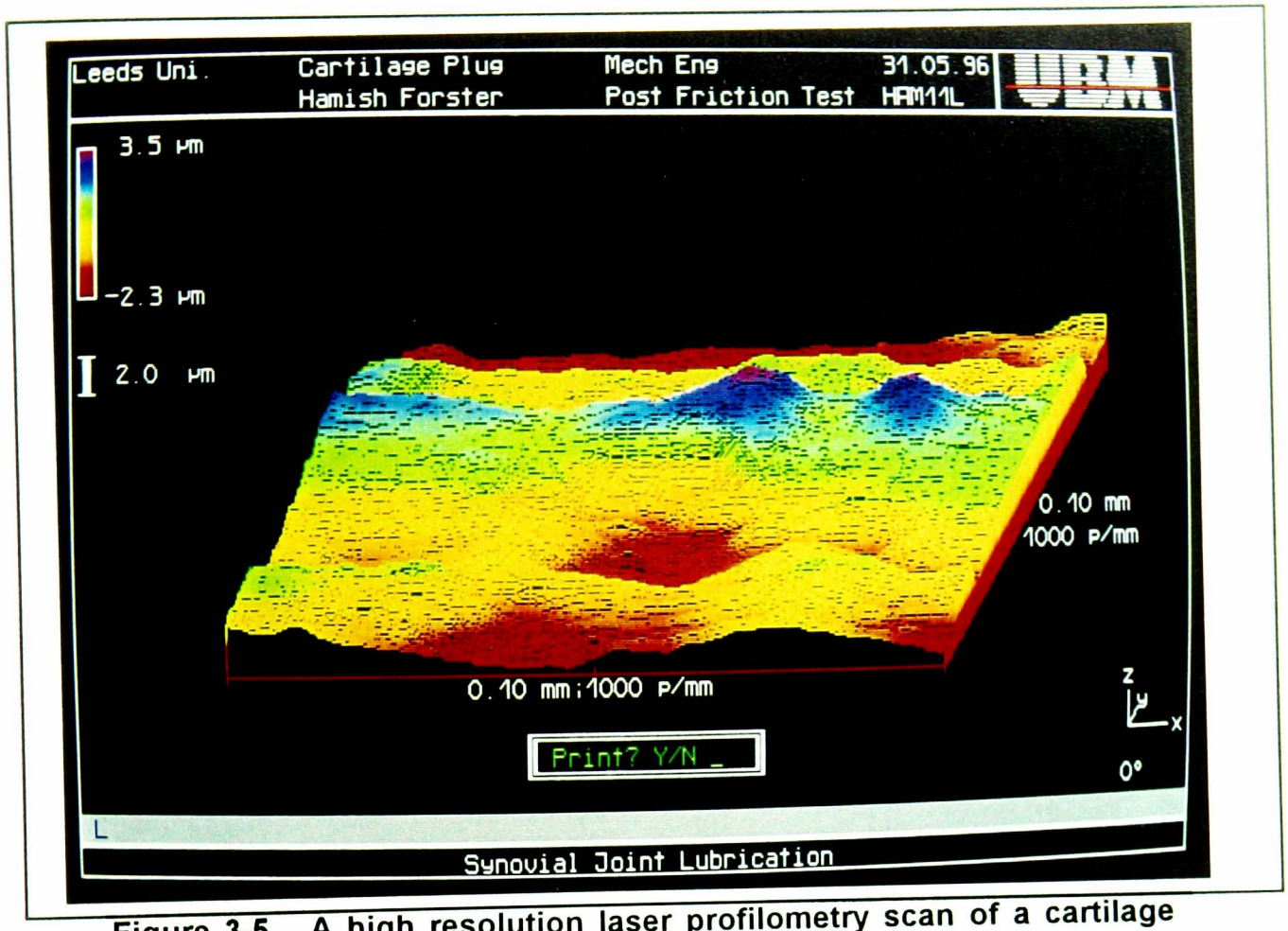


Figure 3-5 A high resolution laser profilometry scan of a cartilage plug after a reciprocating motion friction test. The x and y axes are the same scale of 0.1 mm x 0.1 mm with a 2 µm scale bar to represent the z axis. The z : x & y magnification ratio is ~3:1.

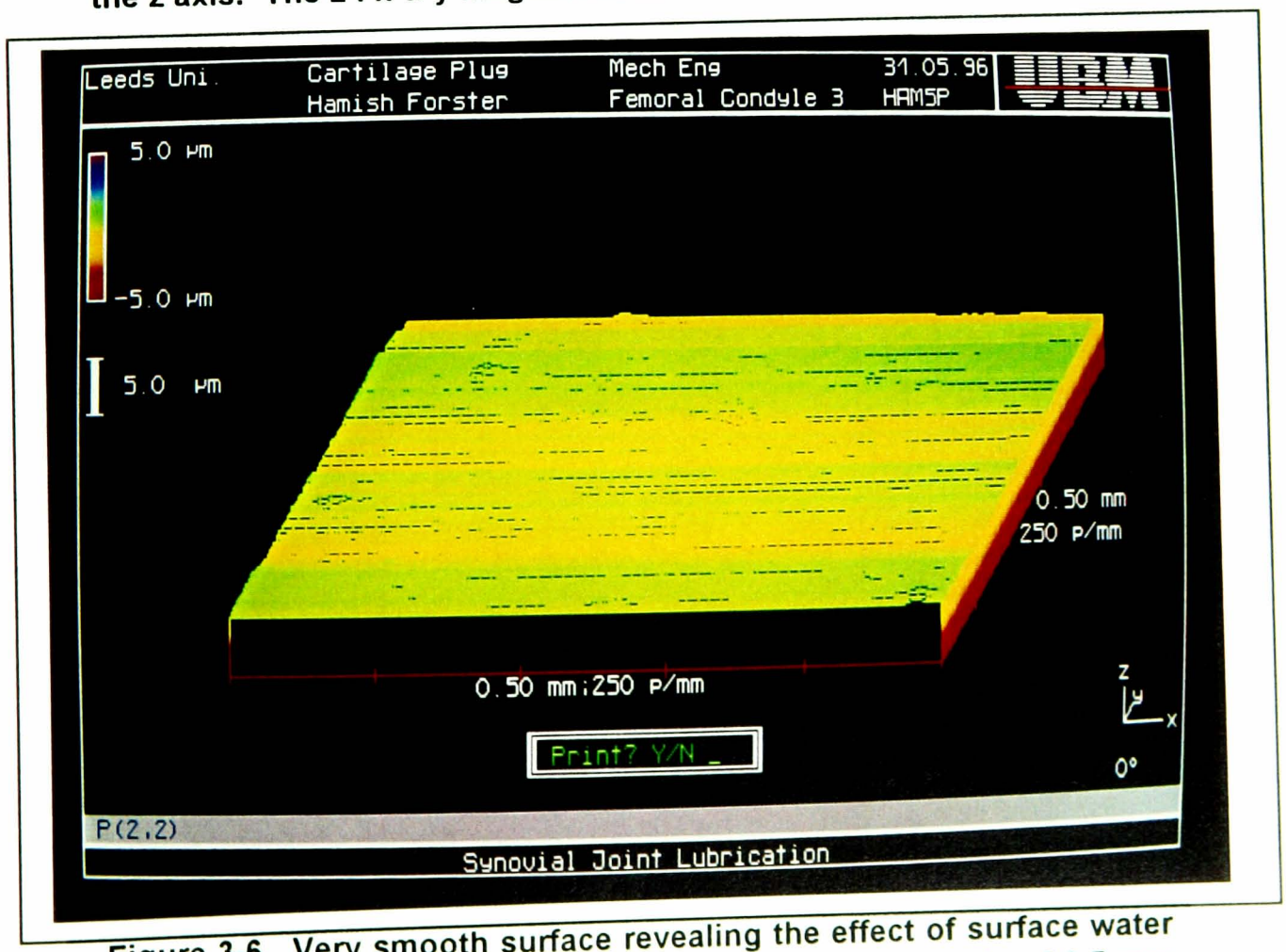


Figure 3-6 Very smooth surface revealing the effect of surface water on the cartilage plug. The x and y axes are the same scale of 0.5 mm x 0.5 mm with a 5 µm scale bar to represent the z axis. The z : x & y magnification ratio is ~8:1.

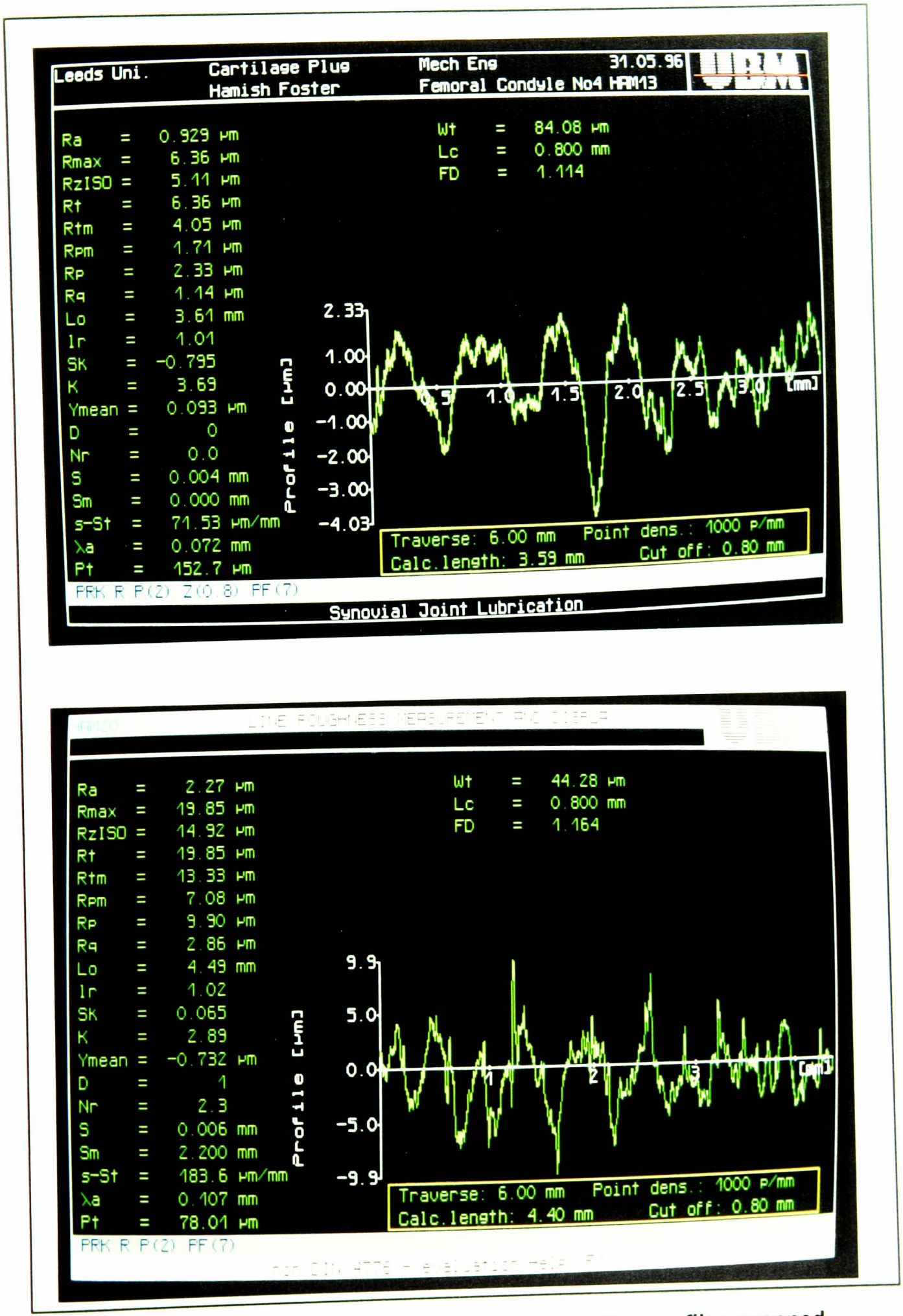


Figure 3-7 Surface parameters calculated from line profiles scanned from a cartilage plug before (top) and after (bottom) a reciprocating motion friction test. Note the change of scale on the y axis between the two line profiles.

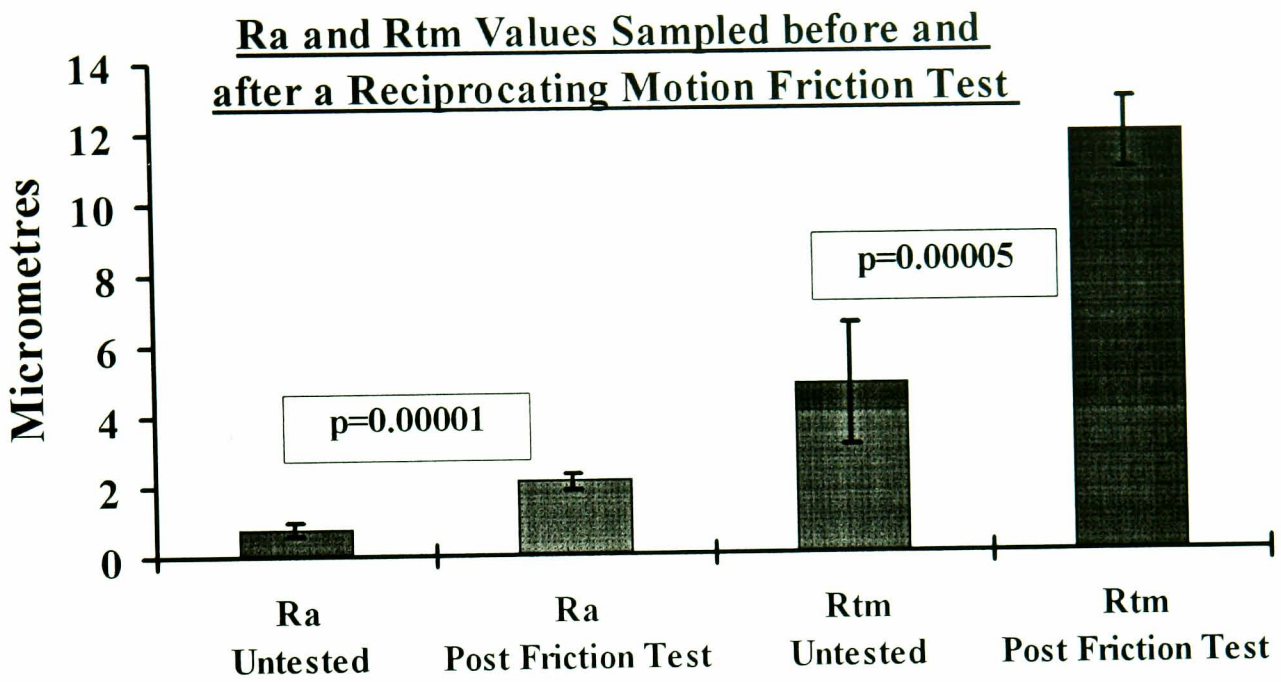


Figure 3-8 Surface parameter Ra and Rtm values calculated from five line profiles sampled from a cartilage plug before and after a reciprocating motion friction test. For both of these parameters t-Tests were conducted and the effect of the reciprocating motion was found to be statistically significant as shown ($p<0.0001$). The sampling length (cut-off) was 0.8 mm.

The results in Figure 3-8 were all from one specimen and as such revealed the good level of repeatability attainable for both the assessed roughness parameters. In Figure 3-8 the mean and standard deviation R_a and R_{tm} values are shown for a cartilage plug analysed before and after a reciprocating motion friction test. The t-Test results proved both these roughness parameters to be statistically significantly higher after the friction test, ($p<0.0001$). The surface roughness R_a (0.8 mm) increased from 0.8 μm to 2.1 μm. Interestingly the R_a and R_{tm} values recorded prior to the friction test were well within the range of the R_a and R_{tm} mean values of 8 cartilage specimens measured using stylus profilometry, comparing those also calculated using a 0.8 mm sampling length. This was a further validation for the use of both techniques in the assessment of cartilage surface roughness.

3.3.3 Scanning Electron Microscopy

The SEM micrographs of conventionally prepared cartilage specimens shown in Figure 3-9 and Figure 3-10 were typical of SEM images presented by other workers (Walker et al., 1969; Draenert and Draenert, 1978; Levanon and Stein, 1991). Surface features such as prominent chondrocytes and orientated collagen fibres were observed. Small white flecks on the cartilage surface were also observed , generally less than 1 micrometer in diameter. These particles have been thought to be a residue of the synovial fluid (Cameron et al., 1976; Bloebaum and Wilson 1980). Walker et al. (1969) went a stage further in describing them as aggregates of complex macromolecules, the smallest particles being consistent with their being hyaluronic acid-protein complex of spherical configuration, 0.2 μm in diameter (Ogston and Stanier, 1951). Larger particles may be contaminates (Bloebaum and Radley, 1995) such as salt deposition from Ringer's solution. The effect of prolonged exposure to the electron beam at high resolution was the removal of the ground substance surrounding the collagen fibres followed by the eventual collapse of the fibres themselves, Figure 3-10.

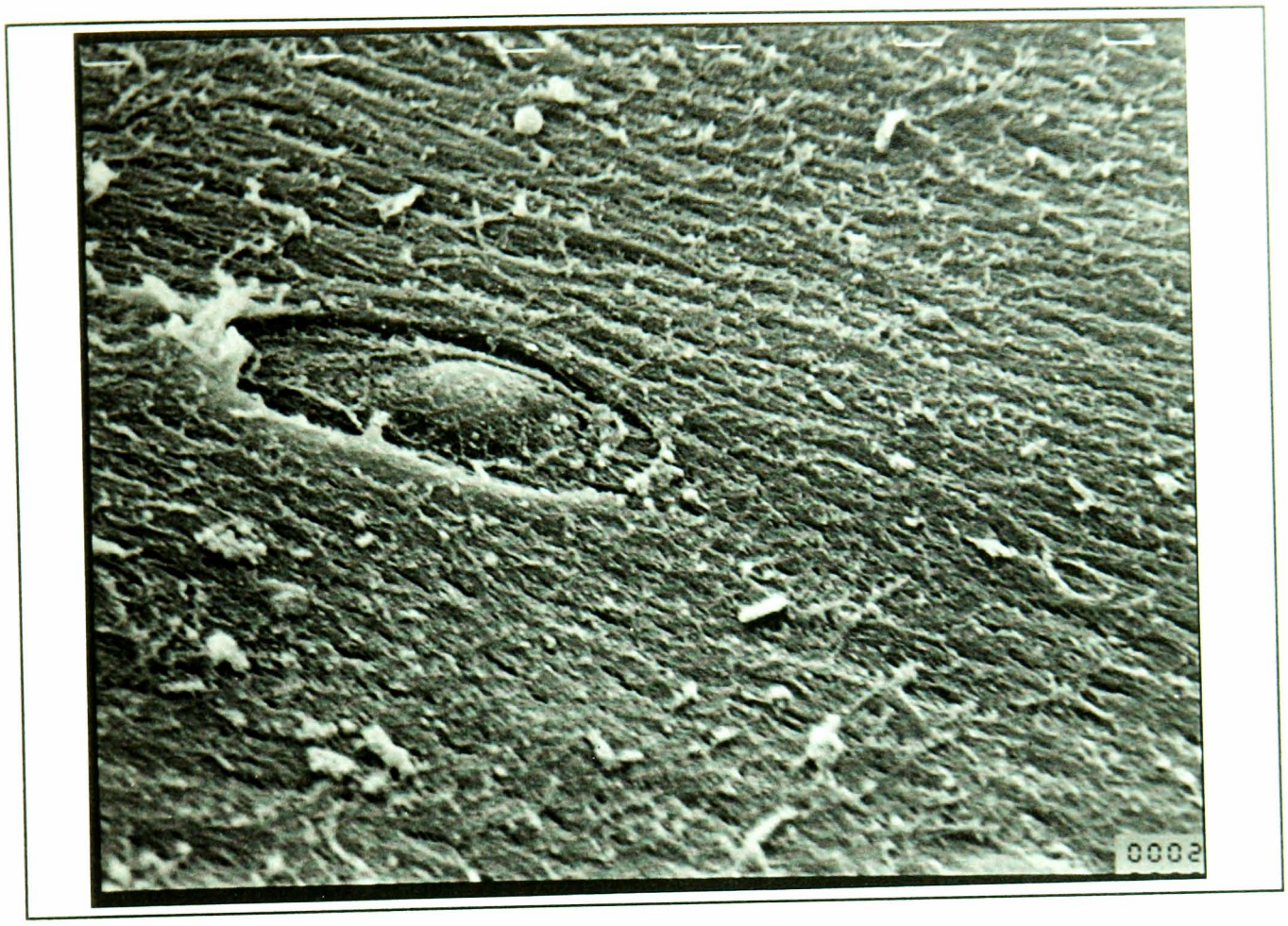


Figure 3-9 Scanning electron microscopy image of the articular cartilage surface. A prominent chrondrocyte is shown alongside orientated superficial collagen fibres. The thin white mark spacers at the top of the image are separated by a distance of 10 μm. Inclination was 60° to the horizontal. (Original magnification was x2 000).

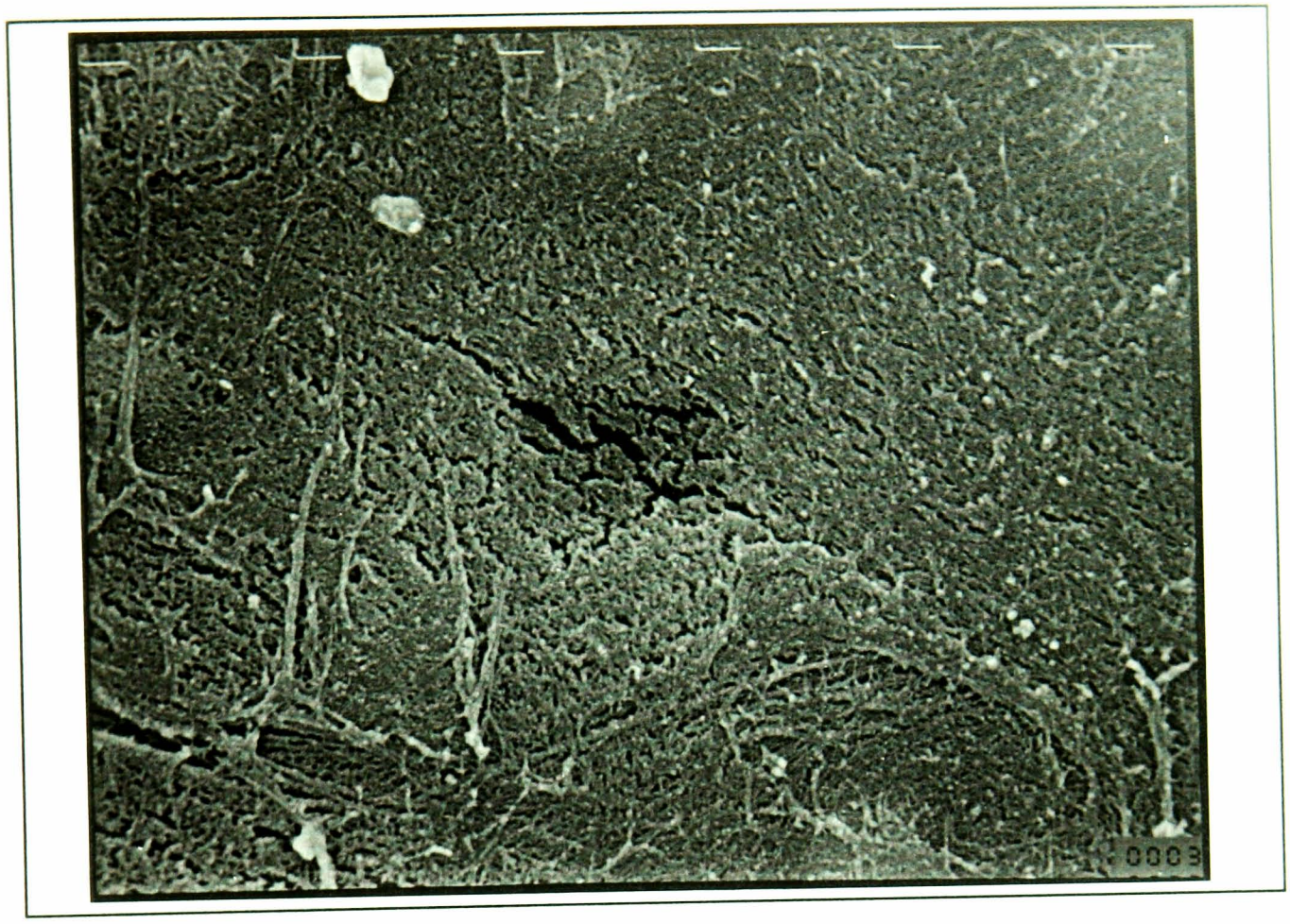


Figure 3-10 Scanning electron microscopy image of the articular cartilage surface. Underlying chondrocytes (light grey spheres) are evident around the edges of the image. In the centre of the image an area of beam damage is apparent. The thin white mark spacers at the top of the image are separated by a distance of 10 μm. Inclination was 18° to the horizontal. (Original magnification was x2 000).

The specimens also provided an opportunity to observe cartilage regions through the thickness of the tissue. In the middle of the superficial tangential zone I region the close packed collagen fibrils (~50-100 nm in diameter) were highly orientated to the cartilage surface and revealed the widespread adherence of granular substance to the fibrils, Figure 3-11, as mentioned by previous workers (Teshima et al., 1995). Little of the interfibrillar solid matrix was evident but the granular substance could have been remnants of the binding sites between the fibrils and proteoglycan network. Further down in the middle zone II the collagen fibres were, as commonly described, noted to be less densely packed and randomly orientated.

In Figure 3-12 SEM micrographs from two different cartilage specimens taken at three equivalent magnifications are compared. Although one of the specimens

had been used for a stationary load friction test prior to preparation for SEM the micrographs generally revealed similar surface topographies. For the previously friction tested specimen at the lowest presented magnification (originally x200) small ripples or ridges due to large superficial collagen fibre bundles were evident but not abundant, top right micrograph Figure 3-12. At higher magnifications, although the cartilage surfaces could not be described as perfectly smooth, due to the prevalence of superficial collagen fibrils, the level of roughness attributable to them from examination of these relatively high resolution images was of sub-micron magnitude. With the use of more elaborate fixing procedures the surface of the articular cartilage has appeared even smoother using scanning electron microscopy (Bloebaum and Wilson 1980; Bloebaum and Radley, 1995) possibly due to the interfibrillar ground substance being better preserved and covering even the uppermost superficial fibrils.

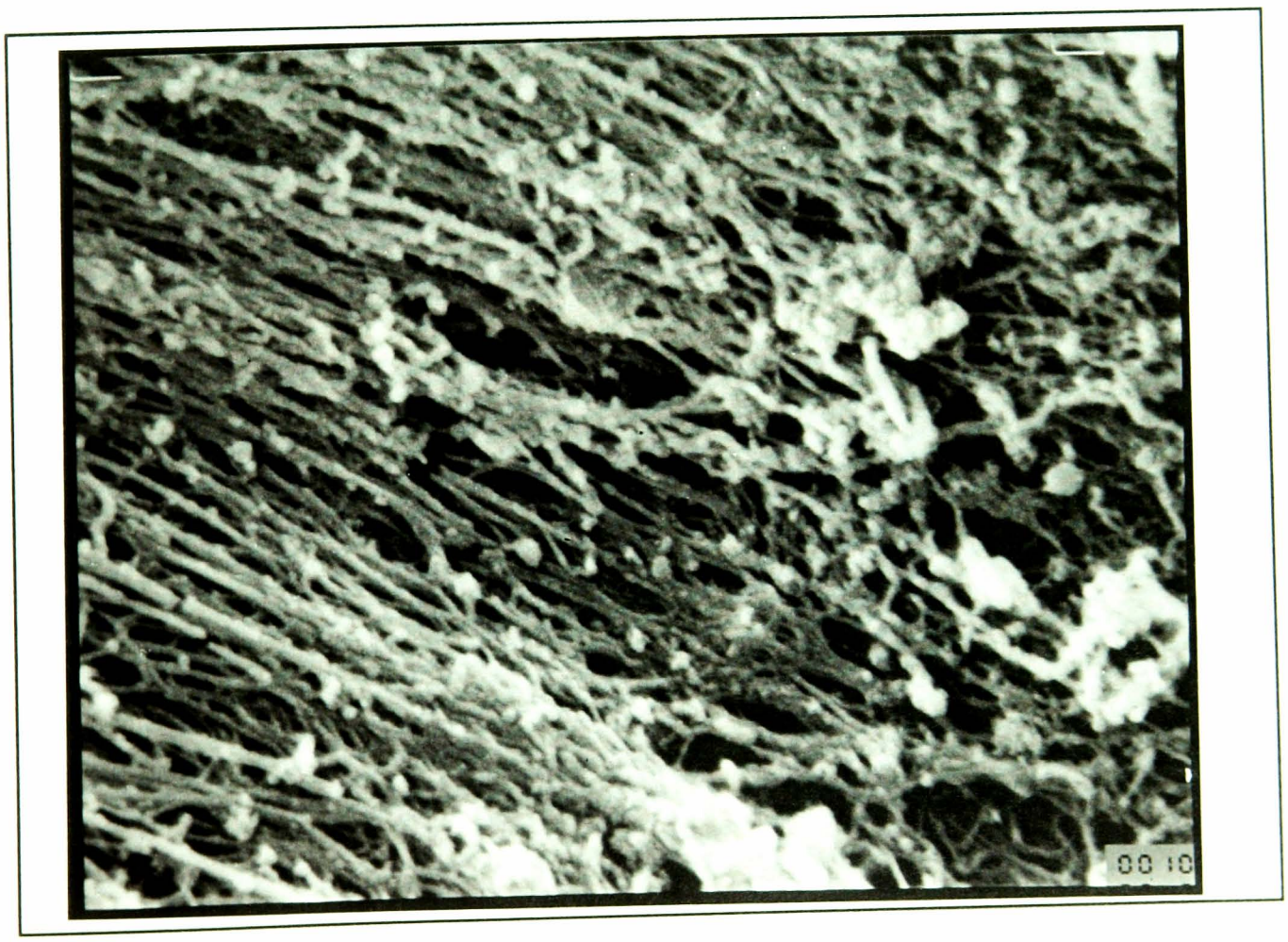


Figure 3-11 Scanning electron microscopy image of the mid-section of the superficial tangential zone I, revealing highly orientated collagen fibrils. The two thin white mark spacers at the top of the image are a distance of 10 μm apart. Inclination was 70° to the horizontal. (Original magnification was x10 000).

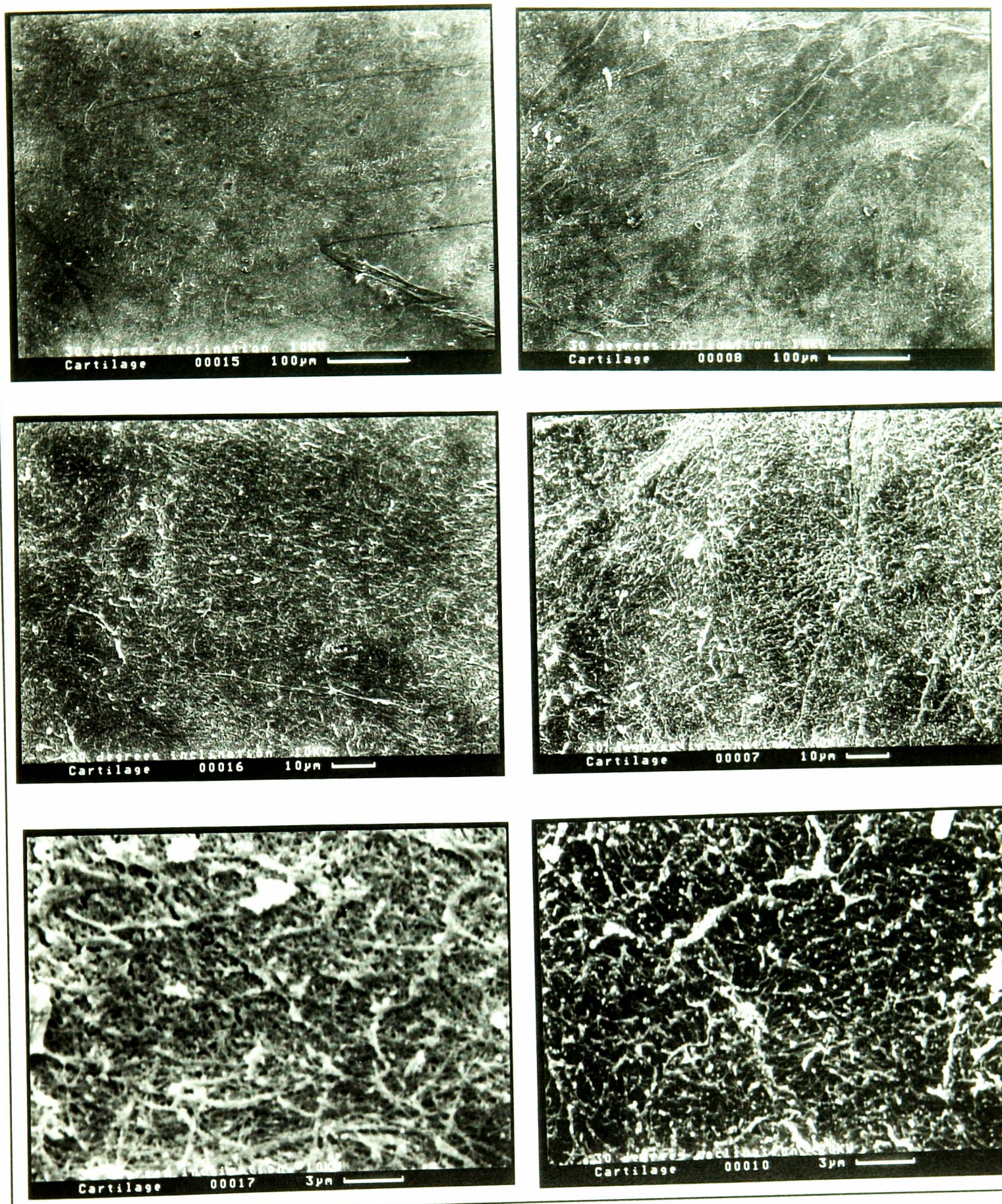


Figure 3-12 Scanning electron microscopy images of two cartilage specimens at three equivalent magnifications (originally x200, x1 000 & x5 000 running to bottom in the figure). Scale bars and inclination to the horizontal are provided on each micrograph. Micrographs on the left are of an untested cartilage specimen and the micrographs on the right are of a cartilage specimen previously used for a stationary load friction test on a metal counterface.

3.3.4 Environmental Scanning Electron Microscopy

The cartilage specimens examined by ESEM were often fresh samples viewed as quickly as possible following their collection from the abattoir, Table 3-3. The cartilage specimens required no preparation and could be viewed in a state of full hydration. After viewing the specimens for up to 45 minutes in the ESEM while trying to maintain a steady dew point, upon removal from the specimen chamber, there were no obvious signs of dehydration although the surface was observed to be dry. This was a necessary condition for clear imaging of the cartilage surface. Even so the specimens were found to be cool and moist to the touch.

Upon initial visualisation of the cartilage surface the obtained images appeared to be 'floating' across the ESEM monitor. This was caused by water on the cartilage surface which evaporated as the dew point was reached. As the image stabilised, aside from the occasional white specks as mentioned in the previous section, a generally featureless articular cartilage surface was evident as described by Kirk et al. (1993b). As all the surface water evaporated, and possible dehydration of the cartilage surface occurred, features in the form of porous holes were revealed, Figure 3-13. Occasionally subsurface light grey spheres were seen, indicative of chondrocytes, as previously observed for conventional SEM, Figure 3-10. Note that in the micrographs the scale bar was the white region only and did not include the black area detailing the length of the scale bar, the original magnifications were also provided on the micrographs.

For specimen 2, a fresh cartilage plug from the carpal/metacarpal anatomical locations, after first observing the specimen at the dew point the specimen chamber temperature was increased in order to slowly dehydrate the specimen. Within a time scale of 20-30 minutes widespread 'pitting' of the cartilage surface was seen, shallow hollows ~20 μm in diameter, and attributed to the collapse of superficial chondrocytes. During this period high resolution focusing (≥ x2 000

magnification) in one area of the cartilage surface, for prolonged periods, created distinct areas of electron beam damage.

As explained in Table 3-3 a cartilage specimen (6) previously prepared for routine SEM using conventional means of tissue fixation and dehydration was also analysed by ESEM. Using the ESEM, features such as those observed with the SEM were evident. Furthermore however a 'surface coat' was observed which may have simply been missed when analysed using the SEM or under the less harsh environment of the ESEM was easier to visualise. A surface layer was shown to be partially removed and have a thickness of ~0.5 μm. Where it was intact it covered the underlying, highly orientated, superficial collagen fibres and a certain amount of porosity could be ascribed to it, Figure 3-14. In Figure 3-14 a micrograph from a fully hydrated cartilage sample at a similar magnification is also displayed. In this bottom micrograph porosities were again apparent in what could well have been a similar surface coat but in its native state of full hydration. Aside from the porosities this surface coat was completely covering the underlying features of the cartilage. Occasionally sub-surfaces features in the form of light grey spheres, or shallow depressions for the mild osteoarthritis specimen, indicative of superficial chondrocytes, were evident, but the surface coat remained intact. Whether or not the porosities were an intrinsic part of the cartilage surface coat or an artefact produced by electron beam damage or partial dehydration was not known.

Also in the bottom micrograph of Figure 3-14 the light grey and white specks observed using conventional SEM, as mentioned, were more numerous and commonly seen to aggregate in clusters. Again these particles could easily have been just the residue of the synovial fluid macromolecules. Alternatively they could have been a more deliberate feature marking the presence of a boundary layer, especially as the cartilage samples were stored in Ringer's solution for at least 24 hours before viewing by ESEM.

Specimen 3 was wiped firmly using a Ringer's solution moistened tissue prior to viewing in the ESEM. The surface coat was largely removed with both superficial chondrocytes and collagen fibres evident. The light specks lying on the surface of the cartilage, as mentioned above, were far less abundant.

Figure 3-13 Environmental scanning electron microscopy micrographs taken at similar magnifications. The top micrograph shows a featureless articular cartilage surface as described by Kirk et al. (1993b). As all the surface water evaporated and possible dehydration of the cartilage surface occurred features in the form of porous holes began to appear (bottom).

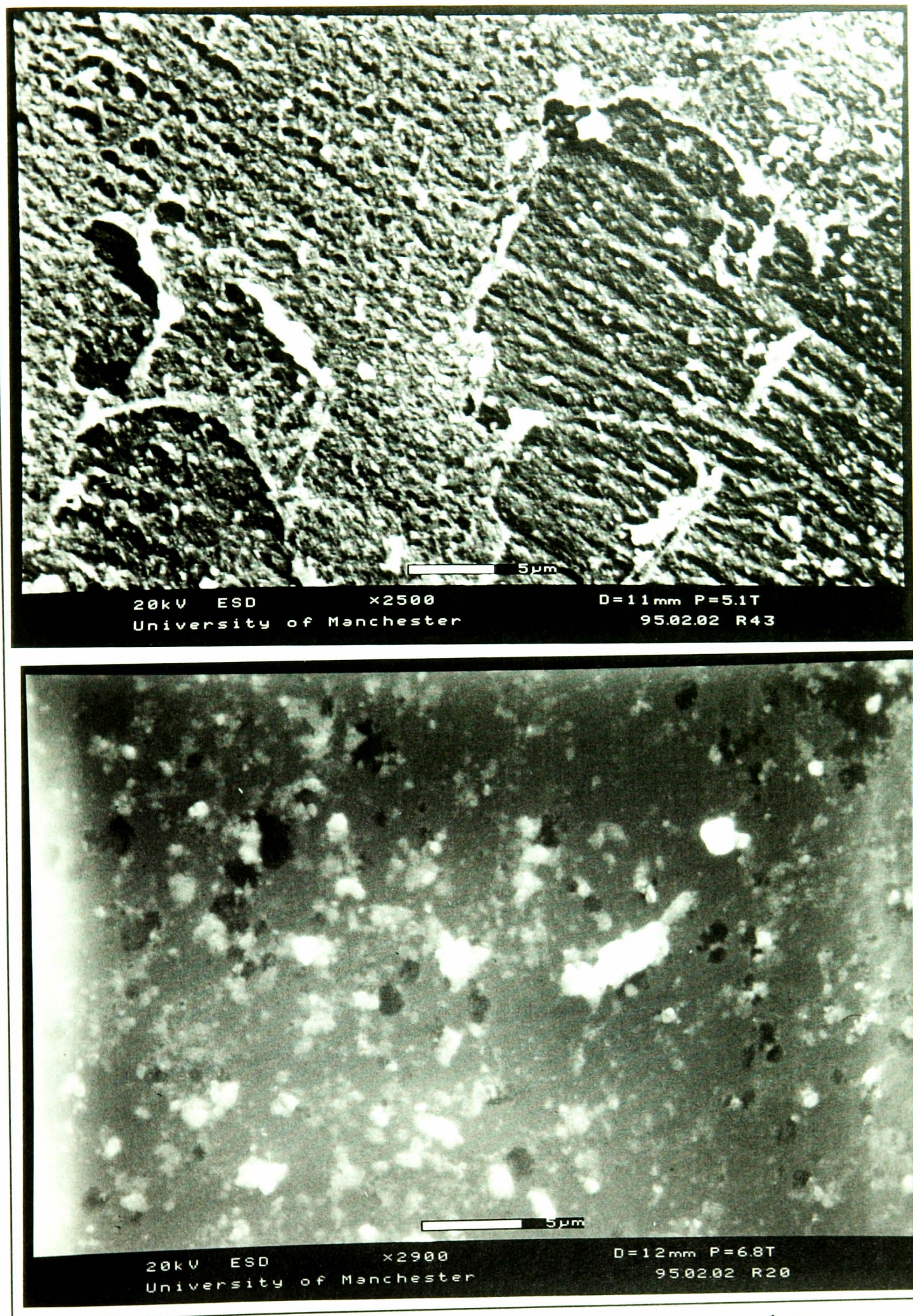


Figure 3-14 Environmental scanning electron microscopy micrographs taken at similar magnifications. The top micrograph shows a conventionally prepared SEM cartilage specimen having a partially removed 'surface coat' and underlying collagen fibres are seen. The bottom micrograph was taken from a cartilage specimen requiring no prior preparation for ESEM. The 'surface coat' has remained largely intact with some porosity evident.

While using the ESEM electron beam damage was still observed to take place, for the conventionally prepared SEM specimen 6, particularly at the higher resolutions. As the image used to record the micrograph in Figure 3-15, of specimen 6, was being focused the highly orientated collagen fibre bundles (≤1 μm thick) could be seen to separate apart as the interconnecting ground substance disintegrated and the fibre bundles themselves began to break down into their constituent fibrils.

For reasons explained in section 3.2.4 the underlying bone had to be sectioned down as much as possible prior to placement in the ESEM specimen chamber. For specimen 1 the sectioning resulted in the accidental removal of all underlying bone at the outer edge of one side of the cartilage plug. Upon viewing this side of the cartilage plug a 'wrinkling effect' was observed, while the remaining areas of the plug remained predominantly featureless. This phenomena was previously witnessed by Ghadially et al. (1982). As the cartilage curled up due to the intrinsic tension in the collagen fibres and no longer being restrained by the subchondral bone, superficial collagen fibre bundles are compacted and forced to protrude above the normally smooth cartilage surface producing the observed ridges, Figure 3-16.

For the specimens (1 & 3) which had previously been examined by stylus profilometry, tracing line profiles of 6 mm in length, no permanent deformation or wear track caused by the 2.5 μm conisphere diamond tipped stylus could be recognised.

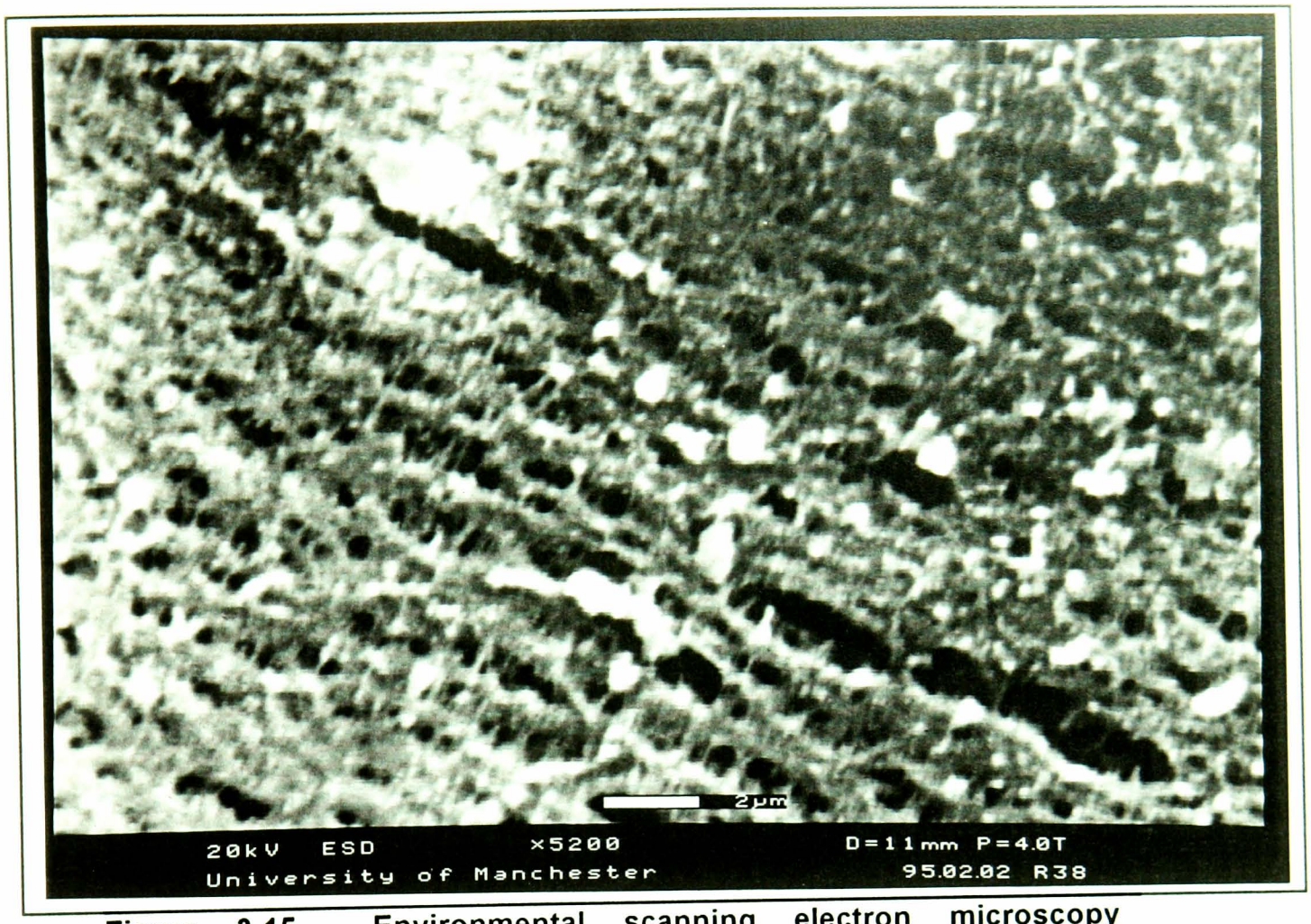


Figure 3-15 Environmental scanning electron microscopy micrograph of the superficial collagen fibrils at high resolution, specimen 6.

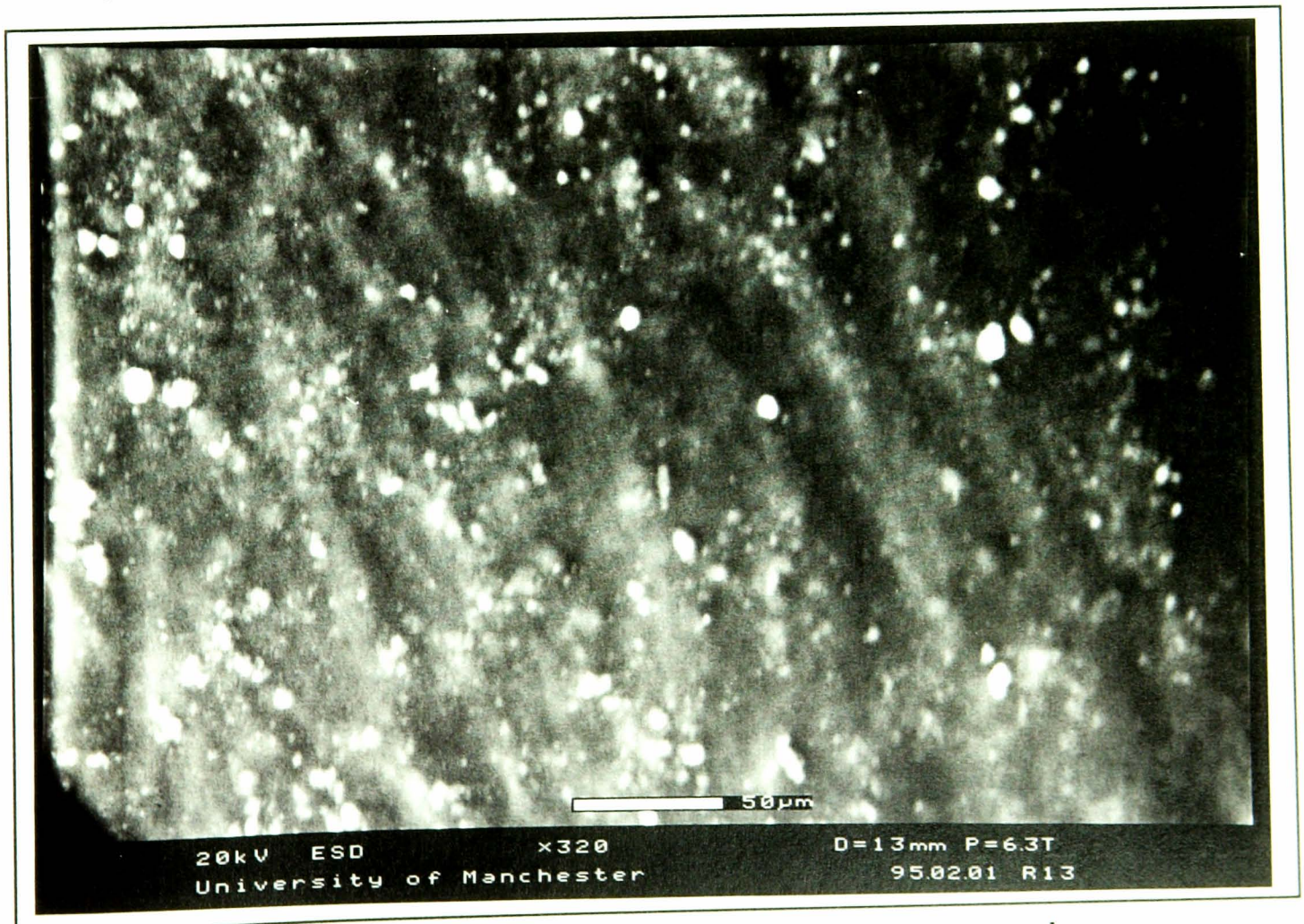


Figure 3-16 Environmental scanning electron microscopy micrograph revealing a wrinkled appearance due to bone detachment as described by Ghadially et al. (1982).

3.3.5 Transmission Electron Microscopy

In Figure 3-17 the TEM micrographs provided an impression of the surface texture of articular cartilage, over a length of approximately 100 μm, from two cartilage specimens at the same magnification of x700. Note that the quoted magnifications are true to the size of the micrographs within ±5%. From these two micrographs the estimated level of roughness appeared to be ~ 1-2 μm. In both of the micrographs typical superficial tangential zone chondrocytes are shown, positioned at least 10 μm below the cartilage surface. They were oval in appearance, flattened parallel or 'tangentially' to the cartilage surface.

The existence of a osmiophilic transitory boundary layer, ranging from 0.03-0.50 μm in thickness, was readily identified, Figure 3-18. This layer may have been the surface coat or alternatively part of what was visualised as nodular white specks resting on the surface of the articular cartilage, both of which were referred to in sections 3.3.3 and 3.3.4. A third possibility is that of introduced artefact, however support from previous studies, as discussed below, points to a distinctive layer existing uppermost on the cartilage surface, above the superficial tangential zone. In both of the micrographs in Figure 3-18 the layer appears to be somewhat detached from the underlying collagen matrix. This can be accounted for by the lack of staining for the inter-fibrillar ground substance of the cartilage's solid matrix, which incidentally allowed the collagen fibres to be so clearly defined. The ground substance is thought to cover the superficial collagen fibres at the surface with a thin coat of ~0.2-0.4 μm in thickness (Orford and Gardner, 1985), as depicted schematically in Figure 3-20. This overlay by the ground substance could therefore be the 'surface coat' displayed in Figure 3-14 and explain the small gap between the boundary layers in Figure 3-18 and the underlying collagen matrix, by its failure to stain.

In the top micrograph of Figure 3-18 an osmiophilic electron dense lipidic body was evident, ~0.2 μm in diameter, lying about 1 μm from the surface. These bodies, by-products of cellular metabolism, were also sometimes evident at the cartilage surface, as previously reported (Meachim and Roy, 1969; Ghadially et

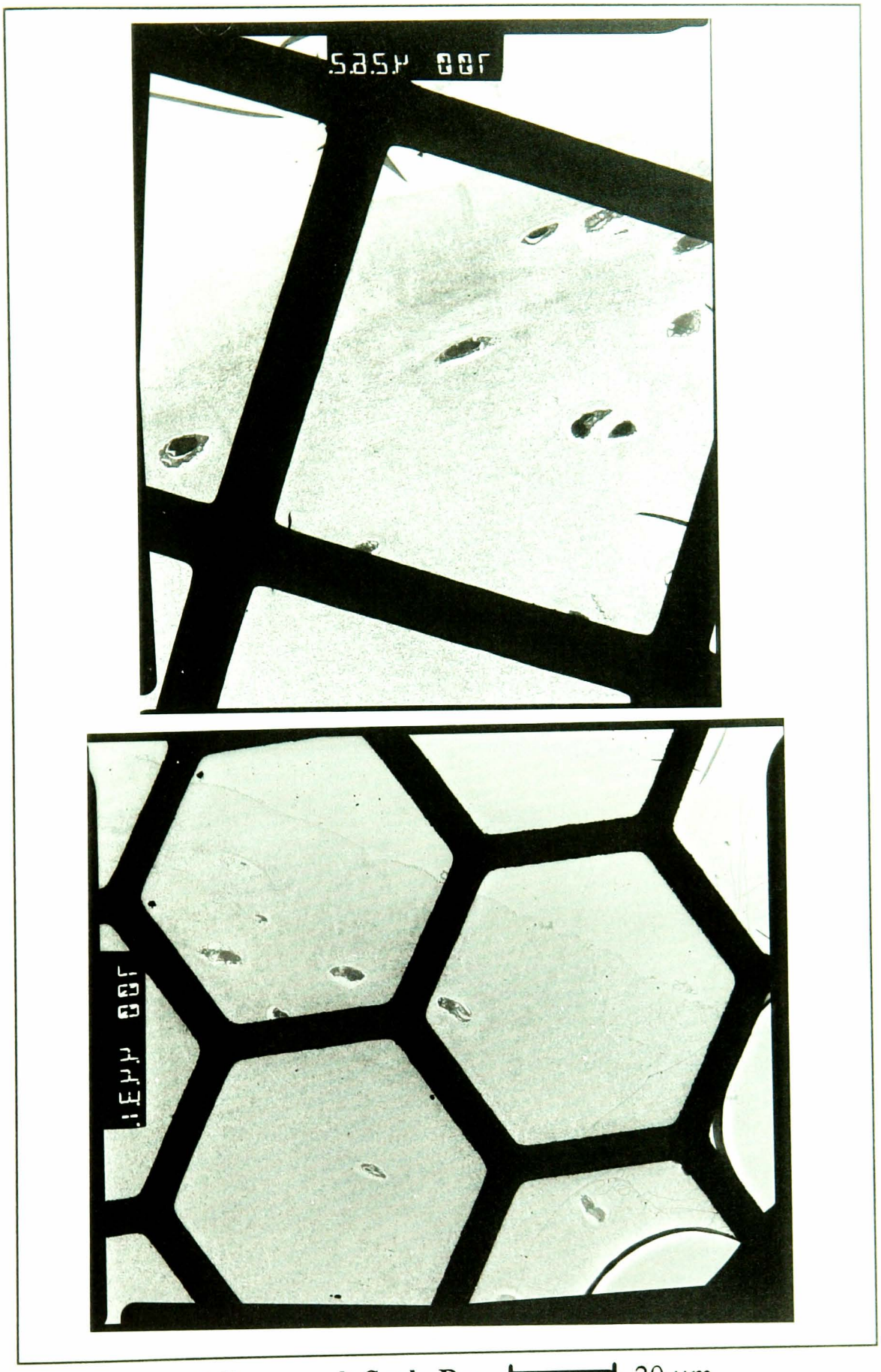
al., 1982), and may contribute to the constituents of the boundary layer. The bottom micrograph of Figure 3-18 revealed the closely packed, highly orientated arrangement of the superficial collagen fibrils, ~20-30 nm in diameter. Electron-dense granular elements were evident as small black dots, slightly smaller in diameter than the fibrils themselves, and were seen in close association with the collagen fibrils. These features may be evidence of binding sites between the proteoglycan ground substance and the collagen fibrils (Meachim and Stockwell, 1979). This was in good agreement with previous SEM images. Although for the SEM images the fibrils were quoted as being 50-100 nm in diameter, from a micrograph taken at x10 000, Figure 3-11. The fibril diameters quoted from TEM analysis, ~20-30 nm, were considered to be more accurate as higher magnification was possible and individual collagen fibrils were more easily distinguished.

As for most surfaces when imaging at sufficiently high magnifications, as in Figure 3-18, the surface could generally appear to be very smooth. However, occasionally even at these high magnifications the prominence of superficial collagen fibre bundles was evident, as seen in Figure 3-19. In Figure 3-19 the pits and humps observed at the surface were 1 μm in magnitude. It was arguable as to whether or not these features were artefacts, created by the specimen preparation procedures, or inherent to the cartilage surface. The boundary layer was shown not to be a continuous attribute of the surface, Figure 3-19. From the three bovine femoral condyle specimens investigated it was estimated that approximately 15-30% of the articular surface was covered by the boundary layer. Again whether or not the layer is fully intact *in vivo* remains an uncertainty.

In 1951 MacConaill was the first to mention the existence of a distinct noncollagenous surface layer resting above what is now commonly referred to as the superficial tangential zone. Using phase contrast light microscopy a bright line was observed at the cartilage surface of sub-micron thickness. It was named the lamina splendens which literally means 'shining layer'. It was later described as an optical effect artefact caused by the phase contrast microscopy (Aspden and

Hukins, 1979). However many independent studies have since confirmed the presence of a distinct noncollagenous surface layer using both SEM (Jeffery et al., 1991; Kobayashi et al., 1995) and TEM (Davies et al., 1962; Barnett et al., 1963; Balazs et al., 1966; Weiss et al., 1968; Meachim and Roy, 1969; Stanescu and Leibovich, 1982; Ghadially et al., 1982; Orford and Gardner, 1985; Kirk et al., 1993a; Shaw and Molyneux, 1994). These studies generally ascribed a thickness of less than one micrometer to the layer. Some studies have however referred to the layer as acelluar but consisting of collagen fibrils and being ~5 μm thick, (Teshima et al, 1995).

For such a volume of literature it must now be generally recognised that a surface layer exists uppermost on the articular cartilage surface and is distinct from the underlying superficial tangential zone I region of the tissue. This layer as well as potentially playing a key role in synovial joint lubrication, especially boundary, must also be considered when discussing both the permeability and nutrition of the articular cartilage.



Micrograph Scale Bar ⟝⟞ 20 μm

Figure 3-17 Transmission electron microscope images taken from different cartilage specimens. The magnification was x700 for both micrographs. The oval appearance of the uppermost chondrocytes in the superficial tangential zone is demonstrated and a reasonable interpretation of surface roughness across ~100 μm can be made.

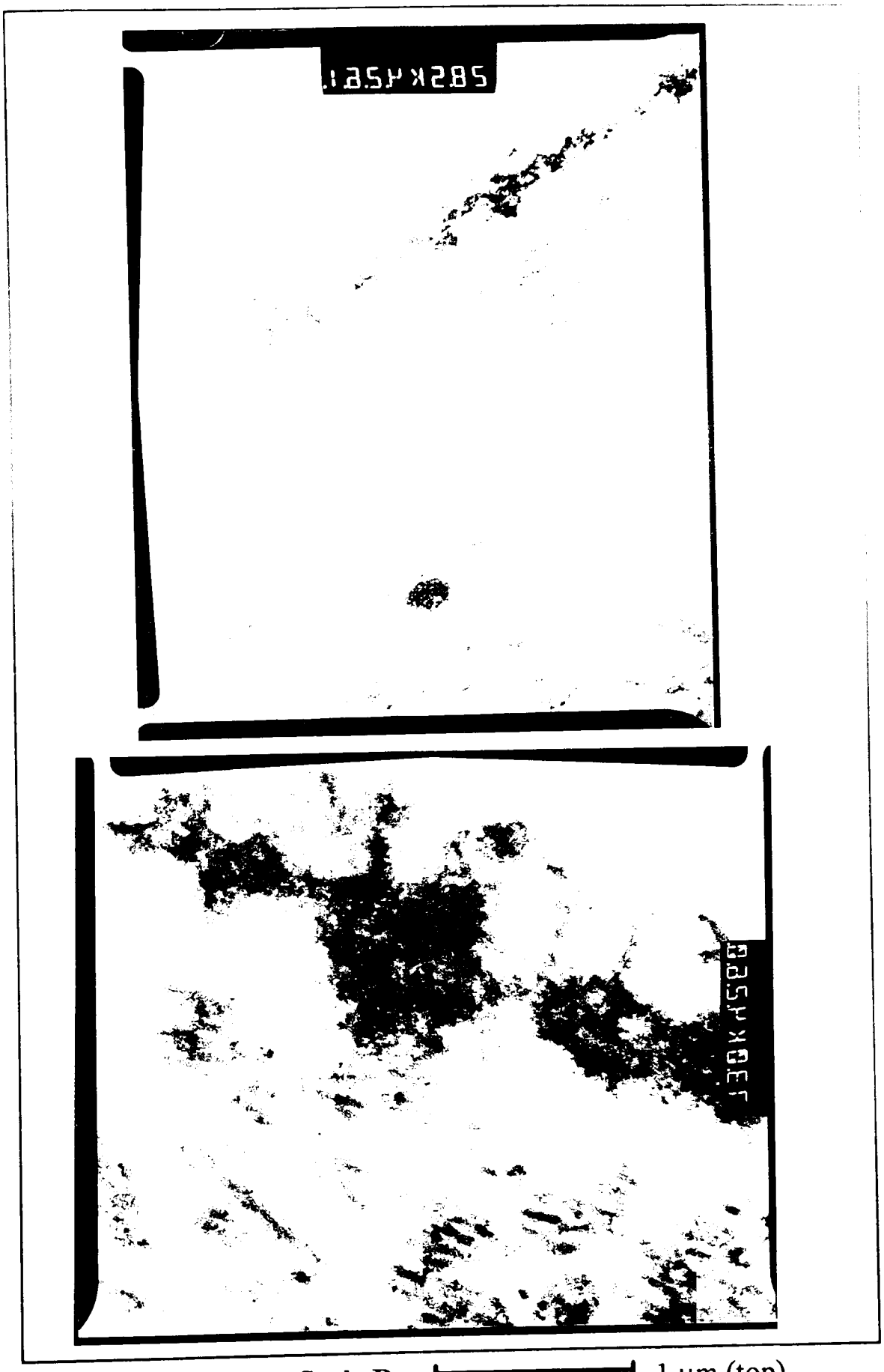
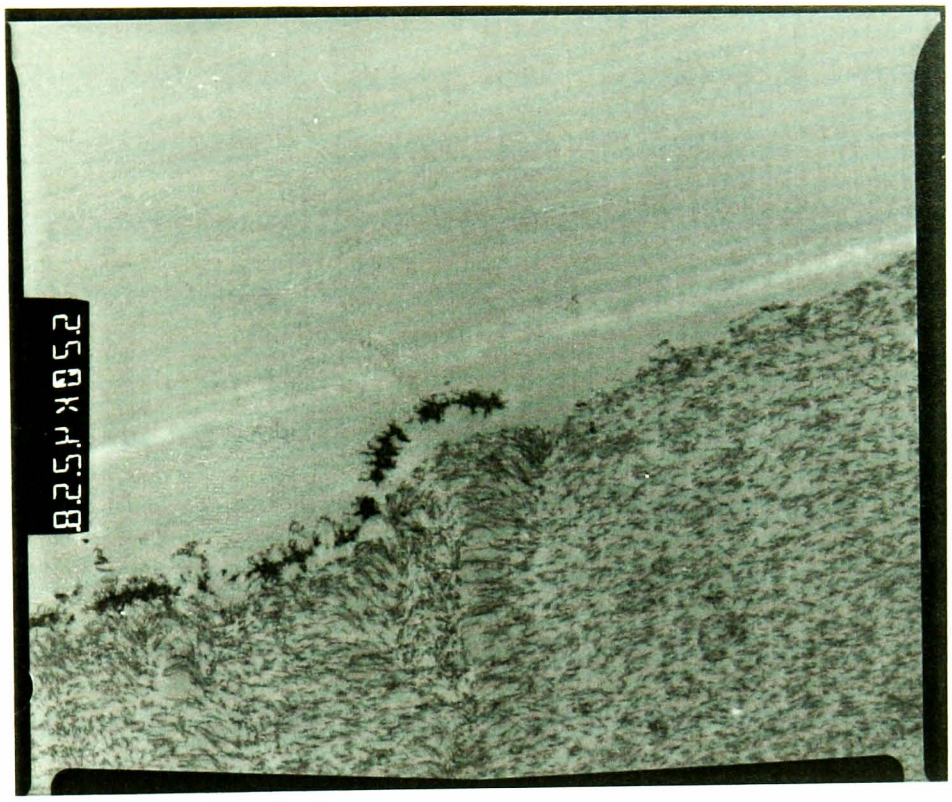


Figure 3-18 Transmission electron microscope images of the 'surface coat' or boundary layer of bovine articular cartilage at magnifications of x28 500 (top) and x73 000 (bottom). The boundary layer in these TEM micrographs was approximately 0.25-0.50 μm thick.



Micrograph Scale Bar ⊢——⊣ 5 μm

Figure 3-19 Transmission electron microscope image of the cartilage surface at x5 200 magnification. As shown, the boundary layer was found not to cover the entire surface area of the specimens prepared for TEM.

3.3.6 Overview of Previous Studies

The characteristics of the superficial tangential zone of articular cartilage are critical to the understanding of the lubrication and wear processes occurring in diarthroses and perhaps the anomalies associated with arthritic disorders. Literature on the morphology and composition of the articular surface is at present inconsistent and contradictory (Kirk et al., 1993a). In order to elucidate the observations of this investigation some prominent studies in this field have been briefly reviewed.

The cartilage surface has traditionally been described as being extremely smooth when examined by a number of techniques from a hand lens (Hunter, 1743) and a light microscope (Toynbee, 1849) to transmission electron microscopy, TEM, (Barnett et al., 1961). However, studies using scanning electron microscopy, SEM, (Walker et al., 1968; Gardner, 1972; Gardner et al., 1981), reflected light interference microscopy (Gardner and McGillivray, 1971) and surface profilometry (Walker et al., 1968; Thomas et al., 1980) revealed a large number of surface features present, such as pits, humps and ridges.

From these studies it became generally accepted that normal (i.e. young healthy adult) articular cartilage had an average surface roughness, Ra, of ~1-6 μm. Gardner (1972) and Clarke (1971) described the pits and humps on the surface as the result of the necrosis and swelling, respectively, of underlying chondrocytes. While Walker et al. (1968) attributed the cartilage surface roughness to prominent collagen fibre bundles in the superficial zone.

Further investigations have shown that the cartilage surface is sensitive to dehydration (Bloebaum and Wilson, 1980; Speer et al., 1990; Kirk et al., 1993b); separation from the subchondral bone (Ghadially et al., 1982); wiping or dabbing of the surface and immersion in hyperosmolar solutions (Bloebaum and Wilson, 1980); and integrity of adjacent cartilage (Ghadially, 1983). The effects of age (Longmore and Gardner, 1975) and species (Ghadially, 1983) have also been noted.

Articular cartilage is constituted of approximately 60-87% water by weight (Mow et al., 1991) and dehydration, without appropriate substitution, is very likely to affect the surface morphology. There is, consequently, concern over the authenticity of observed surface features from fixed, dehydrated tissue (Draenert and Draenert, 1978; Hippe-Sanwald, 1993) or tissue in an uncertain state of hydration, particularly when using the SEM (Levanon and Stein, 1991; Clark and Rudd, 1991). Various measures have been taken to overcome such difficulties but the conclusions remain conflicting (Bloebaum and Wilson, 1980; Gardner et

al., 1981; Ghadially, 1983). Some workers clearly believe that the 'irregularities' on cartilage surfaces are entirely attributable to artefact and not present *in vivo* (Ghadially, 1983; Kirk et al., 1993b), perhaps for both loaded and unloaded conditions.

Using a method referred to as Environmental SEM, ESEM, (Kirk et al., 1993b), not requiring extensive specimen preparations, it was noted that the surface of fully hydrated, non-weight bearing articular cartilage was 'smooth' at all magnifications and that surface features quickly formed upon dehydration. It was further commented that examination at too high an accelerating voltage causes sub-surface, rather than surface, features to appear, explaining other workers' observations.

Such findings would appear to discount lubrication theories based on a rough articular surface, weeping and boosted lubrication for instance, unless of course this transient uppermost layer is capable of forming tiny depressions under load.

For normal articular cartilage a *surface coat* of generally 0.03-0.1 μm thickness, but reaching 1 μm at times, has been described (Ghadially, 1983) and further supported by similar observations (Orford and Gardner, 1985; Kirk et al., 1993a). This layer must be of primary importance when considering boundary lubrication mechanisms in diarthrodial joints. If it does exist it must either be the boundary layer itself or be responsible for the adsorption and/or maintenance and protection of the boundary layer.

From a meticulous TEM analysis of rabbit articular cartilage (Ghadially et al., 1982), it was apparently shown that the surface was remarkably smooth. The 'defects' or 'asperities' were described as being confined to the surface coat (which is a transient structure anyway) and only rarely did they attain a depth of 0.03 μm. On rarer occasions, defects reaching a depth of ~0.15 μm were found. It was concluded that even small surface irregularities described by some workers (Davies et al., 1962; Barnett et al., 1963; Weiss et al., 1968) were artefacts.

Weiss et al. (1968), for example, had mentioned the presence of somewhat larger depressions "usually less than 0.3 micron (μm) but occasionally up to 1.5 micra (μm) in depth".

The 'stable' or 'permanent' cartilage surface was thought to be the one bounded by the most superficial collagen fibrils. While the surface coat was an ephemeral or transient structure which varied in thickness and density and could be shed or rubbed off (Ghadially, 1983). For this reason it was termed the surface coat.

The composition of this electron-dense particulate and filamentous coat was unknown but it was suggested (Ghadially, 1983) that this coat may be a combination of substances derived from the synovial fluid and/or material extruded from the cartilage (e.g. matricial lipidic debris (Ghadially et al., 1965) and degraded metabolites). The studies of Balazs et al. (1966) on bovine articular cartilage suggest that the surface coat may contain hyaluronate, but Meachim and Stockwell (1979) quote several studies, including their own, which casts doubt on this thesis.

A TEM study (Orford and Gardner, 1985) involving staining with Cupromeronic blue and digestion with testicular hyaluronidase revealed two collagen deficient ultramicroscopic layers above the superficial tangential zone I (Figure 3-20). The uppermost layer, ~50 nm thick, was thought to be either glycoprotein or protein in nature. The deeper layer, ~100-400 nm thick, was a chrondroitin sulphate rich proteoglycan layer and structurally similar to the interfibrillar matrix of zone I. It was suggested that these layers may be important in lubrication, permeability and compression resistance of the superficial cartilage zone.

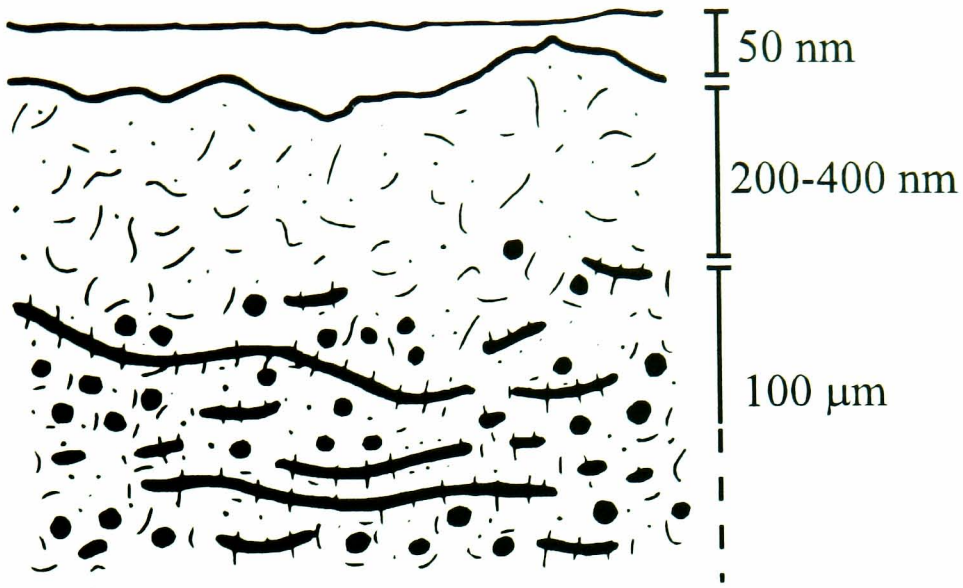


Figure 3-20 The articular cartilage surface as described by Orford and Gardner (1985). Thick lines and heavy dots represent collagen fibrils. Thin lines and light dots represent glycosaminoglycan complexes, generally associated with the interfibrillar network of the cartilage solid matrix.

Recent work (Kirk et al., 1993a) has independently reported similar surface layers to that described by Orford and Gardner (1985). Using light microscopy and TEM with energy dispersive spectroscopy (EDS) a 10 μm thick acellular "superficial stratum" on the surface of articular cartilage was reportedly observed. This stratum consisted of short randomly orientated collagen fibres and interstitial substances, c.f. Figure 3-20. It was found that the collagen fibres were much more sparsely distributed than in the underlying cartilage matrix.

A very thin transitory layer, ~100-400 nm thick, uppermost on the acellular stratum was described, containing high concentrations of phospholipids. Oligolamellar phospholipid has previously been described as evident on the cartilage surface (Hills, 1990), and proposed as the boundary lubricant as mentioned earlier in Chapter 1.

The latest SEM studies (Kobayashi et al., 1995; Kobayashi et al., 1996) have employed cryo-SEM. For this technique pig and human articular cartilage specimens from knee joints were frozen in liquid nitrogen, often during application of load via a flat-ended cylindrical indenter 3.5 mm in diameter, and

freeze-fractured perpendicularly to the articular surface. The specimens were defrosted with radiant heat (thermal etching - to remove the loose bound interstitial fluid), gold coated, and observed, using a 5 kV accelerating voltage, within the microscope specimen chamber which was kept below -160°C. This method was considered to reduce the introduction of artefacts due to fixation and drying necessary for conventional SEM and TEM. Kobayashi et al. (1995, 1996) again identified a distinct amorphous, noncollagenous, acellular surface layer which they called the 'surface amorphous layer'. While in the previous reports, mentioned above, the thickness of the surface layer ranged from several nanometres to several micrometres, it ranged from 2 to 200 μm in the specimens examined by cryo-SEM. It was speculated that the discrepancy was caused by dehydration of the layer or dissolution of its constituents during preparation of specimens for examination by other methods. Undulations on the surface due to superficial chondrocytes and fibre bundles were reported not to be present, being covered by the surface amorphous layer. The morphological appearance of the layer and its thickness were found to be closely related, both being dependent upon the surface layer's fluid content. The response of the surface layer to and recovery from indentation was largely instantaneous and elastic. It was considered to be highly permeable, absorbing or discharging fluid rapidly when a load or pressure gradient was applied to (or removed from) it. Around the indenter, the surface layer expanded forming a triangular wedge shaped appearance in cross-section, possibly due to fluid expressed from beneath the indented region of the cartilage solid matrix. The surface layer, although substantially compressed under load, was always present beneath the indenter covering the articular surface. Upon load removal the previously compressed surface layer rapidly expanded to cover over the indented cartilage solid matrix which was slower to recover its fluid and form following load removal. Similar observations have previously been made, again using a freeze-drying method (Walker et al., 1970). In this study the layer was described as an aggregation of the synovial fluid's hyaluronic acid protein complex which formed on the cartilage surface. The layer's perceived behaviour under load again matched that described by Kobayashi and co-workers. Kobayashi et al. (1995, 1996), having rinsed and

stored their cartilage specimens in physiological saline solution prior to freeze-fracturing, without the presence of synovial fluid, considered the layer to be an inherent feature of the articular cartilage surface.

A study by Gardner et al. (1981) using a cyro-SEM to view unfixed, fully hydrated frozen cartilage surfaces still described the cartilage surface as rough, with linear ridges of diameter ~1-2 μm. This was somewhat contradictory to the observations of Kobayashi et al. (1995, 1996). Surface features were generally commented to be similar as those previously observed by conventional SEM (Longmore and Gardner, 1975). Ghadially et al., (1982), believing the cartilage surface to be essentially smooth, questioned these findings stating that dehydration under the electron beam or distortion due to freezing may have occurred. During viewing in the cyro-SEM dehydration and crystal formation artefacts were discounted (Kobayashi et al., 1996). However, due to the inhomogeneous, anisotropic properties of articular cartilage, concerns regarding the authenticity of SEM images from rapidly frozen cartilage specimens, which may cause considerable distortion, have been put forward (Sayles et al., 1979; Ghadially et al., 1982).

In indentation studies it has been discovered that upon load removal rapid fluid resorption only occurs if both the load and indenter are completely removed. If the load is removed but the indenter is permitted to remain on the surface, leaving only the smallest of loads upon the cartilage surface, fluid resorption is suppressed (Torzilli et al., 1983). The compression of the surface layer, which is probably caused by only the smallest of loads, therefore may be decreasing the permeability of the cartilage surface, and consequently reducing fluid uptake. This conclusion has both important biomechanical and biotribological implications.

The studies carried out to chemically analyse the articular cartilage surface have all focused on the proteoglycans and more specifically the glycosaminoglycans (inherent to the interfibrillar part of the cartilage solid matrix), of the superficial tangential zone I. These studies were probably not, therefore, sensitive to the boundary layer. Several biochemical techniques (Maroudas et al., 1973; Franzén

et al., 1981; Bayliss et al., 1983; Kiviranta et al., 1987; Manicourt and Pita, 1988) as well as X-ray microprobe analysis (Maroudas, 1972) have been utilised in these investigations.

3.3.7 Analytical Technique Summary

Table 3-4, below, briefly summarises the advantages and disadvantages of the various techniques used in this study to analyse the surface of articular cartilage.

Technique	Advantages	Disadvantages
Stylus Profilometry	Fully quantitative data from 2-D line profiles. The bulk and surface of the cartilage tissue may remain hydrated. Constant irrigation by Ringer's solution or synovial fluid was allowed.	Possibly not sensitive to the surface layer due to instantaneous elastic deformation caused by the stylus tip. Some elastic deformation of the cartilage matrix may also occur. Does not provide 3-D images of the cartilage surface.
Laser Profilometry	Non-contacting profilometry. Fully quantitative information and data from area and individual line scans of the surface. 3-D quantitative visualisation of the surface. The bulk of the cartilage tissue may remain hydrated.	Surface must be water free. Scans require time to perform; limited to 5 minutes to prevent surface dehydration. Cannot 'look over' surface as readily as with other methods. One micrometer spot size limits resolution, making individual surface features ambiguous and difficult to discern. Reflectivity artefacts apparent.
SEM	High resolution with good depth of field. Fully manipulative 3-D images obtained while viewing specimens.	Meticulous specimen fixing, dehydration, drying and gold coating required. Dehydration without substitution by another material may significantly distort features, (Ghadially et al., 1982). Electron beam damage. Sub-surface features may appear at high accelerating voltages.
ESEM	Same as for SEM with the benefits of being able to view specimens fully hydrated without any prior preparation.	Surface must be water free. Difficult to control the dew point - consequently either slowly dehydrating or flooding the cartilage surface. Specimen chamber temperature and pressure must be constantly monitored. Electron beam damage. Sub-surface features may appear at high accelerating voltages. Focusing often difficult.
TEM	Highest resolution capabilities of all the techniques used. The water removed from the cartilage matrix is substituted by another material, lessening the effects of dehydration. Features underneath the cartilage surface also visualised, providing they stain. Qualitative inspection of roughness possible.	Meticulous specimen fixing, dehydration, infiltration, embedding, sectioning and staining required. Ultrathin sections through the thickness of the articular cartilage provided essentially only a 2-D perspective. Sections were of limited size. Features are only seen if they stain sufficiently with the applied staining agent. This however is also an advantage as selective use of staining agents can indicate the histochemical composition of various features.

Table 3-4 Summary of advantages and disadvantages of techniques used to characterise the surface of articular cartilage.

3.3.8 Implications for Joint Lubrication Friction Studies

Previously biotribologists have tended to think of the articular cartilage boundary layer as being present solely due to adsorption of macromolecular constituents from the synovial fluid. Some workers considered that this assumption made the investigation of synovial joint boundary lubrication by using synovial fluid as a lubricant and synthetic rubbing materials only in friction tests a viable option, despite remoteness from the physiological condition (McCutchen, 1966; Wilkins, 1968; Reimann et al., 1975; Davis et al., 1979; Jay, 1992; Williams et al., 1993). This alternative presumed that articular cartilage itself could be dispensed with, being too time consuming and lacking reproducibility (Davis et al., 1979; Hills and Butler, 1984). However the surface layer observed and discussed in this study is most likely to be directly involved with the mechanism of boundary lubrication, if it is indeed not the boundary layer itself. Studies which seek to address the nature of boundary lubrication in synovial joints must therefore utilise cartilage specimens in their friction testing. The strong possibility that a boundary layer exists which is inherent to the articular cartilage has only rarely been acknowledged by biotribologists (Little et al., 1969; Forster et al., 1995; Higaki et al., 1995). Previous studies, even those using cartilage specimens, have looked at ways of chemically degrading or altering specific components of the synovial fluid in order to identify the boundary lubricant. Thus ignoring the possibility of an intrinsic cartilage boundary layer and incidentally sometimes ignoring the effect of additive chemicals in the synovial fluid upon the cartilage layer and the bulk of the cartilage tissue itself.

An understanding of cartilage surface roughness during the design of friction experiments investigating mixed and boundary synovial joint lubrication is necessary in order, for example, to theoretically predict when fluid film breakdown and asperity contact occurs (Unsworth, 1993). Some analytical models of joint lubrication also require quantitative assumptions regarding surface texture parameters, e.g. micro-EHL (Dowson and Jin, 1986). In the

interpretation of results from friction studies cartilage roughness is again an important consideration. Previous lubrication theories, notably boosted lubrication (Walker et al., 1968), were based upon the assumption of a rough cartilage surface, ~2-4 μm in magnitude.

By comparing surface integrity between fresh and previously frozen cartilage specimens additional validation of the adopted collection and storage procedures for articular cartilage used in the friction tests could be made.

3.4 General Comments and Conclusions

The observations for each particular technique employed in this study were often found to coincide and confirm findings from other studies. Any discrepancies which came to light when comparing results between each of the techniques can be understood by reflecting upon the different benefits and drawbacks of each particular method, Table 3-4. The conclusions detailed below were deemed to be the most appropriate and objective appraisal of the various features and results referred to in section 3.3, including issues raised by other workers. The quantitative surface roughness of cartilage and characteristics of the surface layer *in vivo*, when unloaded and loaded, have been speculated upon. It must be remembered that in this study only bovine articular cartilage was investigated but as mentioned in section 3.3, similar features were evident comparing with other studies examining several other species including human, porcine, rat and also bovine articular cartilage.

There appeared to be no ideal technique for the examination of the cartilage surface. Each technique had its own merits and drawbacks but when used together an improved understanding of the cartilage surface was achieved. Using SEM collagen fibrils were estimated as being approximately 50-100 nm in diameter, in line with other SEM studies (Kobayashi et al., 1996; Teshima et al., 1995). However, the fibril diameters quoted from TEM analysis, ~20-30 nm, were considered to be more accurate as higher magnification was possible and individual collagen fibrils were more easily distinguished. Other TEM studies have also acknowledged the fibril diameters to be approximately 10-30 nm in the superficial tangential zone, (Barnett et al., 1963; Weiss et al., 1968). This anomaly between the two EM techniques was a good illustration of the benefits of using more than one analytical technique in obtaining a more objective characterisation of articular cartilage.

The *'surface layer'* (first referred to as the *lamina splendens*) is presumed to be a distinct ~0.1-3 μm thick acellular, noncollagenous layer, composed of interfibrillar

proteoglycan ground substance, which covers even the uppermost superficial collagen fibrils and chondrocytes in its native state. In this respect it could be considered as merely an extension of the interfibrillar cartilage solid matrix, see Figure 3-20. A further *'boundary' layer* (≤1 μm) may well also be present, resting upon the surface layer, having perhaps a more transient or ephemeral presence. Although the boundary layer was still considered to be intrinsic to the cartilage surface. The boundary layer constituents (phospholipids, glycoproteins) being derived from the synovial fluid and cartilage matrix cellular metabolites. The surface layer probably functions to prevent fibrillation of the superficial collagen fibrils while the boundary layer in turn minimises friction by boundary lubrication. From observations made using cryo-SEM of loaded and non-loaded cartilage, it was apparent that the boundary and/or surface layer may also possess biphasic lubrication properties, being able to both expand and collapse rapidly, as a result of fluid flow, independently of the underlying cartilage matrix. This will occur over short time scales of less than a few seconds.

The surface coat or lamina splendens is still not a commonly referred to attribute of the articular cartilage surface. This is despite its presence being fairly well documented, although under different guises, for the past 45 years. This is because of the many uncertainties, conflicts between studies and unresolved issues which surround the characteristics of the layer. As yet the layer(s) has not even been bestowed with an agreed nomenclature. The observed differences within the literature are due to differing analytical tools (TEM, SEM, ESEM, cyro-SEM, Light Microscopy), differing specimen preparation requirements and the various anatomical locations, age and species as well as the collection, storage and handling procedures used to obtain the cartilage samples. It is therefore quite difficult to produce a coherent impression of the surface layer. For the surface layer to become a common 'textbook feature' of articular cartilage a suitable name and some considered speculation regarding its characteristics must be broadly agreed upon. A widespread acknowledgement of the presence of a distinct surface layer seems an important issue due to implications it may have

regarding cartilage nutrition, permeability and biphasic properties, and of course lubrication.

For general analyses of synovial joint lubrication it is still probably appropriate to ascribe a quantitative value of roughness of 1-2 μm, as determined by R_a calculations at 0.8 mm cut-offs. However this 'roughness' was attributed to the general form or waviness of the cartilage over the adopted 0.8 mm sampling length. A more realistic value of inherent cartilage surface roughness of ~0.1-0.3 μm was provided from line profiles filtered at 0.08 mm cut-offs. Cartilage lubrication theories involving microscopic analysis of surface roughness and asperity contact would perhaps be better suited to use this R_a (0.08) value. The lack of sensitivity to the surface layer and possible elastic deformation (for stylus profilometry) must still be considered as potential sources of error in the measurements, especially for the R_a (0.08) values. While deformation of the surface would not of course have been a problem resulting from non-contacting laser profilometry, the necessity for a dry surface may have temporarily altered the layer in some manner. The surface layer would presumably cover surface features creating an even smoother surface. However due to it's presumed transient and highly compliant nature, significant compression, during joint loading, creating sub-surface asperities to be prominent will probably still occur.

An increase in roughness of ~ 1 μm was identified by laser profilometry due to reciprocating motion. This increased roughness may have been partially or solely caused by the removal of the 'surface coat', commonly quoted as being ~1 μm thick. Furthermore, the small increases in friction coefficient between the *initial* and *repeat* values may have been caused by the removal of the 1 μm thick surface layer associated with boundary lubrication, and not simply due to the slight roughening of the compliant cartilage surface, see Chapter 7. Note that in these friction studies, for the *repeat* results the synovial fluid values were still statistically lower than those of the Ringer's solution. Therefore while substantial surface wear had occurred the boundary or boosted lubrication provided by the

synovial fluid remained active. From analysis of specimens previously used for stationary load friction testing no substantial wear was evident.

The commonly adopted cartilage specimen storage at -20°C in Ringer's solution prior to friction testing was shown not to compromise the surface integrity of the tissue. This was shown using stylus profilometry and ESEM whereby fresh and previously frozen specimens were examined without any obvious differences witnessed. Some of these specimens had been frozen considerably longer than the maximum 3 month period sanctioned for the friction tests.

From the observations of the articular cartilage surface implications for the correct design of friction tests investigating mixed and boundary lubrication of synovial joints and interpretation of the results have been proposed.

Both the advantages and disadvantages of each of the techniques employed in analysing the surface of articular cartilage have been discussed and further summarised in Table 3-4. Non-contacting 3-D laser profilometry has been successfully employed for the first time in examining the articular cartilage surface.

4. Stationary Loading Friction Tests - Cartilage on Metal Contacts

4.1 Introduction

This is the first chapter detailing the results of friction measurements. Within this chapter the tribological behaviour of two contact configurations, namely 3 mm diameter cartilage disc on metal counterface and 9 mm diameter cartilage plug on metal counterface, are discussed. Friction of the cartilage/metal contacts, within a mixed lubrication regime, was recorded using synovial fluid, Ringer's solution and in some instances with no lubricant present. The main test variable was the period of stationary loading prior to sliding which ranged from 5 seconds to 45 minutes. This was the amount of time that the cartilage spent loaded against the metal counterface prior to sliding the metal counterface horizontally and measuring friction. The constant stationary load was applied in order to establish contacts in the mixed lubrication regime, producing very thin lubricating films due to squeeze film action. By subsequently using very low sliding velocities the contribution made by elastohydrodynamic entraining action, to the film thickness, was very small and the mixed lubrication regime, which results from the constant stationary load, remained (see Chapter 2). By varying the period of stationary loading both the tribological conditions of the cartilage/metal contact and the biphasic condition of the cartilage could be altered. The resulting effects of these changes were then assessed by friction measurements. Both Ringer's solution and synovial fluid were studied in order to assess whether or not the constituents of the synovial fluid, not present in the Ringer's solution, reduced friction by providing a boundary lubricating action.

It is worthwhile to point out at this juncture that an equal loading time/load removal time procedure was adopted for all of the stationary loading tests as explained in Chapter 2 (except for the load removal tests, see page 118).

4.2 Experimental Procedure

4.2.1 Three millimetre diameter cartilage disc on metal counterface contact configuration

4.2.1.1 Preliminary Tests

The 3 millimetre diameter cartilage disc on a metal counterface contact configuration was the first to be investigated. A few preliminary test runs were first conducted using four specimens with both Ringer's solution and synovial fluid as the lubricant and adopting stationary loading times from 5 seconds up to 45 minutes. For two of the cartilage specimens three friction measurements were recorded at 5 seconds, 5 minutes, 10 minutes and 15 minutes respectively, one specimen (1) with Ringer's solution as the lubricant and the other specimen (2) with synovial fluid. For the other two specimens three friction measurements were recorded at 5 seconds, 15 minutes, 30 minutes and 45 minutes respectively, one specimen with Ringer's solution (3) as the lubricant and the other specimen (4) with synovial fluid, Table 4-1.

		Stationary Loading Time					
Specimen	**Lubricant**	**5 s**	**5 min**	**10 min**	**15 min**	**30 min**	**45 min**
1	**Ringer's Solution**	✓	✓	✓	✓		
2	**Synovial Fluid**	✓	✓	✓	✓		
3	**Ringer's Solution**	✓			✓	✓	✓
4	**Synovial Fluid**	✓			✓	✓	✓

Table 4-1 Test conditions for the preliminary tests. ✓ indicates the measurement of friction three times.

4.2.1.2 Load application tests

The first series of friction testing involved friction measurements after 5 second and 2 minute stationary loading periods, Table 4-2. Fourteen 3 mm diameter cartilage

specimens were used, 7 specimens each for the two respective lubricants, Ringer's solution and synovial fluid. Seven friction measurements were undertaken at both the 5 second and 2 minute loading times for all the specimens.

A similar series of tests were then conducted applying 5 second and 5 minute stationary loading times. For both lubricants a further 7 specimens were used and again 7 friction measurements were undertaken at both the 5 second and 5 minute loading times for all the specimens.

Friction tests after 5 second and 45 minute stationary loading periods were also undertaken. Both Ringer's solution and synovial fluid were used as lubricants, with fourteen cartilage specimens being tested for each lubricant. For each specimen single friction readings were recorded after 5 seconds and 45 minutes of stationary loading.

Test 5 second/2 minute

Lubricant	Number of Specimens	Stationary Loading Time	Readings taken per Specimen
Ringer's Solution	7	5 seconds	7
		2 minutes	7
Synovial Fluid	7	5 seconds	7
		2 minutes	7

Test 5 second/5 minute

Lubricant	Number of Specimens	Stationary Loading Time	Readings taken per Specimen
Ringer's Solution	7	5 seconds	7
		5 minutes	7
Synovial Fluid	7	5 seconds	7
		5 minutes	7

Test 5 second/45 minute

Lubricant	Number of Specimens	Stationary Loading Time	Readings taken per Specimen
Ringer's Solution	14	5 seconds	1
		45 minutes	1
Synovial Fluid	14	5 seconds	1
		45 minutes	1

Table 4-2 Tabulated summary of the test conditions for the 5 second/2 minute, 5 second/5 minute and 5 second/45 minute stationary loading friction tests.

4.2.1.3 Load removal tests

The effect of load removal was examined after forty five minutes loading as described below. For these tests the lubrication conditions were: i.) synovial fluid, ii.) Ringer's solution and also iii.) no lubricant present. For each specimen single friction readings were recorded after 5 seconds and 45 minutes of stationary loading, this was immediately followed by a period of load removal and another 5

second loading friction reading was taken, again followed by the same period of load removal and a 2 minute loading friction reading. Of the 14 cartilage specimens used for each of the three lubricating conditions (and therefore 42 specimens in total), 7 specimens were given load removal times of 1 second and 7 specimens were given load removal times of 1 minute. The load removal time of 1 second was selected to allow the fluid to fully re-enter the contact, while allowing minimum cartilage rehydration. The 1 minute load removal period allowed some rehydration of the cartilage, in addition to re-introducing fluid between the two surfaces. By using these two load removal periods the combined and also isolated effects of changes in the tribological conditions of the contact and biphasic properties of the cartilage could be observed.

		Loading and Load Removal Sequence					
Lubricant	**Number of Specimens**	**Stationary Loading Times**		*Load Removal Period*	**Stationary Loading Time**	*Load Removal Period*	**Stationary Loading Time**
Ringer's Solution	7	5 s	45 min	*1 s*	5 s	*1 s*	2 min
	7	5 s	45 min	*1 min*	5 s	*1 min*	2 min
Synovial Fluid	7	5 s	45 min	*1 s*	5 s	*1 s*	2 min
	7	5 s	45 min	*1 min*	5 s	*1 min*	2 min
No Lubricant	7	5 s	45 min	*1 s*	5 s	*1 s*	2 min
	7	5 s	45 min	*1 min*	5 s	*1 min*	2 min

Table 4-3 Tabulated summary of the test conditions for the load removal tests.

4.2.2 Nine millimetre diameter cartilage plug on metal counterface contact configuration

For the 9 millimetre diameter cartilage plug on metal counterface contact configuration 16 cartilage specimens were tested, 8 with Ringer's solution and 8 with synovial fluid as the lubricant. Friction readings were recorded for each specimen after 5 seconds, 2 minutes, 5 minutes, 10 minutes, 20 minutes and 45

minutes stationary loading times. Following the 45 minute test the specimens were immersed in the lubricant with no load for another 45 minutes and then the same friction readings were repeated. The first set of friction results were referred to as the *initial* results while the subsequent identical set of results were referred to as the *repeat* results. For this contact configuration both start-up and steady state friction values were recorded.

Lubricant	**Ringer's Solution**	**Synovial Fluid**
Number of Specimens	**8**	**8**
	***Initial* Friction Readings taken after Stationary Loading Times:**	
	5 seconds	5 seconds
	2 minutes	2 minutes
	5 minutes	5 minutes
	10 minutes	10 minutes
	20 minutes	20 minutes
	45 minutes	45 minutes
	***Repeat* Friction Readings taken after Stationary Loading Times:**	
	5 seconds	5 seconds
	2 minutes	2 minutes
	5 minutes	5 minutes
	10 minutes	10 minutes
	20 minutes	20 minutes
	45 minutes	45 minutes

Table 4-4 Summary of test conditions for the 9 mm cartilage on metal contact configuration.

4.3 Results

4.3.1 Three millimetre diameter cartilage disc on metal counterface

Figure 4-1 and Figure 4-2 show friction traces for the 3 mm cartilage on metal counterface contact configuration sampled after varying periods of stationary loading prior to sliding. Upon the initiation of sliding, following a given period of stationary loading, the friction rose instantaneously to a certain value and then generally remained at this value while the cartilage was sliding across the metal counterface, during the 5 second data sampling. Hence the friction at start-up and during steady state sliding were similar and both were influenced primarily by the duration of stationary loading prior to sliding. Although occasionally the start-up friction exhibited a small peak in relation to the steady state friction. The preliminary tests conducted on four specimens Figure 4-3, plainly demonstrated the rise in the coefficient of friction for articular cartilage when subject to increasingly higher periods of stationary loading, but also indicated rather small differences between synovial fluid and Ringer's solution. Both these effects are investigated in tests in the next sections.

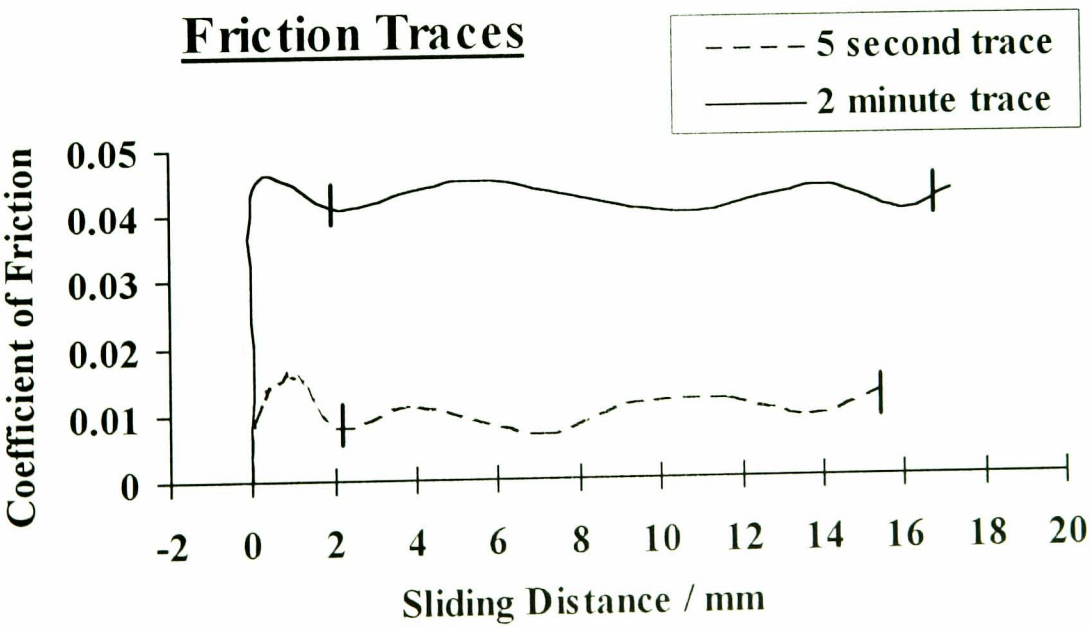


Figure 4-1 Friction traces recorded on a single specimen after stationary loading times of 5 seconds and 2 minutes were applied. Steady state friction was assessed over the trace length marked by the two vertical spacers.

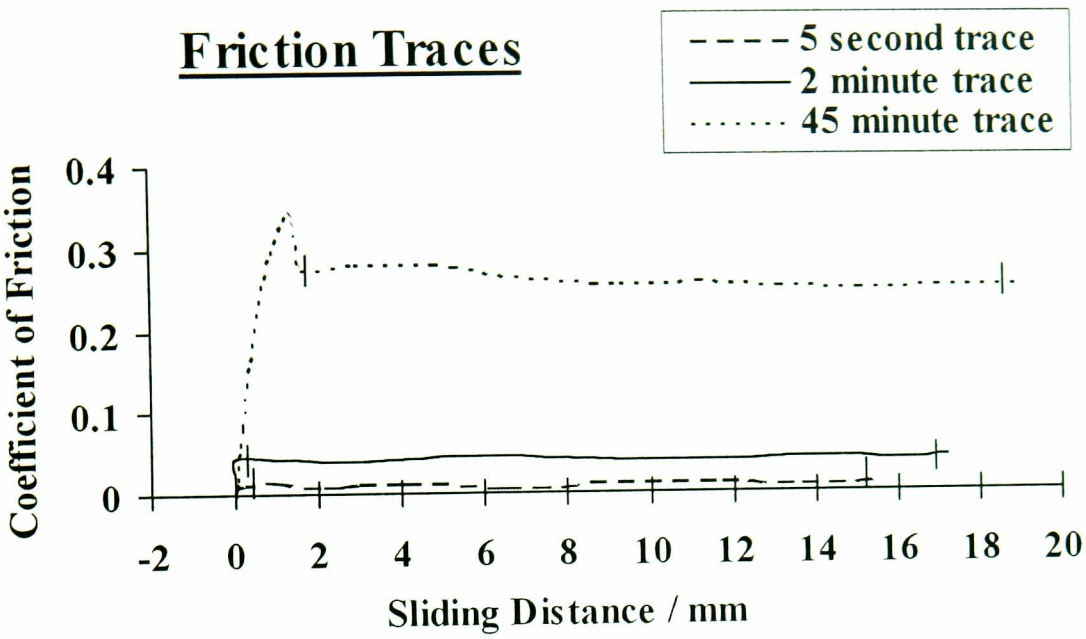


Figure 4-2 Friction traces recorded on a single specimen after stationary loading times of 5 seconds, 2 minutes and 45 minutes were applied. Steady state friction was assessed over the trace length marked by the two vertical spacers.

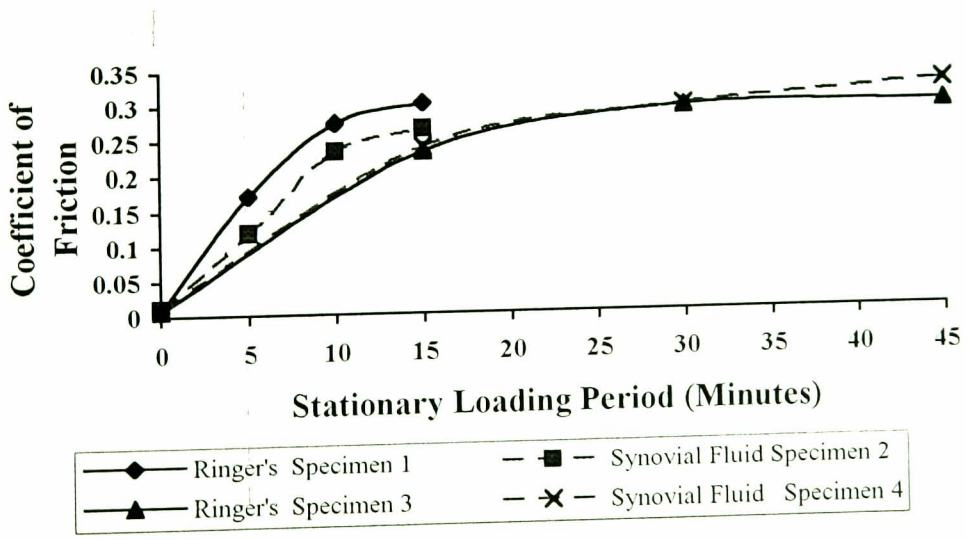


Figure 4-3 Coefficient of friction plotted against stationary loading period for four cartilage specimens. Each point represents an average of 3 readings.

The results for the friction tests carried out after 5 second, 2 minute, 5 minute and 45 minute loading times are shown in Table 4-5 and Figure 4-4. The dramatic increase in friction with loading time was very clear. For each loading time the friction values for the two lubricants were compared using a Student's t-Test.

For the 2 minute loading time synovial fluid was found to produce a statistically significantly reduced coefficient of friction compared to the corresponding value for Ringer's solution ($p<0.05$). No other loading time showed any statistically significant difference between the two lubricants.

	Synovial Fluid			Ringer's Solution		
Loading Time	Mean	StDev	n	Mean	StDev	n
5 seconds	0.012	0.008	26	0.010	0.005	28
2 minutes	0.043	0.014	6	0.071	0.010	7
5 minutes	0.117	0.017	6	0.098	0.020	7
45 minutes	0.270	0.033	15	0.277	0.030	16

Table 4-5 Results of the 5 second/2 minute, 5 second/5 minute and 45 minute stationary loading period tests are shown here for the 3 mm cartilage on metal contact configuration. As all the tests involved an initial 5 second reading they were pooled together according to the lubricant used.

In Figure 4-4, see page 124, all the results from the 2, 5 and 45 minute loading tests have been included, pooling together all the 5 second results for the two respective lubricants. From Figure 4-4 the nature of the rise in the coefficient of friction with stationary loading time is vividly depicted. For these cartilage/metal contacts there was an initial sharp rise in friction up to the 5 minute loading point and then up to the 45 minute loading time the rise became much more gradual. For the 3 mm cartilage specimens on metal there was generally no statistical difference in the results for the two lubricants, except for the 2 minute loading time ($p<0.05$). The most dominant feature influencing the coefficient of friction for these contacts was indeed the duration of stationary loading, prior to sliding, that the articular cartilage specimens underwent.

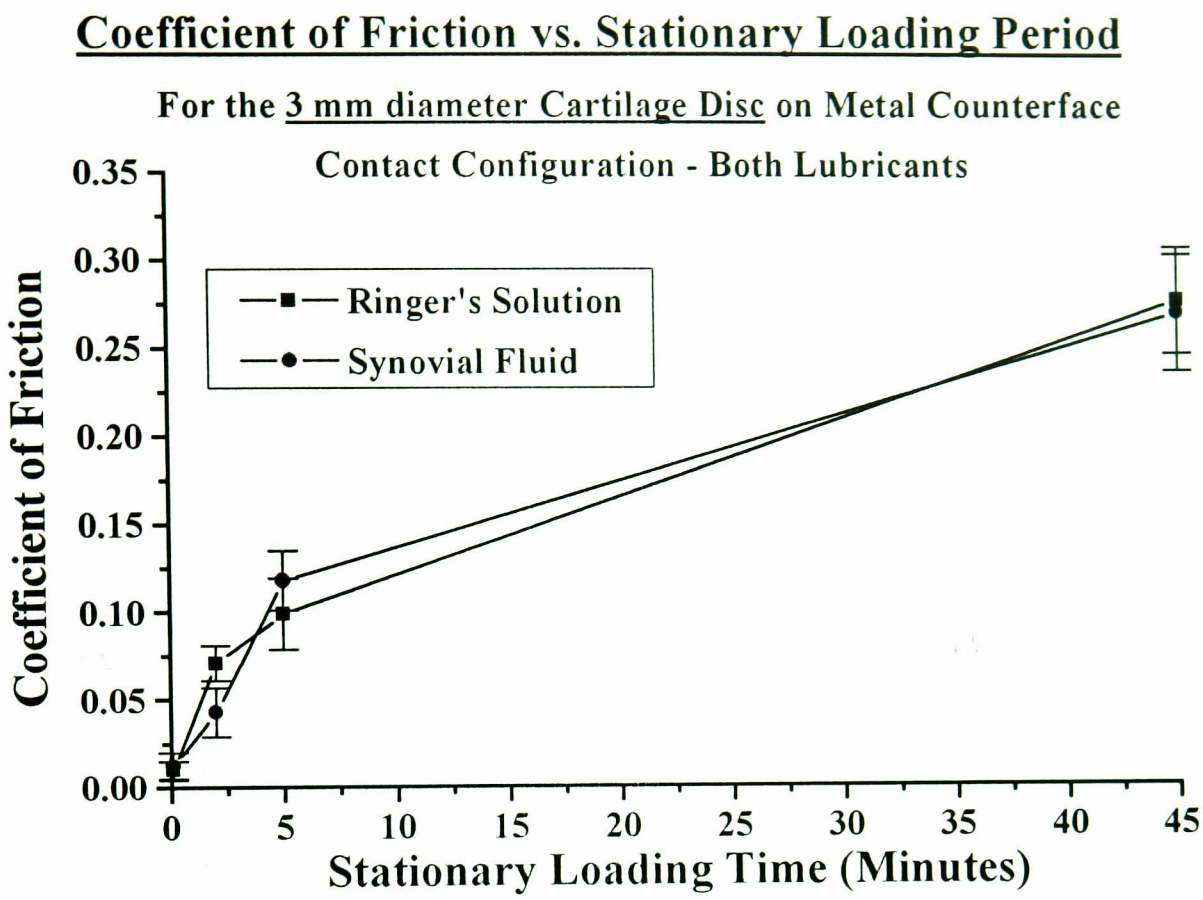


Figure 4-4 Comparison of Ringer's solution and synovial fluid coefficient of friction values at 5 seconds, 2, 5, and 45 minute stationary loading periods. The graph displays the nature in the rise of friction with loading time for the 3 mm cartilage on metal contact. The plotted data was taken from Table 4-5.

4.3.2 Nine millimetre diameter cartilage plug on metal counterface

The results for the 9 mm diameter cartilage plug on metal contact configuration are displayed in Table 4-6, page 127. For a comparison with the 3 mm cartilage on metal, Figure 4-4, the steady state initial results were plotted, Figure 4-5, on exactly the same axis scale. This 9 mm contact configuration showed the same marked rise in friction as the 3 mm configuration. For this contact configuration differences between the coefficient of friction values for the two lubricants was only statistically significant ($p<0.05$) for the lower loading times of both the steady state initial and repeat values, Table 4-6. The results for steady state friction were generally similar to the 3 mm cartilage on metal configuration, Figure 4-6, except that for the 9 mm diameter plugs, Figure 4-5, the rise in the coefficient of friction was more uniform with loading time than for Figure 4-4. Also the rise in the friction was perhaps better defined by the two extra loading period readings taken at 10 and 20 minutes. On the whole the results for these two contact configurations were very similar and indeed there was no substantial difference between them for the 5 second loading time, Figure 4-6. Nonetheless at both the 2 and 5 minute loading periods the 3 mm cartilage disc on metal counterface results were higher than for the 9 mm cartilage plug on metal counterface results; this was true for both lubricants. While at the 45 minute loading period the 9 mm cartilage plug on metal counterface results were higher than for the 3 mm cartilage disc on metal counterface results; and this was again true for both lubricants, Figure 4-6. A similar range of standard deviation for the coefficient of friction values of the 9 mm cartilage on metal configuration to the 3 mm cartilage on metal configuration, Figure 4-6, was observed.

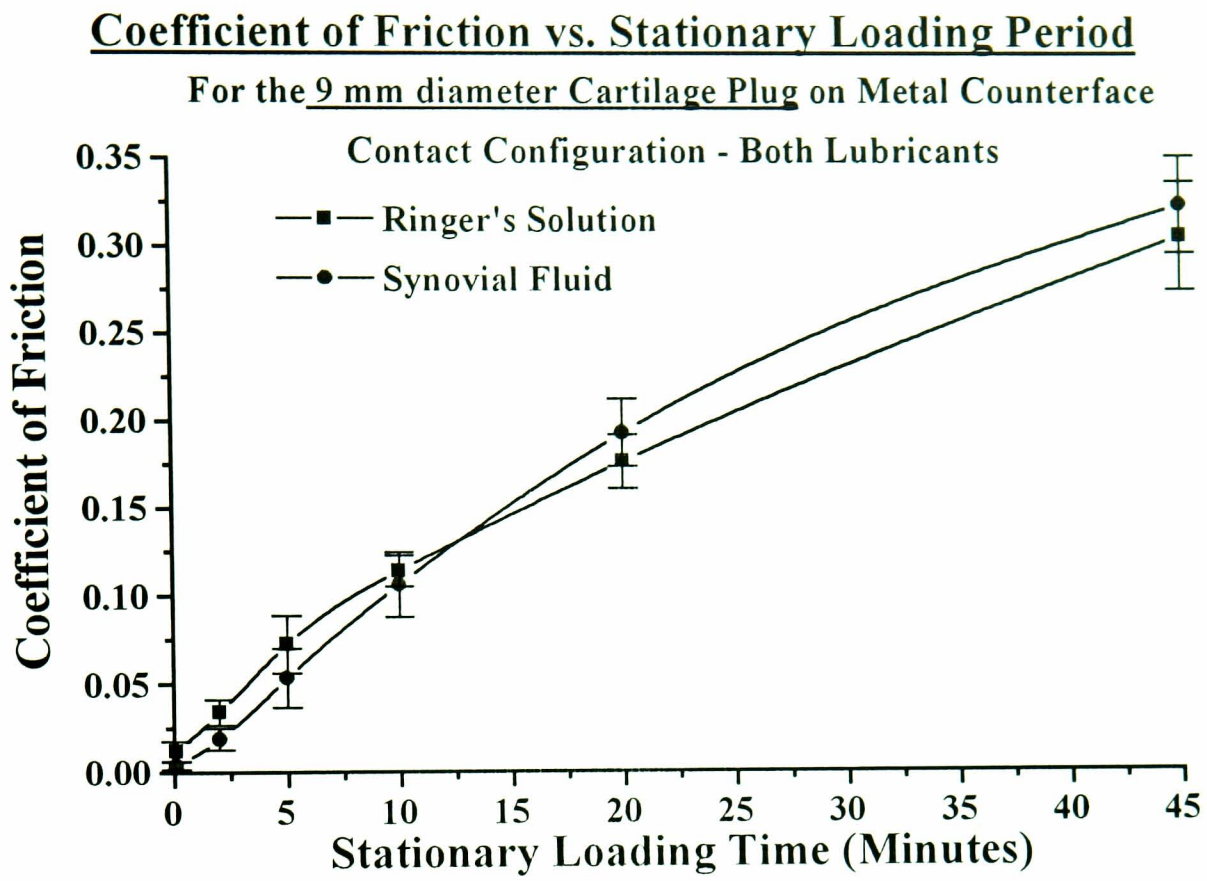


Figure 4-5 Comparison of Ringer's solution and synovial fluid coefficient of friction values at 5 seconds, 2, 5, 10, 20 and 45 minute stationary loading periods. The graph displays the nature in the rise of friction with loading time for the 9 mm cartilage on metal contact. The results shown are for the steady state initial values (Table 4-6).

	Loading Time (Minutes)	Synovial Fluid Mean	StDev	n	Ringer's Mean	StDev	n	t-Test
Steady Initial	**0.08**	0.004	0.002	8	0.012	0.006	8	0.002
Steady Initial	**2**	0.019	0.007	8	0.034	0.008	8	0.001
Steady Initial	**5**	0.054	0.018	8	0.073	0.017	8	0.049
Steady Initial	**10**	0.106	0.019	8	0.114	0.009	8	0.321
Steady Initial	**20**	0.193	0.021	8	0.177	0.017	8	0.102
Steady Initial	**45**	0.324	0.030	7	0.306	0.033	8	0.293
Steady Repeat	**0.08**	0.008	0.007	8	0.018	0.006	8	0.005
Steady Repeat	**2**	0.023	0.008	8	0.036	0.014	8	0.037
Steady Repeat	**5**	0.056	0.013	8	0.080	0.024	8	0.025
Steady Repeat	**10**	0.099	0.018	8	0.123	0.017	8	0.020
Steady Repeat	**20**	0.168	0.018	8	0.185	0.019	8	0.087
Steady Repeat	**45**	0.304	0.029	8	0.305	0.043	8	0.978
Start-up Initial	**0.08**	0.016	0.006	8	0.017	0.006	8	0.623
Start-up Initial	**2**	0.036	0.013	8	0.046	0.015	8	0.185
Start-up Initial	**5**	0.086	0.019	8	0.087	0.030	8	0.937
Start-up Initial	**10**	0.127	0.019	8	0.128	0.023	8	0.936
Start-up Initial	**20**	0.212	0.031	8	0.186	0.027	8	0.099
Start-up Initial	**45**	0.356	0.052	7	0.318	0.052	8	0.182
Start-up Repeat	**0.08**	0.019	0.012	8	0.032	0.015	8	0.070
Start-up Repeat	**2**	0.041	0.013	8	0.056	0.018	8	0.077
Start-up Repeat	**5**	0.094	0.021	8	0.090	0.028	8	0.753
Start-up Repeat	**10**	0.133	0.019	8	0.132	0.017	8	0.924
Start-up Repeat	**20**	0.193	0.019	8	0.185	0.017	8	0.399
Start-up Repeat	**45**	0.333	0.049	8	0.301	0.033	8	0.146

Table 4-6 Results for the 9 mm cartilage on metal contact configuration. Friction readings were recorded after 5 second, 2, 5, 10, 20 and 45 minute stationary loading periods. This procedure was carried out twice for each cartilage specimen, hence the initial and repeat results. Both start-up and steady state values were recorded. Note that 0.08 minutes ≅ 5 seconds. t-Tests conducted between synovial fluid and Ringer's solution results.

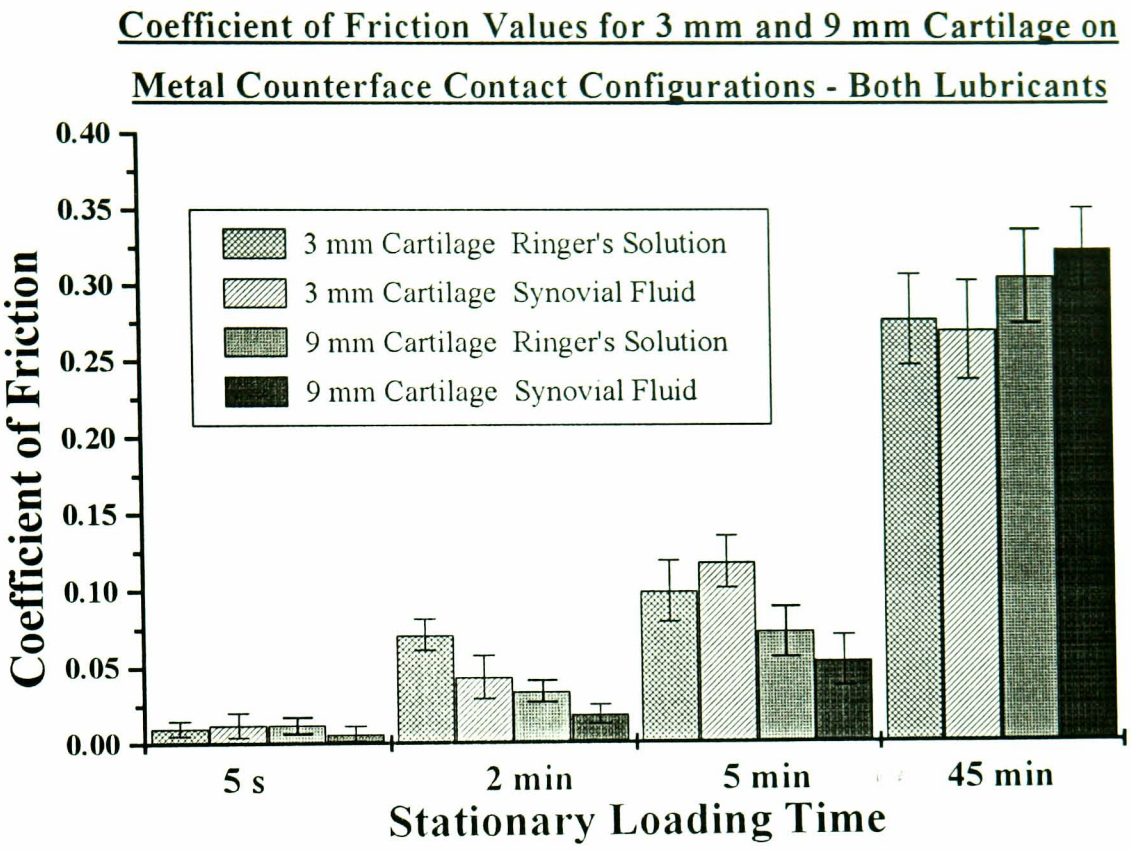


Figure 4-6 Comparison of the 3 mm and 9 mm cartilage on metal results. The results displayed for the 9 mm configuration are the steady state initial values (Table 4-6).

For the 9 mm cartilage on metal contact start-up friction using synovial fluid was generally higher than the steady state friction and thereby indicating the overall presence of a peak (albeit relatively small compared to the steady state value Table 4-6) for most of the friction traces. However for the longer loading periods this start-up peak receded and became insignificant. For the Ringer's solution there was only found to be a marked peak for the 5 second and 2 minute repeat values. The repeat coefficient of friction values in comparison to the initial values showed very little difference, Figure 4-7 and Figure 4-8. Thus indicating that there were no major changes to the specimens during the testing.

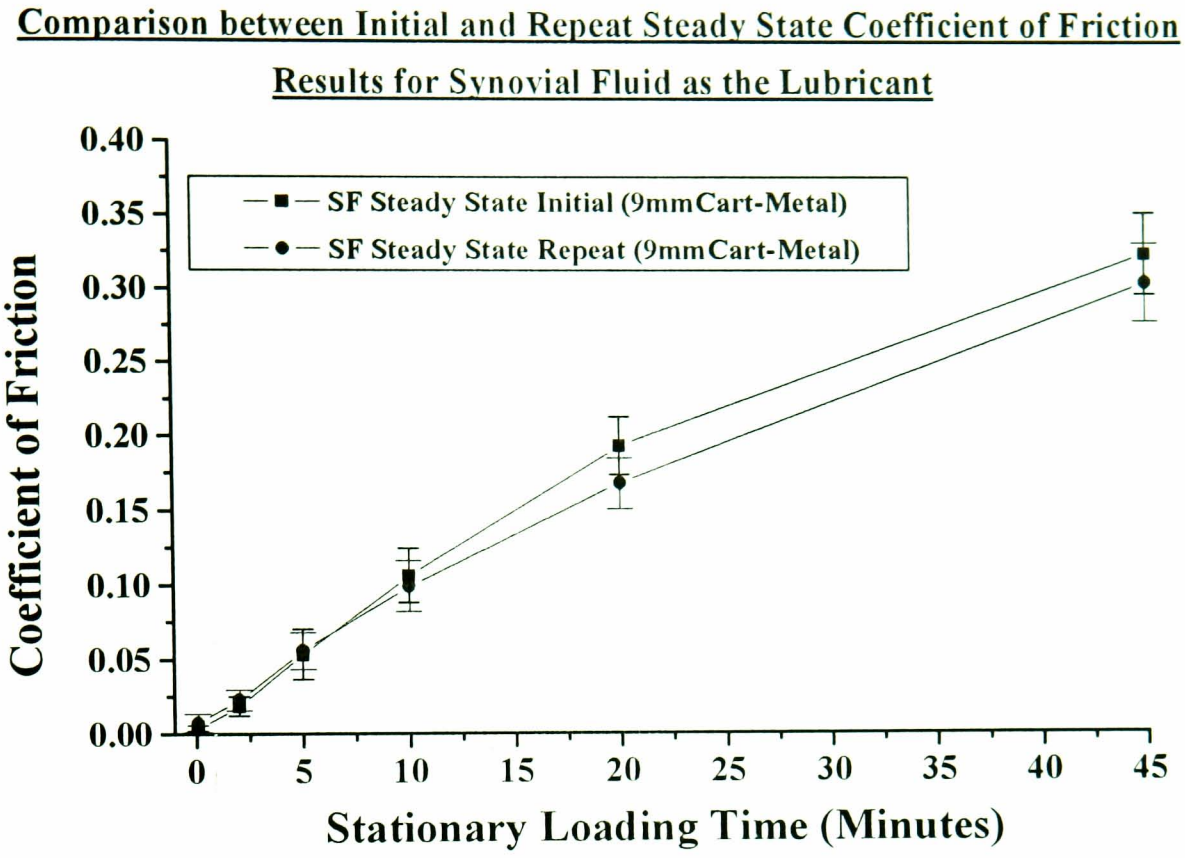


Figure 4-7 Comparison of initial and repeat coefficient of friction values for the 9 mm cartilage on metal configuration. Synovial fluid was the lubricant.

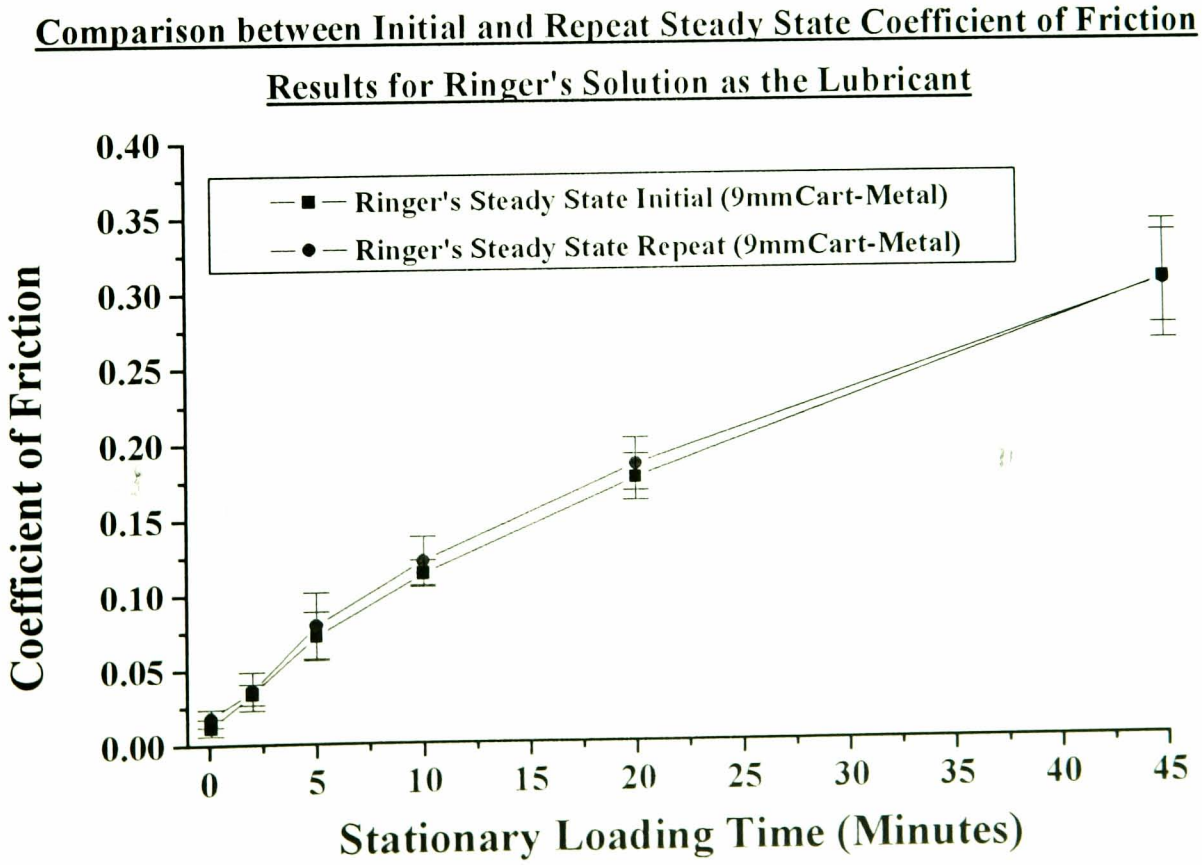


Figure 4-8 Comparison of initial and repeat coefficient of friction values for the 9 mm cartilage on metal configuration. Ringer's solution was the lubricant.

4.3.3 Load removal tests with and without lubricant

Load removal tests were carried out following a period of 45 minutes loading, with the load being removed for both 1 second or 1 minute intervals, Table 4-3. Tests were carried out with both Ringer's solution and synovial fluid and with no lubricant present. The general pattern of the results was similar for all three lubricating conditions, Figure 4-9 and Figure 4-10. It was of particular interest that the no lubricant condition produced similar results. This indicated, as previous results have, that the lubricant was not the most important factor controlling the friction coefficient. The effect of removing load for 1 second, Figure 4-9, after 45 minutes upon a 5 second loading period test was to make a small but significant drop in friction levels compared to the 45 minute value, for all lubricants. The subsequent 2 minute loading tests, with another 1 second load removal in between, brought the friction levels back to the original 45 minute values for Ringer's solution and synovial fluid, and beyond for the no lubricant condition. For an identical series of tests, but for a 1 minute period of load removal instead, the results (Figure 4-10) although again similar for all three lubricating conditions, were quite different than those in Figure 4-9 for the 1 second load removal. The effect of removing load for 1 minute after 45 minutes, upon a subsequent 5 second loading period test was a dramatic reduction in friction to almost the levels found for the original 5 second loading tests. The following 2 minute loading tests, with another 1 minute load removal in between, brought the friction levels to around 0.1 for all the lubricants, being roughly equivalent to the original 5 minute loading period values for the 3 mm cartilage on metal contact configuration.

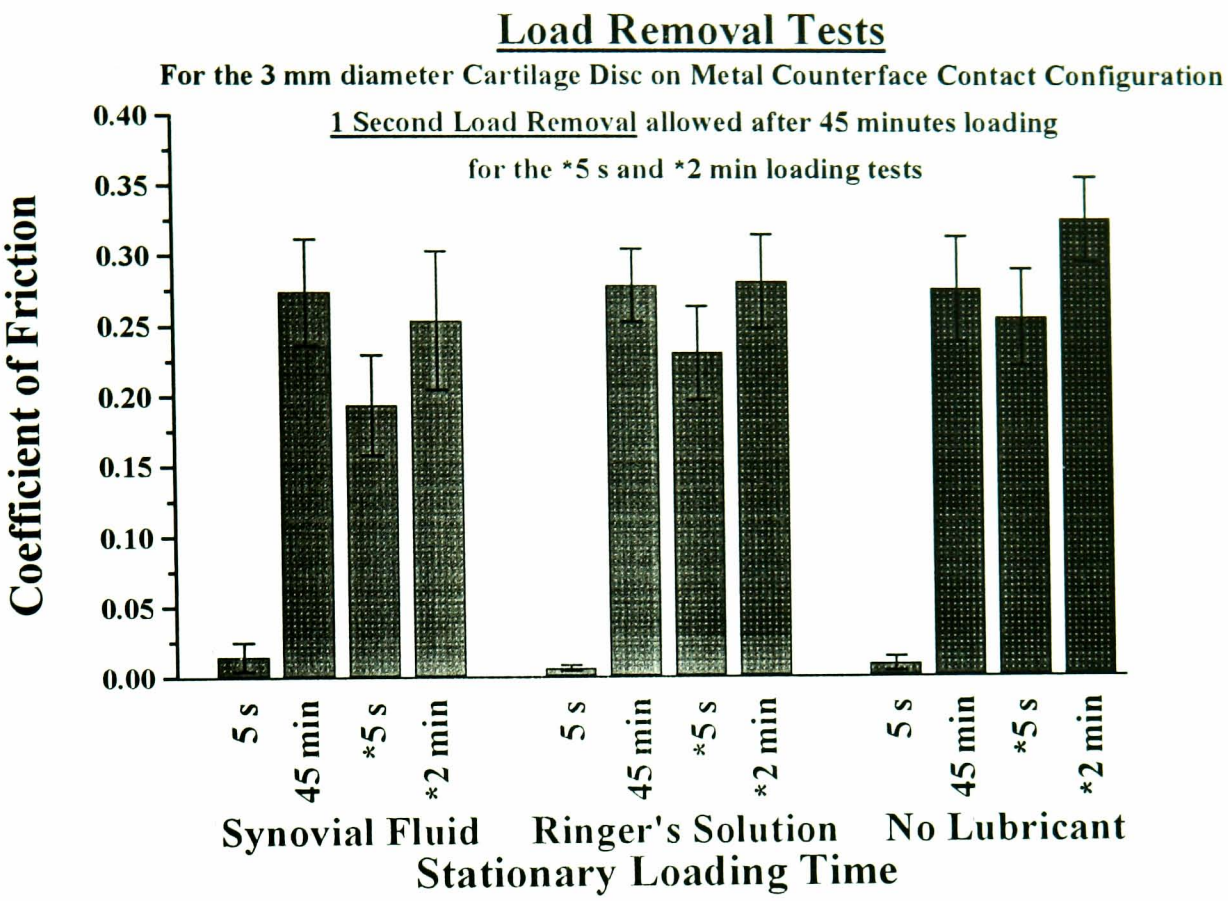


Figure 4-9 Coefficient of friction plotted against stationary loading period for the 5 second and 45 minute loading period tests. Tests were conducted using either synovial fluid or Ringer's solution as the lubricant, and also with no lubricant. At least 7 cartilage specimens were tested for each of the three lubricant conditions. The *5 s and *2 min columns represent readings taken immediately after the 45 minute loading test, following a 1 second removal of load.

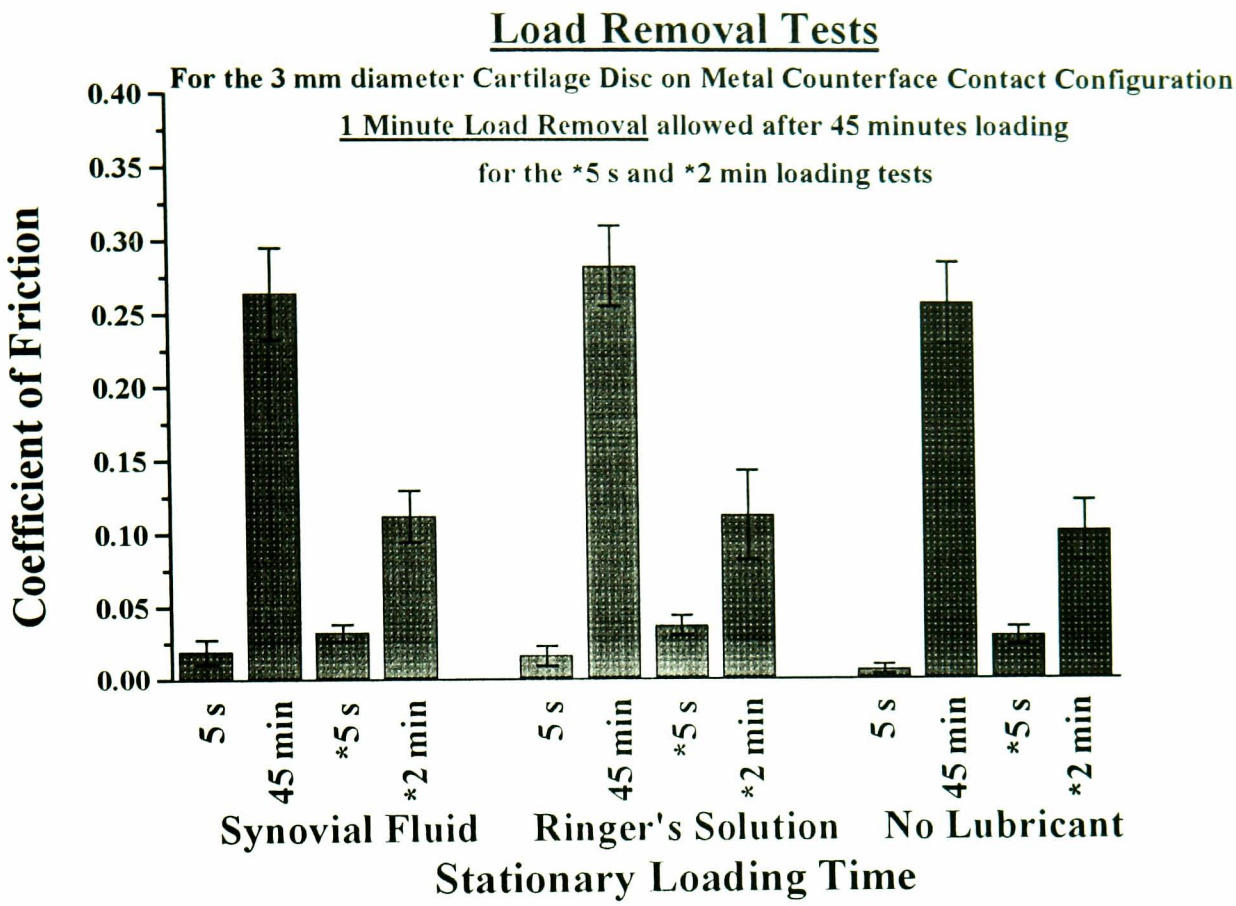


Figure 4-10 Coefficient of friction plotted against stationary loading period for the 5 second and 45 minute loading period tests. Tests were conducted using either synovial fluid or Ringer's solution as the lubricant, and also with no lubricant. At least 7 cartilage specimens were tested for each of the three lubricant conditions. The *5 s and *2 min columns represent readings taken immediately after the 45 minute loading test, following a 1 minute removal of load.

4.4 Discussion

4.4.1 Friction Test Validation

The friction results have shown that the experimental protocol adopted was successful in respect to the following three aspects,

i. lubrication regime
ii. *repeatability* of coefficient of friction readings for any given specimen (as described in chapter 2) and
    reproducibility of coefficient of friction readings between different cartilage specimens
iii. comparison with results from similar studies.

Each of these topics will be discussed below.

4.4.1.1 Lubrication Regime

The purpose of this study was to investigate friction when the contact entered the mixed or boundary lubrication regime for two different lubricants. For the 3 mm cartilage on metal configuration the squeeze film model (see Chapter 2) predicted that some boundary contact had already occurred after a 5 second loading period. The entraining action model (see Chapter 2), for the experimental conditions, predicted that upon sliding, full fluid film lubrication did not occur and boundary contacts in a mixed lubrication regime were predicted. The experimental results showed that both the start-up and steady state friction values were similar. This similarity tended to confirm that the entraining action produced by the low velocity sliding was not sufficient to produce fluid film lubrication and reduce friction by elastohydrodynamic effects. For the 9 mm cartilage on metal contact configuration start-up as well as steady state friction was recorded. Although for several of the shorter duration loading times the start-up was higher than the steady state the differences were never great, Table 4-6, and hence the entraining action was considered not to be producing a full fluid film lubrication regime. This was some evidence that entraining motion was reducing the amount of boundary contact occurring, to some extent, for the lower loading periods and especially when synovial fluid was used as the lubricant.

Therefore overall, the experimental design was considered to be successful in ensuring that the contact remained in the mixed or boundary lubrication regime for both of the contact configurations. The effect of different lubricants and the influence of loading regime could then be investigated with the confidence that the contacts were under the required lubrication regime.

4.4.1.2 Repeatability & Reproducibility

Previous studies investigating boundary lubrication of synovial joints have often tended to use synthetic materials as their contacts (McCutchen, 1966; Wilkins, 1968; Reimann et al., 1975; Davis et al., 1979; Jay, 1992 and Williams et al., 1993), glass/rubber for example, as experiments utilising cartilage specimens had been found to be time consuming and lacking reproducibility (Davis et al., 1979; Hills and Butler, 1984). These studies have only addressed the purpose of synovial fluid in providing boundary lubrication to synthetic surfaces and have then extrapolated their results to discuss boundary lubrication of articular cartilage. However a thorough investigation of boundary lubrication of synovial joints must ultimately include the cartilage surface itself as boundary lubrication involves four main components being the two opposing cartilage surfaces; the mechanism by which the boundary lubricant is deposited and fixed; the source of the boundary lubricant (thought widely to be the synovial fluid); and how the boundary layers repel one another to protect the opposing surfaces. Therefore, the use of cartilage specimens (despite any possible experimental difficulties) for this study was deemed essential. Not only because, as discussed cartilage has, of course, a fully interactive role in boundary lubrication alongside the synovial fluid lubricant, but also the biphasic nature of cartilage has been considered to be an important factor governing friction, in the mixed/boundary lubrication regime, considering such lubrication theories as boosted (Walker et al., 1970) and weeping (McCutchen, 1962) lubrication.

For the 3 mm cartilage on metal contact configuration the 5 second/2 minute and 5 second/5 minute friction tests revealed a satisfactory level of repeatability for

the coefficient of friction values of any particular specimen considering the standard deviations of the 7 friction tests readings for each of the specimens, (see Chapter 2). Once the 5 second/2 minute and 5 second/5 minute results were pooled together to attain a mean value for the particular seven specimens used with respect to lubricant and loading time, Table 4-5, again the standard deviations were reasonably low. These low standard deviations and those of the 45 minute tests, Table 4-5, showed there was good reproducibility for the coefficient of friction values between the cartilage specimens for all the assessed stationary loading times, Figure 4-4. The same was found to be true for the 9 mm cartilage on metal configuration, having a similar range of standard deviations for the mean values of the friction coefficients at the various loading times, Figure 4-5.

Thus, for any particular specimen the friction results were sufficiently repeatable and for the two contact configurations tested there was a satisfactory level of reproducibility between the specimens for the friction values, as discussed in Chapter 2.

Furthermore, for the 9 mm cartilage on metal configuration, only a handful of results showed any marked difference between the initial and repeat coefficient of friction values, Figure 4-7 and Figure 4-8. The overall lack of any difference revealed quite clearly that within the time scale of these tests no appreciable degradation of the cartilage samples was occurring and that no substantial surface or boundary layer wear of the specimens took place. The lack of cartilage surface wear was further confirmed from observations reported in Chapter 3.

4.4.1.3 Comparison with Previous Studies

The results compare well with the literature (see Tables 1-2 and 1-3). Most of the results in the literature, especially the entire joint experiments, Table 1, approximate to the 5 second loading values for the 3 mm and 9 mm cartilage on metal configurations. These coefficient of friction values were, of course, the lowest levels found for both configurations, 5 seconds being the shortest

stationary loading period adopted. This was not surprising as most of the previous studies cited were analysing friction in conditions thought to be representative of physiological walking or running, especially the entire joint experiments. For these previous investigations, therefore, the tribological conditions at the contact regarding load, load duration and sliding speed would have been less severe than this study's contact conditions for which a clear mixed/boundary lubrication regime was established.

For coefficient of friction values very similar to this study's 5 second loading values many of the studies cited in Tables 1-2 and 1-3 proposed a *full* fluid film lubricating mechanism but at the 5 second loading time, for this study, contact between asperities was predicted to have occurred (see Chapter 2). Dowson and Jin (1986) have proposed from theoretical studies that cartilage asperities are flattened by micro-elastohydrodynamic effects, and hence thinner lubricating films can be predicted to be maintained compared to the prediction presented in Chapter 2. Therefore micro-elastohydrodynamic lubrication could be offered as an explanation for the low friction of around 0.01, found with cartilage plugs on metal counterfaces after 5 seconds of loading, Table 4-5 and Table 4-6. An alternative explanation is related to the biphasic properties of the articular cartilage. For this case it is proposed that a large proportion of the applied load is carried by the fluid phase of the biphasic cartilage in the mixed lubrication regime which helps to reduce friction. This rationale is favoured because it not only accounts for the low friction levels at 5 second loading times but also provides for a full interpretation of the nature of the rise in friction with loading times up to 45 minutes, this is further discussed in the following sections. Whereas micro-EHL theory could not reconcile with experimental data as after 5 minutes loading, for example, with coefficients of friction already at ~ 0.1 clearly there was no full fluid film lubrication mechanism taking place. Thus, for reasons explained later (see section 4.4.3) it may then be perfectly possible for cartilage under a mixed/boundary lubrication regime, after short periods of loading, to maintain very low friction levels normally associated with full fluid film lubrication.

4.4.2 Effects of Lubricant and Stationary Loading Time

4.4.2.1 Lubricant

For the 3 mm cartilage on metal contact configuration the only significant difference between the two lubricants occurred at the 2 minute stationary loading time where the synovial fluid results were found to be lower. For the 9 mm cartilage on metal configuration no differences were found between the Ringer's solution and synovial fluid values Figure 4-5, except for the lower loading times of both the steady state initial and repeat values where again the synovial fluid results were statistically lower, Table 4-6. Hence, overall the variation of the coefficient of friction values when using either Ringer's solution or synovial fluid as the lubricant has not been shown to be of significance, except for a few scattered exceptions, whereby synovial fluid was shown to be beneficially lubricating the contacts. Reviewing the data of these two contact configurations the acquired impression was that perhaps for larger groups of specimens the synovial fluid may have been found to have been beneficially lubricating the contacts more often, but the differences would never have been great. Indeed it is fair to say that the two lubricants were certainly not by any means the main variable governing friction under the particular experimental conditions adopted for the two contact configurations investigated.

The small but beneficial lubricating effect of the synovial fluid could be attributed to a number of considerations. In respect to the Ringer's solution the synovial fluid was more viscous and some additional boosted and/or boundary lubrication between the cartilage and metal contacts may have been provided by the synovial fluid.

4.4.2.2 Stationary Loading Time

The preliminary friction tests, provided the first indication of how dependent the coefficient of friction of articular cartilage was to stationary loading time prior to sliding. Clearly the stationary loading time for a mixed/boundary lubrication regime was a highly important variable which required investigation. The influence of loading time on friction was consistent with the early work of

McCutchen (1962) and Walker et al. (1968); however these studies were conducted under reciprocating motion.

4.4.2.2.1 Load and Load Removal Tests

The results of unloading the cartilage for 1 second and 1 minute periods after 45 minutes loading, then reloading and measuring friction again, Figure 4-9 and Figure 4-10, were very helpful in elucidating the frictional behaviour of articular cartilage. A 1 second load removal time, which allowed fluid to cover the surface but not substantially enter into the cartilage, only produced a small reduction in friction for the *5 second loading test. For the *2 minute loading test the friction levels were back up to the levels found after 45 minutes of loading for the two lubricants and beyond for the no lubricant condition, Figure 4-9. For the 1 minute load removal period, where the fluid had an opportunity to re-enter the exuded cartilage, friction was dramatically reduced for the *5 second loading test to almost that of the original 5 second loading friction levels. For the *2 minute loading tests the calculated coefficient of friction values were approximate to the original 5 minute loading period values for this 3 mm cartilage on metal contact configuration, Figure 4-10. The fact that the unlubricated contact results closely matched the results for the two lubricants was also of particular importance. This occurrence, as well as the effect of load removal on friction levels, will be referred to and explained below.

The no surface lubricant condition produced extremely low values of friction after 5 seconds of loading, indicating that bulk fluid on the surface was not required to obtain this low value of friction. Secondly, after 45 minutes of loading the coefficient of friction was extremely high (>0.25), and the entraining action did not reduce friction i.e. micro-EHL should not be a loading time dependent phenomena. Unloading the contact for 1 second, and allowing a lubricant film on the surface, only reduced the friction by a small amount from this high value. It was necessary to unload the contact for 1 minute, and allow the fluid to rehydrate the cartilage, in order to reduce the friction towards the original low value found after the initial 5 seconds of loading. These results question the importance of the

micro-elastohydrodynamic lubrication action in reducing friction in the contacts studied after 5 seconds of load application. Assuming that the low levels of friction after 5 seconds loading were achieved in the mixed or boundary regime, either an excellent boundary lubricant was present on the surface of the cartilage, or the biphasic nature of the cartilage ensures extremely low friction in the fully hydrated condition. The second explanation is favoured, as on initial loading of the biphasic cartilage, a large proportion of the load is carried by the pressure difference in the fluid phase of the cartilage (Ateshian et al., 1994) and this may produce an equally very low coefficient of friction for the synovial fluid, Ringer's solution and no surface lubricant conditions. Also if indeed the low friction at 5 seconds were entirely due to excellent boundary lubrication then it might be expected that the friction levels would remain constant, i.e. be independent of loading time, which of course was not the case.

These particular tests convincingly demonstrated that it was not only the loading time but the duration of load removal that was largely instrumental in influencing the coefficient of friction values. In other words the overall loading history, being reflective of the cartilage's state of hydration, was the key determinant of the friction coefficient for these cartilage/metal contacts.

These tests further illustrated that a gradual increase in boundary contact, within the mixed lubrication regime, with increasing loading time was not the factor increasing the coefficient of friction. For the 1 second load removal tests, allowing the re-introduction of a fluid film, the friction was still very high for the repeated 5 second loading period test, Figure 4-9. This occurrence could not have taken place if increasing boundary contact was the primary factor governing the friction levels.

4.4.2.3 Comment

The overall results in this study showed that the friction was primarily controlled by the duration of loading placed on the cartilage, and therefore by the amount of water content and the biphasic fluid flow in the cartilage at the contact.

4.4.3 *Effect of Biphasic Properties on Friction*

The most important consideration is how the load is carried or transferred in the biphasic cartilage. Upon initial loading a large amount of load is carried in the fluid phase of the cartilage (Ateshian et al., 1994) and it is interesting to speculate that the proportion of load carried by the fluid phase contributes little to the total or aggregate friction force of the two phases. As the stationary loading period is increased the load carried by the solid phase increases and that carried by the fluid phase decreases. Hence, the overall friction and friction coefficient increases. The total friction force is usually expressed as:

$$F_T(t) = \mu_T(t).W$$

This can be expressed further in terms of the solid and fluid biphasic components as follows,

$$F_T(t) = \mu_s W_s(t) + \mu_f W_f(t)$$

where $\mu_s >> \mu_f$

$$\therefore \quad F_T(t) = \mu_s W_s(t)$$

where $W_s(t) = W - W_f(t)$

$\mu_T(t)$ - overall or aggregate friction coefficient

$F_T(t)$ - overall friction force

W - total load

$W_s(t)$ - load carried by the solid phase

$W_f(t)$ - load carried by the fluid phase

μ_s - effective coefficient of friction attributed to the solid phase

μ_f - effective coefficient of friction attributed to the fluid phase

(t) - indicates time function dependency

Following this relationship, for $\mu_s \gg \mu_f$, the coefficient of friction, $\mu_T(t)$, increases as the proportion of the load carried by the solid phase increases with time.

The indentation response of the cartilage to load with time (Figure 2-9) can be

used as an indication of the flow of fluid in the cartilage and the load carriage by the fluid phase. An initial instantaneous elastic deformation in response to load is primarily due to change in shape of the cartilage, (not change in volume). At this point there are maximum pressure differences in the fluid phase, and maximum fluid flow and the largest proportion of the load is carried by the fluid phase. As the loading time progresses the slope of the displacement-time curve reduces, the rate of flow of fluid in the cartilage declines in response to reducing pressure difference in the fluid phase and a reduction in the load carried by the fluid phase, and hence increasing load carried by the solid phase. At 45 to 60 minutes the contact is close to equilibrium, little further deformation or fluid flow is occurring and the pressure difference and load carried by the fluid are close to zero. This is consistent with the long exudation times found by Edwards (1967). The rise in friction in Figure 4-4 and Figure 4-5 with loading time is thought to reflect the relative proportion of load carried by the solid phase of the cartilage, $W_s(t)$. Upon removal of load fluid content is rapidly recovered (Edwards, 1967), explaining the greatly reduced friction levels which were recorded subsequent to 1 minute load removal after 45 minutes loading in Figure 4-10. Following an equal period of load removal to that of previous loading, the indentation test produced a similar displacement-time curve when subsequently reloaded which was indicative of a full recovery of fluid content and this procedure was subsequently used in all friction tests, as described in Chapter 2.

4.4.4 Comparison between the Two Contact Configurations

The 3 mm cartilage disc on metal configuration exhibited the sharpest rise in friction with time due to the higher contact stresses involved, being approximately 4 MNm^{-2}. For the 9 mm cartilage plugs and metal plugs the contact stresses would have been roughly 0.5-2 MNm^{-2}, depending on contact area which would have increased with load duration. It could be argued that for the 9 mm cartilage plugs, friction was rising simply due to the increasing nominal contact area. This was not the case as for the 3 mm cartilage disc tests the nominal contact area was constant and friction, nonetheless, rose sharply with time. For the 3 mm cartilage disc the exudation of fluid from the matrix under load may have been exacerbated

by the exposed surrounding areas of the matrix resulting from sectioning. However water is very tightly bound to the solid matrix by proteoglycans at the molecular level and therefore it could be reasoned that an area of exposed sectioned matrix will release water more readily only within a region of microns from the sectioned region.

For the metal counterface tests at 5 seconds there was little difference between the 9 mm cartilage and 3 mm cartilage results for Ringer's solution but for synovial fluid as the lubricant the 9 mm configuration displayed lower friction compared to the 3 mm configuration. This may have been evidence of a degree of squeeze film lubrication afforded by the viscous synovial fluid to the 9 mm cartilage plugs having a larger nominal contact area and lower stress. At the 2 and 5 minute loading periods the 3 mm cartilage disc configuration exhibited higher friction as it was subject to higher contact stresses and therefore more water exudation would have occurred. At 45 minutes the 9 mm cartilage plugs provided the higher friction values possibly due to there being greater contact area while having a similar state of hydration to the 3 mm cartilage discs after such a long stationary load duration. It should be noted that water loss is not linearly related to applied stress (Edwards, 1967), which may account for the overall similarity of results for the two configurations. It has to be recognised that the study was not developed specifically to look at the effect of contact area or stress on friction, and further work with carefully designed experiments would be needed to investigate this further.

4.4.5 Concluding Comments

- Good reproducibility of friction values between the cartilage specimens was achieved for both contact configurations.

- There was generally found to be little difference between the lubricants.

- The coefficient of friction for these contacts was found to have a strong dependency on the stationary loading time. This relationship has been linked

with the biphasic properties of articular cartilage, as described by Mow and co-workers; in particular the carriage of load by pressure differences within the fluid phase. This argument was further supported by the load removal test results, as well as by comparison to the indentation response of cartilage under load.

5. Stationary Loading Friction Tests - Metal on Cartilage Contacts

5.1 Introduction

In addition to the cartilage on metal counterface tests, the reverse contact configuration of these two materials, i.e. metal plugs sliding on a cartilage counterface was investigated. In this configuration only a small part of the cartilage area directly under the plug was subjected to a constant load, with the remaining cartilage counterface's surface only being loaded for a short period during sliding. The previous cartilage on metal counterface configurations had shown the dramatic effect that the stationary loading time had upon the coefficient of friction values. The effect of reversing the contact configuration was of interest, paying particular attention to any difference this might have upon the biphasic behaviour of the articular cartilage. These studies also served as a useful prelude to the cartilage plug on cartilage counterface tests, detailed in the subsequent chapter.

5.2 Experimental Procedure

Cartilage counterfaces were sectioned from bovine humeral and femoral heads to dimensions similar to those of the metal counterface, approximately 30 x 70 millimetres and having a thickness of 10-20 mm (largely consisting of the underlying subchondral and cancellous bone), in such a manner as to provide as even a cartilage surface area as possible. Fifteen cartilage counterfaces were tested with either of two metal plugs, Table 5-2. Both metal plugs 1 and 2 had surface roughnesses, Ra values, of less than 0.15 μm, for a 0.25 mm sampling interval. Metal plug 1 had a radius of curvature of 66 mm. Metal plug 2 had a radius of curvature of 137 mm. The respective diameters of metal plugs 1 and 2 was 11 and 25 mm. The 25 mm diameter plug 2 was used to ensure that there were no edge effects and no excessive ploughing friction attributing to the measured start-up friction; the plug being too large for edge contacts of the plug against the cartilage to occur, Figure 5-1.

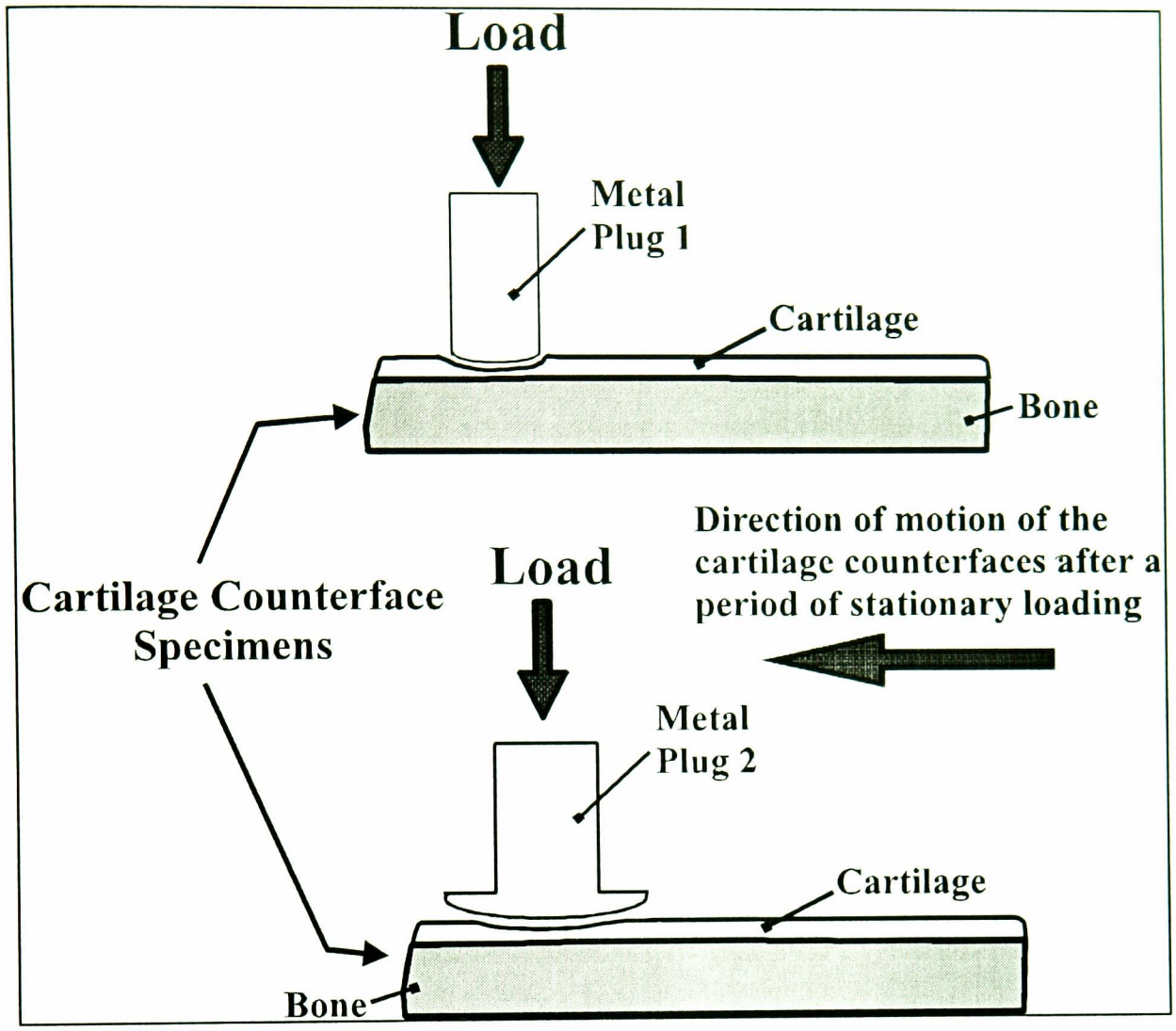


Figure 5-1 Schematic drawing of two cartilage counterfaces, respectively loaded by metal plugs 1 & 2 (not too scale). The 25 mm diameter metal plug 2 ensured no edge effects and no excessive ploughing friction.

Friction of the metal/cartilage contacts was recorded within a mixed lubrication regime, using synovial fluid and Ringer's solution. The main test variable was the period of stationary loading prior to sliding which ranged from 5 seconds to 45 minutes. This was the amount of time that the metal plug spent loaded against the cartilage counterface prior to sliding the counterface horizontally and measuring friction. Seven tests were conducted using Ringer's solution as the lubricant and eight tests using synovial fluid as the lubricant, on 15 different cartilage counterfaces Table 5-2. Each test consisted of the following stationary loading time friction measurements; two readings at 5 seconds, 30 seconds, 2 minutes, 5 minutes and 20 minutes, with a single reading taken after 45 minutes stationary loading. Equal unloading times following each loading period was applied

throughout to allow for rehydration of the cartilage counterface, except following the 45 minute test when after only 1 minute of load removal a 5 second loading period test was performed (*5 seconds, Table 5-1). For this new contact configuration the recording of two readings at 5 seconds, 30 seconds, 2 minutes, 5 minutes and 20 minutes was considered necessary to confirm that significant wear and/or cartilage degradation, affecting the coefficient of friction values, was not occurring by providing data which was hoped to be satisfactorily repeatable for each loading time.

Stationary Loading Times
5 seconds
30 seconds
2 minutes
5 minutes
20 minutes
45 minutes
*5 seconds

Table 5-1 Stationary loading times adopted as friction measurements for the metal on cartilage counterface tests. Two readings were taken at the 5 second, 30 second, 2 minute, 5 minute and 20 minute loading times.

Test	Cartilage Counterface	Metal Plug	Lubricant
1	**i**	1	Ringer's Solution
2	**ii**	1	Ringer's Solution
3	**iii**	1	Ringer's Solution
4	**iv**	1	Ringer's Solution
5	**v**	1	Ringer's Solution
6	**vi**	2	Ringer's Solution
7	**vii**	2	Ringer's Solution
8	**viii**	2	Synovial Fluid
9	**ix**	2	Synovial Fluid
10	**x**	1	Synovial Fluid
11	**xi**	2	Synovial Fluid
12	**xii**	2	Synovial Fluid
13	**xiii**	1	Synovial Fluid
14	**xiv**	1	Synovial Fluid
15	**xv**	2	Synovial Fluid

Table 5-2 Test conditions for the metal plug on cartilage counterface configuration, detailing tests 1-15.

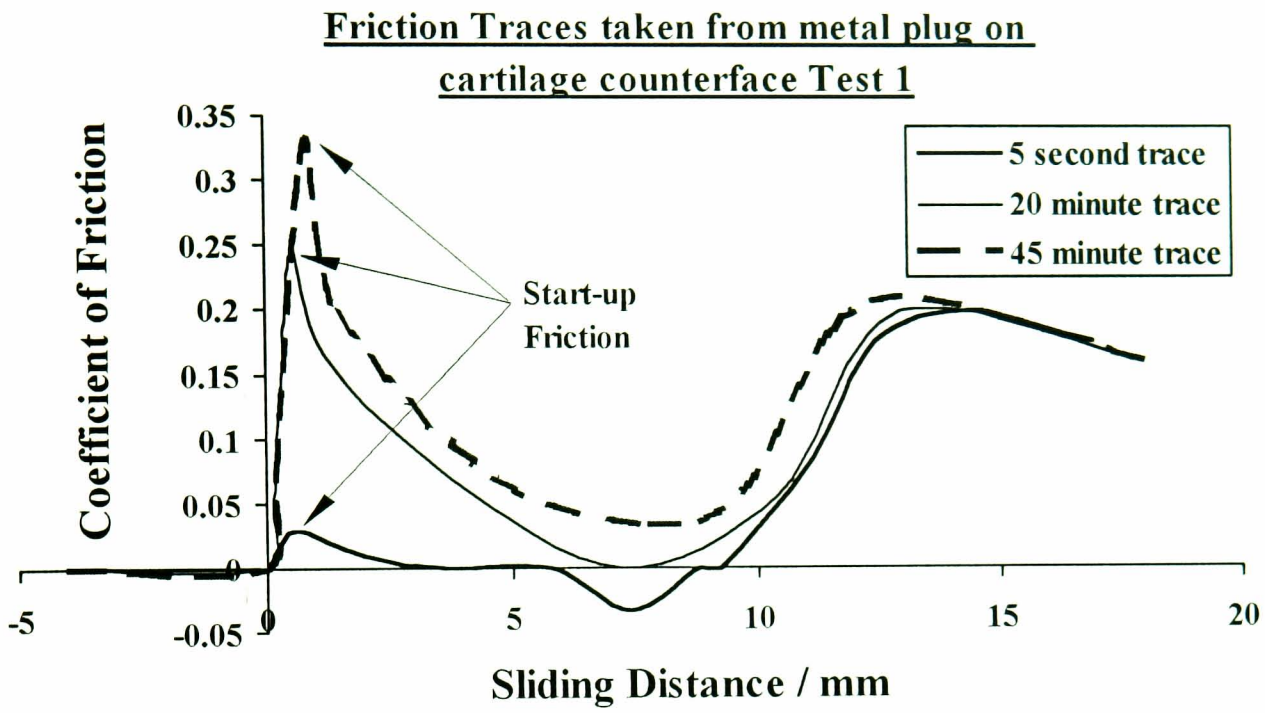
5.3 Results

5.3.1 Steady State and Start-Up Friction

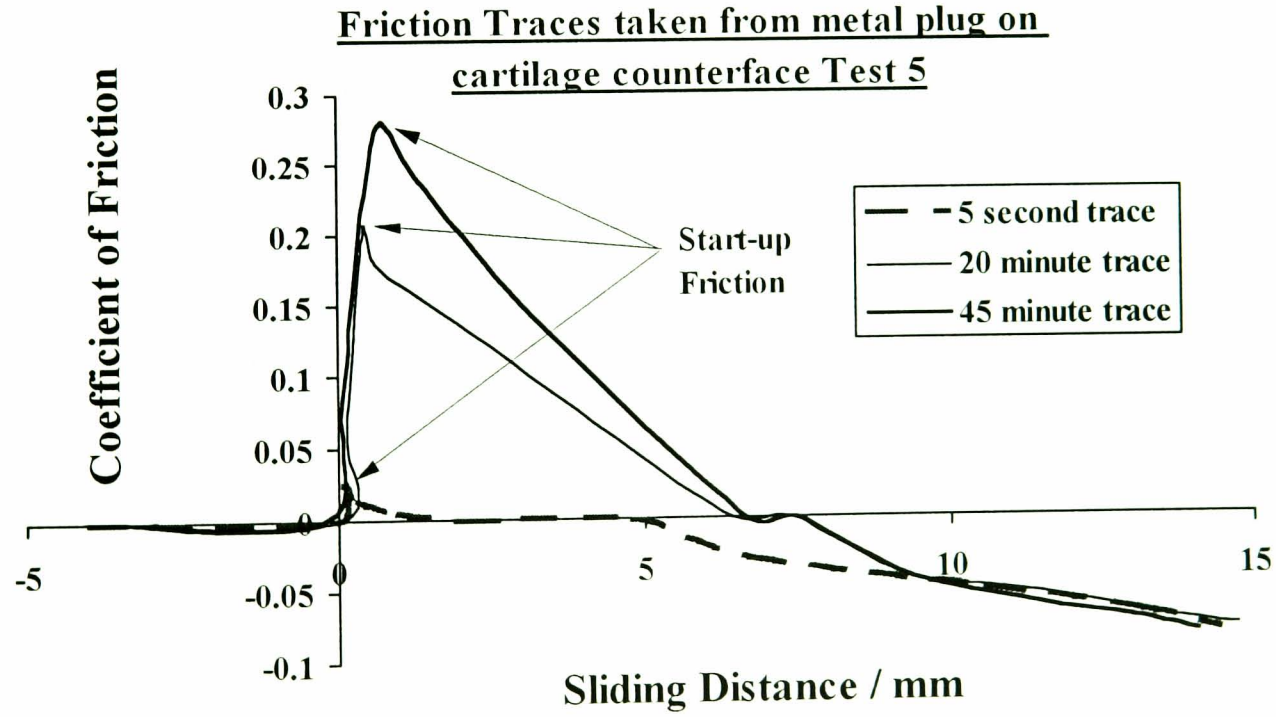
For the tests conducted on the cartilage counterface only the start-up friction was recorded. The reason for this was two-fold, firstly the cartilage counterfaces were not sufficiently even (both in the direction of sliding and perpendicular to the direction of sliding) to record reliable results during sliding and secondly the steady state coefficient of friction was not deemed to be dependent on loading time as after the immediate initiation of the sliding i.e. the start-up, the metal plug was then sliding upon previously unloaded and thereby a fully hydrated part of the cartilage counterface. Therefore the steady state coefficient of friction was not seen to vary for this contact configuration for any particular test, after stationary loading times of between 5 seconds and 45 minutes were applied; see Figure 5-2 in which traces from Tests 1, 5 and 15 are included. This figure shows quite clearly that the start-up friction was increasing with loading time; note the sharp initial rise of the start-up friction peaks followed by a more gradual decline as more and more fully hydrated cartilage entered the contact. However as the metal plug moved away from the loaded cartilage area (which increased in size with loading time), the friction traces all eventually converged, for a particular counterface. This was the case for all the cartilage counterfaces tested. Thus start-up friction was increasing with load duration whereas steady state friction was independent of loading time, as all the traces were meeting up at a point beyond the stationary loaded area of the cartilage. From this point onwards the alteration of the 'steady state' friction, with position along the counterface, merely reflected the nature of the gentle undulations and variations in topography of the particular cartilage counterface being tested.

For metal plugs 1 and 2 a similar contact area, and therefore contact stress, was evident upon the removal of load, especially after 45 minutes of loading. The limiting factors of the contact area were found to be the meeting of two slightly convex surfaces; being the radii of curvature of the metal plug and that of the cartilage surface in the plane perpendicular to the sliding direction (an unavoidable

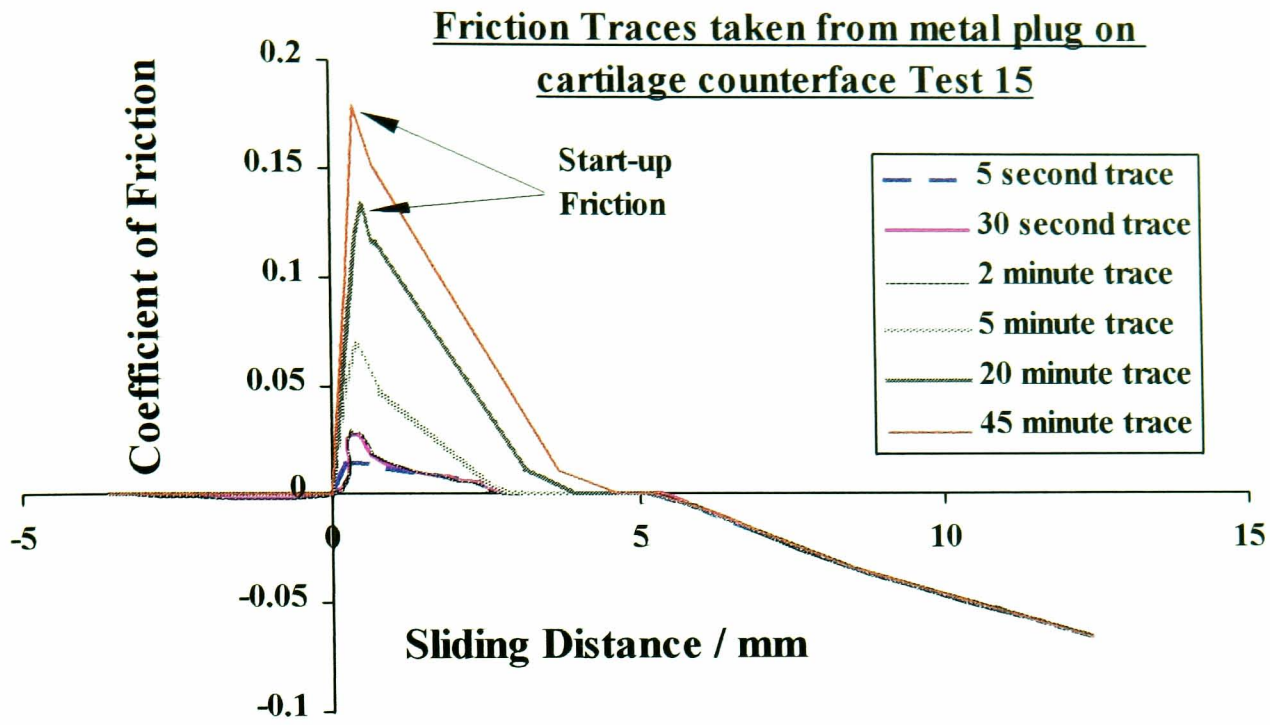
anatomical feature of the bovine femoral and humeral heads), whilst having a counterface compliant layer (i.e. the cartilage) of only 1-2 mm thickness.



a)



b)



c)

Figure 5-2 a),b),c) Plots of coefficient of friction versus sliding distance taken from metal plug on cartilage counterface tests 1, 5 & 15. Start-up friction was measured at the peak of the trace, as indicated.

5.3.2 Start-Up Friction Coefficient Data

Figure 5-3 shows a complete data set taken for a single metal plug on cartilage counterface test. The figure reveals the good repeatability of the results between each of the two particular loading time readings measured at 5 seconds, 30 seconds, 2 minutes, 5 minutes and 20 minutes, which was evident for all the tests. For all the tests (1-15), this data pair was averaged to provide a single value for each loading time. It is this value which is displayed in Table 5-3, and then used for any statistical calculations (i.e. means, standard deviations and t-Tests).

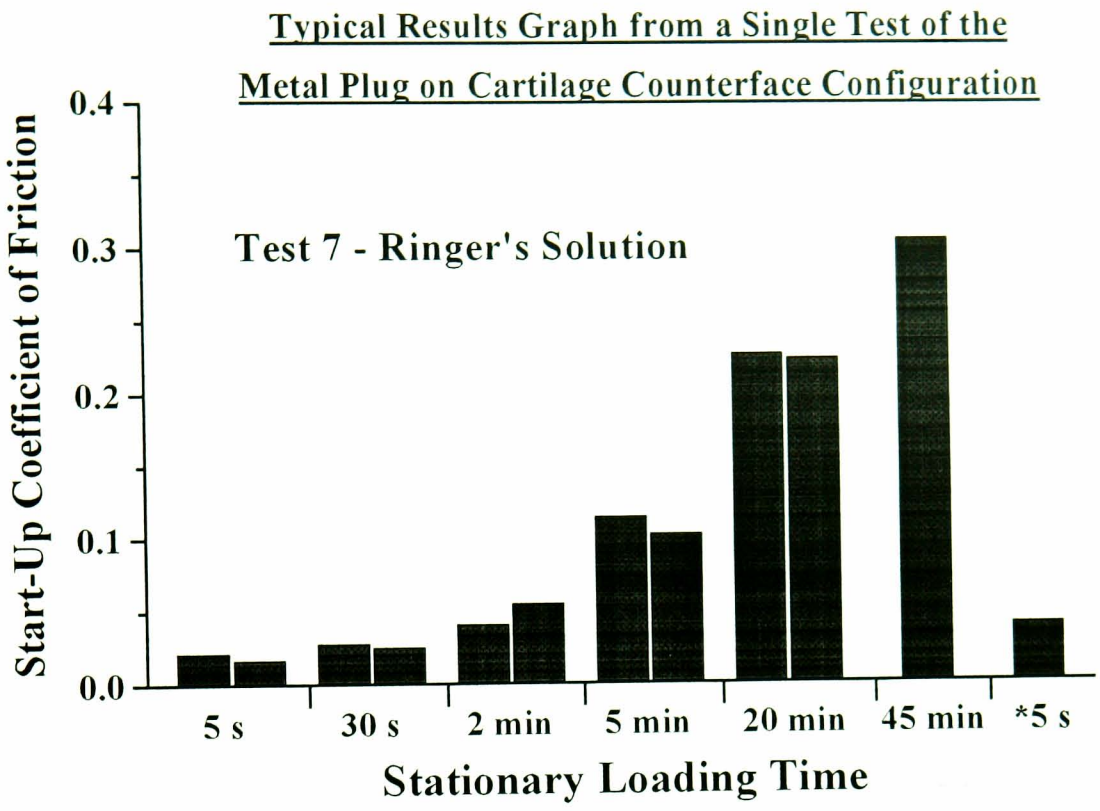


Figure 5-3 Start-up coefficient of friction values from Test 7.

It is important to remember that only start up friction was recorded. Once past the start-up point the friction traces were not analysed for any measurement of friction coefficient. This was because unevenness of the cartilage counterface surface provided erroneous data (probably due to the varying direction of the forces placed on the loading arm of the friction apparatus as the specimen traversed an uneven surface).

	Stationary Loading Times						
	5 s	**30 s**	**2 min**	**5 min**	**20 min**	**45 min**	***5 s**
Test 1 Ringer's Solution	0.038	0.039	0.071	0.097	0.219	0.349	0.131
Test 2 Ringer's Solution	0.025	0.034	0.052	0.093	0.249	0.316	0.064
Test 3 Ringer's Solution	0.025	0.024	0.049	0.083	0.253	0.364	0.063
Test 4 Ringer's Solution	0.015	0.027	0.047	0.078	0.191	0.270	0.040
Test 5 Ringer's Solution	0.039	0.044	0.054	0.097	0.186	0.290	0.050
Test 6 Ringer's Solution	0.004	0.013	0.044	0.075	0.202	0.276	0.037
Test 7 Ringer's Solution	0.020	0.027	0.048	0.109	0.227	0.307	0.040
Mean	**0.024**	**0.030**	**0.052**	**0.090**	**0.218**	**0.310**	**0.061**
Standard Deviation	**0.013**	**0.010**	**0.009**	**0.012**	**0.027**	**0.036**	**0.033**
Test 8 Synovial Fluid	0.035	0.035	0.048	0.068	0.169	0.325	0.110
Test 9 Synovial Fluid	0.032	0.055	0.066	0.096	0.220	0.377	0.170
Test 10 Synovial Fluid	0.028	0.039	0.059	0.115	0.243	0.320	0.042
Test 11 Synovial Fluid	0.024	0.032	0.055	0.093	0.202	0.259	0.036
Synovial Fluid Test 12	0.035	0.039	0.035	0.053	0.125	0.193	0.029
Test 13 Synovial Fluid	0.035	0.034	0.033	0.063	0.159	0.232	0.030
Test 14 Synovial Fluid	0.035	0.039	0.039	0.067	0.146	0.208	0.027
Test 15 Synovial Fluid	0.020	0.029	0.027	0.064	0.138	0.177	0.030
Mean	**0.030**	**0.037**	**0.045**	**0.077**	**0.175**	**0.261**	**0.059**
Standard Deviation	**0.006**	**0.008**	**0.014**	**0.021**	**0.042**	**0.072**	**0.053**

Table 5-3 Start-up coefficient of friction values for the metal plug on cartilage counterface tests. The means and standard deviations, for both lubricants, are provided for all of the loading times tested.

The means and standard deviations from Table 5-3 are displayed in two graphs, Figure 5-4 and Figure 5-5. Figure 5-4 compares the results of both lubricants for

each of the stationary loading times adopted. Student's t-tests were used to compare the means between the two lubricants at each of the loading periods tested. The results of the t-tests are displayed in Table 5-4. From Table 5-4 it can be seen that the only statistical difference between the lubricants was at the 20 minute loading time. At this loading time synovial fluid was found to have a significantly lower mean start-up coefficient of friction than that of the Ringer's solution ($p<0.05$). However for stationary loading times of 2 minutes and above the mean values for synovial fluid as the lubricant were always found to be ~10-20% lower than the Ringer's solution values, Figure 5-4.

The start-up coefficient of friction for the metal plug on cartilage counterface contact configuration was shown to increase with stationary loading time (Figure 5-5), in much the same manner as for the steady state coefficient of friction for the two previous metal counterface tests. The values of friction coefficient were also similar compared to the cartilage on metal configurations. This was not too surprising because as previously discussed in Chapter 4 the start-up and steady state values were found to be quite alike for the two cartilage on metal counterface contacts investigated.

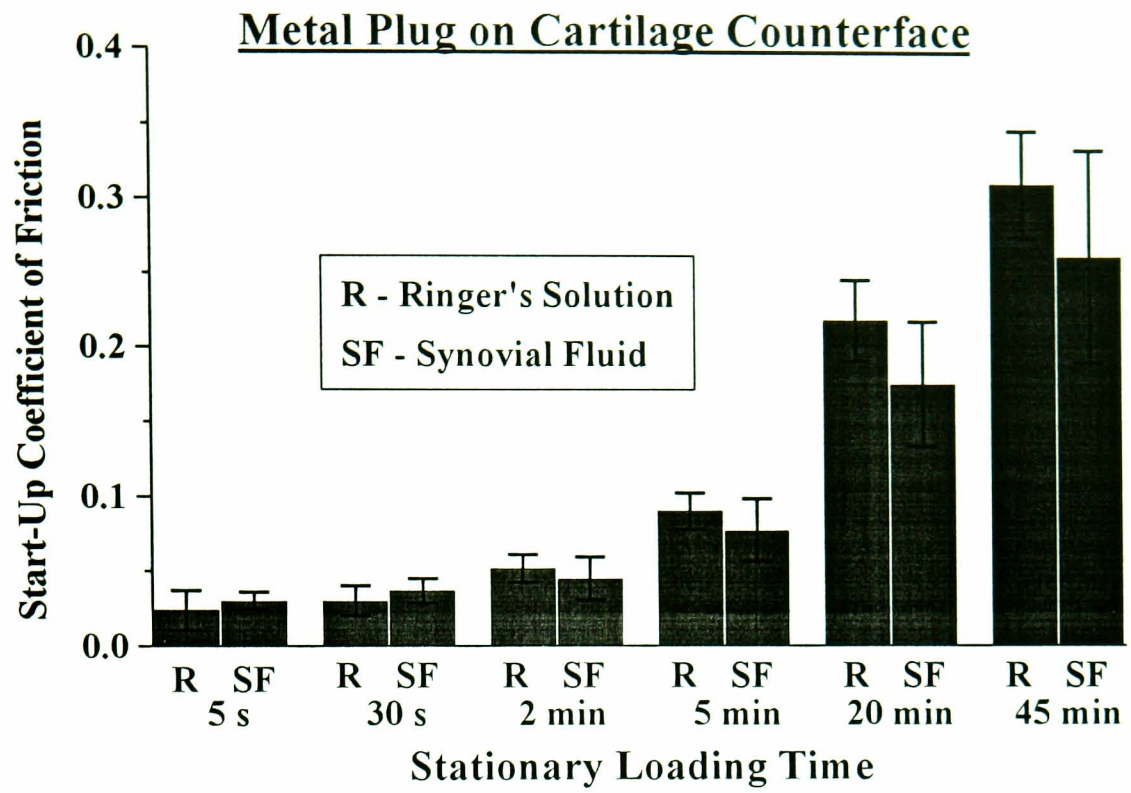


Figure 5-4 Column graph of the metal plug on cartilage counterface data, means and standard deviations are shown. Comparing the results of both lubricants at each loading time.

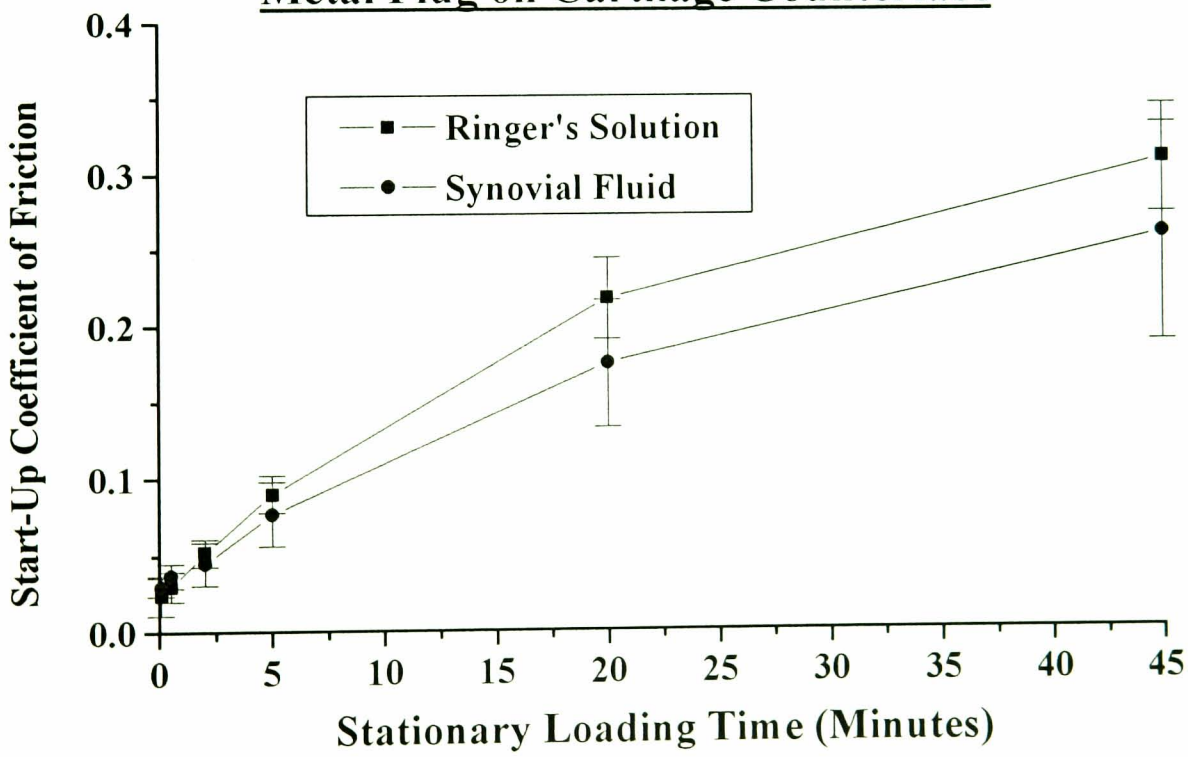


Figure 5-5 Line graph of the metal plug on cartilage counterface data, means and standard deviations are shown. Displaying the nature of the rise in friction coefficient with stationary loading time.

t-Tests							
SF vs. Ringer's	5 s	30 s	2 min	5 min	20 min	45 min	*5 s
	0.191	0.114	0.268	0.185	0.039	0.128	0.950

Table 5-4 t-Test for synovial fluid (SF) versus Ringer's solution for all loading times.

5.3.3 Load Removal Tests

For a load removal period of only 1 minute, immediately after the 45 minute stationary loading tests, the subsequent drop in friction levels, as demonstrated by the further 5 second (*5 s) loading tests, was quite dramatic, Figure 5-6. These *5 second results were again similar for both lubricants. However, from Table 5-3, it can be seen that for both lubricants the respective mean value for the *5 second loading time was still approximately double the mean value for the original 5 second loading time test.

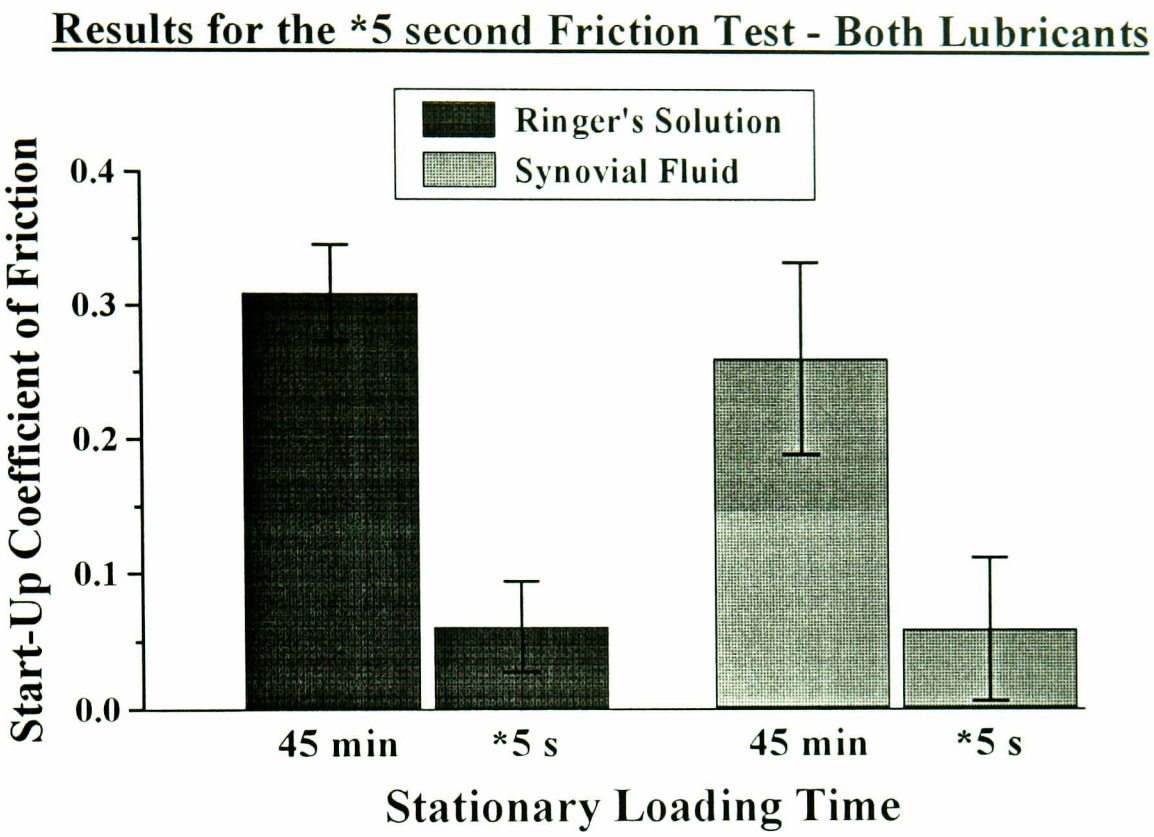


Figure 5-6 After the 45 minute loading test (45 min) the cartilage counterface was left unloaded for 1 minute. Following this 1 minute load removal period a 5 second stationary loading time test was undertaken (*5 s). The mean values are shown along with their respective standard deviations.

5.4 Discussion

5.4.1 Consistency of Results

For all the fifteen cartilage counterfaces tested the repeatability of the friction readings was very good. This was proven by the respective close matching of the two readings taken at 5 seconds, 30 seconds, 2 minutes, 5 minutes and 20 minutes for all the counterfaces. This has been graphically illustrated by plotting a typical set of results from one of the cartilage counterfaces tested, Figure 5-3. By producing data which was satisfactorily repeatable at each loading time, it was evident that no significant wear and/or cartilage degradation affecting the coefficient of friction values was occurring.

For the stationary loading friction measurements taken at 2 minutes and above the reproducibility of the results between tests 1-7, for Ringer's solution, and between tests 8-15, for synovial fluid, were also pleasing (Table 5-3), with the coefficient of variance at each loading time (standard deviation / mean) being ~ 0.1-0.3 for all these results. For the lower stationary loading times of 5 and 30 seconds, as well as the *5 second loading test, the coefficient of variance was higher, ranging from generally 0.2-0.5; (for the synovial fluid *5 second the coefficient of variance was especially high at 0.9). For these lower loading times the start-up friction could have been particularly sensitive to the geometry of the contact between the metal plug and the particular cartilage counterface being tested, which was naturally variable according to the overall topography of the specific counterface involved. For the *5 second tests the variability was further compounded by some difficulty in reapplying the metal plug to exactly the same position, after the 1 minute load removal, as for the prior 45 minute test. For the higher loading times the geometries of the contacts would have become more uniform as the cartilage surfaces complied to the curvature of the metal plug.

Whenever a hard material is in contact with a relatively compliant material the possible occurrence of ploughing friction must be considered. As shown in Table 5-3 the start-up coefficient of friction values were not substantially different when

either metal plug 1 (Tests 1-5, 10, 13, 14) or metal plug 2 (Tests 6-9, 11, 12, 15) was used. Therefore it could be reasonably assumed that the smaller diameter metal plug 1 was not creating depressions in the cartilage surface, during stationary loading, large enough for the outer diameter edges of the plug to contact the cartilage surface upon the start of sliding, even after 45 minutes of stationary loading. In fact, as mentioned in the results section, the two metal plugs produced contact areas with the cartilage counterface of comparable sizes. This was initially of concern for this contact configuration and, of course, the reason why metal plug 2 was used in addition to metal plug 1, (Figure 5-1). It can therefore be stated that none of the obtained friction values were unduly affected by excessive edge contacts and ploughing friction. This statement is further supported by the similarity of the values with those of the cartilage on metal contacts, Chapter 4, which shall be discussed below. Generally in synovial joints the convex/concave geometry of the cartilage mating surfaces is such that ploughing friction is not a prevalent occurrence.

5.4.2 Effect of Lubricant and Loading Time

Synovial fluid reduced the levels of friction by ~10-20% for all the 2 to 45 minute loading times, Figure 5-4 and Figure 5-5. This drop however was only statistically significant for the 20 minute loading time. Nonetheless a trend did appear to be developing which highlighted synovial fluid as the superior lubricant. The synovial fluid was more viscous than the Ringer's solution and some additional boosted and/or boundary lubrication between the metal and cartilage contacts may have been provided by the synovial fluid. But once again, as for the cartilage on metal contacts, the choice of lubricant was not the governing factor influencing friction. Instead friction was primarily influenced by the duration of stationary loading. The start-up coefficient of friction values ranged from approximately 0.03 at 5 seconds to just under 0.3 after 45 minutes of loading for both lubricants.

Load duration, or more specifically the loading history of the tested cartilage specimens, was further emphasised as the primary factor governing friction by the

load removal tests. The load removal tests, Figure 5-6, revealed large decreases in coefficient of friction levels from just under 0.3 at the 45 minutes loading period to approximately 0.06 after only a 1 minute load removal period followed by a further 5 second (*5 second) loading test, again for both lubricants.

From the 5 second loading time to the 5 minute loading time there was a sharp rise in friction, with mean values rising from 0.024 to 0.090 and 0.030 to 0.077 for Ringer's solution and synovial fluid respectively. From the 5 minute to 45 minute loading times the increase in the coefficient of friction values became more gradual, Figure 5-5, with mean values at 45 minutes loading being 0.310 and 0.261 for Ringer's solution and synovial fluid respectively. This non-linear rise, like the drop in coefficient of friction levels for the *5 second tests, was in complete agreement with the nature of the rise in friction levels for the cartilage on metal contacts discussed in the previous chapter. It was therefore concluded that the biphasic properties of the articular cartilage were governing the coefficient of friction results for this metal on cartilage configuration also.

5.4.3 Biphasic Lubrication for Metal on Cartilage Counterface Contacts

So in brief the friction was initially low at the lower loading times due to a high proportion of the load carried by the pressure gradients within the fluid phase, as flow of the fluid phase occurs away from the loaded region (Ateshian et al., 1994). The fluid phase has a negligible associated coefficient of friction and therefore any load which it carries reduces the overall friction by reducing the amount of load carried by the solid phase. For the longer loading times, as more and more fluid is forced away, the pressure gradients become less, and so less load can be supported by the fluid phase. This consequently increases the amount of load carried by the solid phase which, of course, has a particular coefficient of friction attributable to itself. It is this increasing load carriage on the solid phase that therefore increases the overall friction force exerted when the metal and cartilage surfaces slide across one another. Hence the overall measured coefficient of friction values increased with loading time. The non-linear rise in the coefficient of friction values was directly related to the non-linear decline of

the pressure gradients and reduction in flow within the fluid phase while the cartilage specimens were subject to constant stationary loads.

Likewise for the *5 second loading tests, Figure 5-6, the dramatic fall of the start-up coefficient of friction values can be explained by the biphasic properties of articular cartilage. After the 45 minutes loading test the 1 minute load removal allowed a large proportion of the previously exuded fluid back into the cartilage. Therefore upon the reapplication of load for the *5 second loading test the fluid phase would have been capable of carrying a higher percentage of the load, than after the 45 minute loading period, creating a large drop in the friction levels.

5.4.4 Comparison with Cartilage on Metal Configurations

As well as the non-linear nature of the rise of coefficient of friction values plotted against stationary loading time being similar to the cartilage on metal configuration, the respective coefficient of friction values themselves were quite comparable, c.f. Figure 4-4, Figure 4-5 and Figure 5-5. The only discernible difference appeared at the respective 5 second values whereby the start-up coefficient of friction values for the metal plug on cartilage counterface configuration were moderately higher than the steady state coefficient of friction values for the cartilage on metal counterface configurations. This difference would have been somewhat less if the start-up coefficient of friction values for the cartilage on metal counterface configurations had been compared, as occasionally small peaks were observed at start-up relative to the steady state values. Also at the 5 second loading time for the metal on cartilage configuration the start-up values may have been higher due to the lack of perfectly flat mating surfaces, attributable to small incongruities between the metal plug and cartilage counterface.

It is important to re-emphasise the inherent low friction properties of the biphasic cartilage at this point, as discovered by both the cartilage on metal and metal on cartilage configurations. Firstly low values of friction coefficient, of less than or equal to 0.05, were maintained for up to 2 minutes after the application of

stationary loading. Secondly, even after 45 minutes of stationary loading, low coefficient of friction values could again be achieved, of approximately 0.05, after only 1 minute of load removal; which allowed for substantial rehydration of the cartilage matrix. This provides considerable understanding into the mechanism of maintaining low friction in synovial joints under the most severe loading conditions and duty cycles.

5.4.5 Concluding Remarks

In conclusion the effect of reversing the cartilage on metal contact to a metal on cartilage contact revealed very similar findings regarding the effects of lubricant, stationary loading times and load removal, and the influence of the biphasic properties of the cartilage. This was portrayed by similar non-linear rises in friction levels with loading time and similar coefficient of friction values between this configuration and the two contact configurations discussed in Chapter 4.

There was however one unique and important variation. As stated in the results start-up friction was recorded only, one of the reasons being that the steady state coefficient of friction was not deemed to be dependent on loading time as after the immediate initiation of the sliding, i.e. the start-up, the metal plug was then sliding upon previously unloaded and thereby a fully hydrated part of the cartilage counterface.

This finding provides a very useful analogue and insight into what occurs in physiological conditions after prolonged periods of stationary loading in a normal, healthy synovial joint. Movement in a synovial joint will always lead to contact areas translating over previously unloaded cartilage surfaces possessing the normally high amounts of fluid content. Therefore after a period of stationary loading, although the initial start-up friction may be quite high, movement of the contact areas onto previously unloaded regions of the joint surface will produce substantial drops in the friction levels. Start-up friction itself may be countered against by momentarily unloading the joint prior to motion, if possible, e.g. for the hip and knee joints after standing still shifting body weight onto one leg whilst beginning to move the other leg forward. In fact with movement in some joints

the contact areas under load may change and move on both of the opposing cartilage surfaces due to combined rolling and sliding actions. So as soon as motion occurs, just after start-up, neither of the previously loaded areas will remain in contact with the opposing cartilage surface. This is, therefore, one very effective way that nature has designed synovial joints in order to prevent excessively high friction levels to occur. Friction between cartilage on cartilage contacts will be investigated in the next chapter.

6. Stationary Loading Friction Tests - Cartilage on Cartilage Contacts

6.1 Introduction

The successful conduction of the metal plug on cartilage counterface tests provided informative and reproducible data of start-up coefficient of friction values and made similar studies for cartilage on cartilage contacts a viable proposition. For this contact configuration 9 millimetre diameter cartilage plugs were loaded against cartilage counterfaces for periods of between 5 seconds and 45 minutes. After the selected stationary loading time interval was over horizontal sliding of the counterface along the plug was initiated. Start-up coefficients of friction were recorded using both Ringer's solution and synovial fluid as the lubricant.

Naturally tribological experiments involving cartilage/cartilage contacts were of great interest as they had the best potential to further elucidate the frictional characteristics of synovial joints. This would build upon insights already gained from the three previous contact configurations described in chapters 4 and 5. For the two opposing articular cartilage layers important issues to be addressed were 'Is the effect of the biphasic properties of cartilage in carrying load within the fluid phase, and thereby reducing friction, enhanced by the addition of a second biphasic cartilage layer?' and 'Can the beneficial boundary lubricating ability of synovial fluid be shown to be augmented by these cartilage/cartilage contacts as compared to the cartilage/synthetic (metal) contacts?'.

A hydrogel material was also used as a counterface for stationary loading friction tests with a 9 millimetre diameter cartilage plug. This cartilage/hydrogel contact configuration permitted both measurements of start-up and steady state coefficients of friction to be made. The hydrogel counterface was intended to simulate a perfectly flat cartilage layer, possessing similar biphasic attributes to articular cartilage (Corkhill et al., 1990 and Oka et al., 1990).

6.2 Experimental Procedure

Friction of the cartilage/cartilage contacts was recorded within a mixed lubrication regime, using synovial fluid and Ringer's solution. The main test variable was the period of stationary loading, applied prior to sliding, which ranged from 5 seconds to 45 minutes. This was the amount of time that the cartilage plug spent loaded against the cartilage counterface prior to sliding the counterface horizontally and measuring friction. Eight tests were conducted using Ringer's solution as the lubricant and eight tests using synovial fluid as the lubricant. For each test a different 9 mm cartilage plug was used. A new cartilage counterface was generally employed after 2 or 3 tests, using a total of 6 in all, as specified in Table 6-1. Of course a single cartilage counterface could not have been used throughout the study due to deterioration and possible wear effects.

Cartilage counterfaces were sectioned from bovine humeral and femoral heads to dimensions similar to those of the metal counterface, approximately 30 x 70 millimetres and having a thickness of 10-20 mm (largely consisting of the underlying subchondral and cancellous bone). These specimens were prepared in order to provide as even a cartilage surface area as possible.

Test	Cartilage Counterface	Cartilage Plug	Lubricant
1	**i**	1	Synovial Fluid
2	**i**	2	Synovial Fluid
3	**ii**	3	Synovial Fluid
4	**ii**	4	Synovial Fluid
5	**ii**	5	Synovial Fluid
6	**iii**	6	Synovial Fluid
7	**iii**	7	Synovial Fluid
8	**iv**	8	Synovial Fluid
9	**iv**	9	Ringer's Solution
10	**v**	10	Ringer's Solution
11	**v**	11	Ringer's Solution
12	**v**	12	Ringer's Solution
13	**vi**	13	Ringer's Solution
14	**vi**	14	Ringer's Solution
15	**vi**	15	Ringer's Solution
16	**vi**	16	Ringer's Solution

Table 6-1 Test conditions for the cartilage plug on cartilage counterface configuration, detailing tests 1-16.

Each test consisted of the following stationary loading time friction measurements; two readings at 5 seconds, 30 seconds, 2 minutes, 5 minutes and 20 minutes, with a single reading taken after 45 minutes stationary loading. Equal unloading times following each loading period was applied throughout to allow for rehydration of the cartilage counterface, except following the 45 minute test when, after only 1 minute of load removal for the cartilage counterface and 10 seconds load removal for the cartilage plug, a 5 second loading period test was performed (*5 seconds, Table 6-2). For this cartilage/cartilage contact configuration the recording of two readings at 5 seconds, 30 seconds, 2 minutes, 5 minutes and 20 minutes was considered necessary to confirm that wear and/or cartilage degradation, affecting the coefficient of friction values, was not occurring by providing data which was hoped to be satisfactorily repeatable for each loading time.

Stationary Loading Times
5 seconds
30 seconds
2 minutes
5 minutes
20 minutes
45 minutes
*5 seconds

Table 6-2 Stationary loading times adopted as friction measurements for the cartilage plug on cartilage counterface tests. Two readings were taken at the 5 second, 30 second, 2 minute, 5 minute and 20 minute loading times.

For this contact configuration only the initial friction at the start up of motion was recorded, and not steady state friction, as steady state friction was not found to be a function of loading time. This was simply because only the area of the cartilage counterface at the start-up of motion was subjected to any load. The stationary loading upon the cartilage plug did not appear to influence friction during sliding, once past the start-up region.

A 9 millimetre diameter cartilage plug was also tested against a hydrogel[1] counterface, with Ringer's solution as the lubricant. Friction readings were recorded after 5 seconds, 2 minutes, 5 minutes, 10 minutes, 20 minutes and 45 minutes stationary loading times. Following the 45 minute test, after 1 minute of load removal for the hydrogel counterface and 10 seconds load removal for the cartilage plug, a 5 second loading period test was performed (*5 seconds, Table 6-3). Both the hydrogel and cartilage specimen was then left immersed in Ringer's solution with no applied load for one hour, to allow for full rehydration, and then the same friction readings were repeated. The first set of friction measurements were referred to as the *initial* results while the subsequent identical set of measurements were referred to as the *repeat* results. For this contact configuration both start-up and steady state friction values were recorded.

---

[1] The hydrogel material was *N*-vinyl pyrrolidone methyl methacrylate (PC110) as described in Chapter 2.

***Initial* Friction Readings taken after Stationary Loading Times:**
5 seconds
2 minutes
5 minutes
10 minutes
20 minutes
45 minutes
*5 seconds
Hydrogel counterface and cartilage plug unloaded while immersed in Ringer's solution for one hour prior to repeat friction measurements
***Repeat* Friction Readings taken after Stationary Loading Times:**
5 seconds
2 minutes
5 minutes
10 minutes
20 minutes
45 minutes
*5 seconds

Table 6-3 Test Protocol adopted for the cartilage plug on hydrogel counterface configuration. Only one cartilage plug was tested for this configuration.

6.3 Results

6.3.1 Steady State and Start-Up Friction

For the cartilage on cartilage friction tests, like those for the metal on cartilage, only the start-up friction was recorded, for exactly the same reasons as before. Firstly the cartilage counterfaces were not sufficiently even (both in the direction of sliding and perpendicular to the direction of sliding) to record reliable results during sliding and secondly the steady state coefficient of friction was not deemed to be dependent on loading time as after the immediate initiation of the sliding i.e. the start-up, the cartilage plug was then sliding upon previously unloaded and thereby a fully hydrated part of the cartilage counterface. The effect of stationary loading upon the cartilage plug itself did not appear to influence friction during sliding, once past the start-up region, Figure 6-1 and Figure 6-2. Therefore the steady state coefficient of friction was not seen to vary for this contact configuration for any particular test, after stationary loading times of between 5 seconds and 45 minutes were applied; see Figure 6-1 and Figure 6-2 in which traces from Tests 2 and 5 are included. These figures illustrate quite clearly that the start-up friction was increasing with loading time; note the sharp initial rise of the start-up friction peaks followed by a more gradual decline as more and more fully hydrated cartilage entered the contact.

The area of the cartilage counterface in direct contact with the cartilage plug would have increased with loading time up to an approximate maximum of 4 millimetres in radius at the 45 minute loading period. From the individual friction traces it can be seen at what point the cartilage plug began to move away from the stationary loaded region of the cartilage counterface and onto fully hydrated tissue by the distance taken for the trace to eventually coalesce with the other friction traces, for a particular test. This distance was, of course increasing with loading time as the stationary loaded area of the cartilage counterface increased with loading time. This was the case for all the cartilage counterfaces tested.

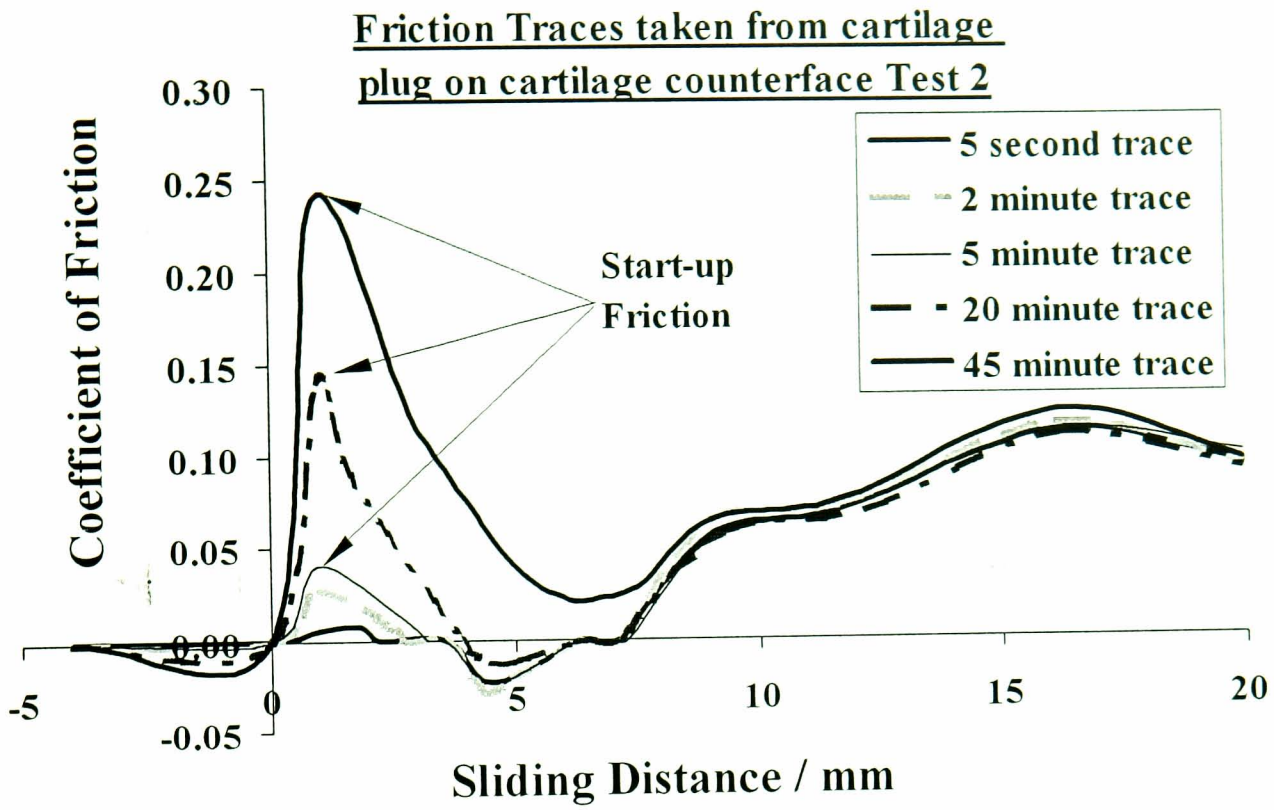


Figure 6-1 Plots of coefficient of friction versus sliding distance taken from the cartilage plug on cartilage counterface test 2. Start-up friction was measured at the peak of the trace, as indicated.

Thus start-up friction was increasing with load duration whereas steady state friction was independent of loading time, as all the traces were meeting up at a point beyond the stationary loaded area of the cartilage. From this point onwards the alteration of the 'steady state' friction, with position along the counterface, merely reflected the nature of the gentle undulations and variations in topography of the particular cartilage counterface being tested (probably due to the varying direction of the forces placed on the loading arm of the friction apparatus as the specimen traversed an uneven surface). As in Chapter 5, (where cartilage counterfaces were also used), it is important to remember that only start up friction was recorded. Once past the start-up point the friction traces were not analysed for any measurement of friction coefficient. This was because unevenness of the cartilage counterface surface provided erroneous data, as explained above.

Generally the stationary loaded region of the cartilage counterface, where start-up friction measurements were recorded from, was the area of the counterface

estimated to be the most even and level. This region would rarely extend beyond a length of 10 to 15 millimetres. However there were a few exceptions and the cartilage counterface used for Test 5 was one of them. The friction traces from Test 5 when plotted against the sliding distance along the counterface, Figure 6-2, provided an excellent account of what was actually happening for these cartilage/cartilage contacts when, after a given period of stationary loading, sliding was started. In Figure 6-2 it can be seen that the start-up friction was increasing with loading time but when the cartilage plug moved away from the stationary loaded region and onto the fully hydrated cartilage the steady state coefficient of friction levels during sliding were of the order of 0-0.01 and evidently unaffected by the amount of loading endured by the cartilage plug.

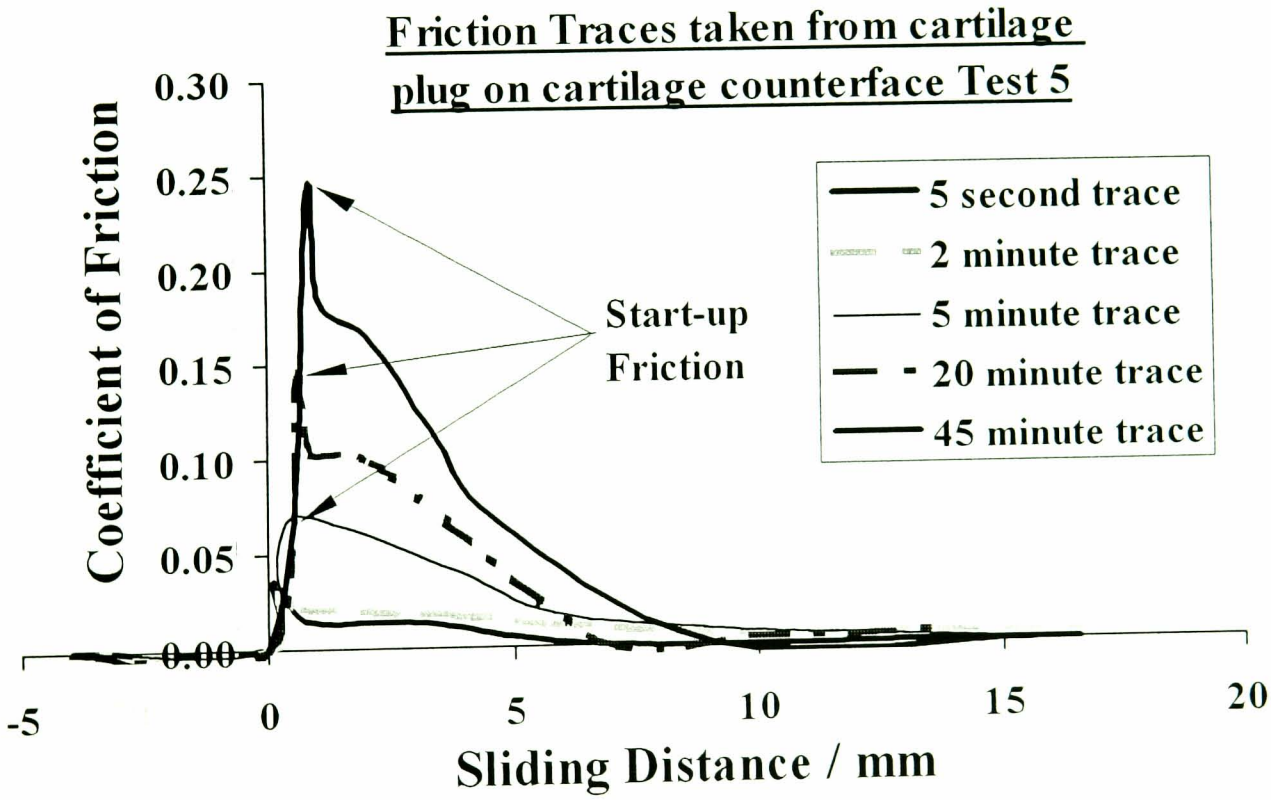


Figure 6-2 Plots of coefficient of friction versus sliding distance taken from the cartilage plug on cartilage counterface test 5. Start-up friction was measured at the peak of the trace, as indicated.

For the friction test conducted on the hydrogel counterface the friction traces of the initial readings have been plotted in Figure 6-3 against the sliding distance along the counterface. Both the start-up and steady state coefficient of friction values were shown to increase with increasing stationary loading time, in much the same manner as for the two cartilage on metal counterface configurations,

Chapter 4. Although the 0.8 mm thick hydrogel counterface was perfectly uniform and even, the friction traces for steady state sliding revealed some systematic variations with sliding distance at all loading times. This was probably caused by the introduction of small amounts of distortion in the hydrogel layer when bonded down onto a hard plastic (polyvinyl chloride) substrate, using cyanoacrylate adhesive, for stability during the tests.

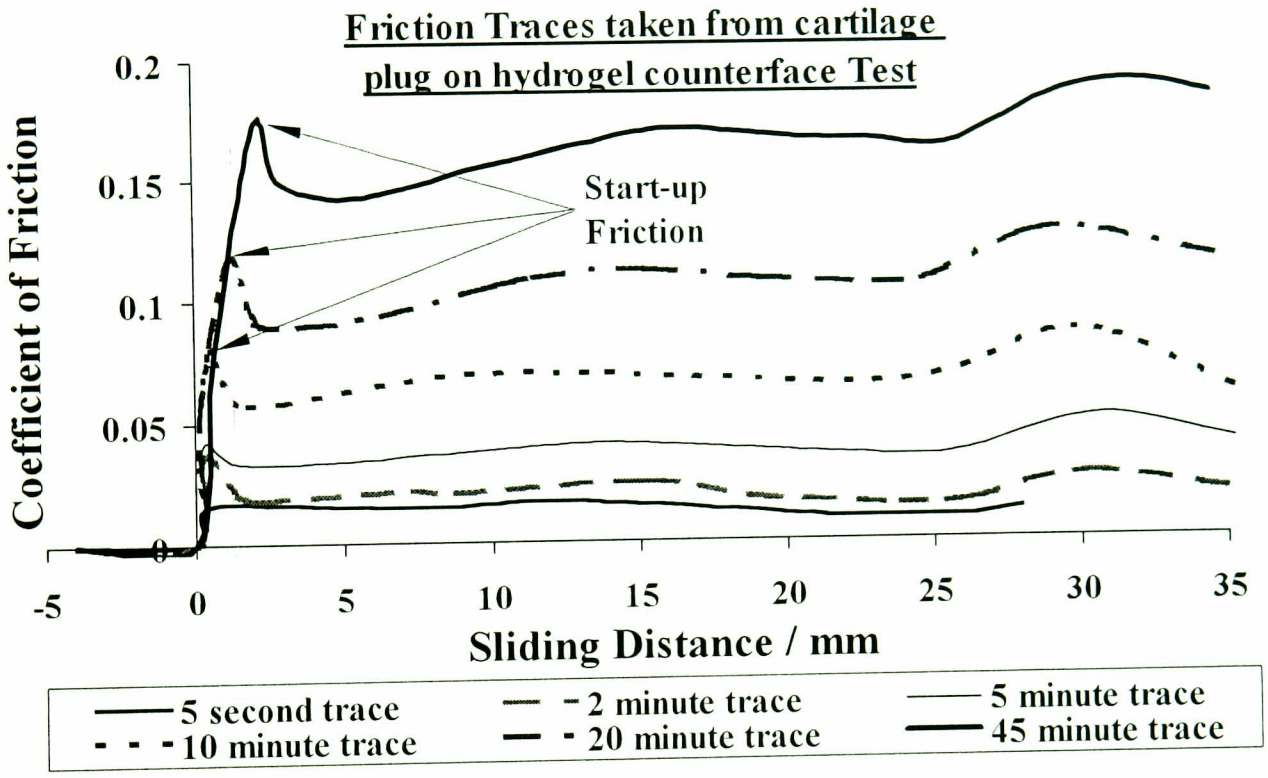


Figure 6-3 Plots of coefficient of friction versus sliding distance taken from the cartilage plug on hydrogel counterface test. Start-up friction was measured at the peak of the trace, as indicated. Steady state friction for each stationary loading time was also recorded.

6.3.2 Start-Up Friction Readings for Cartilage on Cartilage Contacts

Figure 6-4 shows a complete data set for the start-up friction of a single cartilage plug on cartilage counterface test. The figure reveals the good repeatability of the results between each of the two particular loading time readings measured at 5 seconds, 30 seconds, 2 minutes, 5 minutes and 20 minutes, which was evident for all the tests. For all the tests (1-16), this data pair was averaged to provide a single value for each loading time. It is this value which is displayed in Table 6-4, and then used for any statistical calculations (i.e. means, standard deviations and t-Tests).

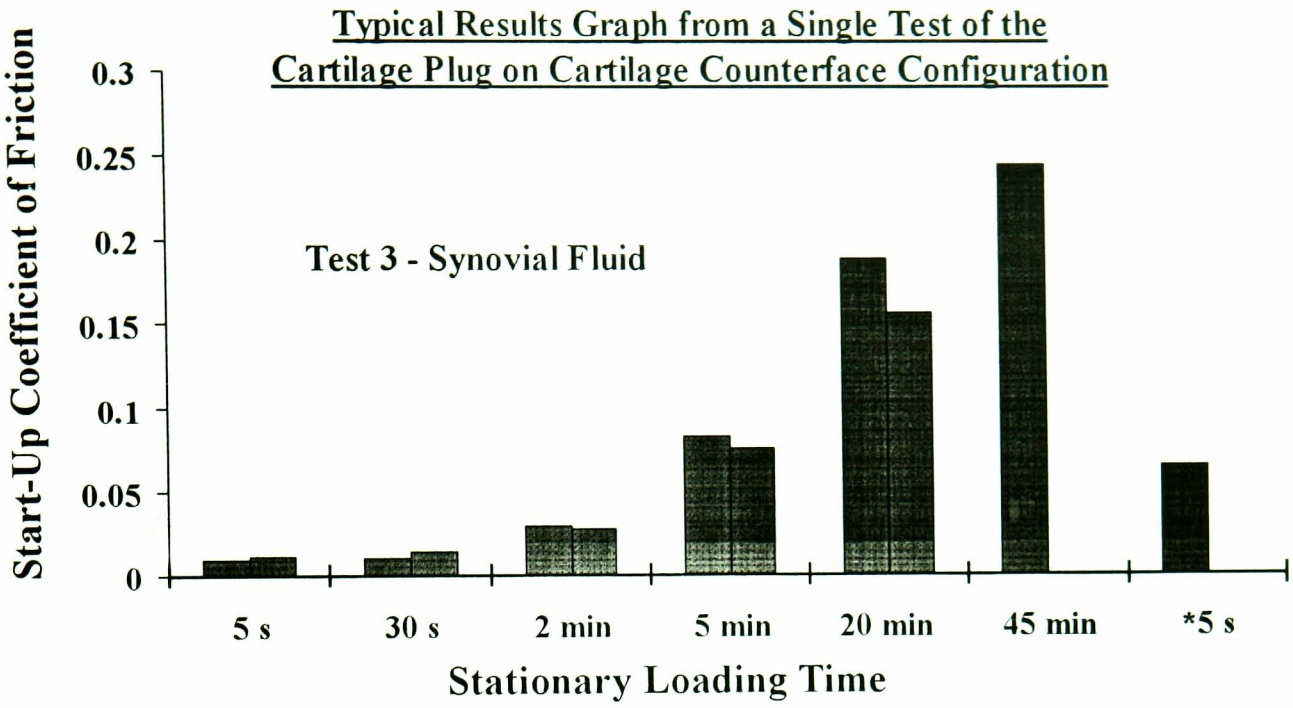


Figure 6-4 Start-up coefficient of friction values from Test 3.

	Stationary Loading Times						
	5 s	**30 s**	**2 min**	**5 min**	**20 min**	**45 min**	***5 s**
Test 1 Synovial Fluid	0.017	0.013	0.022	0.078	0.156	0.220	#
Test 2 Synovial Fluid	0.011	0.014	0.042	0.063	0.145	0.220	#
Test 3 Synovial Fluid	0.010	0.012	0.029	0.079	0.173	0.245	0.065
Test 4 Synovial Fluid	0.016	0.015	0.025	0.064	0.177	0.282	0.074
Test 5 Synovial Fluid	0.028	0.023	0.031	0.044	0.134	0.241	0.052
Test 6 Synovial Fluid	0.006	0.012	0.028	0.075	0.155	0.257	#
Test 7 Synovial Fluid	0.028	0.030	0.028	0.065	0.194	0.290	0.063
Test 8 Synovial Fluid	0.039	0.043	0.039	0.055	0.177	0.277	0.133
Mean	**0.019**	**0.020**	**0.030**	**0.065**	**0.164**	**0.254**	**0.077**
Standard Deviation	**0.011**	**0.011**	**0.007**	**0.012**	**0.020**	**0.027**	**0.032**
Test 9 Ringer's Solution	0.031	0.024	0.031	0.049	0.148	0.265	0.053
Test 10 Ringer's Solution	0.039	0.033	0.040	0.073	0.232	0.380	0.090
Test 11 Ringer's Solution	0.032	0.046	0.040	0.074	0.207	0.401	0.051
Test 12 Ringer's Solution	0.040	0.031	0.038	0.082	0.180	0.288	0.079
Test 13 Ringer's Solution	0.024	0.037	0.030	0.110	0.188	0.240	0.029
Test 14 Ringer's Solution	0.044	0.031	0.057	0.101	0.194	0.277	0.033
Test 15 Ringer's Solution	0.023	0.032	0.049	0.086	0.215	0.285	0.083
Test 16 Ringer's Solution	0.034	0.038	0.045	0.090	0.194	0.299	0.023
Mean	**0.033**	**0.033**	**0.041**	**0.082**	**0.195**	**0.305**	**0.060**
Standard Deviation	**0.008**	**0.007**	**0.010**	**0.020**	**0.027**	**0.061**	**0.025**

Table 6-4 Start-up coefficient of friction values for the cartilage plug on cartilage counterface tests. The means and standard deviations, for both lubricants, are provided for all of the loading times tested. # indicates that no reading was taken.

The means and standard deviations of the results given in Table 6-4 are displayed in two graphs, Figure 6-5 and Figure 6-6. Figure 6-5 compares the results of both lubricants for each of the stationary loading times adopted. Student's t-tests were used to compare the means between the two lubricants at each of the loading periods tested. The results of the t-tests are shown in Table 6-5. From Table 6-5 it can be seen that, for these cartilage on cartilage contacts, synovial fluid was found to have a statistically significant lower mean start-up coefficient of friction than that of the Ringer's solution for all the loading times ($p<0.05$); except for the *5 second load removal test.

The start-up coefficient of friction for the cartilage plug on cartilage counterface contact configuration was shown to increase with stationary loading time (Figure 6-6), in much the same manner as for the metal plug on cartilage counterface configuration, chapter 5, and the steady state coefficient of friction for the two cartilage on metal counterface studies, chapter 4. The values of friction coefficient were also similar compared to the previous contact configurations.

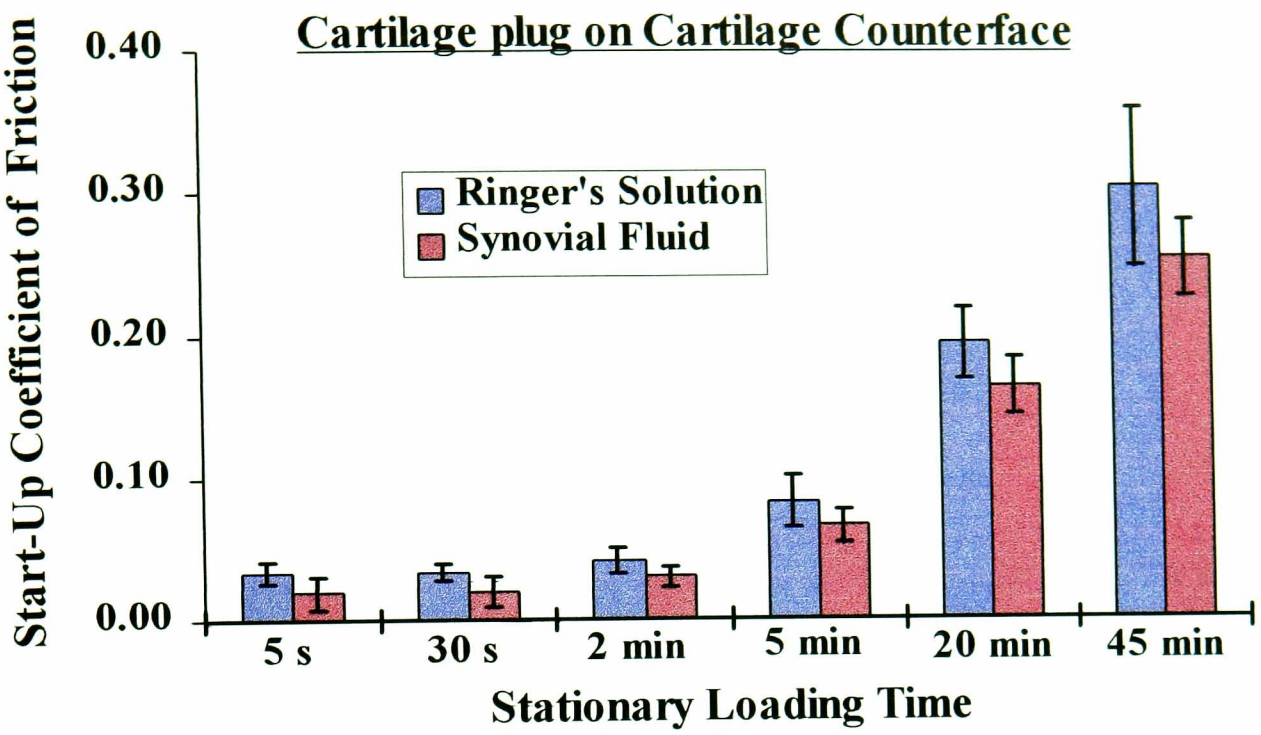


Figure 6-5 Column graph of the cartilage plug on cartilage counterface data, means and standard deviations are shown. Comparing the results of both lubricants at each loading time.

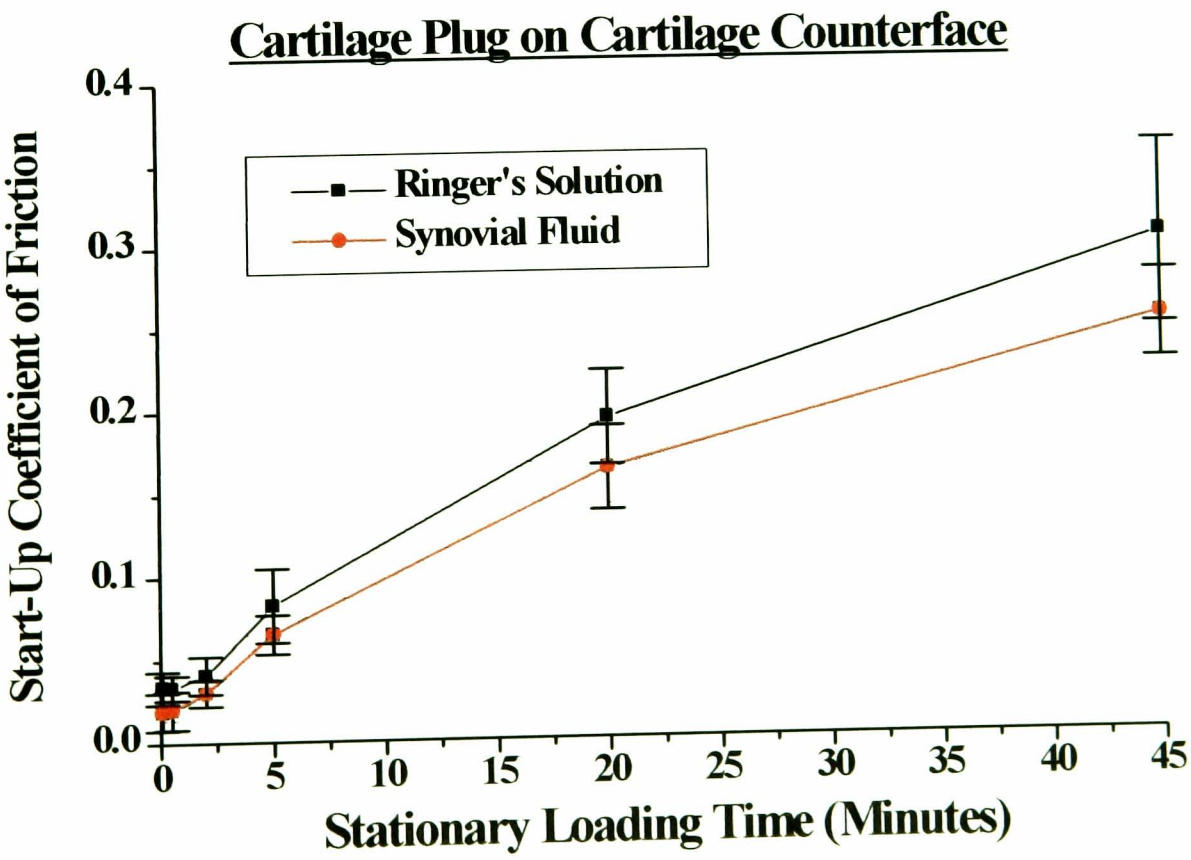


Figure 6-6 Line graph of the cartilage plug on cartilage counterface data, means and standard deviations are shown. Displaying the nature of the rise in friction coefficient with stationary loading time.

t-Tests							
SF vs. Ringer's	**5 s**	**30 s**	**2 min**	**5 min**	**20 min**	**45 min**	***5 s**
	0.012	**0.010**	**0.017**	**0.044**	**0.017**	**0.039**	**0.907**

Table 6-5 Student's t-Tests for synovial fluid (SF) versus Ringer's solution for all loading times.

6.3.3 Load Removal Start-Up Friction Tests

For a load removal period of only 1 minute for the cartilage counterface and 10 seconds for the cartilage plug, immediately after the 45 minute stationary loading tests, the subsequent drop in friction levels, as demonstrated by the further 5 second (*5 s) loading tests, was quite dramatic, Figure 6-7. These *5 second results were similar for both lubricants. From Table 6-4 it can be seen that for Ringer's solution and synovial fluid the particular mean value for the *5 second loading time was approximately double and triple respectively the mean value for the original 5 second loading time test.

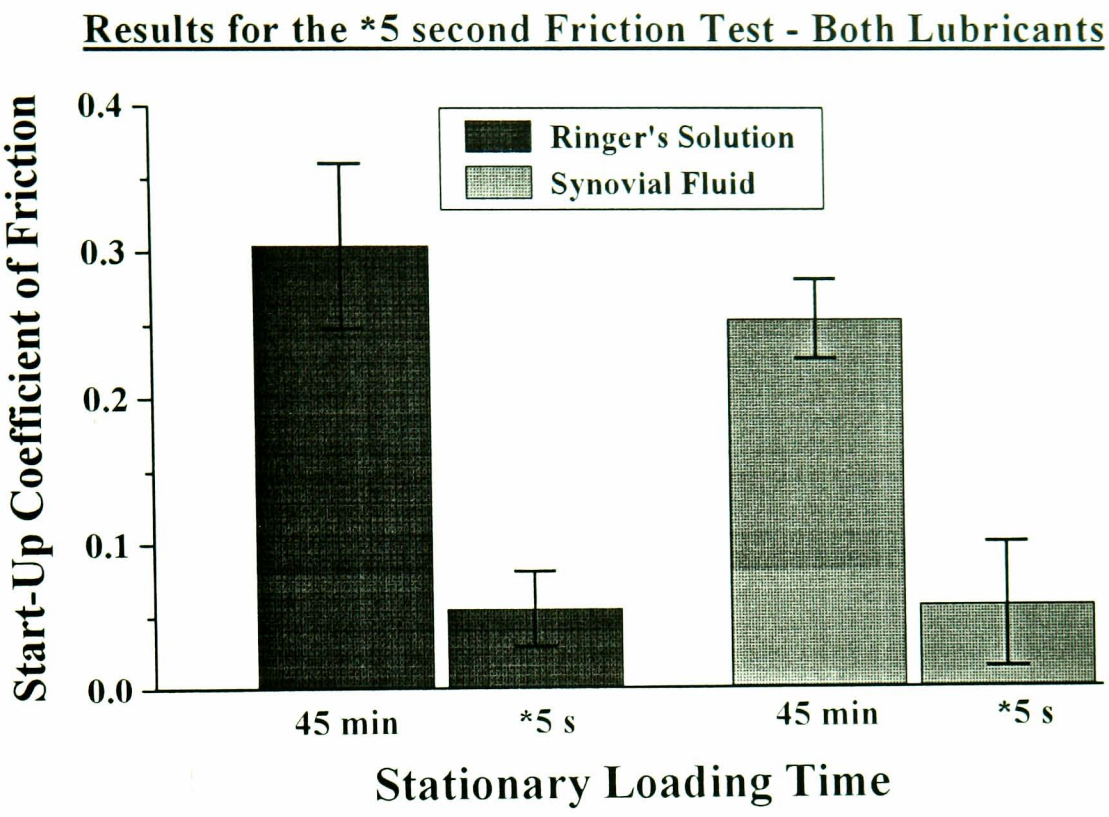


Figure 6-7 After the 45 minute loading test (45 min) the cartilage counterface was left unloaded for 1 minute, while the cartilage plug was unloaded for 10 seconds. Following this 1 minute load removal period a 5 second stationary loading time test was undertaken (*5 s). The mean values are shown along with their respective standard deviations.

6.3.4 Cartilage Plug on Hydrogel Counterface Test

The coefficient of friction results for the cartilage plug on hydrogel counterface have been tabulated in Table 6-6. The average value of the initial and repeat results listed in Table 6-6 was used to plot the cartilage plug on hydrogel counterface data shown in Figure 6-8.

		Stationary Loading Times						
		5 s	2 min	5 min	10 min	20 min	45 min	*5 s
Start-Up	Initial	0.017	0.036	0.041	0.08	0.116	0.173	0.049
	Repeat	0.011	0.016	0.051	0.060	0.098	0.153	0.048
	Average	**0.014**	**0.026**	**0.046**	**0.070**	**0.107**	**0.163**	**0.049**
Steady State	Initial	0.014	0.021	0.039	0.071	0.11	0.168	0.075
	Repeat	0.011	0.014	0.030	0.048	0.083	0.135	0.07
	Average	**0.013**	**0.018**	**0.035**	**0.060**	**0.097**	**0.152**	**0.073**

Table 6-6 Start-up and steady state coefficient of friction values for the cartilage plug on hydrogel counterface test.

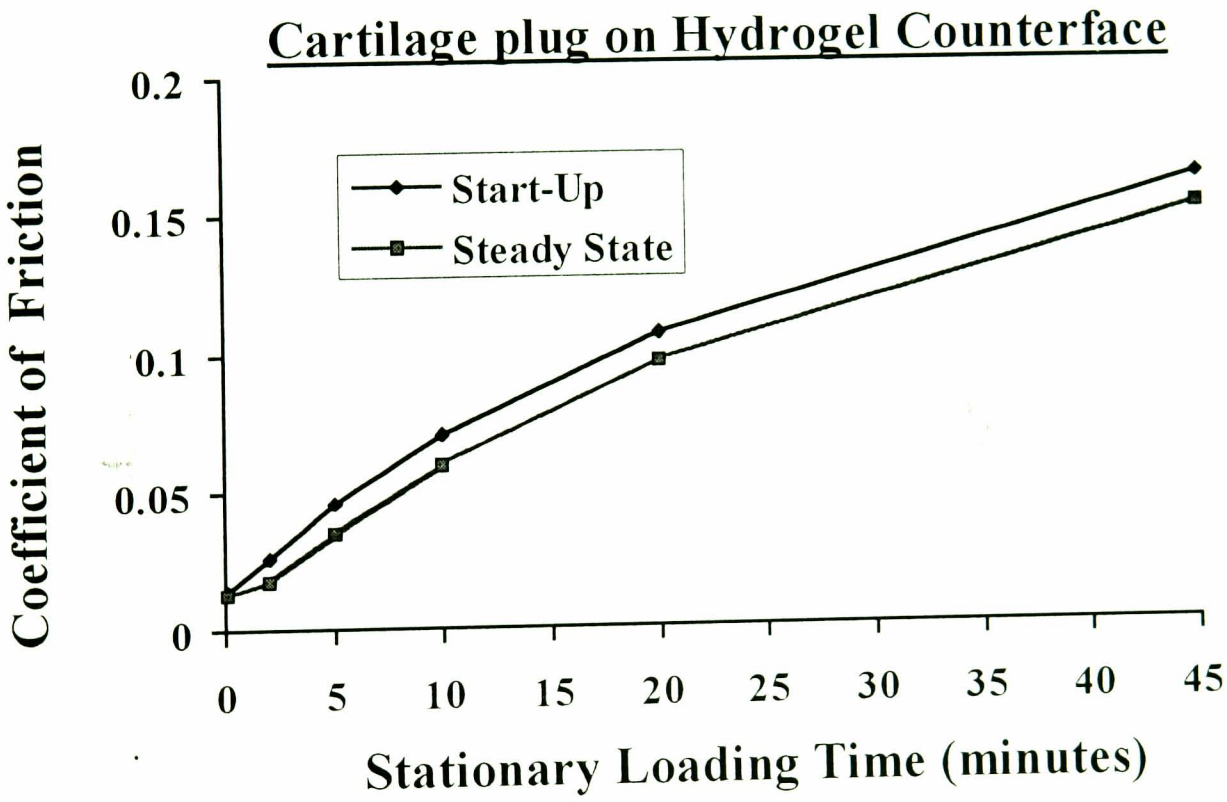


Figure 6-8 Start-up and steady state friction values for the cartilage plug on hydrogel counterface test.

The nature of the rise in friction levels for the cartilage on hydrogel configuration (Figure 6-8) with stationary loading time was similar to the four other contact configurations, except that the increase in friction was less marked. After 45 minutes of stationary loading both the start-up and steady state values were still below 0.18, compared to the more familiar value of around 0.3 at 45 minutes for the other contact configurations (ignoring the steady state friction levels for

the cartilage counterface studies which were independent of loading time, as discussed).

It should be emphasised that only a single cartilage plug was tested using the hydrogel counterface under stationary loading conditions, chiefly to attain reliable measurements for steady state friction. The intention was that the biphasic hydrogel layer would behave like a cartilage counterface in providing lubrication as the cartilage plug moved onto the unloaded region. This was not the case, as demonstrated by the close similarity of values for both start-up and steady state friction coefficients, Figure 6-3, and therefore this model was not pursued.

6.4 Discussion

6.4.1 Start-Up and Steady State Friction

As mentioned in the results section only start-up friction was analysed. One of the reasons was that the cartilage counterfaces were not sufficiently even (both in the direction of sliding and perpendicular to the direction of sliding) to record reliable steady state results during sliding. However there were a few exceptions whereby the cartilage counterfaces were even enough over a sufficient distance to analyse the steady state coefficient of friction also. Cartilage counterface specimen ii, used for tests 3-5 (Table 6-1), was one of these and the friction traces plotted for test 5, Figure 6-2, were similar to those for tests 3 and 4. At the end of the Chapter 5 discussion, based upon the findings of the metal on cartilage configuration, it was speculated that "Movement in a synovial joint will always lead to contact areas translating over previously unloaded cartilage surfaces possessing the normally high amounts of fluid content. Therefore after a period of stationary loading, although the initial start-up friction may be quite high, movement of the contact areas onto previously unloaded regions of the joint surface will produce substantial drops in the friction levels". Figure 6-2 has provided evidence that this hypothesis is correct. Furthermore for the cartilage on cartilage contacts it was shown that once the cartilage plug moved away from the loaded region of the cartilage counterface steady state coefficient of friction levels remained extremely low, between 0-0.01. This steady state coefficient of friction was independent of the duration of stationary loading that the cartilage plug had been subjected to. Therefore it has been established with reasonable confidence that for a healthy synovial joint as long as one of the opposing cartilage surfaces is in a condition of full hydration, with no subjection to recent loading, friction levels during sliding or rolling motion will remain very low, irrespective of the previous loading history and hydrational state of the other opposing cartilage surface.

6.4.2 Start-Up and Steady State Friction - Cartilage on Hydrogel

The friction traces plotted from the cartilage plug on hydrogel counterface test, Figure 6-3, were similar in appearance to friction traces from the two cartilage on

metal counterface configurations. The friction coefficients at start-up and during steady state sliding were similar and both increased with stationary loading time. Unlike the cartilage on cartilage configuration, the duration of stationary loading upon the cartilage plug, prior to sliding, was clearly having a very noticeable effect upon the steady state coefficient of friction. It would seem that the fluid phase of the hydrogel counterface was unable to instantaneously carry load and thereby reduce the coefficient of friction as the cartilage plug moved across. For optimal performance of articular cartilage biphasic lubrication, upon loading of a synovial joint, instantaneous load carriage by the fluid phase is essential. This has been shown to occur for the two contact configurations using a cartilage counterface by the rapid fall of start-up friction as soon as the cartilage or metal plug moved away from the stationary loaded region.

Under similar test conditions, although slightly faster sliding velocities of 8 mm/s, coefficients of friction between metal plugs and hydrogel counterfaces have been recorded (Caravia et al., 1994). In this study it was found that start-up friction was dependent on the duration of the loading period. However steady state friction was always substantially lower than the start-up friction and was independent of the duration of the loading time prior to sliding. This occurrence was attributed to possible elastohydrodynamic lubrication (EHL) between the hard slider and the compliant hydrogel counterface, as well as biphasic lubrication by the unloaded region of the hydrogel during sliding. The start-up coefficient of friction values were reaching their maximum after approximately 5 minutes, ~ 0.2-0.8 depending on surface roughness of the metal plug (Caravia, 1993). Also indentation tests attained equilibrium after approximately 5 minutes of loading. Thus, as evident from the friction and indentation tests reaching equilibrium conditions within 5 minutes, the loosely bound water in the hydrogel matrix has a smaller drag coefficient, when placed under load, than that of the fluid phase in articular cartilage. It is this lower drag coefficient that allows equilibrium to be established quicker for the hydrogel, upon the application of an external load. For the smoothest metal plugs, Ra=0.01 μm, the maximum start-up friction coefficient, reached within 5 minutes, was ~0.2. After 5 minutes loading for the

cartilage on hydrogel configuration the start-up friction coefficient was less than 0.05 and still well below 0.2 after 45 minutes of stationary loading, Figure 6-8. Cartilage, in comparison to the hydrogel material is of course the superior bearing material, as having a higher drag coefficient enables the pressure differentials in the fluid phase to be sustained for longer periods and therefore enables fluid flow and fluid phase load carriage to last longer, which in turn reduces friction levels. In Figure 6-8 friction levels were still increasing after 45 minutes of loading. This was attributed to the cartilage plugs not having yet reached equilibrium conditions. The stationary loaded region of the hydrogel counterface was predicted to have 'wrung-out' after approximately 5 minutes of loading.

It is not immediately clear why steady state friction was independent of loading time for the metal/hydrogel configuration (Caravia et al., 1994) but not for the cartilage/hydrogel counterface investigated here. One possibility was that during sliding for the metal/hydrogel contact EHL alone was in operation at the slightly faster sliding velocity of 8 mm/s, for this 'hard on soft' contact. However for the more compliant cartilage plug and at the lower 4 mm/s sliding velocity EHL could not have been predicted for the cartilage/hydrogel configuration. The duration of stationary loading upon the cartilage plug was clearly increasing the steady state friction but it could have been expected that the unloaded, fully hydrated hydrogel counterface would have reduced this to some degree. Figure 6-8 demonstrates just how close the start-up and steady state values actually were. The cartilage on metal configurations had similar close matching of start-up and steady state values. Another likely explanation is that the fully hydrated hydrogel counterface did not provide a biphasic lubrication mechanism as effective as the fully hydrated cartilage counterfaces. As mentioned above it was possible that the fluid phase of the hydrogel counterface was unable to instantaneously carry load and thereby reduce the coefficient of friction as the water depleted cartilage plug moved across. Despite the failure of the hydrogel counterface to substantially reduce steady state friction coefficients as compared to start-up, when comparing the other four contact configurations with the cartilage plug on hydrogel configuration, it was shown to have much lower start-up friction coefficient

values (Table 6-6). This could possibly be due to a very low intrinsic friction coefficient for the solid phases between the cartilage/hydrogel contact.

The *5 second stationary loading friction test (after 1 minute load removal for the hydrogel counterface and 10 seconds for the cartilage plug) produced similar large decreases in friction to those for the cartilage counterfaces (Table 6-6). This configuration also provided repeatable results as demonstrated by the close matching of the initial and repeat results. Both these factors indicate a recovery of low friction levels due to both cartilage and hydrogel rehydration.

6.4.3 Start-Up Friction Analysis of the Cartilage on Cartilage Contacts

6.4.3.1 Repeatability within each Test and Reproducibility between Tests

For all the cartilage on cartilage tests (1-16) the repeatability of the start-up friction readings was very good. This was proven by the respective close matching of the two readings taken at 5 seconds, 30 seconds, 2 minutes, 5 minutes and 20 minutes for each test. This has been graphically illustrated by plotting a typical set of results from one of the cartilage counterfaces tested, Figure 6-4. By producing data which was satisfactorily repeatable at each loading time, it was evident that no significant wear and/or cartilage degradation affecting the coefficient of friction values was occurring.

For the stationary loading friction measurements taken at 2 minutes and above the reproducibility of the results between tests 1-8, for synovial fluid, and between tests 9-16, for Ringer's solution, were also pleasing (Table 6-4), with the coefficient of variance at each loading time (standard deviation / mean) being ~ 0.1-0.3 for all these results. For the lower stationary loading times of 5 and 30 seconds, as well as the *5 second loading test, the coefficient of variance was higher, ranging from generally 0.2-0.6; (for the synovial fluid *5 second the coefficient of variance was especially high at 0.75). For these lower loading times the start-up friction could have been particularly sensitive to the geometry of the contact between the cartilage plug and the particular cartilage counterface being tested, which was naturally variable according to the overall topography of the

specific counterface and cartilage plug involved. For the *5 second tests the variability was further compounded by some difficulty in reapplying the cartilage plug to exactly the same position, after the 1 minute load removal, as for the prior 45 minute test. For the higher loading times the geometries of the contacts would have become more uniform as the two cartilage surfaces complied to one another.

6.4.3.2 Effect of Stationary Loading and subsequent Load Removal

Synovial fluid produced statistically significant reductions in the levels of start-up friction compared to Ringer's solution for all the 5 second to 45 minute loading times, see Figure 6-5, Figure 6-6 and Table 6-5 (pages 176-177). The reasons why synovial fluid produced systematic reductions in the start-up coefficient of friction for the cartilage on cartilage contacts and not for any of the previous stationary loaded contact configurations, mentioned in Chapters 4 and 5, will be referred to later in this discussion. For now it is worthwhile to remember that despite the statistically significant beneficial lubricating ability of synovial fluid the choice of lubricant was still not the governing factor influencing friction. This can be adequately demonstrated by Figure 6-6, which although it displays lower friction levels for synovial fluid as compared to the Ringer's solution, the overriding factor was clearly the duration of stationary loading. The mean start-up coefficient of friction values ranged from 0.019 at 5 seconds to 0.254 after 45 minutes of stationary loading for synovial fluid and from 0.033 at 5 seconds to 0.305 after 45 minutes of stationary loading for Ringer's solution.

Load duration, or more specifically the loading history of the tested cartilage specimens, was further emphasised as the primary factor governing friction by the load removal tests. The load removal tests, Figure 6-7, revealed large decreases in coefficient of friction levels from 0.254 and 0.305 for synovial fluid and Ringer's solution respectively at the 45 minutes loading period to approximately 0.06, for both lubricants, after only a 1 minute load removal period followed by a further 5 second (*5 second) loading test.

The non-linear rise of the friction levels with stationary loading time is illustrated by Figure 6-6. From the 5 second to 2 minute loading times there was little change in the friction levels for either of the lubricants, Table 6-4. From the 2 minute to 5 minute loading times the coefficient of friction began to rise markedly, with mean values rising from 0.030 to 0.065 and 0.041 to 0.082 for synovial fluid and Ringer's solution respectively. From the 5 minute to 45 minute loading times the increase in the coefficient of friction values became more gradual, with mean values at 45 minutes loading being 0.254 and 0.305 for synovial fluid and Ringer's solution respectively. This non-linear rise, like the drop in coefficient of friction levels for the *5 second tests, was in complete agreement with the behaviour of the coefficient of friction levels for the previous cartilage on metal and metal on cartilage contacts discussed in the two previous chapters. It was therefore again evident that the biphasic properties of the articular cartilage were governing the coefficient of friction results for this cartilage on cartilage configuration.

6.4.4 Cartilage on Cartilage Biphasic Lubrication Mechanism

The non-linear relationship of the friction coefficient to applied stationary load duration for the cartilage on cartilage contacts can be explained in respect to the biphasic characteristics of articular cartilage. The friction was initially low at the lower loading times due to a high proportion of the load carried by the pressure gradients within the fluid phase, as flow of the fluid phase occurs away from the loaded region (Ateshian et al., 1994). The fluid phase has a negligible associated coefficient of friction and therefore any load which it carries reduces the overall friction by reducing the amount of load carried by the solid phase. Indeed for the two lowest loading periods, being 5 and 30 seconds, there were no changes in the friction levels for either of the lubricants, Table 6-4 and Figure 6-5. This would indicate that between the 5 and 30 second loading times the amount of load carried by the fluid phase did not diminish substantially enough for the friction levels to alter for the two opposing cartilage layers in this contact configuration studied. After 2 minutes of stationary loading it can be seen that the friction levels did begin to rise, Table 6-4 and Figure 6-6, but it was not an exceptionally marked increase from the 5 and 30 second coefficient of friction values. Thus for

two opposing cartilage layers under stationary loaded conditions marked increases for the start-up coefficient of friction, upon the instigation of motion, may not occur for up to at least a couple of minutes of loading. It is probable that this observation upon the excellent properties of articular cartilage as a bearing material will also occur within the healthy synovial joint.

For the longer loading times, as more and more fluid is forced away, the pressure gradients become less, and so less load can be supported by the fluid phase. This consequently increases the amount of load carried by the solid phase which, of course, has a particular coefficient of friction attributable to itself. It is this increasing load carriage on the solid phase that therefore increases the overall friction force exerted when the cartilage surfaces slide across one another. Hence the overall measured coefficient of friction values increased with loading time. The non-linear rise in the coefficient of friction values was directly related to the non-linear decline of the pressure gradients and reduction in flow within the fluid phase while the cartilage specimens were subject to constant stationary loads.

Likewise for the *5 second loading tests, Figure 6-7, the dramatic fall of the start-up coefficient of friction values can be explained by the biphasic properties of articular cartilage. After the 45 minutes loading test the 1 minute load removal for the cartilage counterface allowed a large proportion of the previously exuded fluid back into the cartilage. Therefore upon the reapplication of load for the *5 second loading test the fluid phase would have been capable of carrying a substantially higher percentage of the load, than after the 45 minute loading period, creating a large drop in the friction levels.

6.4.5 Comparison with other Contact Configurations

In addition to the non-linear nature of the rise of coefficient of friction values plotted against stationary loading time being similar to the previous contact configurations, the respective coefficient of friction values themselves were quite comparable, c.f. Figure 5-5 and Figure 6-6. Thus overall the four stationary loaded contact configurations studied (Chapters 4-6) revealed similar findings

regarding the effects of lubricant, stationary loading times and load removal, and the influence of the biphasic properties of the cartilage. Figure 6-9, below, compares the metal on cartilage configuration start-up friction coefficients with those of the cartilage on cartilage configuration. It should be remembered that these two contact configurations were not experimentally designed to be directly comparable but some insight may be gained by contrasting the results. It can be seen that the addition of a second biphasic cartilage layer did not have any obvious overall effect upon the start-up coefficient of friction values, Figure 6-9, with values for the cartilage on cartilage contacts being only moderately lower.

In a synovial joint the benefits of having two opposing cartilage layers are maximised during active motion by having a continuous cycle of loaded and unloaded regions. The period of time that any one area of cartilage is loaded will generally be far less than the period of time that it is unloaded. It is speculated that this physiological cyclic loading allows the cartilage layers to maintain almost full hydration preserving very low friction levels due to biphasic lubrication.

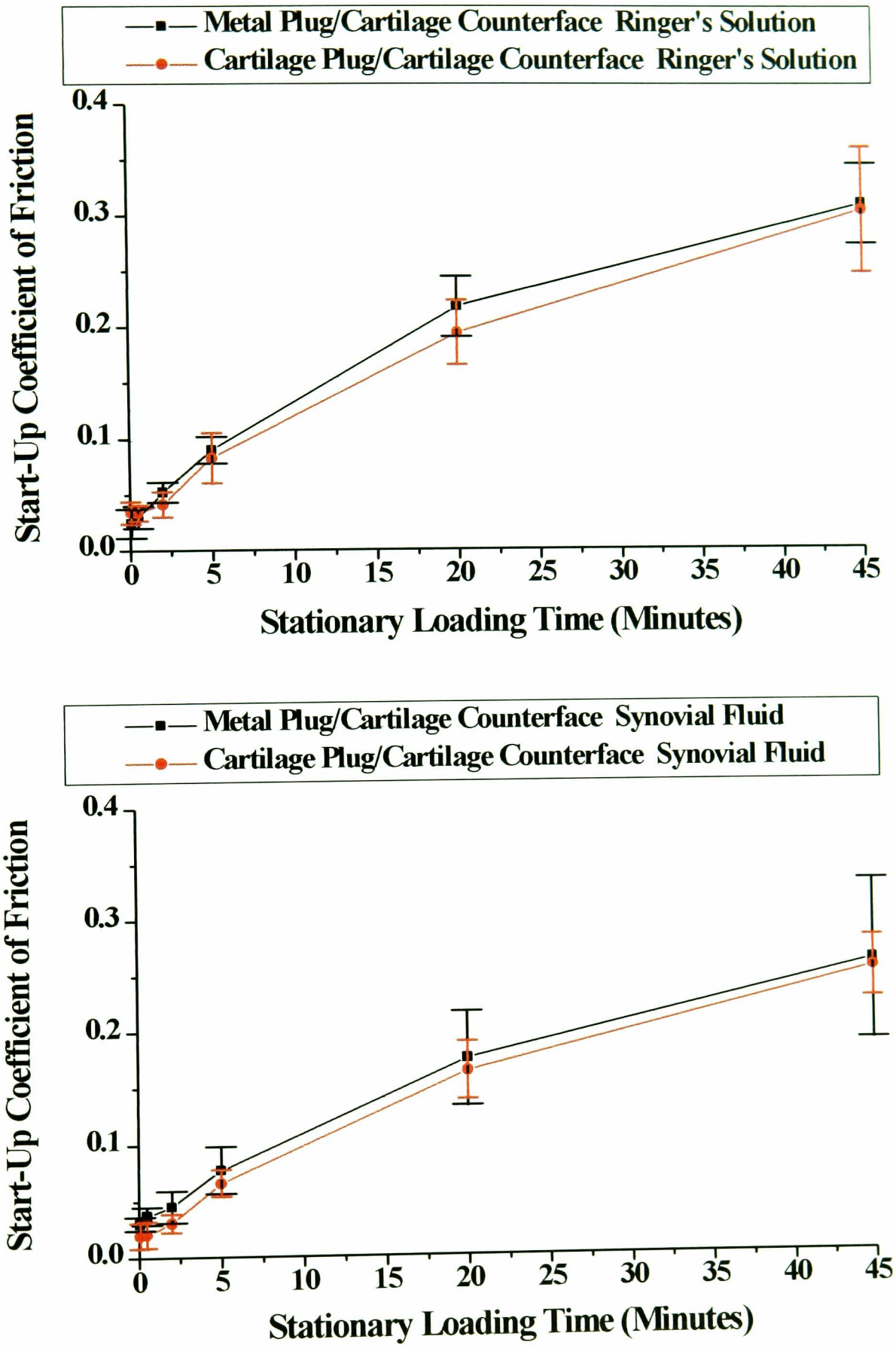


Figure 6-9 Comparison of the metal on cartilage configuration start-up friction coefficients with those of the cartilage on cartilage configuration; for both Ringer's solution (top) and synovial fluid (bottom).

6.4.5.1 Tribological Advantages of the Cartilage on Cartilage Contacts

There were however certain differences in the findings for the cartilage plug on cartilage counterface configurations. Unlike the other configurations studied the cartilage on cartilage contacts demonstrated scarcely any rises in friction coefficient up to the 2 minute loading period. In previous chapters it has been revealed that low values of friction coefficient, of less than or equal to 0.05, were maintained for up to 2 minutes after the application of stationary loading. Thus very low coefficients of friction can be maintained for cartilage / synthetic (metal) contacts for up to 2 minutes of stationary loading. This particular configuration moreover revealed that for cartilage / cartilage contacts there was no considerable increase in the friction coefficient for up to a stationary load duration of 2 minutes (Table 6-4). So there was an advantage of having two opposing cartilage surfaces by maintaining start-up coefficient of friction values at minimum levels for at least a couple of minutes after loading, for this study. In order to explain this occurrence it was postulated that sufficient drops in the fluid phase load carriage of the two biphasic layers, in order to increase friction, did not occur for up to two minutes after loading.

The cartilage on cartilage contacts also showed systematic reductions for friction levels when synovial fluid was the lubricant, which were statistically significant at all loading times. Therefore the beneficial lubricating ability of the synovial fluid was augmented by having two opposing cartilage surfaces, as used in this investigation. The previous contacts were either cartilage on metal or metal on cartilage configurations. These contacts generally pointed to a trend of synovial fluid providing lower coefficients of friction than the Ringer's solution but statistical significance was rare. The addition of a second cartilage surface could have enhanced the synovial fluid's capacity to provide boosted and/or boundary lubrication. If boundary lubrication of articular cartilage relies on more than just a tenacious macromolecular layer adhering to the surface then this might well be the case. Previous studies (Roberts, 1971; Stanescu and Leibovich, 1982) have suggested that the boundary lubricant works by repulsive hydration forces or electrostatic repulsion, with separation of the boundary surfaces having to be

below 3 nm. Therefore an opposing cartilage counterface, with synovial fluid to provide and replenish the boundary lubricant, would be needed for optimum performance of the boundary lubricant. The boosted lubrication mechanism is also very much dependent upon the presence of synovial fluid and two opposing cartilage layers. Under stationary loaded conditions, synovial fluid is first trapped between the two porous cartilage surface layers and then the solute is forced into the cartilage layers, leaving behind a highly viscous gel which is better able to sustain the higher loads (Walker et al., 1968).

6.4.6 Concluding Remarks

- Generally the results possessed good repeatability within each test and good reproducibility between tests.
- For a healthy synovial joint as long as one of the opposing cartilage surfaces is in a condition of full hydration, with no subjection to recent loading, friction levels during sliding or rolling motion will remain very low, irrespective of the previous loading history and hydrational state of the other opposing cartilage surface.
- The findings of non-linear rise of friction levels with loading time and large drops after relatively short durations of load removal were similar to the previous configurations tested. It was therefore again evident that the biphasic properties of the articular cartilage were governing the coefficient of friction results.
- For the cartilage on cartilage contacts there was very little increase in the coefficient of friction for up to 2 minutes after loading. It was postulated that sufficient drops in the fluid phase load carriage of the two biphasic layers, in order to increase friction, did not occur for up to 2 minutes after loading.
- The cartilage on cartilage contacts showed systematic reductions for friction levels when synovial fluid was the lubricant, which were statistically significant at all loading times. Therefore the beneficial lubricating ability of the synovial fluid was augmented by having two opposing cartilage surfaces. The addition of a second cartilage surface could have enhanced the synovial fluid's capacity to provide boosted and/or boundary lubrication.

- The hydrogel counterface was not effective in reducing steady state coefficient of friction levels relative to the start-up values, unlike the cartilage counterfaces. Under the contact conditions tested the hydrogel counterface was not therefore shown to possess a biphasic lubricating ability. The start-up coefficient of friction values were however much lower than for the other contact configurations investigated. This could possibly be due to a very low intrinsic friction coefficient for the solid phases between the cartilage/hydrogel contact.

7. Reciprocating Motion Friction Tests - Cartilage on Metal Contacts

7.1 Introduction

The stationary, constant load friction tests discussed so far in chapters 4-6 have revealed the importance of loading time upon the friction of articular cartilage contacts. The main reason for carrying out reciprocating motion tests was to ascertain the contribution of the additional effect of continuous sliding between cartilage/metal contacts while again subject to a constant load. Under these conditions cartilage wear was anticipated (Lipshitz and Glimcher, 1979). A test protocol was designed in order to identify this potential occurrence and investigate both the effect of loading time and possible cartilage surface wear on friction. This was achieved by an *initial* 120 minutes sliding friction test, followed by a minimum 120 minute load removal period while immersed in lubricant, and then a *repeat* set of identical friction measurements were taken on the same cartilage specimen. Any alteration between the first and second set of friction readings would indicate the effect of wear or surface damage upon the friction coefficient.

For these tests the continual measurement of friction under constant load and reciprocating motion more closely resembled previous studies conducted by other workers (Walker et al., 1969; McCutchen, 1962); as compared to the previous tests whereby sliding was only instigated and friction recorded after a given period of stationary loading.

During the course of one of the reciprocating motion friction tests the cartilage plug was lifted away from the metal counterface for a period of one minute, while still immersed in lubricant, in order to assess the effect of load removal upon friction levels. Non-contacting laser profilometry was performed upon the cartilage plug both before and after this particular friction test to analyse possible signs of wear.

A reciprocating motion cartilage on hydrogel counterface test was also conducted. The hydrogel material was the same used for the previous stationary loaded cartilage on hydrogel test. The benefits of having a hydrogel, as opposed to a metal, counterface were investigated.

7.2 Experimental Procedure

7.2.1 Stationary Loading Friction Test Checks

For each 9 mm cartilage plug to be tested under reciprocating motion conditions, stationary loading friction tests were first conducted upon the metal counterface. Two readings were recorded after 5 seconds of stationary loading and one reading was recorded after 2 minutes of stationary loading. These stationary loading friction tests were used in order to quickly assess the condition of the cartilage specimens prior to the start of the reciprocating motion test; lasting 6 hours in total. A particular 9 mm cartilage specimen was considered viable for testing, being healthy and unmarked, if the friction coefficients from the stationary loading checks were approximately within the standard deviations about the mean values of the previously obtained 9 mm cartilage on metal stationary loaded contact configuration results. There were occasions when potential specimens were not used for a reciprocating motion test due to the recording of comparatively high friction levels during the stationary loading friction test checks.

7.2.2 Friction Measurement for the Reciprocating Motion Nine Millimetre Cartilage Plug on Metal Counterface Contact Configuration

Friction of the cartilage/metal contacts was recorded within a mixed lubrication regime, using synovial fluid and Ringer's solution. The 9 mm cartilage plugs were gently lowered onto the metal counterface which was continually reciprocating. The first friction measurements were taken 5 seconds after the placement of the cartilage plugs onto the moving metal counterface. Further friction measurements were then recorded from 2 minutes up to 120 minutes following the placement of the cartilage plugs onto the moving metal counterface, see Table 7-1. Nine tests were conducted using Ringer's solution as the lubricant and nine tests using synovial fluid as the lubricant, see Table 7-2. For each test a different 9 mm cartilage plug was used.

***Initial* Steady State Friction Readings taken at Set Intervals after the Start of Reciprocating Motion under a 30 N Load:**	***Repeat* Steady State Friction Readings taken at Set Intervals after the Start of Reciprocating Motion under a 30 N Load:**
5 seconds	5 seconds
2 minutes	2 minutes
5 minutes	5 minutes
10 minutes	10 minutes
15 minutes	15 minutes
20 minutes	20 minutes
25 minutes	25 minutes
30 minutes	30 minutes
35 minutes	35 minutes
40 minutes	40 minutes
45 minutes	45 minutes
50 minutes	50 minutes
55 minutes	55 minutes
60 minutes	60 minutes
70 minutes	70 minutes
80 minutes	80 minutes
90 minutes	90 minutes
100 minutes	100 minutes
110 minutes	110 minutes
120 minutes	120 minutes

Table 7-1 Test Protocol adopted for the reciprocating motion cartilage plug on metal counterface configuration. The cartilage plug was unloaded while immersed in lubricant fluid for a minimum period of 120 minutes between the *initial* and *repeat* measurements, for every test.

For each test an *initial* set of friction measurements taken at regular intervals from 5 seconds up to 120 minutes was followed by a minimum 120 minute load removal period while immersed in lubricant. An identical set of *repeat* friction measurements was then recorded, Table 7-1. For this reciprocating motion cartilage on metal contact configuration the recording of an *initial* and *repeat* set of results was considered necessary to assess whether or not wear and/or cartilage degradation, affecting the coefficient of friction values, was occurring due to sliding. This was especially warranted for this configuration as the tests were conducted over a minimum period of 6 hours, 4 hours of which were spent under continual sliding, between the cartilage plugs and metal counterface.

Test	Cartilage Plug	Lubricant
1	1	Ringer's Solution
2	2	Ringer's Solution
3	3	Ringer's Solution
4	4	Ringer's Solution
5	5	Ringer's Solution
6	6	Ringer's Solution
7	7	Ringer's Solution
8	8	Ringer's Solution
9	9	Ringer's Solution
10	10	Synovial Fluid
11	11	Synovial Fluid
12	12	Synovial Fluid
13	13	Synovial Fluid
14	14	Synovial Fluid
15	15	Synovial Fluid
16	16	Synovial Fluid
17	17	Synovial Fluid
18	18	Synovial Fluid

Table 7-2 Test conditions for the reciprocating motion cartilage plug on metal counterface configuration, detailing tests 1-18. A metal counterface was used throughout.

For this study only steady state friction was recorded during sliding. There was some evidence of 'start-up' peaks as the counterface changed directions during reciprocation, Figure 7-1 and Figure 7-2. However, these were not strictly start-up friction peaks as referred to in previous stationary loaded contact configurations, whereby start-up peaks were those produced immediately after a given period of stationary loading.

7.2.3 Reciprocating Motion Load Removal Test

A single reciprocating motion 9 mm diameter cartilage plug on metal counterface test was performed with both initial and repeat friction measurements monitored over 45 minute loading times. Ringer's solution was the lubricant. After the initial 45 minutes of loading, a load removal period of 1 minute was allowed while immersed in lubricant, prior to the start of the repeat 45 minutes of loading, Table

7-3. The surface analysis of the 9 mm cartilage plug was undertaken by non-contacting laser profilometry before and after the reciprocating motion test in order to assess possible signs of wear.

Initial Steady State Friction Readings taken at Set Intervals after the Start of Reciprocating Motion under a 30 N Load:	**Repeat Steady State Friction Readings taken at Set Intervals after the Start of Reciprocating Motion under a 30 N Load:**
5 seconds	5 seconds
2 minutes	2 minutes
5 minutes	5 minutes
10 minutes	10 minutes
15 minutes	15 minutes
20 minutes	20 minutes
25 minutes	25 minutes
30 minutes	30 minutes
35 minutes	35 minutes
40 minutes	40 minutes
45 minutes	45 minutes

Table 7-3 Test Protocol adopted for the reciprocating motion load removal test. The cartilage plug was unloaded while immersed in lubricant fluid for 1 minute between the initial and repeat measurements.

7.2.4 Friction Measurement for the Reciprocating Motion Nine Millimetre Cartilage Plug on Hydrogel Contact Configuration

A 9 millimetre diameter cartilage plug was also tested against a hydrogel[1] counterface under reciprocating motion conditions, with Ringer's solution as the lubricant. Friction readings were recorded in exactly the same manner as for the 9 mm cartilage on metal reciprocating motion tests, Table 7-1.

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[1] The hydrogel material was *N*-vinyl pyrrolidone methyl methacrylate (PC110) as described in Chapter 2.

7.3 Results

7.3.1 Stationary Loading Friction Test Checks

The results of the stationary loading friction test checks are provided in Table 7-4. These stationary loading checks were conducted under identical conditions to those of the 9 mm cartilage plug on metal counterface stationary loaded contact configuration. The relevant results from this latter configuration have also been included in Table 7-4 for comparison. In Table 7-4 it can be seen that the stationary loading friction test checks closely matched those of the 9 mm cartilage plug on metal counterface stationary loaded contact configuration. This was especially true for the steady state results as opposed to the more erratic nature of the start-up friction.

Stationary Loading Time		**5 seconds**		**2 minutes**	
Checks prior to Reciprocating		**Start-Up**	**Steady State**	**Start-Up**	**Steady State**
Ringer's Solution	**Mean**	0.013	0.008	0.045	0.029
	StDev	0.005	0.005	0.016	0.011
Synovial Fluid	**Mean**	0.010	0.005	0.023	0.013
	StDev	0.003	0.004	0.009	0.008

Stationary Loading Time		**5 seconds**		**2 minutes**	
Stationary 9 mm Cartilage on Metal Configuration		**Start-Up**	**Steady State**	**Start-Up**	**Steady State**
Ringer's Solution	**Mean**	0.017	0.012	0.046	0.034
	StDev	0.006	0.006	0.015	0.008
Synovial Fluid	**Mean**	0.016	0.004	0.036	0.019
	StDev	0.006	0.002	0.013	0.007

Table 7-4 The top table displays the means and standard deviations of the stationary loading friction tests conducted prior to starting the *initial* 120 reciprocating motion friction tests. For a direct comparison the bottom table contains the relevant results from the stationary loaded 9 mm cartilage plug on metal counterface contact configuration.

7.3.2 *Effect of Loading Time*

7.3.2.1 Reciprocating Motion Friction Traces

Typical friction traces for this 9 mm cartilage on metal reciprocating motion contact configuration are displayed in Figure 7-1 and Figure 7-2. The results for both of these figures were supplied from a test using Ringer's solution (Test 8). To provide a full and clear account of how friction was again increasing with loading time, in this instance under reciprocating motion, two plots were required. Figure 7-1 details the friction traces from between 5 seconds to 15 minutes after the start of reciprocating motion friction test while Figure 7-2, overlapping to some extent, details selected friction traces sampled from between 5 minutes to 120 minutes. In Figure 7-2, for the sake of clarity, it is worthwhile to note that many of the friction traces recorded between the 5 minute and 120 minute times have been omitted, see Table 7-1.

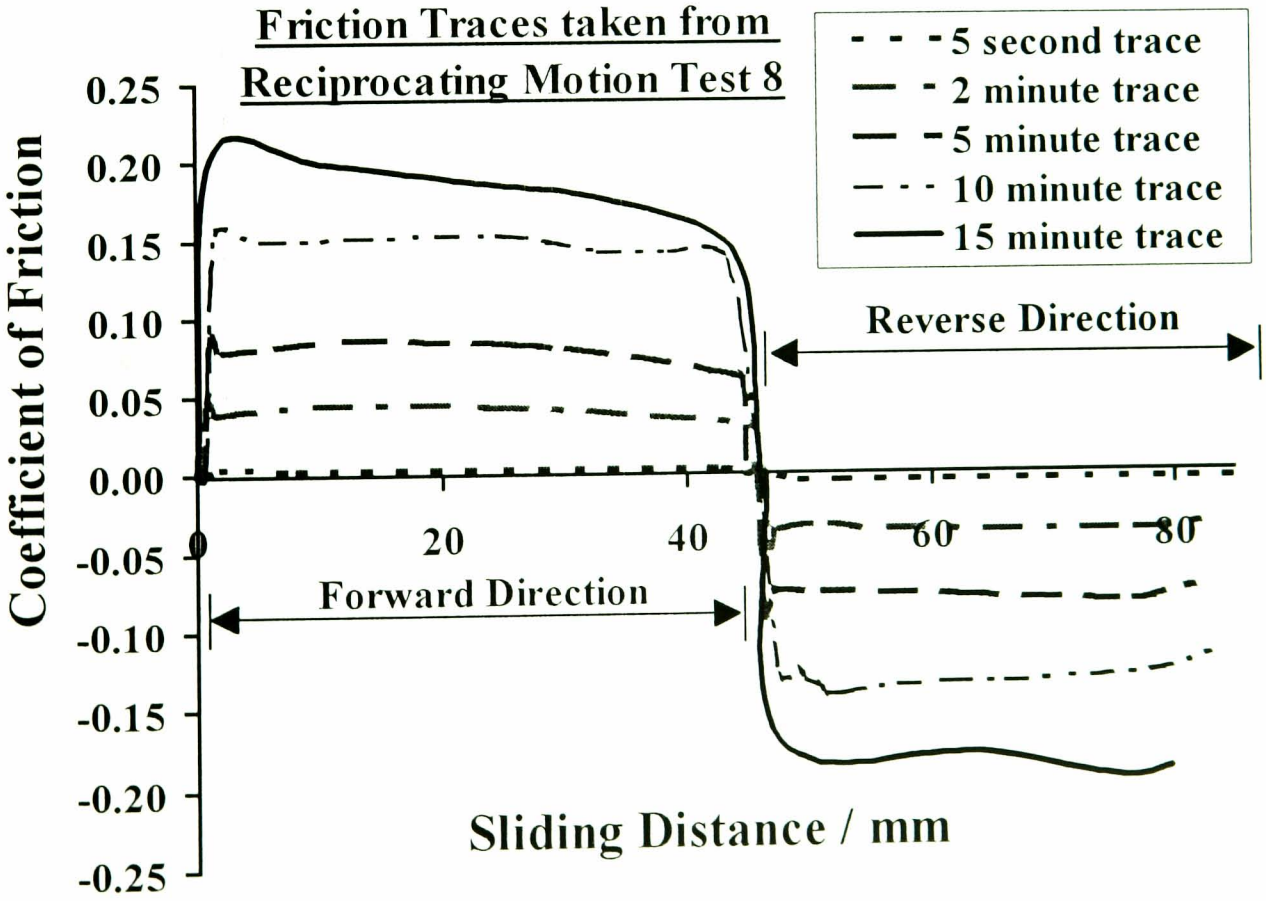


Figure 7-1 Plots of coefficient of friction versus sliding distance taken from the 9 mm cartilage plug on metal counterface reciprocating motion contact configuration; Test 8 - 5 second to 15 minute traces.

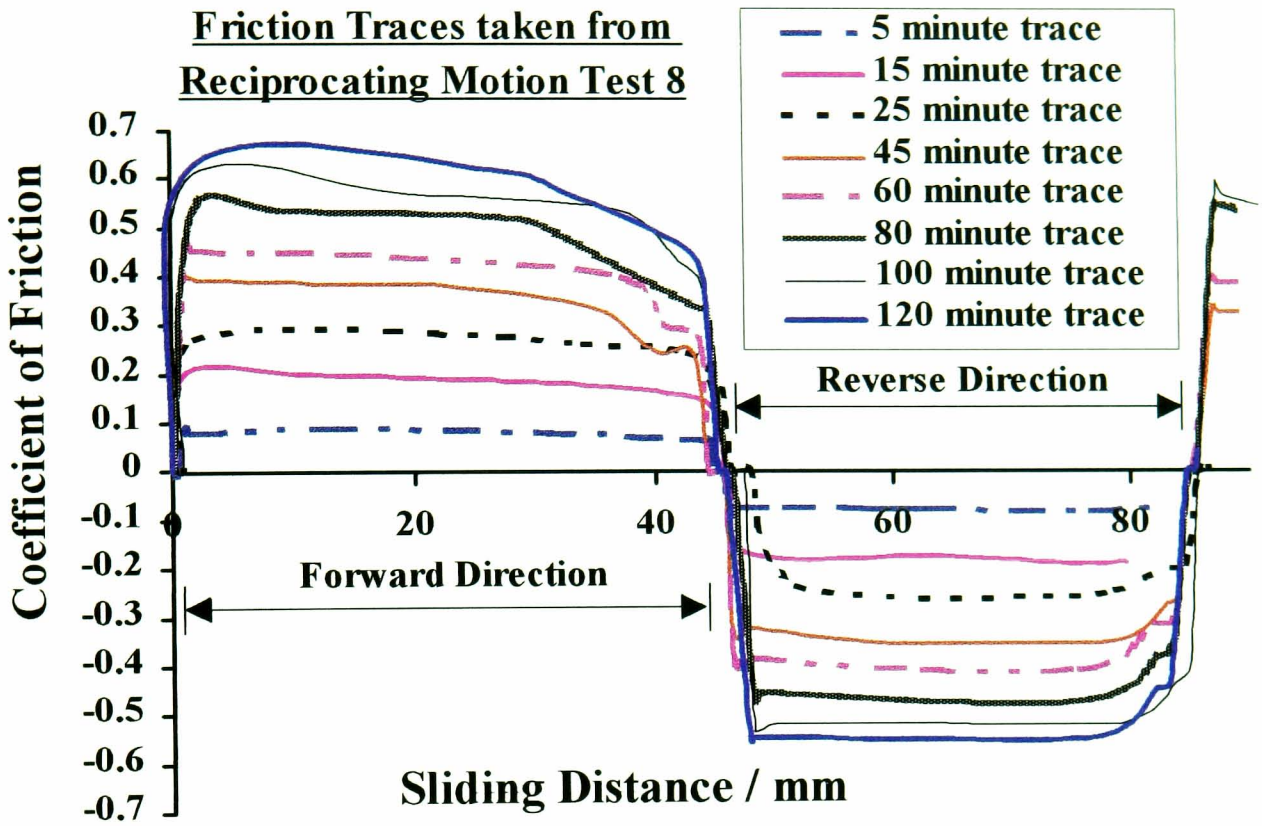


Figure 7-2 Plots of coefficient of friction versus sliding distance taken from the 9 mm cartilage plug on metal counterface reciprocating motion contact configuration; Test 8 - selected traces from 5 minutes to 120 minutes.

Steady state friction was monitored during both forward and reverse direction sliding (Figure 7-1 and Figure 7-2). For consistency however and because the friction rig was only calibrated for positive (forward direction) friction values, only steady state friction values sampled during sliding in the forward direction were used for analysis of the data. While undergoing reciprocation the change of direction was clearly evident from the friction trace, changing from a positive (forward direction) to a negative (reverse direction) value and visa versa.

7.3.2.2 Reciprocating Motion Test Plots - Coefficient of Friction vs Loading Time for Individual Tests and Pooled Test Data

The steady state friction values were recorded at set intervals after the start of reciprocating motion under a load of 30 N as previously described, Table 7-1.

tests could be constructed. Such plots from two tests of the reciprocating motion cartilage plug on metal counterface configuration are depicted in Figure 7-3; using Ringer's solution and synovial fluid respectively as the lubricant. From these individual tests the close similarity of the form of both the initial and repeat plots was noted, with the repeat value being generally consistently higher than the initial.

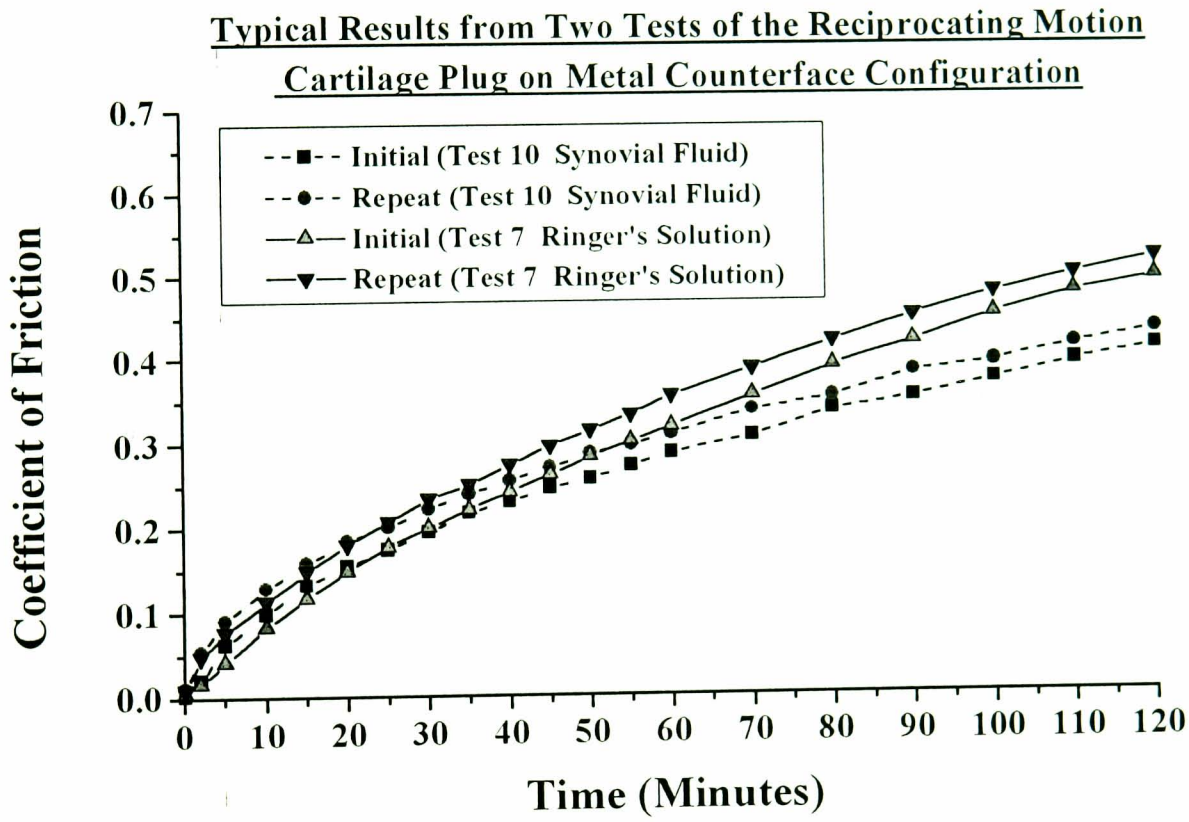


Figure 7-3 Steady state coefficient of friction values from Tests 7 & 10, using Ringer's solution and synovial fluid respectively.

The coefficient of friction values for the 9 tests conducted using Ringer's solution and the 9 tests using synovial fluid, respectively, as the lubricant were then pooled together in order to obtain mean values and standard deviations at each time interval. These means and standard deviation are provided in Table 7-5.

	Ringer's Solution				Synovial Fluid			
Time	***Initial***		***Repeat***		***Initial***		***Repeat***	
(Minutes)	**Mean**	**StDev**	**Mean**	**StDev**	**Mean**	**StDev**	**Mean**	**StDev**
0.08	0.007	0.005	0.033	0.021	0.006	0.005	0.028	0.026
2	0.035	0.012	0.076	0.022	0.019	0.010	0.053	0.032
5	0.069	0.019	0.113	0.030	0.045	0.020	0.078	0.033
10	0.121	0.027	0.158	0.035	0.080	0.031	0.113	0.036
15	0.166	0.031	0.199	0.039	0.113	0.034	0.144	0.038
20	0.204	0.031	0.234	0.042	0.147	0.034	0.177	0.041
25	0.235	0.033	0.262	0.046	0.177	0.036	0.203	0.042
30	0.267	0.035	0.293	0.047	0.206	0.036	0.231	0.051
35	0.294	0.039	0.322	0.053	0.235	0.042	0.256	0.057
40	0.321	0.042	0.348	0.057	0.261	0.052	0.282	0.064
45	0.344	0.045	0.373	0.058	0.286	0.059	0.304	0.071
50	0.367	0.047	0.398	0.060	0.307	0.067	0.325	0.078
55	0.387	0.049	0.420	0.062	0.328	0.073	0.345	0.082
60	0.407	0.053	0.442	0.063	0.346	0.074	0.364	0.086
70	0.447	0.055	0.474	0.065	0.382	0.081	0.401	0.089
80	0.479	0.056	0.504	0.068	0.415	0.087	0.427	0.092
90	0.506	0.059	0.529	0.068	0.437	0.089	0.452	0.093
100	0.528	0.055	0.551	0.064	0.458	0.090	0.467	0.093
110	0.550	0.061	0.570	0.061	0.479	0.091	0.481	0.094
120	0.567	0.061	0.585	0.059	0.498	0.088	0.497	0.092

Table 7-5 Steady state coefficient of friction values for the reciprocating motion cartilage plug on metal counterface tests. The means and standard deviations, for both lubricants, are supplied for all the loading time intervals.

From the data in Table 7-5 graphs of coefficient of friction against loading time were constructed for both lubricants, Figure 7-4 and Figure 7-5.

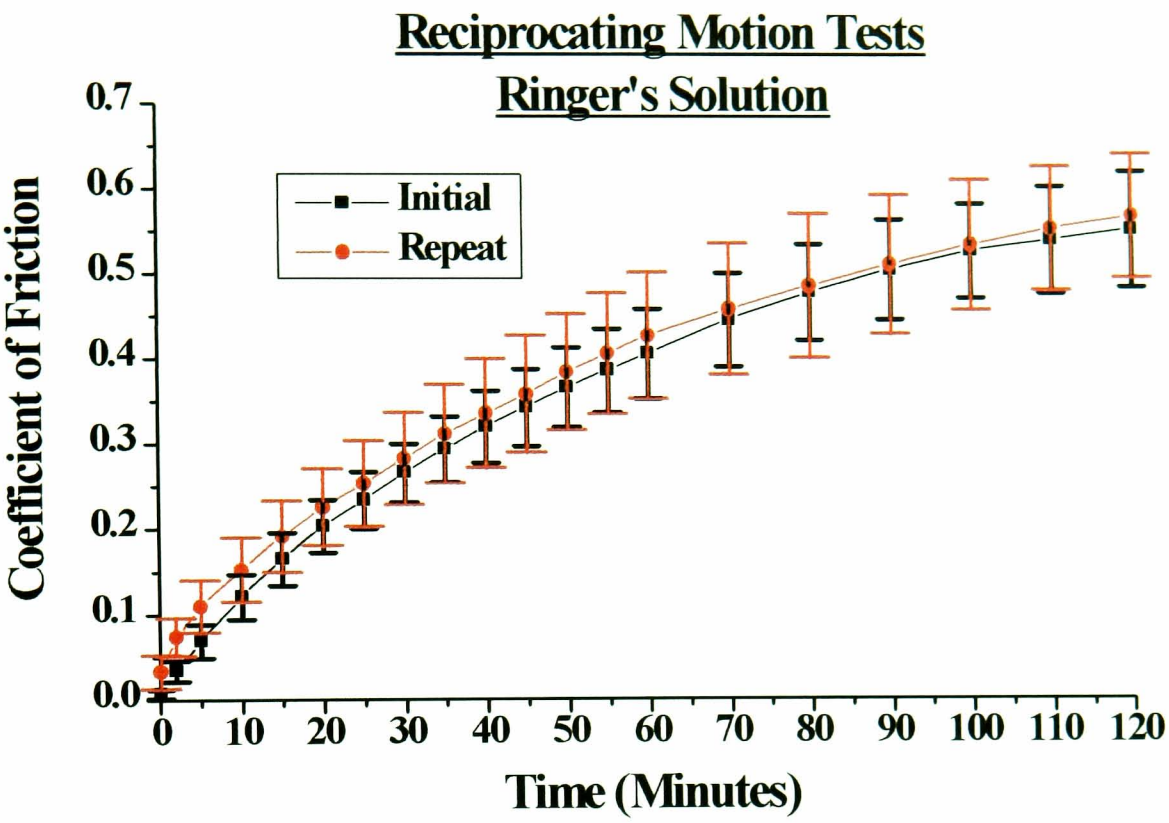


Figure 7-4 Means and standard deviations of the steady state coefficient of friction values for Ringer's solution. Both the *initial* and *repeat* set of results are shown.

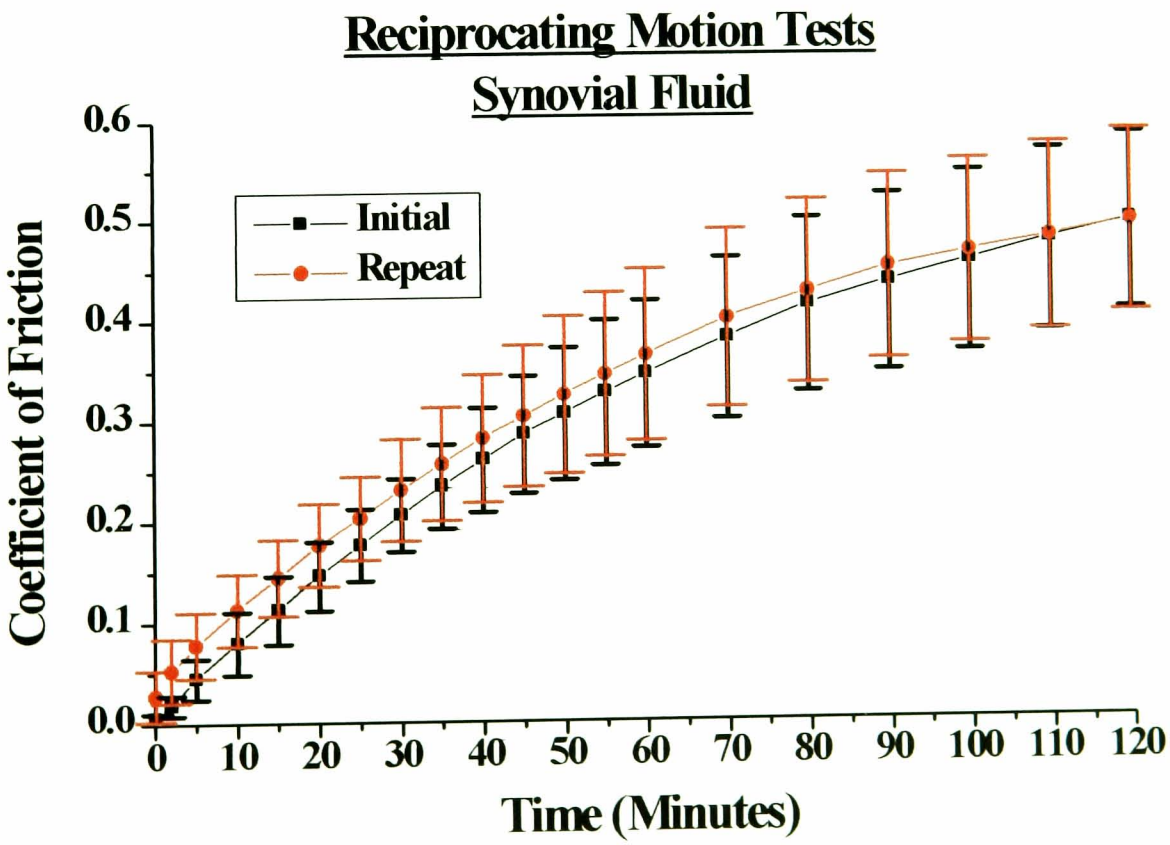


Figure 7-5 Means and standard deviations of the steady state coefficient of friction values for synovial fluid. Both the *initial* and *repeat* set of results are shown.

From Figure 7-4 and Figure 7-5 the major influence of loading time upon the friction of cartilage was again evident. The coefficient of friction values were found to increase all the way up to the 120 minute loading time; at which point very high friction coefficients of up to 0.6 were recorded. Following an equal 120 minute period of load removal while immersed in lubricant, the *repeat* friction coefficients were on the whole only slightly higher than the *initial* values. For the lower loading times however, 15 minutes and below, the *repeat* values were appreciably higher than those of the *initial* values for both lubricants, Table 7-5.

7.3.3 Effect of Lubricant

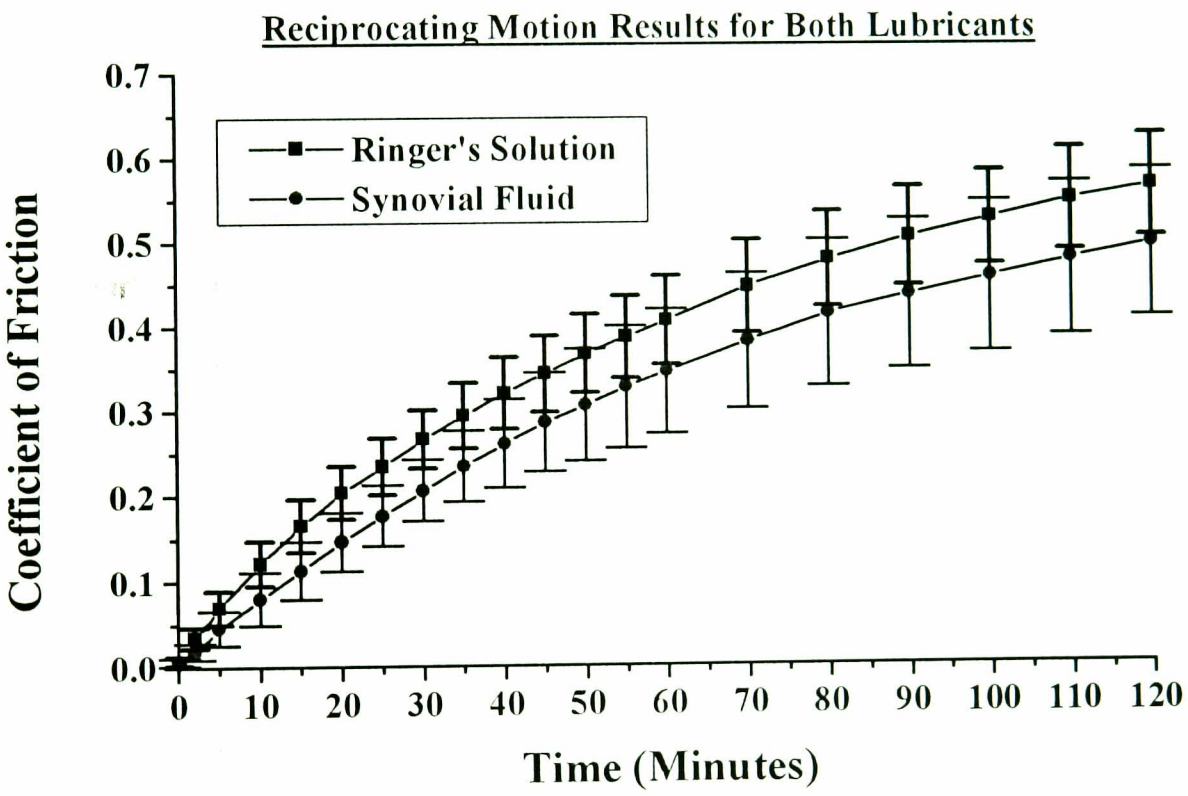


Figure 7-6 Comparison of the Ringer's solution and synovial fluid results for the reciprocating motion cartilage plug on metal counterface tests. Only the *initial* results for each lubricant is displayed.

The initial results for both lubricants have also been plotted together in Figure 7-6. From this graph the beneficial lubricating ability of synovial fluid was clearly illustrated, for this reciprocating motion configuration. The mean values of the friction coefficients obtained using synovial fluid as the lubricant were found to be statistically significantly lower than those of the Ringer's solution in many

instances, using Student t-Tests ($p<0.05$), Table 7-6. For the *initial* results significance was established for all loading times between 2 to 50 minutes. For the *repeat* results significance was established for all loading times between 5 to 40 minutes and also at the 110 and 120 minute sampling times.

Student t-Tests Ringer's Solution vs. Synovial Fluid		
Time (Minutes)	***Initial***	***Repeat***
0.08	0.657	0.624
2	0.009	0.105
5	0.023	0.041
10	0.010	0.023
15	0.004	0.014
20	0.002	0.016
25	0.003	0.017
30	0.003	0.024
35	0.008	0.032
40	0.019	0.045
45	0.036	0.051
50	0.044	0.056
55	0.061	0.056
60	0.066	0.056
70	0.069	0.081
80	0.087	0.077
90	0.078	0.079
100	0.069	0.053
110	0.088	0.042
120	0.090	0.039

Table 7-6 Results of t-Tests conducted between Ringer's solution and synovial fluid *initial* and *repeat* data sets - as displayed in Table 7-5.

7.3.4 Comparison with the Stationary Loaded Nine Millimetre Cartilage on Metal Contact Configuration

As stated in the introduction of this chapter a main objective for carrying out these reciprocating motion tests was to ascertain the significance of the additional effect of continuous sliding between the cartilage/metal contacts while under a

constant load. Figure 7-7 and Figure 7-8 below compare both the stationary and reciprocating motion (up to 60 minutes only) 9 mm cartilage on metal friction tests for the respective lubricants. This was for a general comparison only as the two configurations were not experimentally designed for direct comparison; but more rather designed to compare the influence of the two lubricants with loading time having undergone a given period of stationary loading or reciprocating motion under load within the particular configuration.

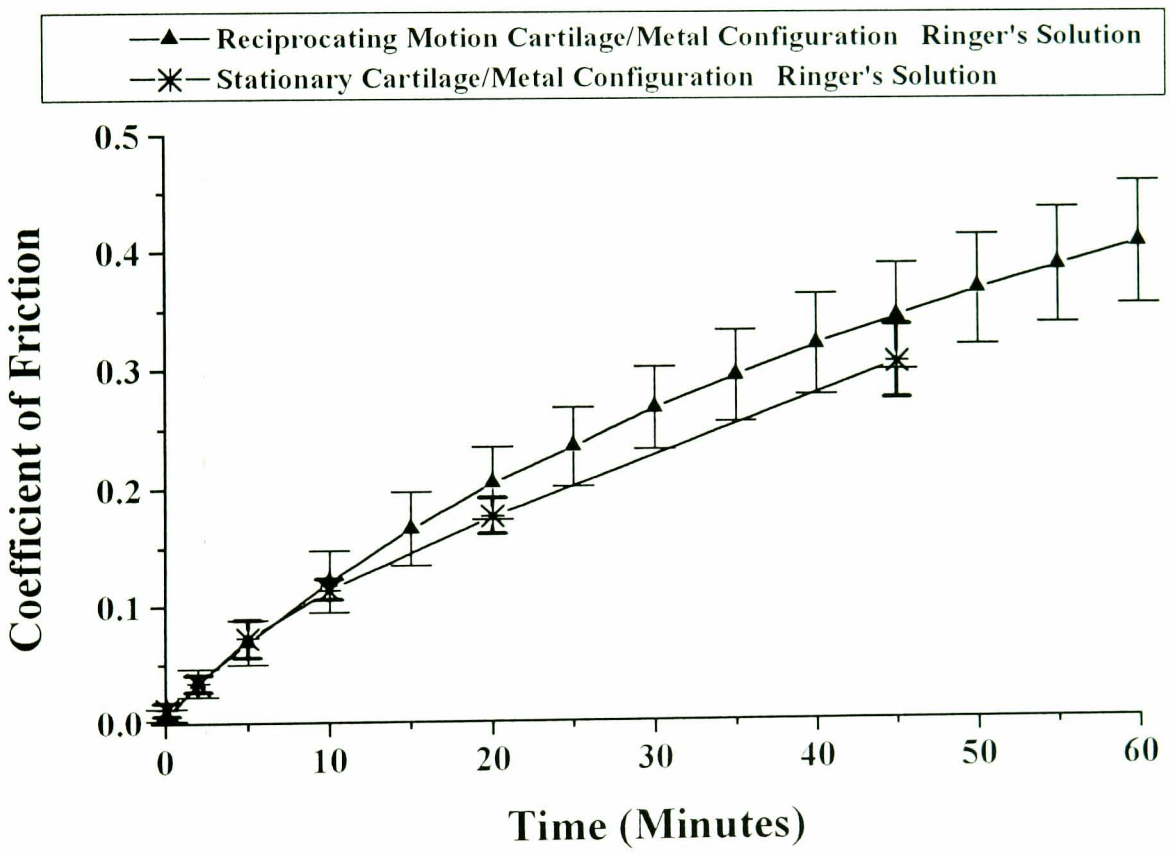


Figure 7-7 Steady state *initial* values plotted for reciprocating motion cartilage plug on metal counterface and stationary cartilage plug on metal counterface configurations. Ringer's solution was the lubricant.

In Figure 7-7, comparing the two configurations when using Ringer's solution as the lubricant, it can be seen that there was little difference between the two configurations up to 5 minutes of loading. After 10 minutes of loading and beyond, the reciprocating motion configuration began to exhibit slightly higher values up to the maximum 45 minute loading period tested for the stationary configuration. This trend, in comparing the two configurations, was different

when synovial fluid was the lubricant, Figure 7-8. Again, there was little difference between the two configuration for up to 5 minutes of loading but after 10 minutes of loading and beyond, the reciprocating motion configuration began to exhibit slightly lower values.

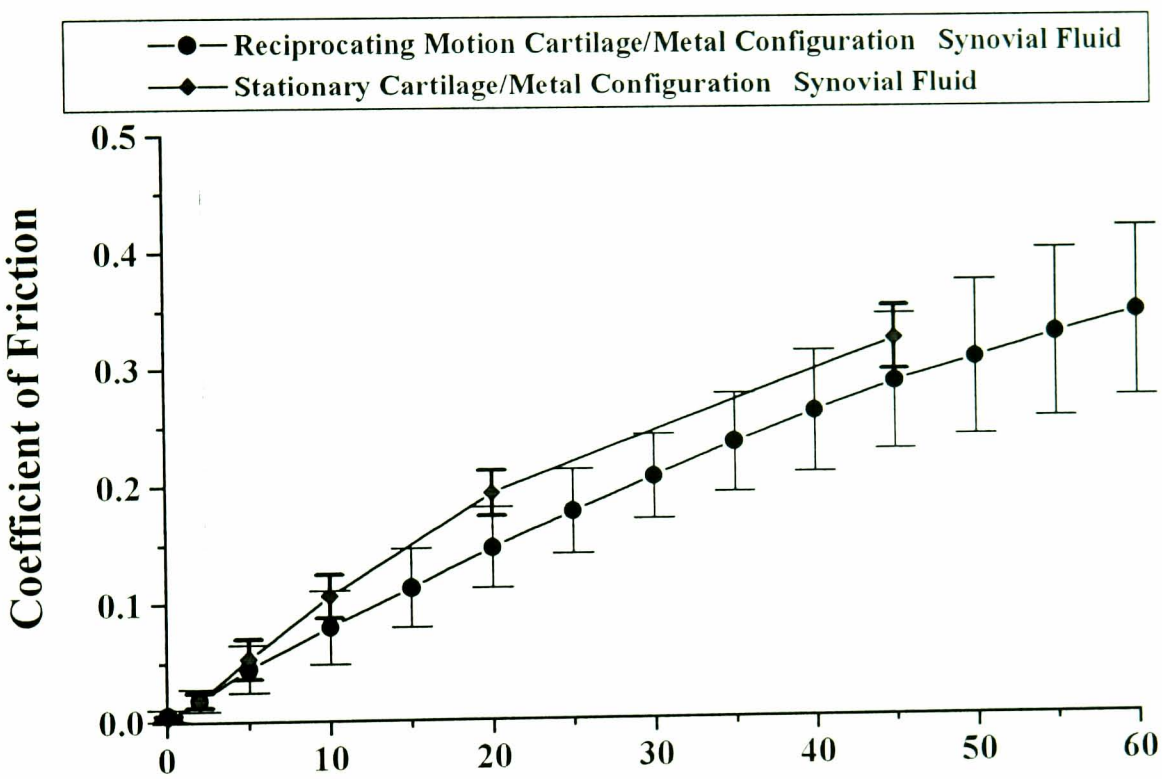


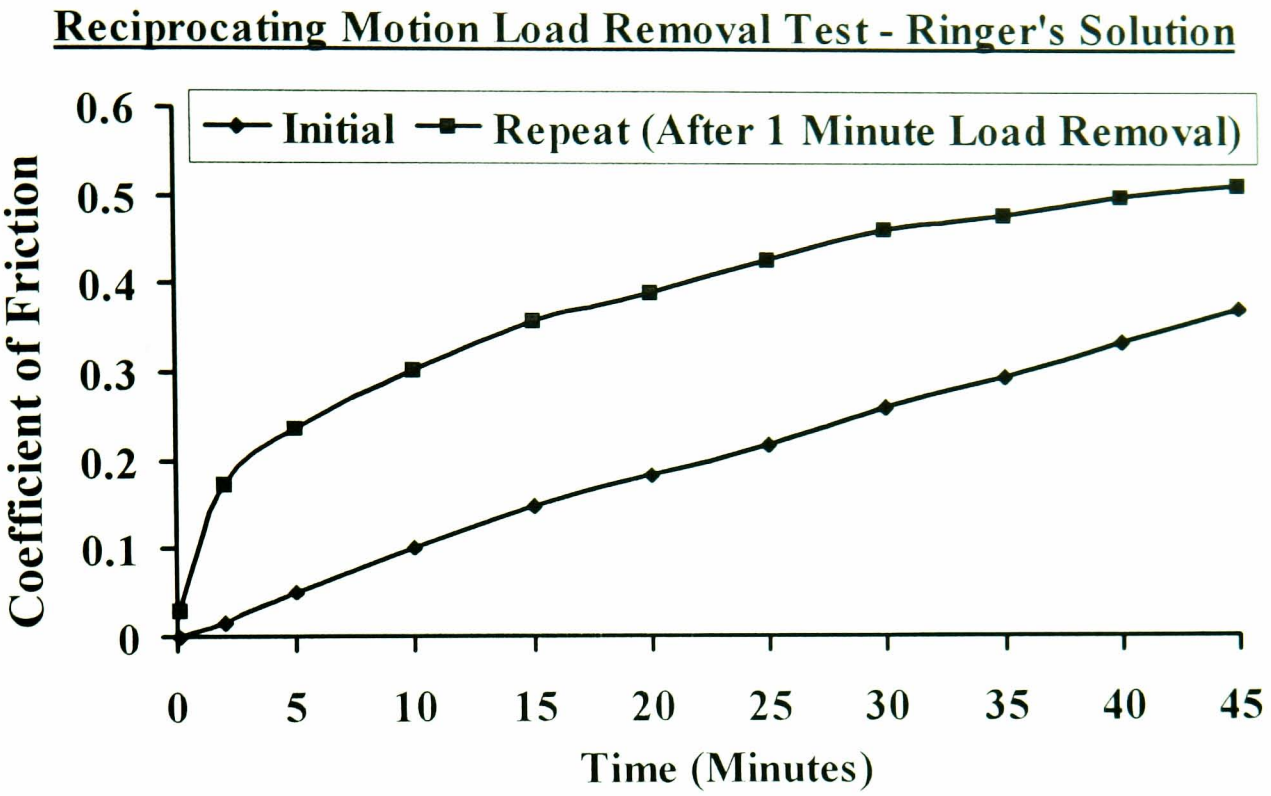
Figure 7-8 Steady state *initial* values plotted for reciprocating motion cartilage plug on metal counterface and stationary cartilage plug on metal counterface configurations. Synovial fluid was the lubricant.

7.3.5 Reciprocating Motion Load Removal Test

For this test the effect of the 1 minute period of load removal, following 45 minutes of loading while monitoring friction, was a sharp drop in friction coefficient from 0.373 at 45 minutes to 0.030 after 5 seconds of loading for the repeat measurements, Figure 7-9. For the repeat readings however the friction coefficient rose very sharply to 0.173 at just 2 minutes, subsequent increases becoming more gradual.

The repeat friction coefficients were found to be markedly higher at all loading times as indicated in the top graph of Figure 7-9. The bottom graph in Figure 7-9 demonstrates the effect of the 1 minute load removal period upon subsequent friction readings, in relation to the mean results of the *initial* reciprocating motion cartilage on metal configuration for Ringer's solution. After the 1 minute of unloading the distinct drop in friction coefficient was diminished by sharp increases occurring within a couple of minutes. The friction rise then became more gradual. At 75 minutes the effect of the load removal in reducing the friction coefficients expired as the readings approached those of the *initial* reciprocating motion cartilage on metal configuration.

The R_a (0.8 mm) values, from the 9 mm cartilage plug tested, rose from 0.8 μm (sampled prior to test) to 2.1 μm (after the reciprocating load removal test). Thus, as demonstrated in section 3.3.2, the non-contacting laser profilometry analysis confirmed that the cartilage surface had become rougher; the R_a (0.8 mm) means of the line profiles sampled before and after the test showed statistical significance ($p<0.0001$).



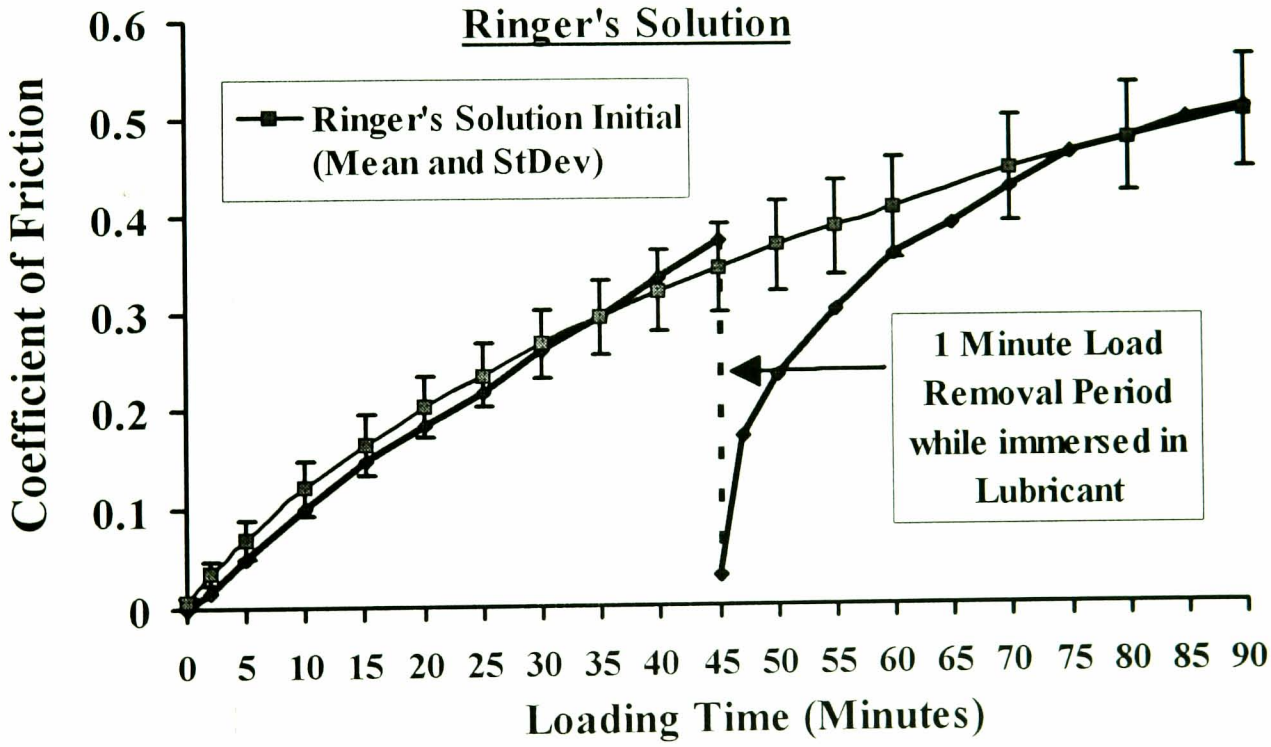


Figure 7-9 Reciprocating Motion Load Removal Test. Ringer's solution was the lubricant. The top graph plots the initial readings side by side with the repeat readings following a 1 minute load removal period. The bottom graph plots the initial and repeat results continuously, contrasting the effect of the 1 minute period of load removal with the mean results of the Ringer's Solution Reciprocating Motion Tests.

7.3.6 Reciprocating Motion Nine Millimetre Cartilage Plug on Hydrogel Contact Configuration Data

7.3.6.1 Comparison with Reciprocating Motion Nine Millimetre Cartilage Plug on Metal Contact Configuration

The friction coefficient results for the single test of the 9 mm cartilage plug on hydrogel counterface, under reciprocating motion and constant 30 N load, are listed in Table 7-7. These results have also been plotted in Figure 7-10, along with those of the cartilage on metal reciprocating motion tests for contrast.

Time (Minutes)	***Initial***	***Repeat***
0.08	0.010	0.014
2	0.017	0.024
5	0.021	0.044
10	0.045	0.069
15	0.068	0.090
20	0.086	0.109
25	0.096	0.122
30	0.110	0.136
35	0.124	0.145
40	0.134	0.158
45	0.14	0.165
50	0.148	0.171
55	0.155	0.177
60	0.159	0.182
70	0.177	0.186
80	0.185	0.195
90	0.183	0.188
100	0.185	0.189
110	0.186	0.191
120	0.185	0.187

Table 7-7 Nine millimetre cartilage plug on hydrogel counterface test friction coefficient *initial* and *repeat* results, under reciprocating motion.

The beneficial effect of having a hydrogel counterface instead of a metal counterface was abundantly clear for these reciprocating motion tests, Figure 7-10. The friction coefficients were always considerably lower for the hydrogel

counterface configuration and they attained a maximum value of approximately 0.185 within 70 to 80 minutes. For the metal counterface configuration friction coefficients were approximately 0.5 to 0.6 and still apparently raising, not having attained a peak value, after 120 minutes of loading. The *repeat* values of the hydrogel counterface configuration closely matched the *initial* values but increases were noticeable especially for loading times of 60 minutes and below.

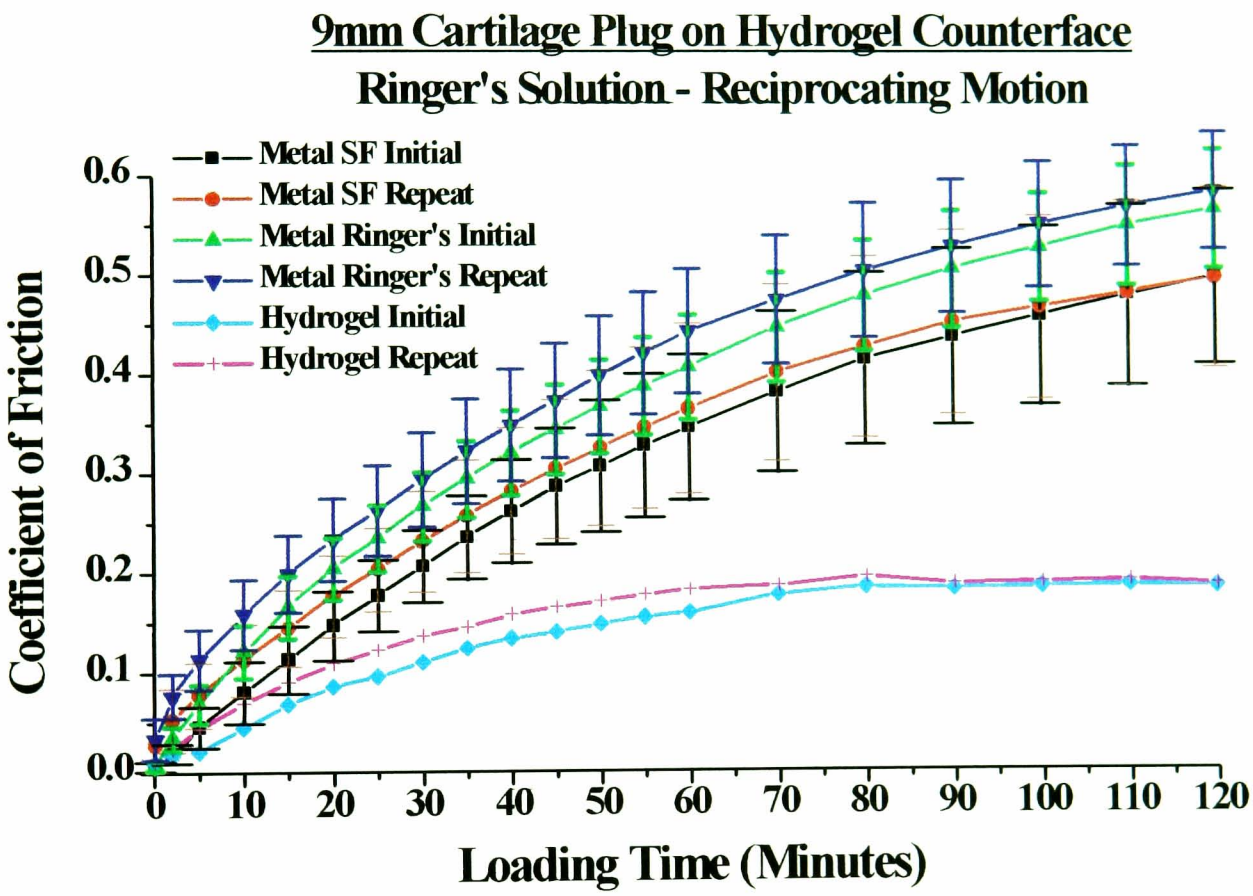


Figure 7-10 Comparison of the Reciprocating Motion 9 mm Cartilage Plug on Hydrogel Contact Configuration with the Reciprocating Motion 9 mm Cartilage Plug on Metal Contact Configuration.

7.3.6.2 Comparison with the Stationary Loaded Nine Millimetre Cartilage Plug on Hydrogel Contact Configuration

It was discovered that the introduction of reciprocating motion had no major influence on friction coefficients for the cartilage/hydrogel contacts, Figure 7-11. In making this observation it should be understood that Figure 7-11 is only contrasting the results of two separate tests using only a single cartilage plug in each case.

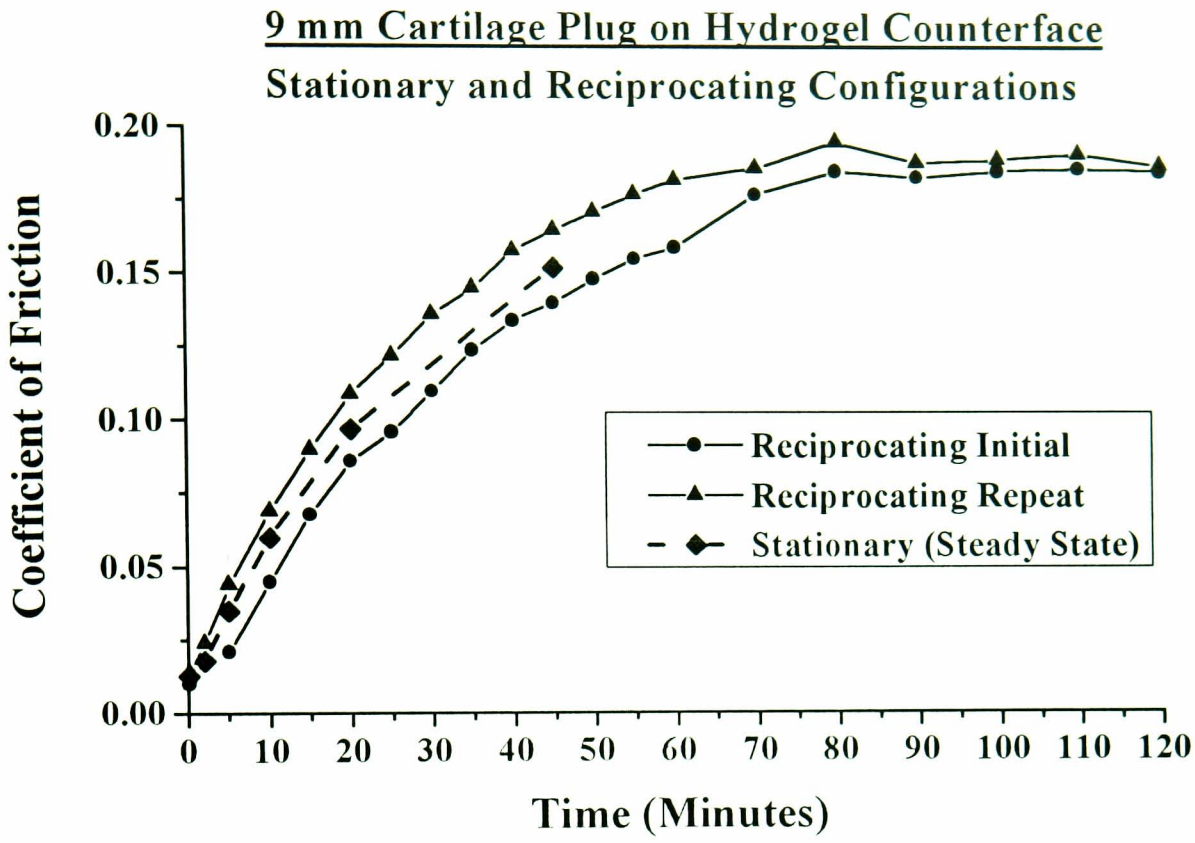


Figure 7-11 Comparison of the Reciprocating Motion 9 mm Cartilage Plug on Hydrogel test with the Stationary Loaded 9 mm Cartilage Plug on Hydrogel test.

7.4 Discussion

7.4.1 Reciprocating Motion Cartilage on Metal Friction Analysis

7.4.1.1 Reciprocating Motion Friction Traces

The rise in the coefficient of friction with loading time for this reciprocating motion configuration was immediately apparent from the friction traces obtained for each test. A typical set of friction traces for a particular test has been presented, comprehensively covering both the lower (Figure 7-1) and higher (Figure 7-2) loading time friction trace outputs. The steady state friction coefficient values were found to increase right up to the 120 minute loading time, Figure 7-2. Start-up friction coefficients could not be recorded as from the beginning of the test the contact was subject to continuous motion and therefore, by definition, there was no 'start-up' friction.

7.4.1.2 Repeatability within each Test and Reproducibility between Tests

For any particular test conducted for this reciprocating motion configuration the *repeat* results always followed the same form, when plotted, as the *initial* results; although generally they possessed moderately higher friction coefficients, Figure 7-3. Therefore the *initial* values, while displaying a good similarity to the *repeat* values, were not absolutely repeatable for the particular test protocol adopted. It must be concluded that the *initial* 2 hour reciprocating motion friction test altered the cartilage specimens in such a way as to increase subsequent identical friction measurements, despite an interim unloaded period of 2 hours minimum while immersed in lubricant. This phenomenon will be addressed soon within this discussion.

For the loading times of 10 minutes and above the reproducibility of the *initial* and *repeat* coefficient of friction values for both lubricants was favourable with coefficients of variance (standard deviation / mean) ranging from 0.4 to 0.1, Table 7-5. The higher loading times had the smallest coefficients of variance, as the mean values tended to increase faster in relation to the increase in

standard deviation for each progressive loading time. Also the Ringer's solution data was more reproducible by virtue of its coefficients of variance, in relation to the synovial fluid data, possessing generally higher friction coefficient means with smaller standard deviations for any given loading time. For the 2 and 5 minute loading periods the coefficients of variance were adequate at between 0.3-0.5. The 5 second loading time coefficients of variance however were approaching 1. This general analysis of friction readings being more reproducible at higher loading times and Ringer's solution results exhibiting less variability between tests mirrored the trend from previous configurations. A fuller analysis was appropriate here in respect of the greater number of loading times from which friction was sampled.

7.4.1.3 Influence of Loading Time

As with all the previous stationary configurations loading time was found to be a dominant factor influencing friction, refer to Figure 7-4 and Figure 7-5. The mean coefficient of friction values, after 120 minutes of reciprocating motion under a 30 N load, were approximately 0.5 and 0.6 for Ringer's solution and synovial fluid respectively. From the form of the coefficient of friction against loading time plots, further increases in friction would have been predicted if even longer loading times had been adopted, for both lubricants Figure 7-4 and Figure 7-5.

7.4.1.4 Initial versus Repeat Friction Coefficient Results

Following a minimum load removal of 120 minutes the *repeat* values were found to be somewhat higher, with percentage increases falling from around 20% at 20 minutes to 4% and 0.2% for Ringer's solution and synovial fluid respectively. For the lower loading times the percentage increases between the *initial* and *repeat* coefficient of friction values were much higher, as illustrated in Figure 7-12.

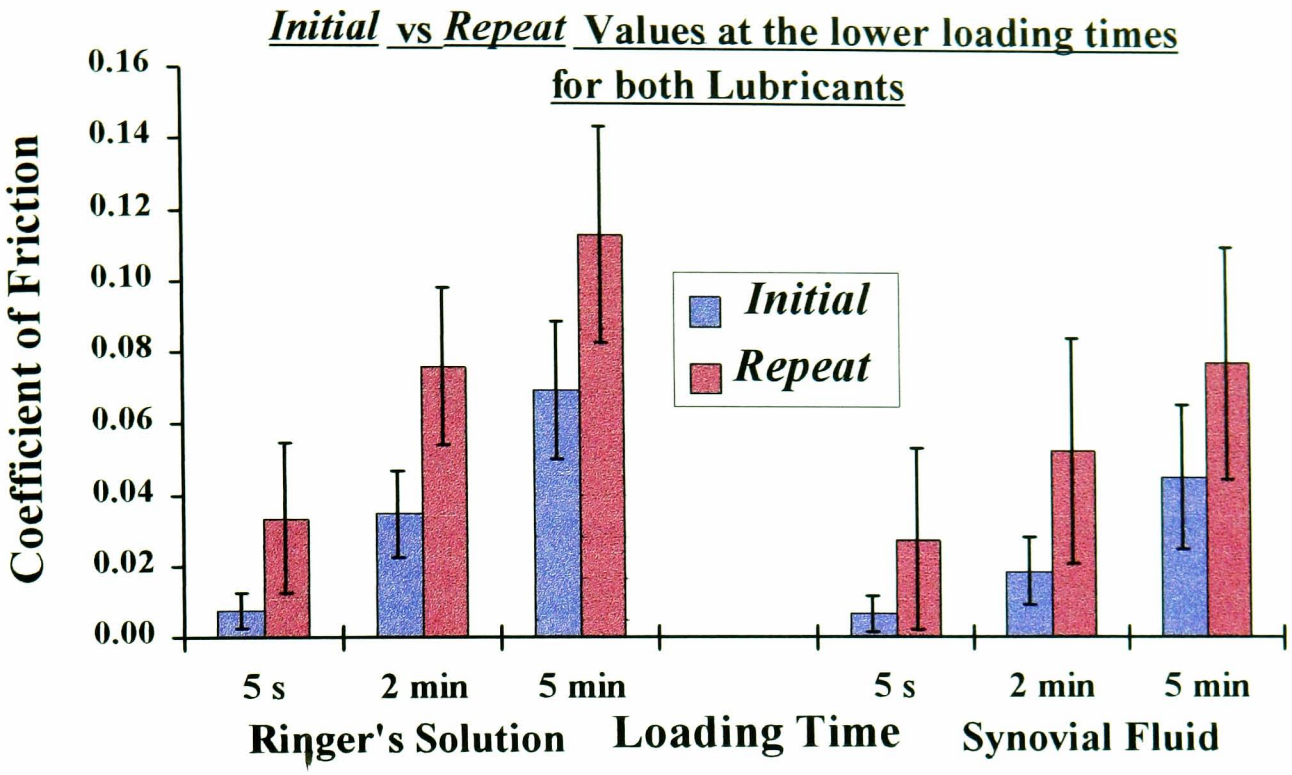


Figure 7-12 Comparison of the *initial* and *repeat* Reciprocating Motion 9 mm Cartilage Plug on Metal Contact Configuration values at the lower loading times for both lubricants.

The *initial* 120 minute reciprocating motion friction test was therefore found to increase subsequent friction readings, especially for the lower loading times, even after an equal period of load removal while immersed in lubricant. In order to effect subsequent friction readings, during the *initial* test, wear of the superficial boundary layer may have occurred (Stachowiak et al., 1994; Lipshitz and Glimcher, 1979) and/or the load removal period was not sufficient to fully rehydrate the cartilage specimens. It also must be remembered that by the time of beginning the *repeat* friction measurements the cartilage specimens had been exposed to room temperature for a minimum period of 4 hours and the possible occurrence of biological degradation, and its consequences, must be considered. For the previous stationary configurations investigated there was overall little sign of friction increases for repeat measurements, although repeat friction measurements were usually undertaken collectively before progressing to higher loading times and therefore there is no like with like similarity; except for the 9 mm cartilage on metal stationary configuration whereby marked differences between the initial and repeat results were mentioned as being infrequent and modest, refer to Figures 4-7 and 4-8. Obviously with sliding only occurring for

approximately 10 seconds during friction measurement for the stationary configurations the effect of wear was kept to a minimum.

7.4.2 Biphasic Lubrication under Reciprocating Motion

The discovery that friction coefficients were still rising after 2 hours of loading (Figure 7-4 and Figure 7-5), albeit successively more gradual, requires a re-examination of the biphasic lubrication hypothesis for articular cartilage under reciprocating motion. The load carriage of the fluid phase has been predicted to be much the same for both stationary compressive loading (Ateshian et al., 1994) and under compressive rolling motion (Ateshian and Wang, 1995). However , upon first consideration, it was uncertain as to whether or not there were still sufficient changes in fluid phase load carriage occurring beyond the 60 minute loading time to substantiate the increases in friction coefficient according to the biphasic lubrication hypothesis. Previous studies (Edwards, 1967; Kempson, 1980) have generally shown that equilibrium loading conditions, upon application of an external load to articular cartilage, were attained within approximately 60 minutes. Other workers have however provided data that indicated equilibrium conditions being established at and beyond 3 hours after loading (Maroudas, 1980; Higginson and Unsworth, 1981). The precise time constant to achieve equilibrium loading conditions, where $p_{applied} = p_{swelling}$-$p_{elastic}$ (see Chapter 2), will depend upon the applied load, condition of the cartilage (healthy/diseased; pH levels; irrigation fluid etc.), the size of the cartilage specimen and contact geometry with the applied load. In overall consideration, granted that equilibrium conditions, where the net pressure differential $\Delta p = 0$ and no further fluid flow occurs (Maroudas, 1980), may not be established for up to 240 minutes, then the results presented in this chapter do appear in keeping with the biphasic lubrication hypothesis.

7.4.3 Boundary Layer Wear

Also for this reciprocating motion configuration a wear component has been identified by the *repeat* results failing to perfectly match the *initial* results and by Ra (0.8 mm) values increasing from 0.8 μm (prior to test) to 2.1 μm (after the test), as referred to above. The *repeat* results were systematically higher at all

loading times, although this was less evident with increasing load duration, Figure 7-4 and Figure 7-5. Therefore the already acknowledged existence of a wear component was likely to have had a role to play in increasing the friction coefficients up to 120 minutes, in conjunction with the declining fluid phase load carriage. The declining fluid phase load carriage with loading time was still thought to be the dominant factor as the relatively small differences between the *repeat* and *initial* results did not point towards a catastrophic wear failure of cartilage lubrication for the *repeat* results compared to the *initial* results. Stachowiak et al. (1994) using a pin-on-disc apparatus to monitor friction of cartilage on metal contacts under continuous load and sliding stated that progressive wear of the superficial boundary lubricating layer was primarily responsible for increasing friction coefficients with loading time. The findings of this study dispute that explanation, having concluded that the decline of fluid phase load carriage was increasing friction coefficients and following an equal cartilage unloading period the *repeat* friction readings were generally the same; in line with the biphasic lubrication hypothesis. Although indeed a wear component had been recognised.

The finding that the *repeat* results at the lower loading times were substantially higher than the *initial* results does however warrant further attention, Figure 7-12. The lower loading times represent the more realistic physiological loading times that any particular articular cartilage area may be subject to. Although after two minutes of loading the friction coefficients were still below 0.1 for the *repeat* results the increase from the *initial* results did appear alarming. The fact that synovial fluid failed to prevent this discrepancy also would indicate that the wear effect was more than just removal of the boundary layer or that within this test protocol the synovial fluid could not sufficiently replenish and maintain the boundary layer. Therefore it appeared possible that after 2 hours of reciprocating motion under constant load the cartilage specimens had suffered a possibly permanent loss of lubrication performance due to wear. Despite this conclusion it should be remembered that this occurred for cartilage on metal contacts and wear

effects, within such a time scale and similar test conditions, for cartilage/cartilage contacts would not necessarily be anticipated.

7.4.4 Interpretation of High Friction Coefficients

Although the friction coefficients did become very high, ~ 0.6 after 120 minutes loading, it should again be considered that this was for cartilage/metal contacts. This was a necessary limitation of the reciprocating motion configuration as for cartilage/cartilage contacts steady state friction could not be reliably recorded, as explained in the previous chapter. It is envisaged that for cartilage/cartilage contacts, under the same test conditions, friction coefficients would never have become as high because the cartilage counterface would be subject to a cyclic loading phase. The sliding areas along the cartilage counterface would have experienced only transient loading, with unloaded time greater than the transient loading time. The reciprocating motion would therefore have maintained a proportionally high amount of water content within the cartilage counterface, ensuring relatively low coefficients of friction. Even for a stationary loaded cartilage/cartilage configuration, after 2 hours of loading, start-up friction may not have climbed so high as the cartilage/cartilage contact may well have afforded a better boundary and/or boosted lubrication mechanism and possess inherently lower friction. Although this prediction is less assured, even so for physiological cartilage/cartilage contacts, upon movement the combined sliding and rolling motion attributed to synovial joints, (Dowson, 1981b), will provide instantaneously lower steady state friction after the start-up as the two opposing cartilage layers translate upon, previously unloaded, alternative cartilage areas of the joint. In respect of these high friction coefficients it should further be noted that values as high as 0.9, for cartilage on glass contacts, have previously been quoted, (Dowson et al., 1968).

7.4.5 Influence of Lubricant during Reciprocating Motion

Synovial fluid provided statistically significantly lower coefficients of friction, compared to Ringer's solution, at loading times 2 to 50 minutes and from 5 to 40 minutes then 110 and 120 minutes for the *initial* and *repeat* results respectively ($p<0.05$), Table 7-6. The beneficial lubricating capacity of the synovial fluid was

plainly illustrated in Figure 7-6. In Figure 7-6 only the *initial* results of the two lubricants were contrasted for clarity. Systematic friction reductions for synovial fluid were therefore evident (Figure 7-6) and found to be statistically significant over broad ranges of loading time (Table 7-6).

The tribological advantages of using synovial fluid as the lubricant have thus been comprehensively proven for this reciprocating motion cartilage on metal contact configuration. Such conclusive evidence of the benefits of synovial fluid were not forthcoming from any of the previous stationary loaded cartilage plug on metal counterface or metal plug on cartilage counterface configurations. From these observations it is believed that the continuous sliding between the cartilage /metal contacts generated appreciable boundary layer wear, as previously mentioned. When synovial fluid was employed as the lubricant this boundary layer was replenished by macromolecular constituents adsorbed from the synovial fluid; constituents which were not available from the Ringer's solution.

It would appear however that for the harsh loading conditions adopted in this test configuration the synovial fluid was not able to fully maintain and replenish the boundary layer wear or perhaps more serious wear down into the superficial collagen fibres occurred. This was assumed because of the higher friction coefficients, especially at the lower loading times, found for the *repeat* values for both lubricants which was mentioned earlier. Although again it must be acknowledged that this could have been, instead, due to incomplete rehydration of the cartilage specimens.

7.4.5.1 Role of Synovial Fluid in Boundary Lubrication

The constituents within the synovial fluid thought to provide boundary lubrication are phospholipids (Hills, 1989; Williams et al., 1993; Kirk et al., 1993a), glycoproteins (Davis et al., 1979; Swann et al., 1981; Jay, 1992) or a complex of these macromolecules (Higaki and Murakami, 1995). Prior to this work it was widely believed that the boundary layer was adsorbed from the synovial fluid and furthermore that no synovial fluid present during friction testing inferred no boundary layer. However this present work has indicated that there is a boundary

layer commonly associated with or inherent to the articular cartilage surface. The function of the synovial fluid has been demonstrated to be the maintenance and replenishing of the boundary layer; for this particular test configuration under harsh loading conditions leading to possibly substantial superficial wear. Within a healthy synovial joint this function of synovial fluid is presumed to be far less demanding with wear of the boundary layer considerably less for cartilage/cartilage contacts under normal physiological activity. Nonetheless this role, over the lifetime usage of a synovial joint, is considered to be essential and its malfunction may well, over periods of weeks, months or years, lead to a gradual increase in superficial cartilage wear, causing friction levels to increase, spiralling more wear and the potential onset of arthritis. It is noteworthy to point out that constituents released from the cartilage matrix have also been cited as responsible for the articular cartilage boundary layer (Ghadially, 1983) and the synovial fluid's role in boundary lubrication may well be in coalition with that of the cartilage matrix.

One of the few friction studies to recognise that the boundary layer is more than probably inherent to the articular cartilage surface was recently reported, (Higaki et al., 1995). These workers acknowledged that the boundary layer must first be removed (or worn away during the course of testing as in the present study) in order to make objective assessments concerning the influence of potential bio-synthetic boundary lubricants.

7.4.5.2 Comparison with Stationary Loaded Cartilage/Metal Configuration

The additional effect upon friction coefficients produced by sliding, compared to stationary loading, for metal/cartilage contacts can be seen in Figure 7-7 and Figure 7-8. These graphs compare both the stationary and reciprocating motion (up to 60 minutes only) 9 mm cartilage on metal friction tests for the respective lubricants. For the Ringer's solution, Figure 7-7, beyond the 5 minute loading time the wear effects caused due to sliding were made apparent by the higher friction coefficients of the reciprocating motion configuration. For the synovial fluid, Figure 7-8, the *initial* values of the reciprocating motion configuration failed

to indicate any additional signs of wear due to sliding, in relation to the stationary configuration. Both these findings were in agreement with the conclusions regarding the nature of articular cartilage boundary lubrication stated above. In Figure 7-8, it can be seen, moreover, that the reciprocating motion configuration began to reveal lower friction coefficients beyond 5 minutes of loading. This can be attributed to some beneficial hydrodynamic lubrication effects, while still within the mixed lubrication regime, for the viscous synovial fluid under entraining action conditions.

7.4.6 Load Removal during Reciprocating Motion

The 1 minute load removal period allowed between two identical sets of friction readings taken over loading times of 45 minutes was evidently not sufficient to fully rehydrate the cartilage plug. This was shown by the repeat results being substantially higher than the initial results, Figure 7-9. As for the stationary loaded configurations load removal was shown to dramatically drop friction levels but the low levels could not be maintained, with a sharp increase in friction occurring within a couple of minutes of loading, Figure 7-9. However at the 45 minute loading time, when the 1 minute load removal was permitted, the friction coefficient was clearly reduced for a further 30 minutes compared to the friction coefficients obtained when no load removal was allowed, Figure 7-9 (bottom graph). These observations again support the hypothesis that fluid phase load carriage within the cartilage reduces the coefficient of friction. The 1 minute load removal period allowed a substantial amount of fluid back into the cartilage, vividly reducing friction levels in the short term and having a notably beneficial effect for a further 30 minutes upon the resumption of loading.

In the synovial joint loading regions of the articular cartilage surface are subject to varying amounts of cyclic load, especially during walking and running (Unsworth, 1993). A given area of the cartilage surface will experience both changing amounts of load and relatively long periods of no loading, in the course of day-to-day activities. It was of interest within the nature of this work to ascertain what amounts of loading and load removal were necessary to adequately maintain low

friction levels for a particular region of the articular cartilage surface. This was investigated by conducting cyclic loading friction tests described in the following chapter.

7.4.7 Reciprocating Motion with Hydrogel Counterface

Table 7-8 has listed the coefficient of friction results obtained for the 9 mm cartilage plug on hydrogel counterface configuration. Only one test was undertaken for this configuration and Ringer's solution was the lubricant. It was immediately apparent that the hydrogel counterface was lowering the friction coefficient values at all loading times. This was vividly demonstrated in Figure 7-10, comparing the *initial* and *repeat* reciprocating motion hydrogel results with the *initial* and *repeat* results of the reciprocating motion 9 mm cartilage on metal configuration for both lubricants. Therefore the hydrogel counterface did, as might be expected, appreciably lower friction coefficients when used in place of a metal counterface. After approximately 80 minutes of loading a peak friction coefficient value was reached, ~ 0.19, and maintained during longer loading times, Figure 7-10. Hence, the constant loading upon the cartilage plug and the consequent loss of fluid content and decline of fluid phase load carriage increased friction levels up to 80 minutes. However, beyond the 80 minutes loading time it was presumed that the friction coefficients had been stabilised at a maximum value of ~ 0.19 by fluid phase load carriage of the hydrogel counterface. It should be noted that the reciprocating motion and resultant cyclic loading experienced by the hydrogel counterface would have allowed the fluid content of the hydrogel to remain quite high. Therefore the friction coefficient was not rising beyond this asymptote value, irrespective of further cartilage plug loading which for the metal counterface continued to increase friction right up to 120 minutes. Fluid phase load carriage by the hydrogel counterface was deemed responsible for this suppression of increasing friction levels.

For the stationary loaded hydrogel configuration, referred to in the previous chapter, it was believed that biphasic lubrication by the hydrogel counterface was not occurring and the reduction of fluid phase load carriage within the cartilage

plug was the dominant factor increasing friction. The lower friction levels were thought to be established simply due to a very low intrinsic friction coefficient between the solid phases of the cartilage/hydrogel contacts. For this reciprocating motion configuration the biphasic influence of the hydrogel may have been brought into play as a result of the longer loading times employed, as the cartilage plugs became more and more depleted of fluid. It may well have been that the stationary configuration would have produced similar results at higher loading times, as the close matching of the two configurations up to 45 minutes loading would suggest, Figure 7-11.

The *repeat* values were observed to be higher than the *initial*, Table 7-7, especially for the lower loading times, in much the same manner as for the reciprocating motion metal counterface configuration. This would indicate a degree of boundary layer wear having occurred, as already described for the metal counterface.

7.5 Conclusions

- As for the previous stationary loaded configurations, loading time was the dominant factor affecting friction coefficients. Biphasic lubrication by articular cartilage was predicted to be in operation for this reciprocating motion configuration. The biphasic lubrication hypothesis was emphasised by the effect of load removal, upon friction coefficients, during the course of a reciprocating motion test. Boundary layer wear was also believed to be influencing friction coefficients but this was not as overriding a parameter as the decline of fluid phase load carriage in increasing friction.

- The small increase for the *repeat* friction coefficients compared to the *initial* values, especially evident at the lower loading times, has suggested the development of boundary layer wear for the cartilage plugs during the *initial* sliding friction tests. There was also a credible possibility that the increase was caused by incomplete rehydration of the cartilage specimens between the *initial* and *repeat* measurements. By the time the *repeat* friction measurements began the cartilage specimens had been exposed to room temperature for a minimum period of 4 hours and the possible occurrence of biological degradation, and its consequences, must be considered. This increase also took place for the reciprocating motion hydrogel counterface test.

- Systematic friction reductions for synovial fluid was plainly demonstrated, compared to Ringer's solution, and found to be statistically significant over broad ranges of loading time. From observations made in this study the role of synovial fluid in providing articular cartilage boundary lubrication has been indicated.

- The hydrogel counterface was lowering the friction coefficient values at all loading times. While the intrinsic friction coefficient between the cartilage/hydrogel contact was believed to be very low, biphasic lubrication by the hydrogel counterface was also assumed to be active.

For the hydrogel counterface the friction coefficient was not rising beyond an approximate 0.19 asymptote value obtained within 80 minutes of loading. This was irrespective of further cartilage plug loading, which for the metal counterface continued to increase friction right up to 120 minutes. Fluid phase load carriage by the hydrogel counterface was deemed responsible for the suppression of increasing friction levels. For this reciprocating motion configuration the biphasic influence of the hydrogel may have been brought into play as a result of the longer loading times employed (compared to the stationary configuration which indicated no obvious biphasic lubrication mechanism), as the cartilage plugs became more and more depleted of fluid.

8. Cyclic Loading Friction Tests - Cartilage on Metal Contacts

8.1 Introduction

In the synovial joint loading regions of the articular cartilage surface are subject to varying amounts of cyclic load, especially during walking and running (Unsworth, 1993). A given area of the cartilage surface will experience both changing amounts of load and relatively long periods of no loading, in the course of day-to-day activities and periods of rest. It was of interest within the nature of this study to ascertain what relative amounts of loading and load removal were necessary to adequately maintain low friction levels for a particular load bearing region of the articular cartilage surface. This was investigated by conducting cyclic loading friction tests upon articular cartilage plugs extracted from the load bearing areas of the bovine knee joint. The cyclic loading of the cartilage plugs consisted of set periods of loading (loading/load on phase) and complete load removal (unloading/load off phase). For each particular test a different cartilage plug was used and subjected to a particular load on/load off loading cycle. During the loading phase one or more friction measurements were recorded at predetermined time intervals.

During these tests, coefficients of friction, for the 9 mm diameter cartilage plug on metal counterface contacts studied, rose from the first loading cycle measurement to gradually approach a maximum asymptote value over the assessed number of applied loading cycles. These values were respectively referred to as the initial and equilibrium friction coefficient values. In the results section all the friction data has been displayed in the format of friction coefficient against number of loading cycles for each test. Also, to provide a clearer and more concise illustration of the results, graphs showing the initial and approximated equilibrium values only have been included.

8.2 Experimental Procedure

The cyclic loading friction tests involved sampling friction during the loading phase of set periods of load on/load off intervals for each specific test. Immediately after the end of the load off phase the loading cycle was repeated for a given number of times, usually twenty, Table 8-1. The tests were most frequently conducted under stationary loading conditions. The stationary loading, then sliding to record friction at a given time, method made the friction tests easier to conduct[1] and eliminated the occurrence of superficial cartilage wear, (Lipshitz and Glimcher, 1979; Stachowiak et al., 1994), as discussed in the previous chapter.

Nine millimetre diameter cartilage plug on metal contacts were used as they were the most comprehensively characterised over various tribological conditions. Use of one metal counterface reduced variability and allowed for the measurement of steady state and start-up friction. A cartilage on cartilage contact configuration could have been used but limited to stationary loading and start-up friction measurements only. Furthermore, different cartilage counterfaces would have had to be used for each test (to prevent wear and degradation), increasing variability between tests as well as being time consuming. Ringer's solution was always the lubricant. For the load off or unloading phase of the load cycle the cartilage remained immersed in the Ringer's solution, for all the tests.

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[1]During reciprocating motion the sliding had to be stopped to reset the pieozelectric transducer to zero, prior to restarting the sliding and taking a friction measurement. This procedure could take up to 10 seconds.

Test	Cartilage Specimen	Loading Cycle Load On/Load Off	Number of Cycles	Test Duration (Minutes)	Friction Sampling Times (seconds)	Reciprocating/ Stationary
1A	i	5 minutes/5 minutes	12	120	5, 120 & 300	Reciprocating
1B	i	1 minute/1 minute	20	40	30	Reciprocating
2A	ii	1 minute/30 seconds	20	30	30	Stationary
2B	ii	1 minute/45 seconds	20	35	30	Stationary
2C	ii	1 minute/1 minute	20	40	30	Stationary
3A	iii	2 minutes/1 minute	20	60	30 & 90	Stationary
3B	iii	2 minutes/90 seconds	20	70	30 & 90	Stationary
3C	iii	2 minutes/2 minutes	20	80	30 & 90	Stationary
4A	iv	3 minutes/10 seconds	20	63	30 & 120	Stationary
4B	iv	3 minutes/90 seconds	20	90	30 & 120	Stationary
4C	iv	3 minutes/3 minutes	20	120	30 & 120	Stationary
5A	v	3 minutes/10 seconds	20	63	30 & 120	Stationary
5B	v	3 minutes/10 seconds	20	63	30 & 120	Reciprocating

Table 8-1 Test protocol for the cyclic loading friction tests 1-5, for 9 mm cartilage plugs on a metal counterface. Note that for each test (1-5) a different 9 mm cartilage plug was used. Test 1 was essentially a preliminary investigation of cyclic load friction testing. Tests 2-4 each had a given period of loading, 1 minute, 2 minutes and 3 minutes respectively, with varying amounts of unloading times for each A, B and C condition. Test 5 compared a stationary, cyclic load test (5A) with a reciprocating motion cyclic load test (5B), under identical experimental conditions.

For each new cartilage plug used in this investigation standard stationary loading friction tests, two after 5 seconds of loading and one after 2 minutes of loading, were performed to confirm the integrity of the specimen. This was done before beginning the cyclic loading test. For each test (1-5) a different cartilage plug was used, Table 8-1. Between cyclic loading friction tests, using the same cartilage specimen, an unloading time of at least the total test duration of the previous test was routinely allowed, while immersed in Ringer's solution.

The loading cycle and other variables applied for each test will now be described in the ensuing sections. These details are summarised in Table 8-1. The applied load was 30 N throughout.

8.2.1 Test 1

Test 1 was essentially a preliminary investigation of cyclic load friction testing and sought to determine an appropriate test protocol for the friction apparatus and cartilage specimens involved. For test 1A the cartilage plug underwent a repetition of twelve loading cycles of 5 minutes loading followed by 5 minutes unloading. The 5 minute loading period allowed friction to be sampled at three time intervals being 5 seconds, 2 minutes and 5 minutes. It should be pointed out that recording friction at the 5 minutes loading time required an extra 20 seconds of loading while friction data was sampled. Therefore the actual loading cycle was closer to 5 minutes 20 seconds loading followed by 4 minutes 40 seconds unloading. Test 1B was conducted in order to establish the minimum practical load cycle time, consisting of equal duration loading and unloading phases. This minimum value was found to be 1 minute loading/1 minute unloading which permitted friction data to be recorded during the loading phase and the data file saved to the hard drive prior to the start of the subsequent loading cycle. So the loading cycle for test 1B was 1 minute loading/1 minute unloading. Coefficients of friction were calculated from readings obtained at 30 seconds during the loading phase of each of the twenty loading cycles carried out. Both tests 1A and 1B were conducted under reciprocating motion.

8.2.2 Tests 2, 3 and 4

The loading cycles for tests 2, 3 and 4 consisted of loading times of 1, 2 and 3 minutes respectively. For each test condition (A, B, C) the adopted unloading times were ½, ¾ and equal that of the loading time, Table 8-1. This was except for test 4A when a relatively harsh loading cycle of 3 minutes loading/10 seconds unloading was in operation. Cyclic load friction tests 2-4 were all conducted under stationary loading conditions. This permitted both steady state and start-up friction coefficients to be calculated. For each test condition the cartilage plugs underwent 20 loading cycles, with a subsequent 20 friction readings taken during the loading phase at each designated sampling time, Table 8-1.

8.2.2.1 Test 2

For this test series the loading phase of the cycle lasted for 1 minute with unloading times lasting for ½ minute, ¾ minute and 1 minute respectively for tests 2A, 2B and 2C. Friction was sampled at 30 seconds during the loading phase of each of the 20 loading cycles conducted.

8.2.2.2 Test 3

For this test series the loading phase was increased to 2 minutes with unloading times of 1 minute, 1½ minutes and 2 minutes respectively for tests 3A, 3B and 3C. The 2 minute loading phase was sufficient to enable two friction sampling times, being 30 seconds and 90 seconds.

8.2.2.3 Test 4

For this test series the loading phase was again increased to 3 minutes. The prescribed unloading times were 10 seconds, 1½ minutes and 3 minutes for tests 4A, 4B and 4C respectively. Friction was sampled at 30 seconds and 120 seconds (2 minutes) during the 3 minute loading phase.

8.2.3 Test 5

As tests 2-4 were all conducted under stationary loading conditions it was of interest to determine what difference, if any, the effect upon friction coefficients reciprocating motion would have. For test 5A friction measurements were recorded after 30 seconds and 120 seconds (2 minutes) of stationary loading, while the cartilage plug was subject to a 3 minute loading/10 seconds unloading cycle. Subsequently, the same cartilage plug, following an unloaded period while immersed in Ringer's solution of approximately 1 hour and 20 minutes, had the same loading time friction measurements taken under the same loading cycle, but on this occasion during reciprocating motion (test 5B). The repeatability of the friction readings was therefore assessed between a stationary and a reciprocating motion configuration, for the same cartilage specimen. The 3 minute loading/10 seconds unloading cycle was chosen because it produced the most substantial increases in friction levels over the assessed 20 loading cycles. Also, as this loading cycle had been used for a previous test (4A) it provided an opportunity to evaluate the reproducibility of the tests between the two cartilage specimens used for test 4 and 5 respectively.

During the reciprocating motion test 5B a 'start-up' friction peak was recorded, as the sliding direction changed. This was not conventionally true start-up friction as recorded for the stationary loading test 5A. The start-up results for tests 5A and 5B have however been compared with one another for this particular test in the results section, as have the steady state values.

8.3 Results

8.3.1 Test 1

The coefficients of friction recorded at 2 minutes and 5 minutes for test 1A increased between every loading cycle until about the sixth or seventh cycle, Figure 8-1. The first loading cycle friction coefficients were 0.036 and 0.075 for the 2 and 5 minute loading times respectively. This was in good agreement with the mean results obtained for the equivalent Ringer's solution reciprocating motion cartilage plug on metal counterface tests of the previous chapter, being 0.035 (±0.012) and 0.069 (±0.019) for the 2 and 5 minute loading times respectively. Equilibrium values were then established of approximately 0.065 and 0.110 for the 2 and 5 minute loading times respectively. No prominent rise in the friction coefficient was detectable at the 5 second loading time. There were changes in the friction coefficients over the assessed 12 loading cycles for this 5 second loading time but they were within the established level of repeatability.

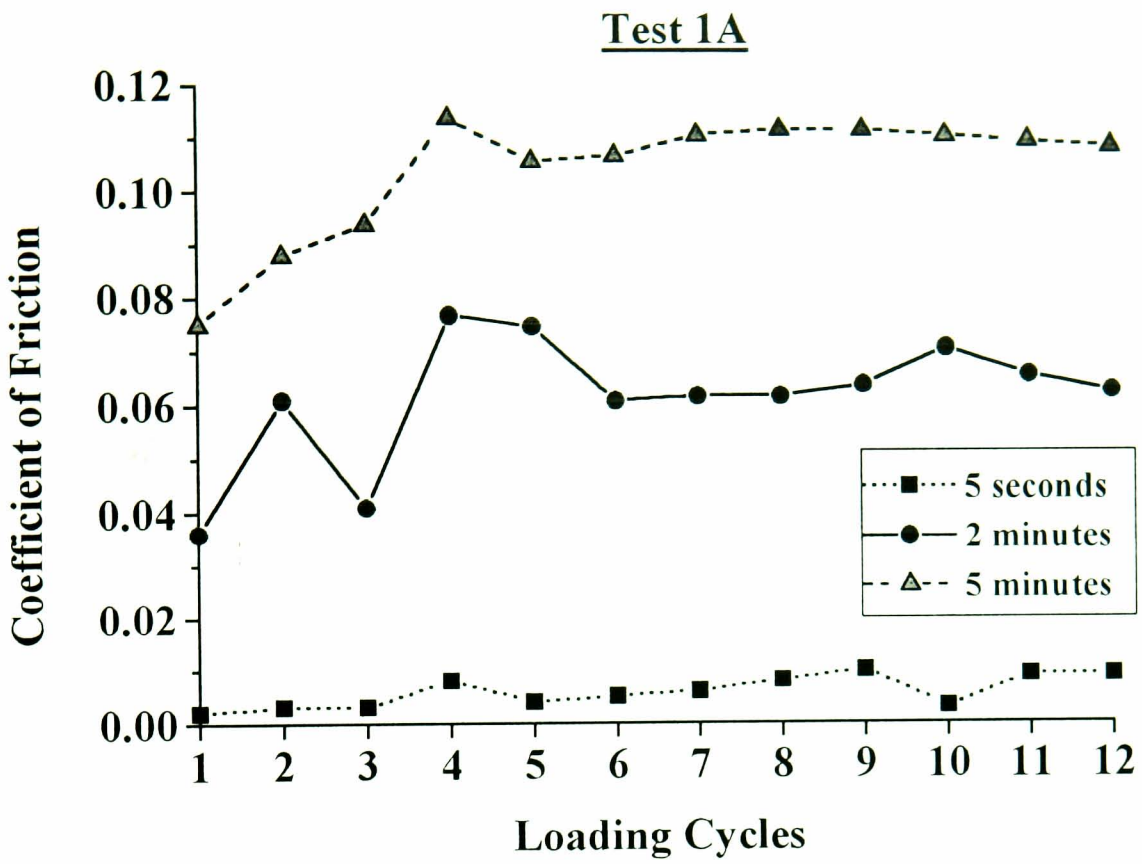


Figure 8-1 Coefficient of friction results for Test 1A. The loading cycle was planned to be 5 minutes loading/5 minutes unloading, although actually 5 min 20 s loading/4 min 40 s unloading. This was due to an extra 20 seconds of loading required to measure friction at the 5 minute loading time, which was not be taken into account prior to the start of the test. As displayed in the legend, friction measurements were taken at 5 seconds, 2 minutes and 5 minutes during the loading phase of the cycle.

For the 5 second loading time it was probable that this very short loading time was not sufficient to detect changes in friction due to any difference in the hydrational state of the cartilage and/or general tribological conditions for equal duration loading and unloading cycles. Therefore, in subsequent tests a 30 second loading time was adopted as the minimum friction sampling time.

The coefficients of friction recorded at 30 seconds for test 1B increased between every loading cycle up to the fifth cycle, Figure 8-2. The first friction coefficient value was 0.015, with a fluctuating equilibrium value being established at around 0.025.

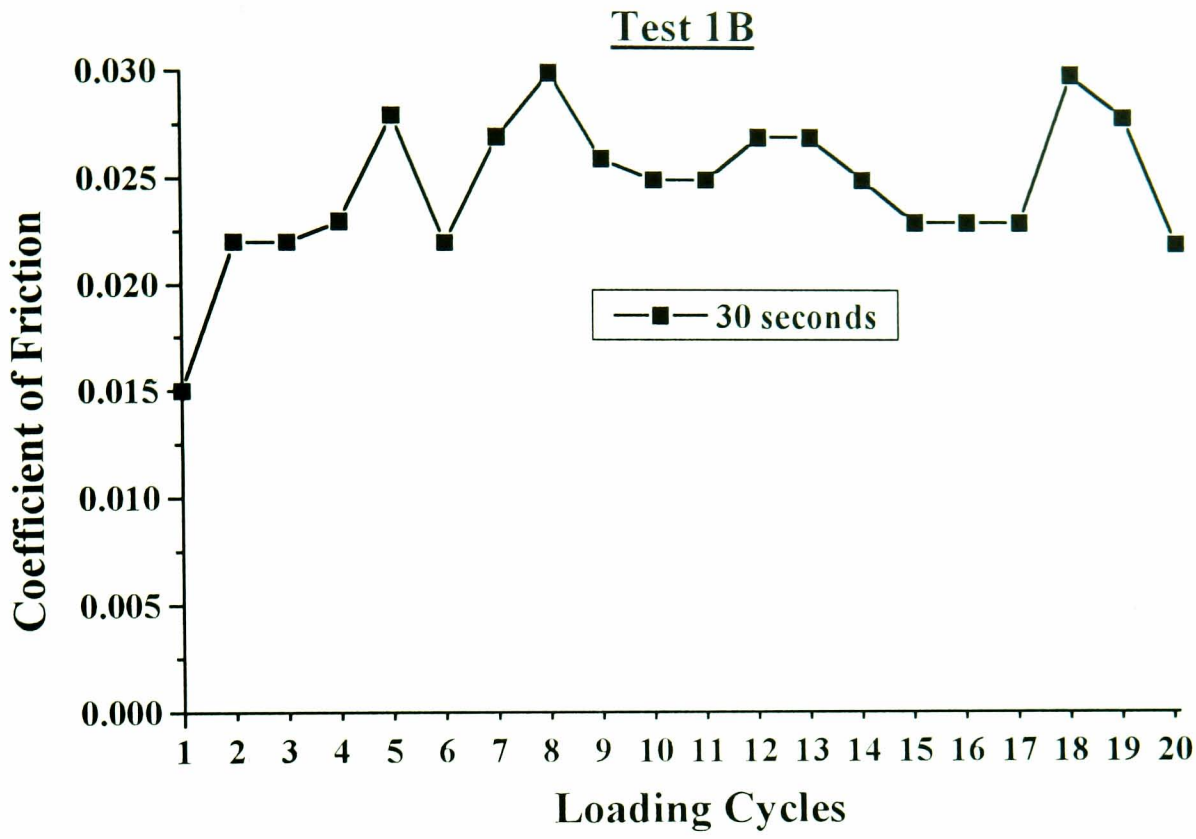


Figure 8-2 Coefficient of friction results for Test 1B. The loading cycle was 1 minute loading/1 minute unloading. Friction measurements were sampled at 30 seconds during the loading phase of the cycle.

8.3.2 Test 2

For tests 2A, 2B and 2C (all 1 minute loading phases with 30 s, 45 s and 1 minute unloading phases respectively) steady state friction coefficients recorded at 30 seconds demonstrated a small gradual increase for each loading cycle, Figure 8-3. From the seventh loading cycle test 2A, having the shortest load removal period, consistently provided the highest friction coefficient value for every loading cycle. These values were slightly higher than those of the close matching results of tests 2B and 2C.

Moderate increases in start-up friction coefficient values recorded at 30 seconds were observed over the 20 loading cycles evaluated for tests 2A, 2B and 2C, Figure 8-3. These start-up friction coefficients revealed a little less distinction between tests 2A, 2B and 2C. However the start-up friction coefficients for test

2A were consistently higher than those of test 2C; being the tests with the lowest and highest load removal periods respectively, Figure 8-3. The friction coefficients for test 2B, having the intermediate load removal period, meandered between those of tests 2A and 2C. After twenty loading cycles the coefficient of friction values were still quite low at <0.02 and <0.04 for steady state and start-up friction respectively.

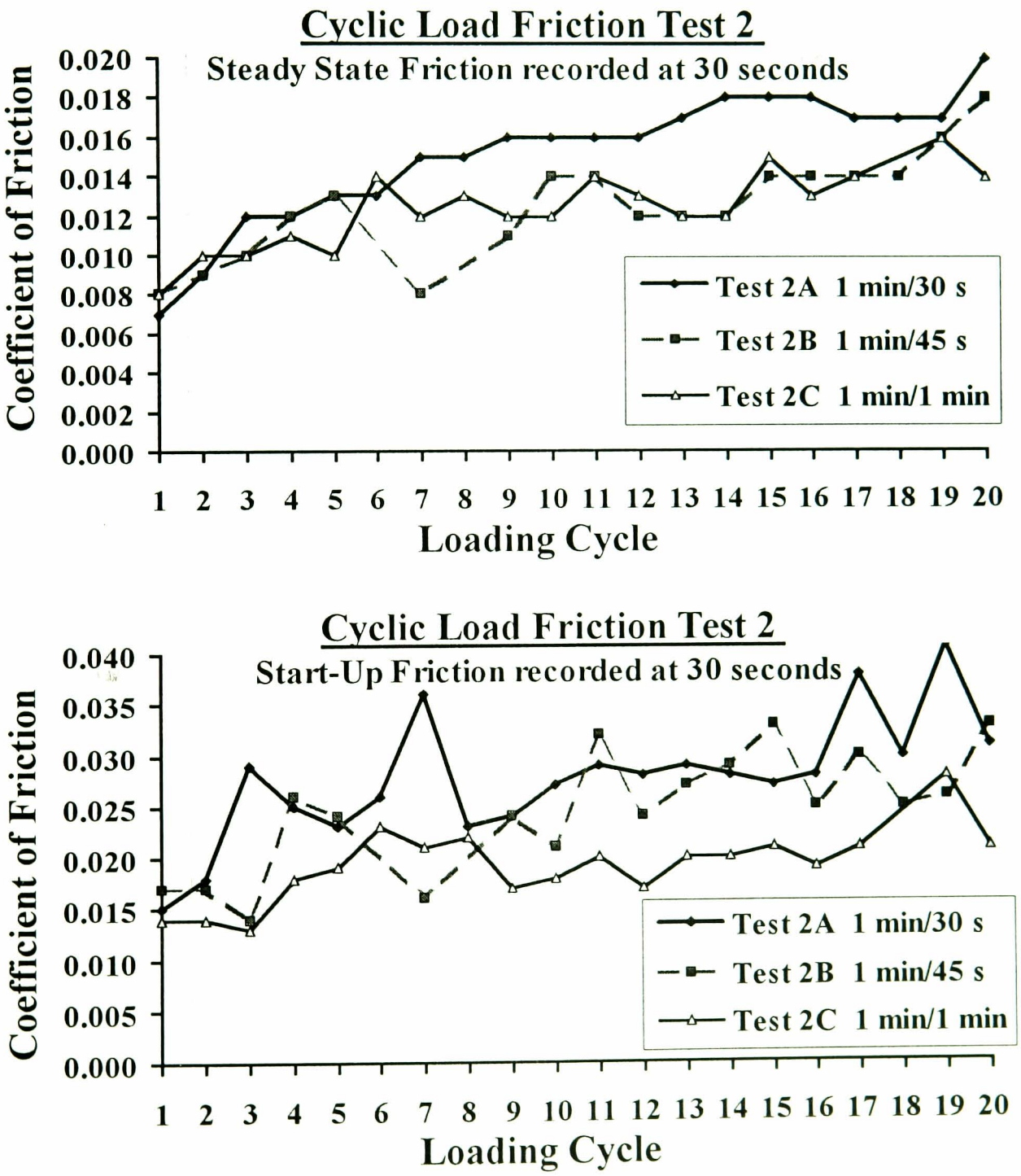


Figure 8-3 Steady state (top graph) and start-up (bottom graph) coefficient of friction results sampled at 30 seconds during the loading phase for Tests 2A, 2B and 2C. The adopted loading cycle for each of these tests is shown in the legend.

8.3.3 Test 3

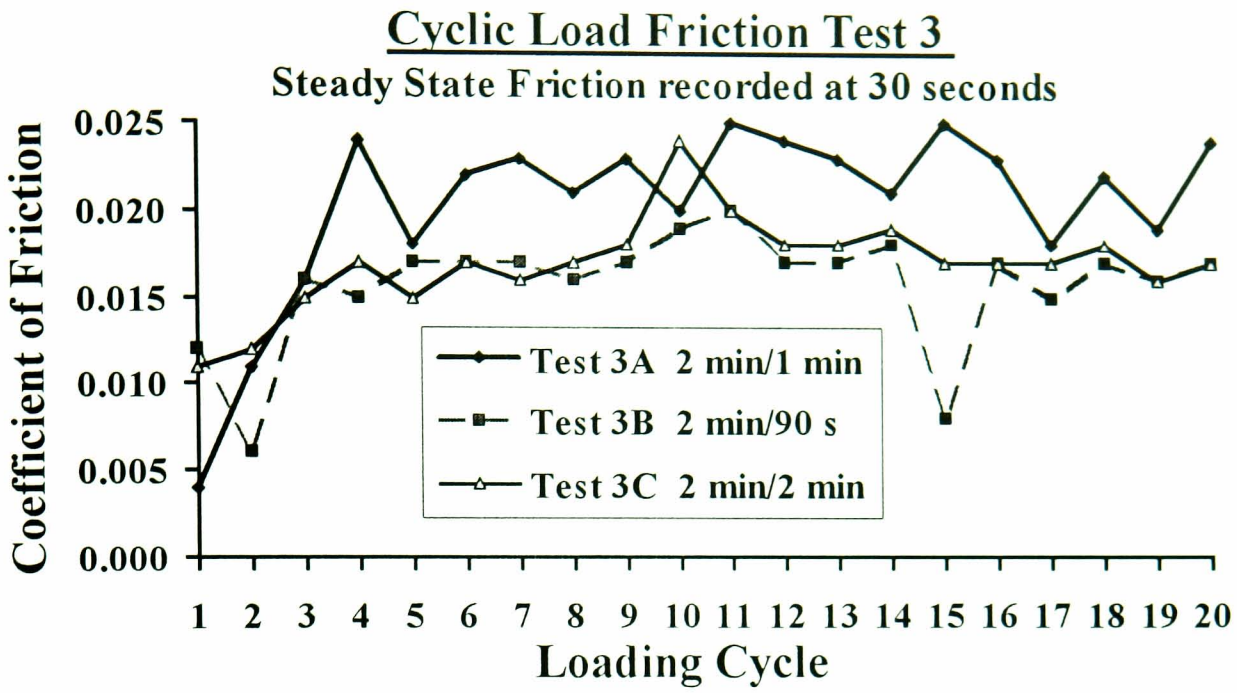
For tests 3A, 3B and 3C (all 2 minute loading phases with 1 minute, 90 s and 2 minute unloading phases respectively) steady state friction coefficients recorded at 30 seconds demonstrated increases for each loading cycle, Figure 8-4. This was especially true for test 3A, having the shortest unloading period per cycle. From the fourth loading cycle test 3A consistently provided the highest friction coefficient value for every loading cycle, with only one exception at loading cycle 10. These values were slightly higher than those of the close matching results of tests 3B and 3C. Coefficient of friction values for test 3A rose from 0.004 for the first loading cycle to an equilibrium value of approximately 0.022, established by the sixth cycle. Coefficient of friction values for tests 3B and 3C rose from 0.012 and 0.011 for the first loading cycle to an equilibrium value of approximately 0.017, again established by the sixth cycle.

Increases in start-up friction coefficient values recorded at 30 seconds were found over the 20 loading cycles evaluated for tests 3A, 3B and 3C, Figure 8-4. These start-up friction coefficients revealed no discernible difference between tests 3A, 3B and 3C. A rough equilibrium value of 0.025-0.030 was attained by the sixth loading cycle.

The 1 minute unloading phase of test 3A was clearly not enough to maintain friction coefficients at the original first cycle values for the steady state 90 second results, Figure 8-5. For test 3A the steady state friction coefficient initially rose from 0.015 to attain an equilibrium value of about 0.045 by the seventh loading cycle. The steady state friction coefficient 90 second values for tests 3B and 3C, having 1½ minutes and 2 minutes unloading phases respectively, matched closely and did not alter much over the 20 loading cycles evaluated. Their values of ~0.015 for tests 3B and 3C were similar to those found at 30 seconds.

Start-up friction coefficients at 90 seconds were also found to be higher for the test 3A results compared to the once again close matching and little changing

results of tests 3B and 3C, Figure 8-5. The test 3A start-up friction coefficients at 90 seconds increased from an initial 0.023 to reach an equilibrium value of ~0.045. For tests 3B and 3C the start-up values of ~0.033 were substantially higher than their steady state results of ~0.015 found at 30 seconds, causing the difference between test 3A and tests 3B & 3C to be less distinctive when looking at the start-up friction data. Generally though the start-up and steady state results were much the same.



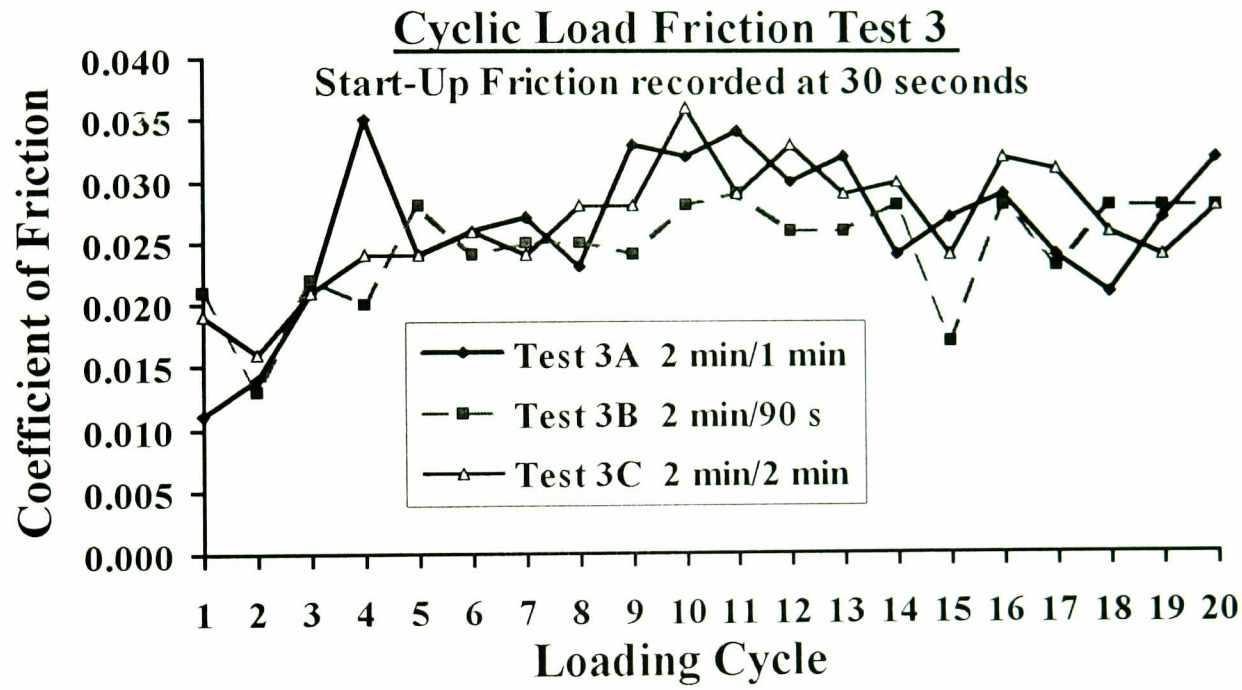
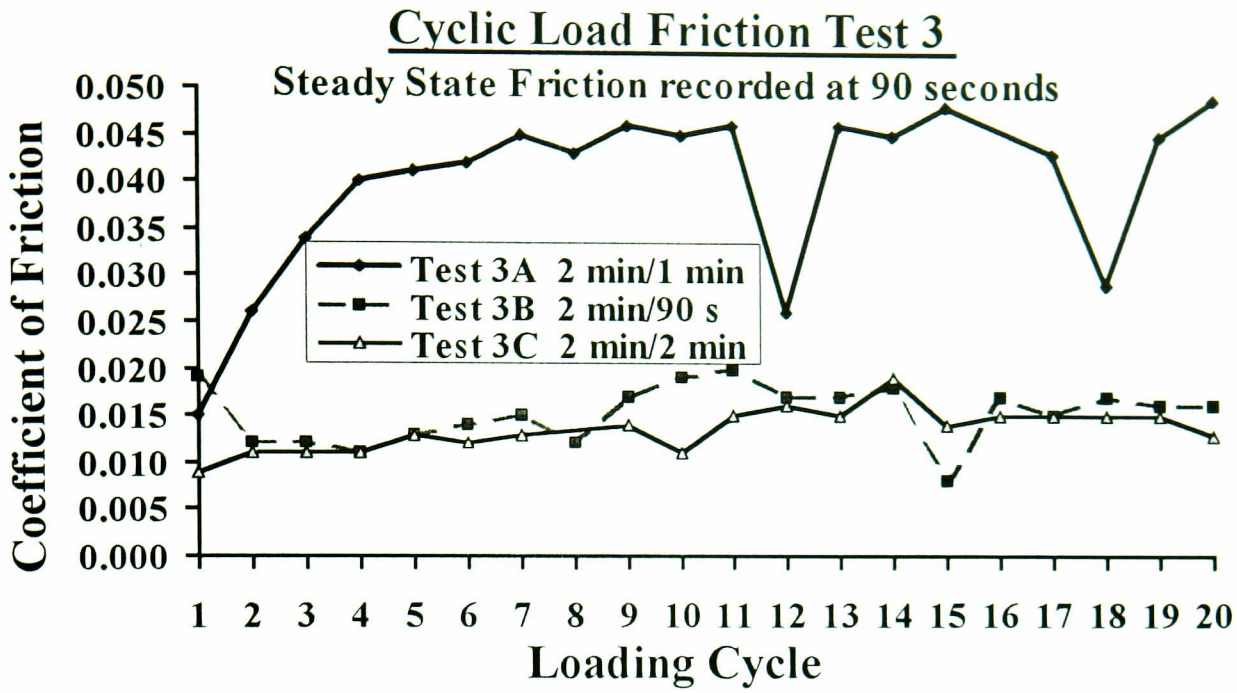


Figure 8-4 Steady state (top graph) and start-up (bottom graph) coefficient of friction results sampled at 30 seconds during the loading phase for Tests 3A, 3B and 3C. The adopted loading cycle for each of these tests is shown in the legend.



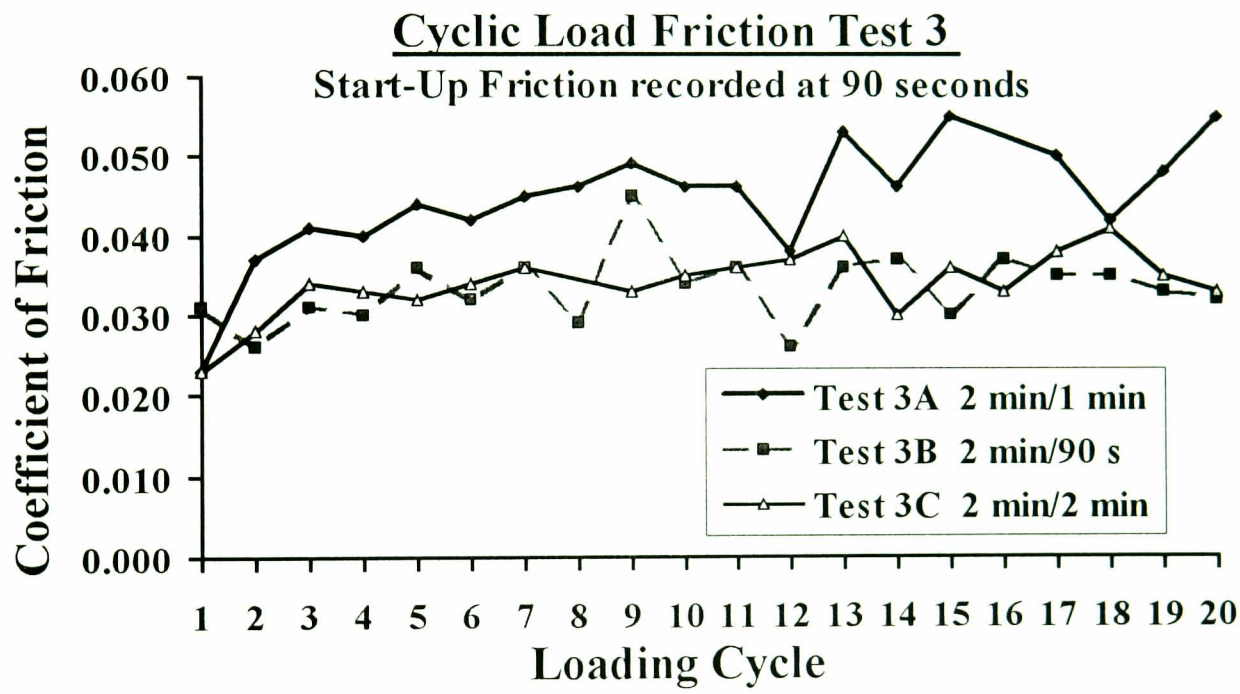


Figure 8-5 Steady state (top graph) and start-up (bottom graph) coefficient of friction results sampled at 90 seconds during the loading phase for Tests 3A, 3B and 3C. The adopted loading cycle for each of these tests is shown in the legend.

8.3.4 Test 4

Any differences between the steady state and start-up coefficients of friction recorded at 30 seconds for tests 4A, 4B and 4C (all having 3 minute loading phases) were found to be generally insignificant, Figure 8-6. The steady state friction coefficients for test 4C, having the longest duration unloading phase of 3 minutes, were surprisingly moderately higher than test 4B, having an unloading phase of 1½ minutes. Their respective equilibrium values, rising from identical first cycle values of 0.002, were ~0.022 and ~0.010, reached by the twelfth cycle. The start-up coefficients of friction for tests 4B and 4C revealed a little less difference, Figure 8-6. Both the steady state and start-up coefficients of friction recorded at 30 seconds for test 4A rose sharply, from respective first cycle readings of 0.004 and 0.005, up to the twelfth cycle at which point an equilibrium value of ~0.070 was attained. Test 4A had a very short load removal time of 10 seconds per cycle compared to the adopted loading time of 3 minutes.

The steady state coefficients of friction recorded at 2 minutes for tests 4B and 4C rose from respective initial values of 0.008 and 0.016 to equilibrium values of ~0.025 and ~0.030, attained by the second or third cycle, Figure 8-7. Similarly, the start-up coefficients of friction recorded at 2 minutes for tests 4B and 4C rose from respective initial values of 0.009 and 0.016 to equilibrium values of ~0.050 and ~0.042, attained by the third and twelfth cycle respectively, Figure 8-7.

For test 4A the steady state and start-up coefficients of friction recorded at 2 minutes rose from a very low identical initial value of 0.006 to equilibrium values of ~0.065 and ~0.085 attained by the eleventh and fifth cycle respectively, Figure 8-7. At the 2 minute loading time the mean steady state and start-up friction coefficients obtained for the equivalent Ringer's solution stationary loaded cartilage plug on metal counterface tests were 0.034 (±0.008) and 0.046 (±0.015) respectively. Clearly the equilibrium values for test 4A at 2 minutes had risen outside the range of these quoted values. This was especially significant as it was

noted that the stationary load friction test checks revealed that the cartilage specimen used for test 4 provided unusually low steady state (and start-up friction) coefficients of 0.002 after 5 seconds of loading (0.005 for start-up) and 0.006 after 2 minutes of loading (0.006 again for start-up).

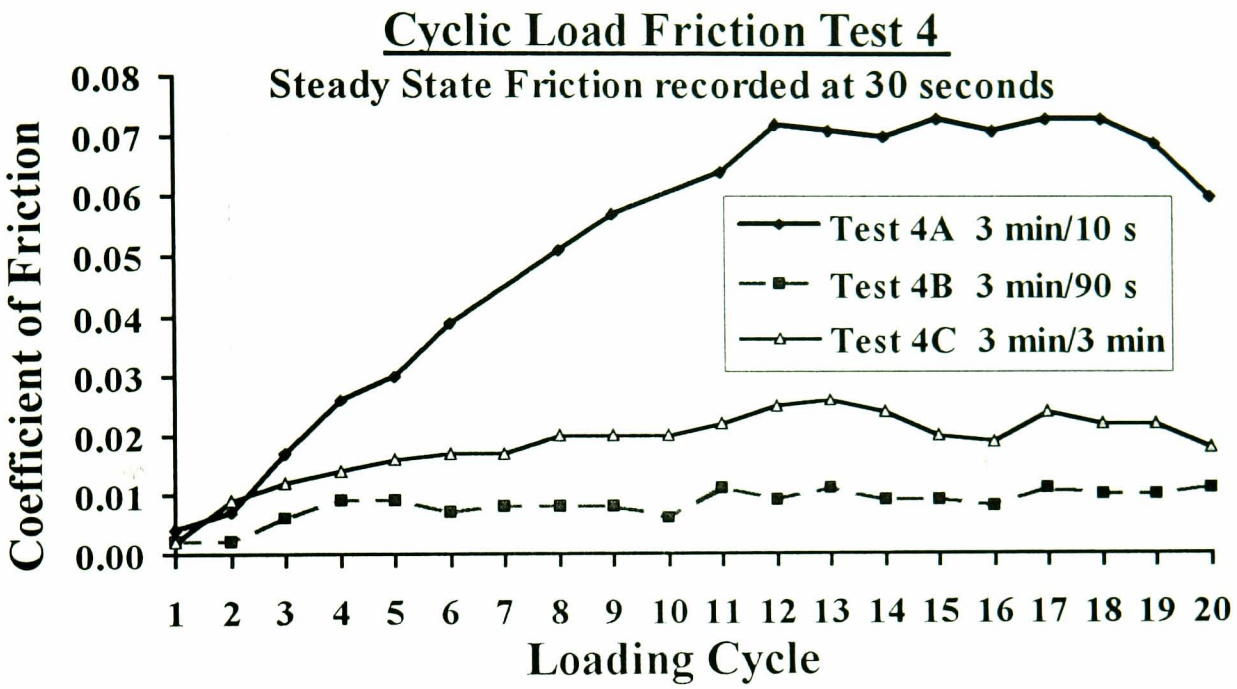


Figure 8-6 Steady state (top graph) and start-up (bottom graph) coefficient of friction results sampled at 30 seconds during the loading phase for Tests 4A, 4B and 4C. The adopted loading cycle for each of these tests is shown in the legend.

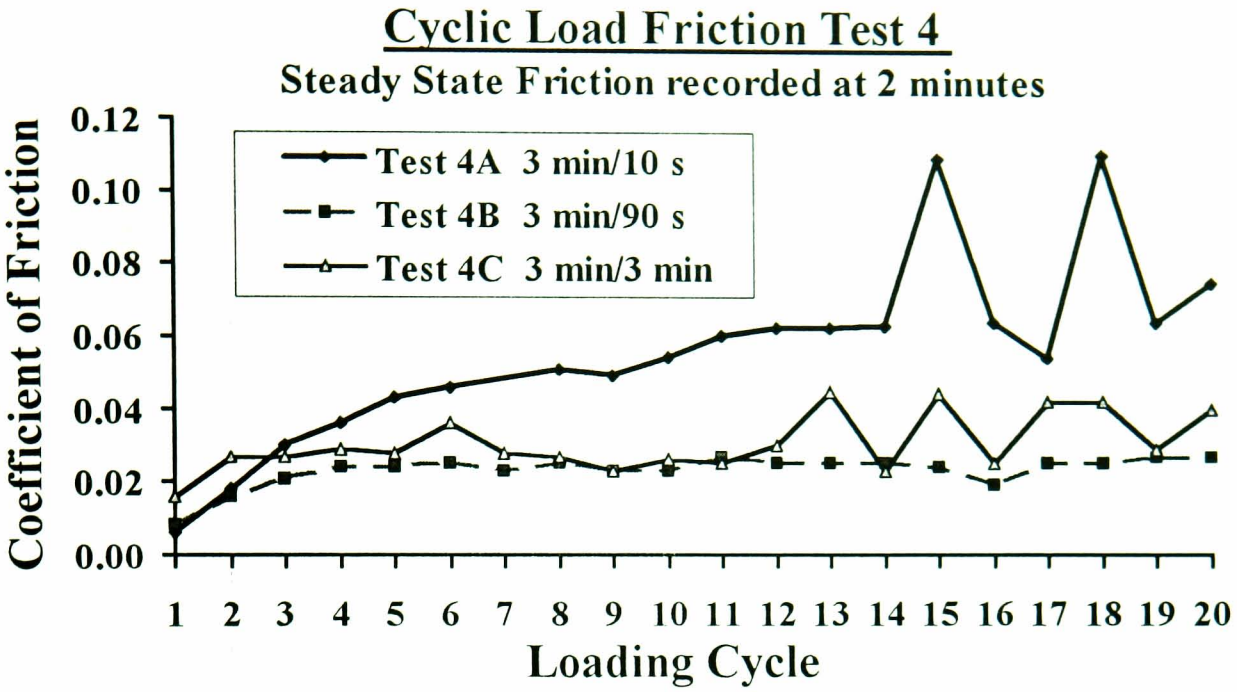
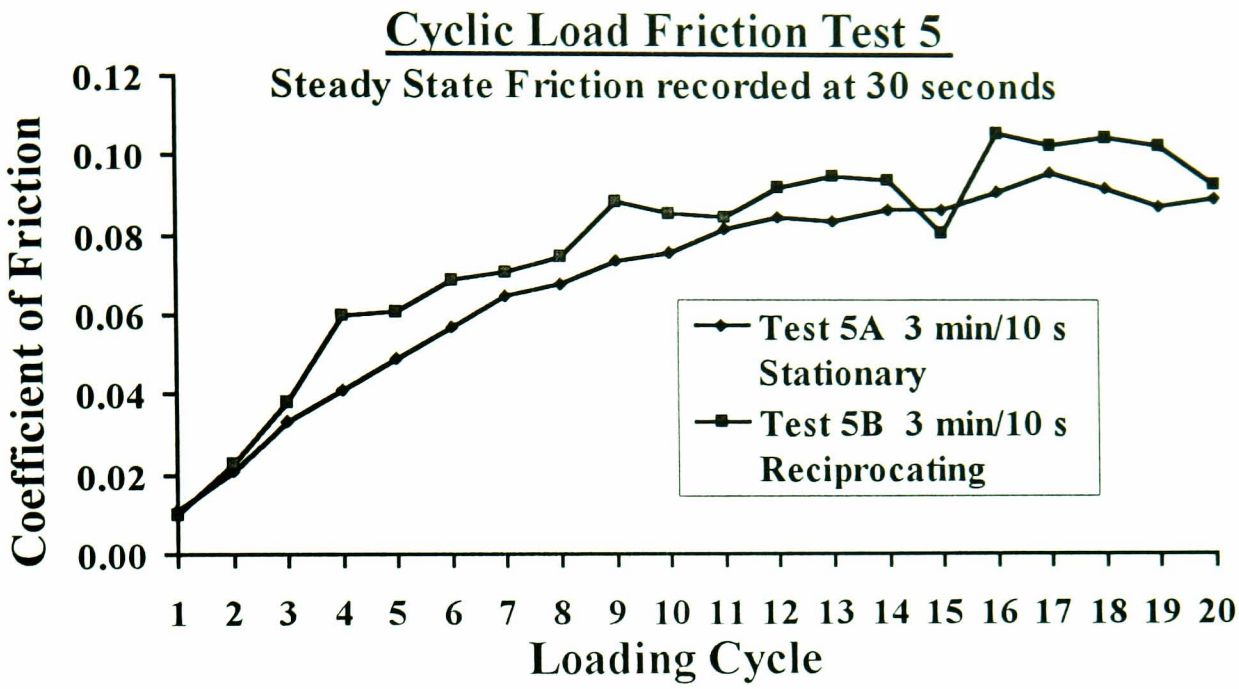


Figure 8-7 Steady state (top graph) and start-up (bottom graph) coefficient of friction results sampled at 2 minutes during the loading phase for Tests 4A, 4B and 4C. The adopted loading cycle for each of these tests is shown in the legend.

8.3.5 Test 5

The reciprocating motion friction coefficients calculated from test 5B were much the same as those calculated for the stationary loaded test 5A, although generally the test 5B values were slightly higher, Figure 8-8 and Figure 8-9. By comparing Figure 8-8 and Figure 8-9 it can be seen that the steady state and start-up values were also quite similar. For both tests 5A and 5B steady state friction coefficients recorded at 30 seconds were ~0.010 at the first cycle rising to a possible equilibrium value of ~0.095 by the sixteenth cycle. For tests 5A and 5B steady state friction coefficients recorded at 2 minutes were ~0.015 at the first cycle rising to an equilibrium value of ~0.135 by the fourteenth cycle, although for test 5A the results were conspicuously rather erratic. It should be acknowledged that there was a possibility that the friction values for this cartilage specimen had not attained equilibrium within the evaluated 20 loading cycles and may have risen still higher for further loading cycles.



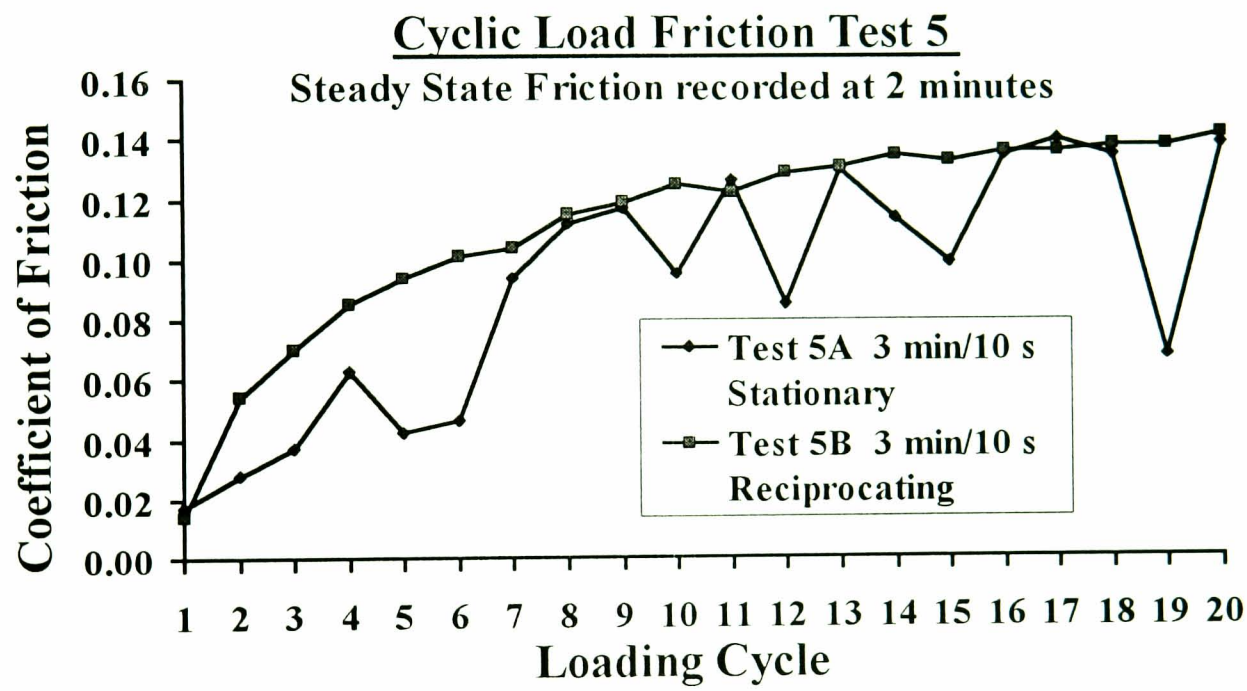
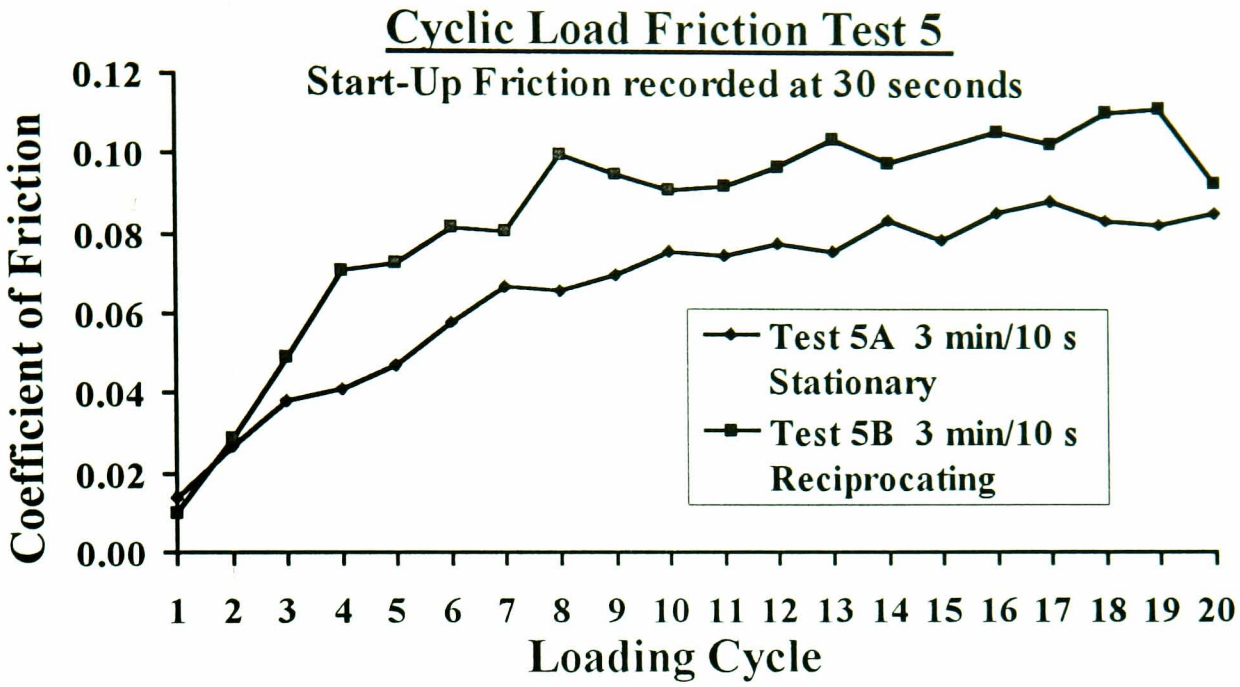


Figure 8-8 Steady state coefficient of friction results sampled at 30 seconds (top graph) and 2 minutes (bottom graph) during the loading phase for Tests 5A and 5B. The adopted loading cycle for these two tests was the same. The difference was friction measurements for Test 5A were sampled after periods of stationary loading whereas for Test 5B friction measurements were taken at set intervals during reciprocating motion.



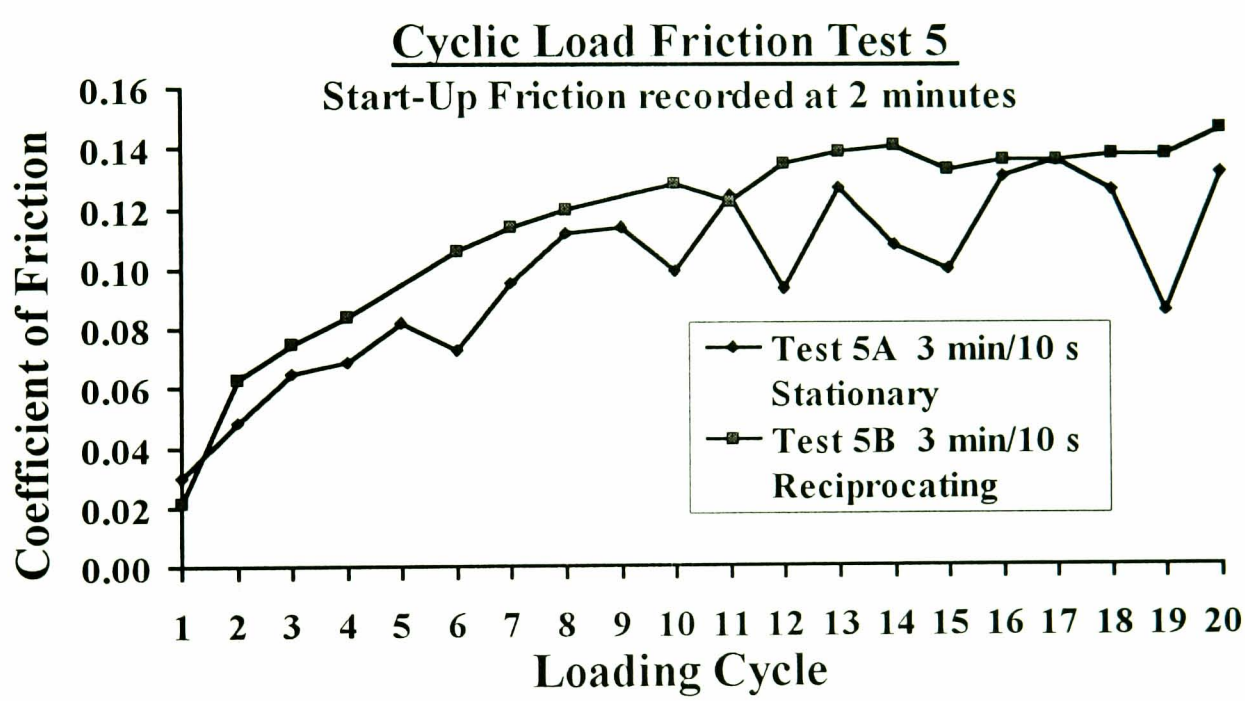
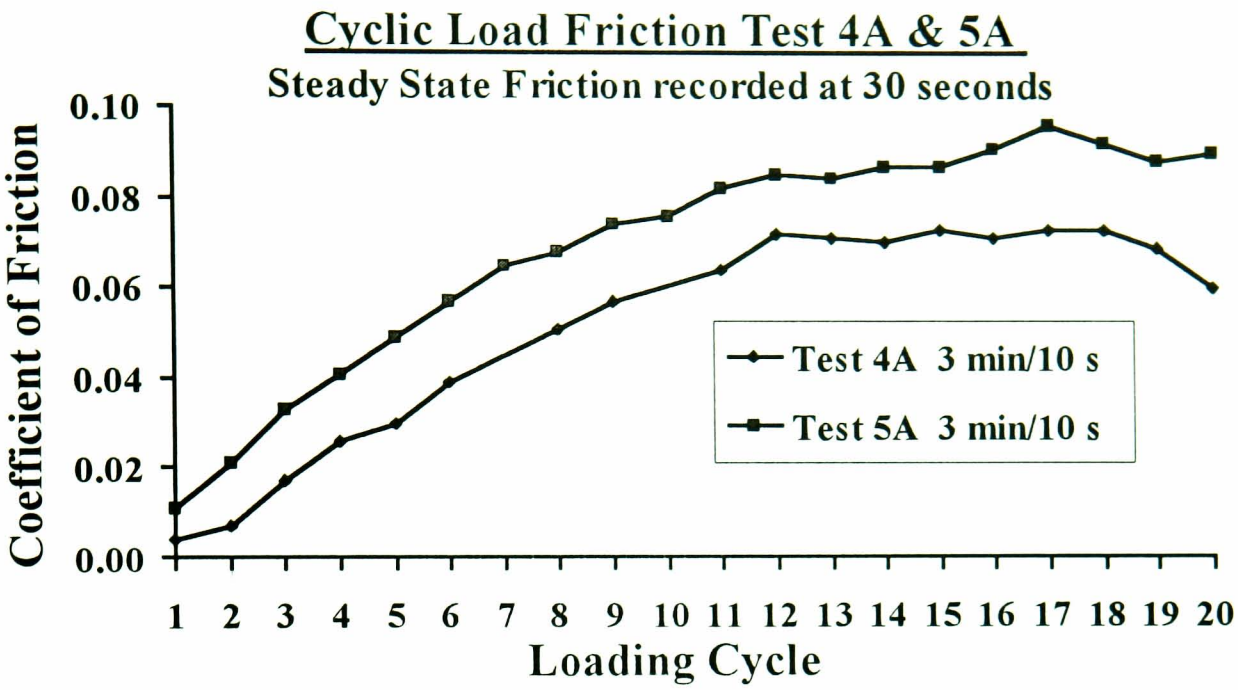


Figure 8-9 Start-up coefficient of friction results sampled at 30 seconds (top graph) and 2 minutes (bottom graph) during the loading phase for Tests 5A and 5B. Note that 'start-up' friction for the reciprocating motion Test 5B was the recorded friction peak occurring as the direction of sliding changed and is not the conventional start-up friction taking place after periods of stationary loading.

8.3.6 Assessment of Cyclic Loading Friction Test Reproducibility

The experimental conditions of tests 4A and test 5B, having the same 3 minute loading/10 seconds unloading cycle, were identical, except for the use of different cartilage specimens. In order to estimate the reproducibility of the results for these cyclic loading friction tests the results from these two tests have been plotted together, Figure 8-10 and Figure 8-11. The friction coefficients recorded at 30 seconds illustrated a very pleasing level of reproducibility, Figure 8-10. The friction coefficients recorded at 2 minutes again demonstrated good reproducibility for the start-up values, but less so for the steady state values, Figure 8-11. This was due to the unusually low steady state values relative to the start-up values evident at 2 minutes for test 4A. The steady state and start-up values for test 5A were, as commonly found, quite similar.



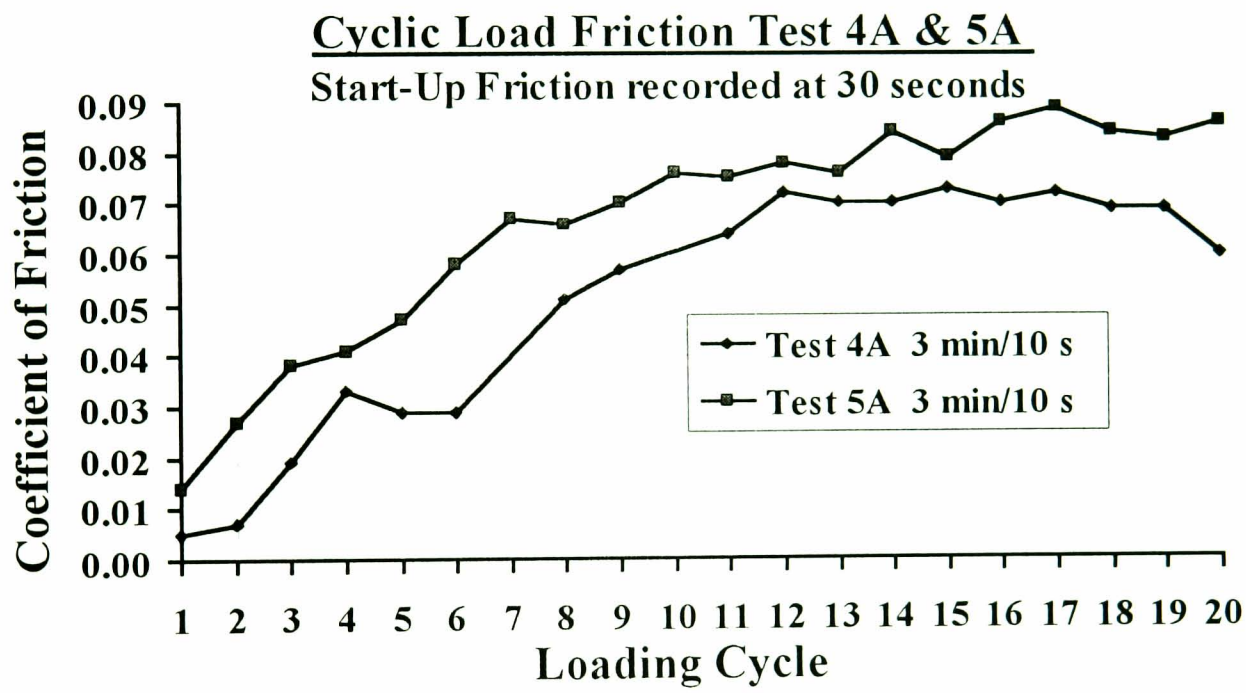
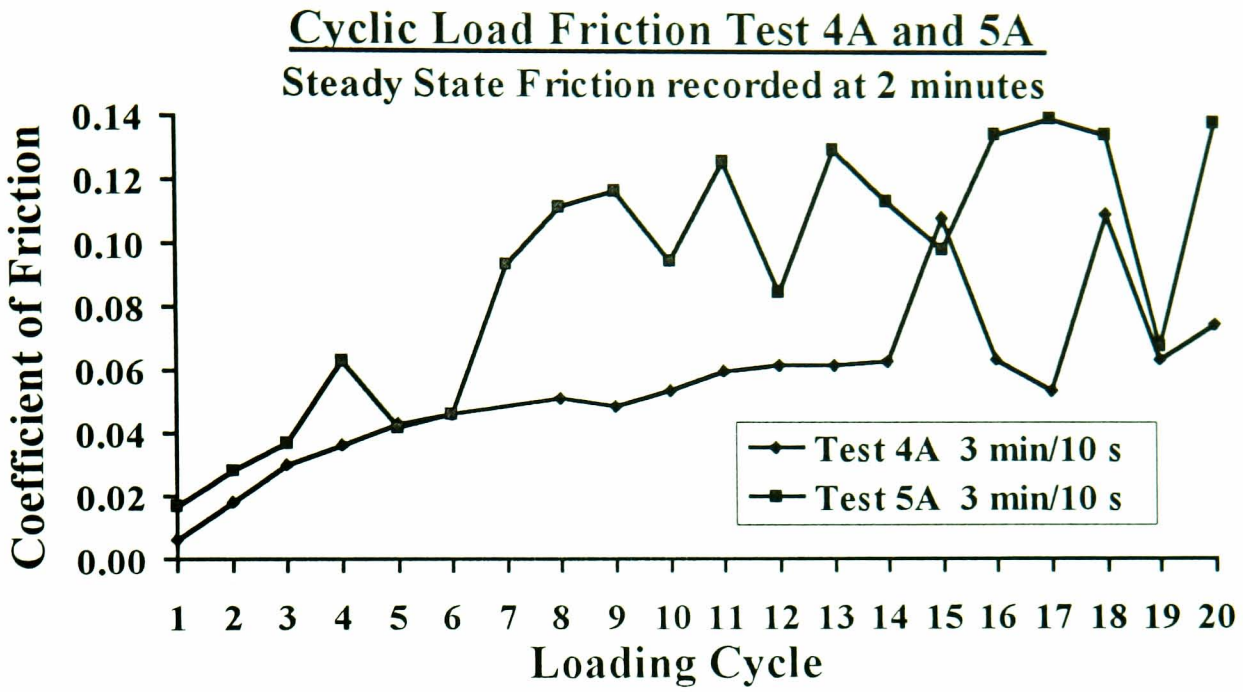


Figure 8-10 Steady state (top graph) and start-up (bottom graph) coefficient of friction results sampled at 30 seconds during the loading phase for Tests 4A and 5A. This data has been compared as it was recorded under identical experimental conditions but using different cartilage specimens.



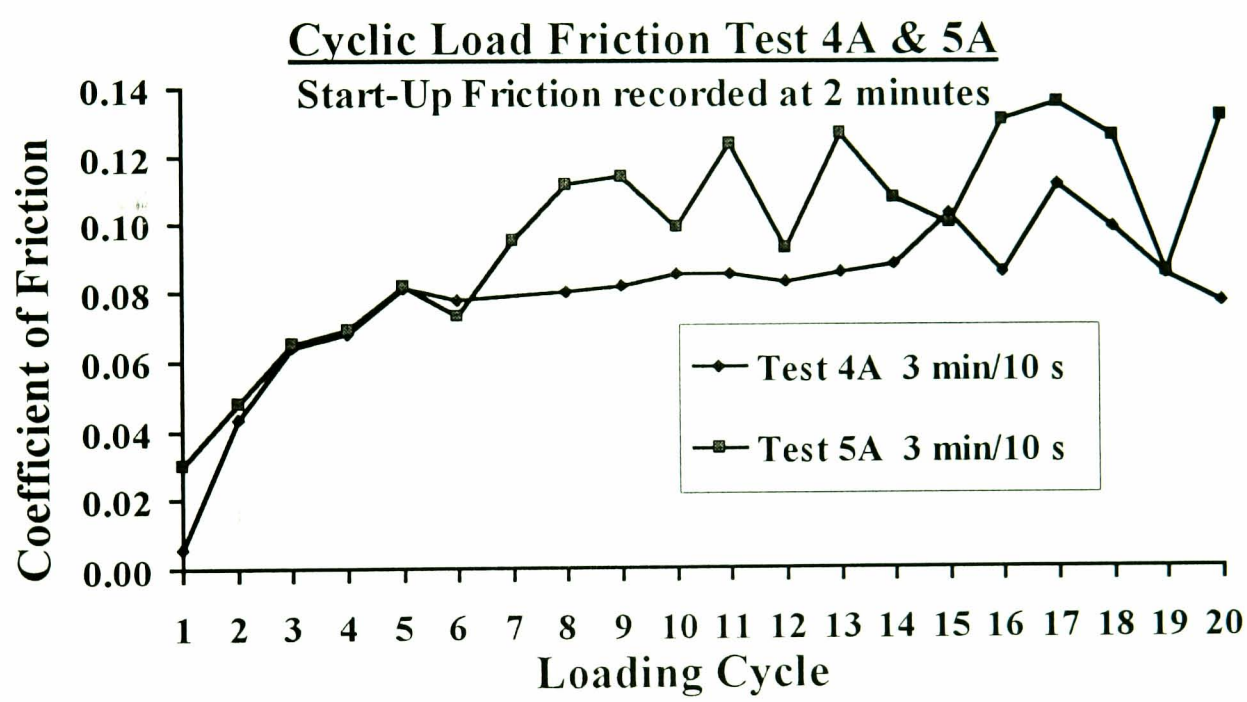


Figure 8-11 Steady state (top graph) and start-up (bottom graph) coefficient of friction results sampled at 2 minutes during the loading phase for Tests 4A and 5A. This data has been compared as it was recorded under identical experimental conditions but using different cartilage specimens.

8.3.7 Initial and Equilibrium Friction Coefficient Graphs for Tests 1-5

Figures 8-1 to 8-11 all plotted the raw data friction coefficient results for tests 1-5. This data format while providing a good illustration of how the friction levels were rising per loading cycle required summarising to allow easier interpretation. In order to accomplish this the initial loading cycle and approximated equilibrium friction coefficients were plotted in separate graphs for tests 1-5; refer to Figure 8-12, Figure 8-13, Figure 8-14, Figure 8-15 and Figure 8-16 respectively. These figures only display the steady state results, as the start-up results were generally very similar. These results for steady state friction were therefore chiefly addressed in the discussion.

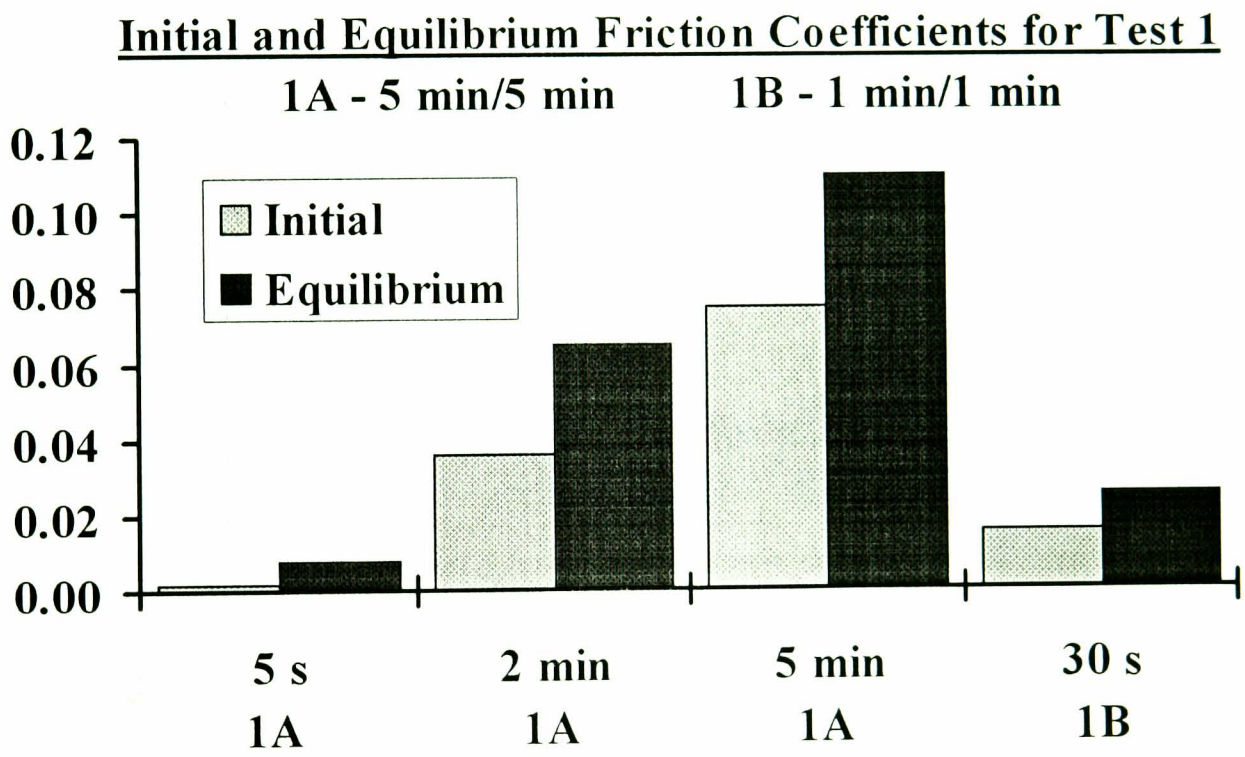


Figure 8-12 Initial and equilibrium friction coefficients for cyclic loading friction test 1. The friction sampling times for tests 1A and 1B are indicated on the x axis. The adopted loading time/unloading time cycles for tests 1A and 1B are also shown at the top of the figure.

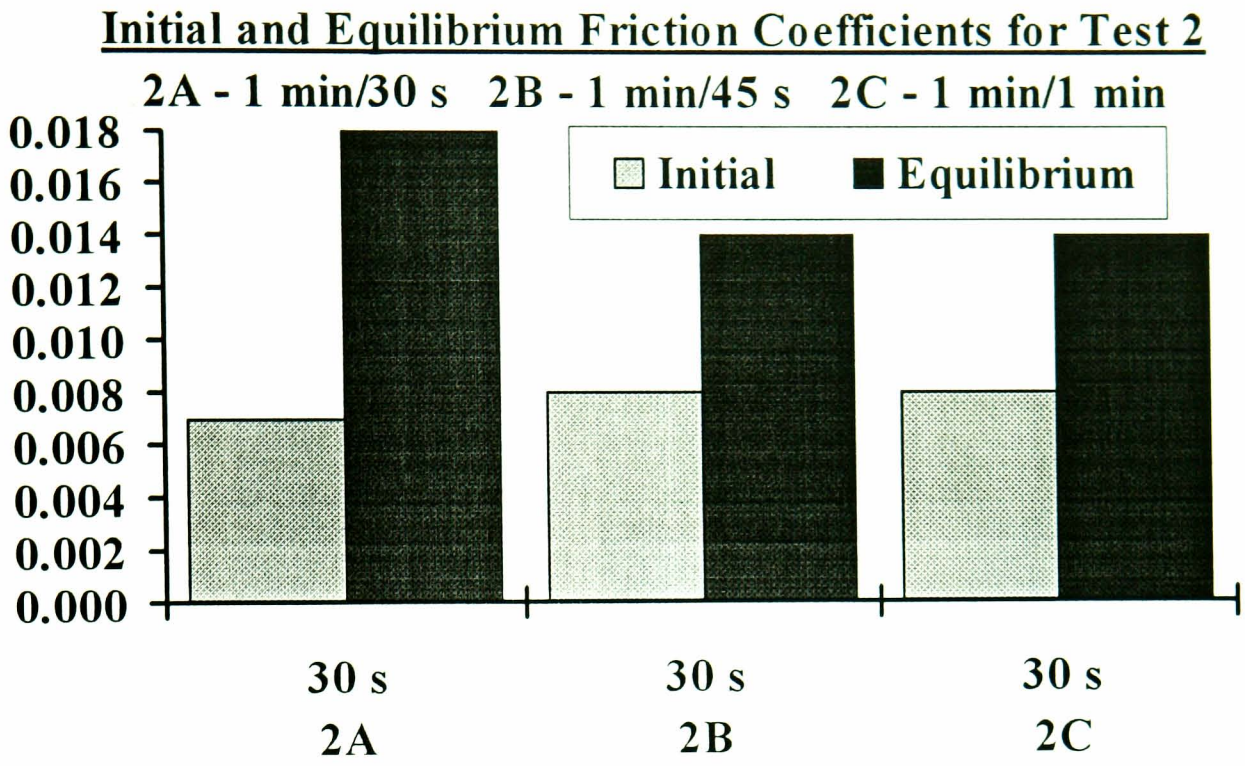


Figure 8-13 Initial and equilibrium friction coefficients for cyclic loading friction test 2. The friction sampling time for test 2 is indicated on the x axis. The adopted loading time/unloading time cycles for tests 2A, 2B and 2C are also shown at the top of the figure.

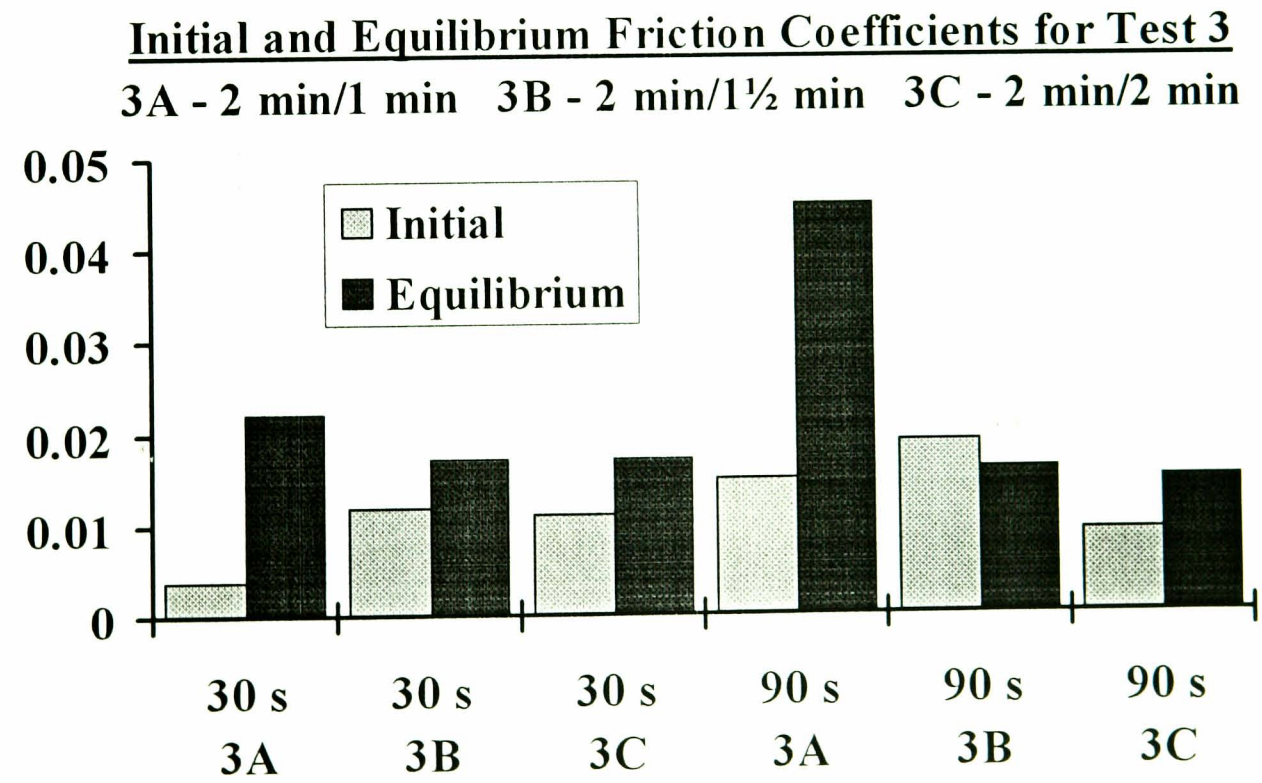


Figure 8-14 Initial and equilibrium friction coefficients for cyclic loading friction test 3. The friction sampling times for test 3 are indicated on the x axis. The adopted loading time/unloading time cycles for tests 3A, 3B and 3C are also shown at the top of the figure.

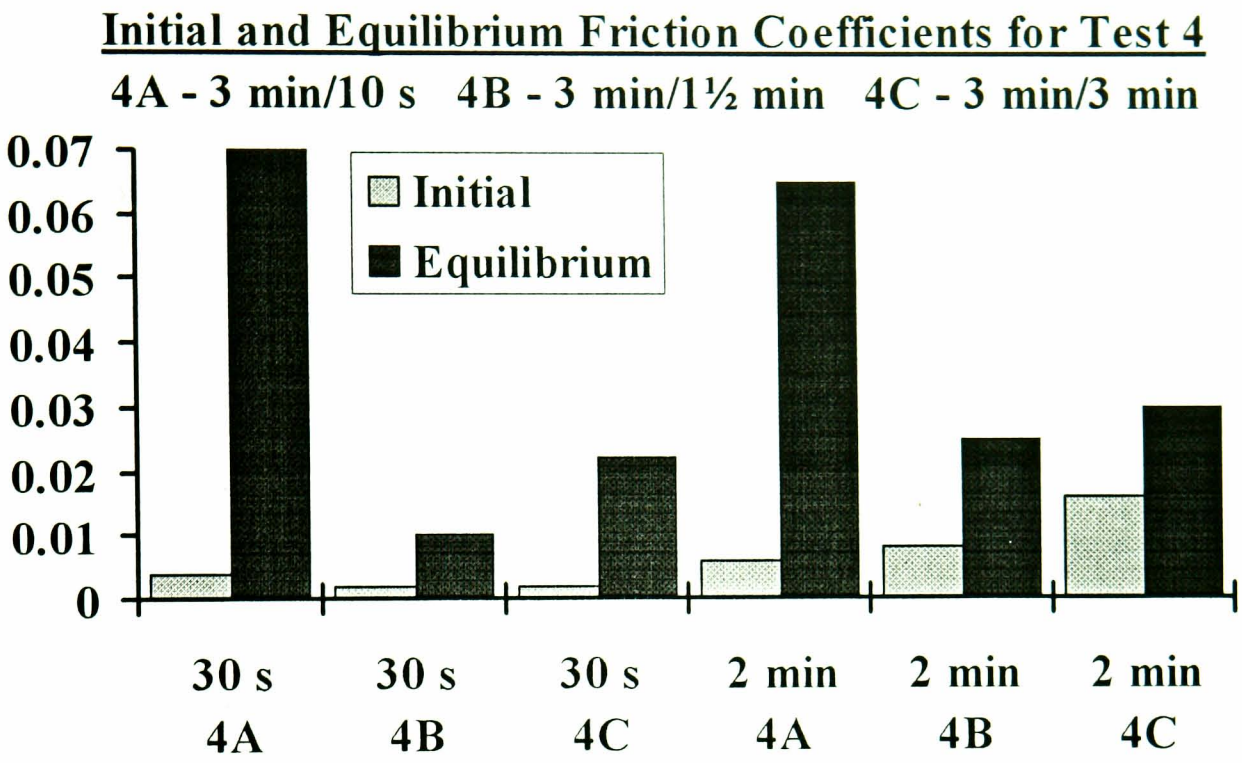


Figure 8-15 Initial and equilibrium friction coefficients for cyclic loading friction test 4. The friction sampling times for test 4 are indicated on the x axis. The adopted loading time/unloading time cycles for tests 4A, 4B and 4C are also shown at the top of the figure.

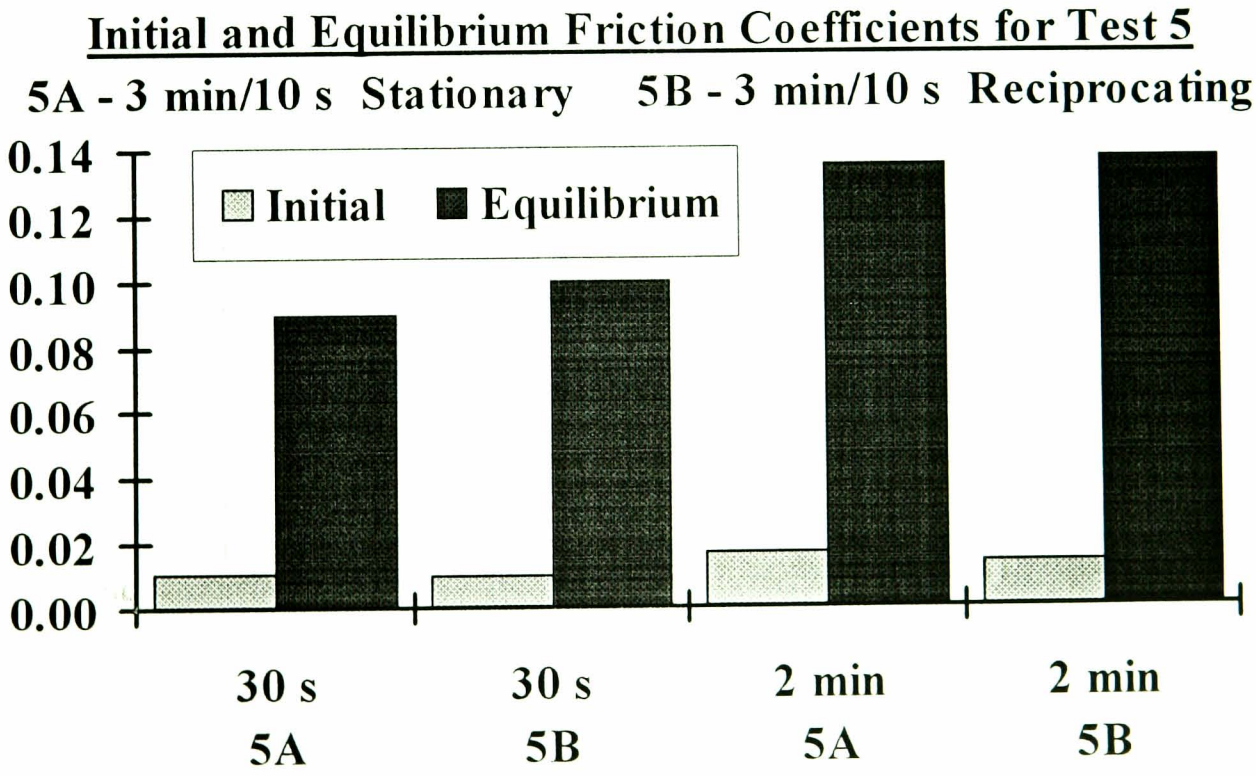


Figure 8-16 Initial and equilibrium friction coefficients for cyclic loading friction test 5. The friction sampling times for test 5 are indicated on the x axis. The adopted loading time/unloading time cycles for tests 5A and 5B are also shown at the top of the figure.

8.4 Discussion

8.4.1 Repeatability of friction readings for a particular cartilage specimen and Reproducibility of friction readings between different cartilage specimens

As only one cartilage specimen was tested for each of tests 1-5 it was not thought appropriate to compare in too much depth the results for the various tests. In order to do that properly would have required testing around seven cartilage specimens for each of these tests. However this type of comparison was not the main aim of the study. The cyclic load friction tests were specifically designed to address how the friction coefficients altered for the various loading cycles applied within a particular test for an individual cartilage specimen. Tests 2-4 in particular were experimentally set-up with that purpose in mind.

Tests 4A and 5A were however worthy of comparison as they were conducted under identical test conditions, except for the use of different cartilage specimens, Table 8-1. This permitted a qualitative inspection of reproducibility to be undertaken by plotting the results of the two tests together, Figure 8-10 and Figure 8-11. The friction coefficients recorded at 30 seconds illustrated a very pleasing level of reproducibility, Figure 8-10. The friction coefficients recorded at 2 minutes again demonstrated good reproducibility for the start-up values, but less so for the steady state values, Figure 8-11. This was due to the unusually low steady state values relative to the start-up values evident at 2 minutes for test 4A. The steady state and start-up values for test 5A were, as commonly found, quite similar.

The friction coefficient results of test 5A in relation to test 5B, Figure 8-16, revealed the strong level of repeatability obtainable for these cyclic load friction tests, for a particular cartilage specimen. The small variance between tests 5A and 5B could be attributable to one or other or in part both, of the intrinsic variance of friction readings for the cartilage specimen and any small changes upon friction readings due to the use of a stationary loading or reciprocating motion configuration during the testing. Therefore test 5 revealed good

repeatability of friction coefficients and that the use of a stationary loading or reciprocating motion configuration caused little difference upon friction readings. The slight increases for the reciprocating motion test may have also been due to superficial cartilage wear (Lipshitz and Glimcher, 1979; Stachowiak et al., 1994), or possible degradation as it was conducted after the stationary load test.

In a way the cyclic load friction tests for equal unloading/loading times by repeatedly taking friction measurements at a given loading time were also repeatability assessments. Any substantial changes in the friction coefficients, beyond accepted levels of friction variability, were caused by the build up effect of fluid loss caused by continuously applied loading cycles, as described below, (section 8.5, page 258).

8.4.2 Analysis of Tests 1-5

8.4.2.1 Test 1

For test 1A, which applied a 5 minute loading/5 minute unloading cycle, the friction coefficients sampled at 5 seconds, 2 minutes and 5 minutes all increased over the evaluated 12 loading cycles, Figure 8-12. For the 5 second loading time the increases were within the established level of variability for any particular cartilage specimen. For the 2 and 5 minute results however the initial friction coefficients of 0.036 and 0.075 rose to equilibrium values of 0.065 and 0.110. These equilibrium values were well above and outside the range of the mean results obtained for the equivalent Ringer's solution reciprocating motion cartilage plug on metal counterface tests of the previous chapter, being 0.035 (±0.012) and 0.069 (±0.019) for the 2 and 5 minute loading times respectively. For the 1 minute loading/1 minute unloading cycle, test 1B, the friction coefficient sampled at 30 seconds rose from 0.015 initially to an approximate equilibrium value of 0.025. Therefore despite being subject to an equal loading/load removal cycle, (although as mentioned the loading cycle for test 1A was closer to a 5 minute 20 seconds loading/4 minute 40 seconds unloading), there were substantial increases in friction occurring due to the repeated subjection of the particular loading cycle for tests 1A and 1B.

8.4.2.2 Test 2

The friction readings taken at 30 seconds during the loading phase for each of tests 2A, 2B and 2C were initially very similar, Figure 8-13. Upon subsequent loading cycles test 2A, having the shortest unloading phase of 30 seconds, did however attain the highest equilibrium value of 0.018. The 1 minute loading/45 seconds unloading and 1 minute loading/1 minute unloading cycles (tests 2B and 2C) both attained an equilibrium friction coefficient of 0.014. Thus, all these tests had an equal 1 minute loading phase and as might be expected the 1 minute loading/30 seconds unloading cycle of test 2A produced the highest equilibrium value, with the 1 minute loading/45 seconds unloading and 1 minute loading/1 minute unloading cycles (tests 2B and 2C) producing similar results to one another.

For test 2A the increase in friction coefficient from 0.007 to ~0.018 over the assessed 20 loading cycles (total test duration of 30 minutes) was still only a very modest rise. This was an important observation considering the loading phase time was twice that of the unloading phase time.

8.4.2.3 Test 3

For the 30 second friction coefficients test 3A was found to attain a moderately higher equilibrium value than those of test 3B and 3C, Figure 8-14. For the 90 second friction coefficients the effect of the 2 minute loading/1 minute unloading cycle of test 3A became much more apparent compared to the 2 minute loading/1½ minute unloading and 2 minute loading/2 minute unloading cycles of tests 3B and 3C. For test 3A the equilibrium friction coefficient of 0.045 was considerably higher than its own initial value and again considerably higher than the equilibrium values of tests 3B and 3C at ~0.015.

So for test 3 the 2 minute loading/1½ minute unloading and 2 minute loading/2 minute unloading cycles (tests 3B and 3C) had little effect on the friction coefficients. Although increases were evident they were within the limits of friction variability for a particular cartilage specimen, Figure 8-4 and Figure 8-5.

Moreover, even the 90 second results showed little increase compared to the 30 second results. The 2 minute loading/1 minute unloading cycle (test 3A), however, demonstrated an influence on the friction readings at the 30 second friction sampling time and even more so at the 90 second friction sampling time.

8.4.2.4 Test 4

For the especially harsh 3 minute loading/10 seconds unloading cycle of test 4A, substantial increases from the first loading cycle friction coefficients at 30 seconds occurred, reaching an equilibrium value of 0.070. This equilibrium value was much higher than those for tests 4B and 4C, subject to far less severe loading regimes. The tests 4B and 4C equilibrium values were, even so, reasonably higher than their initial values. For the friction measurements taken at 2 minutes the values and general trends were very similar, Figure 8-15, especially for test 4A with the test 4B and 4C results being only slightly higher. Test 4B often exhibited lower friction coefficients than test 4C, but test 4C experienced a longer unloading phase time. Such discrepancies can be attributed to the testing sequence were factors such as wear and degradation may have played a part in increasing friction levels of the latter test 4C. By the start of test 4C it should be acknowledged that the cartilage specimen had already been in use for at least 5 hours on the friction apparatus for tests 4A and 4B.

8.4.2.5 Test 5

In test 5 the influence of the 3 minute loading/10 second unloading cycles upon friction readings was very obvious, for both a stationary (test 5A) and reciprocating motion (test 5B) test protocol, Figure 8-8 and Figure 8-9. For both tests 5A and 5B the first loading cycle friction coefficients were ~0.010 at 30 seconds and ~0.016 at 2 minutes rising to respective equilibrium values of ~0.095 and 0.136, Figure 8-16. These values were well outside the range of values recorded at 2 minutes for the respective equivalent stationary and reciprocating motion Ringer's solution 9 mm cartilage plug on metal counterface tests under constant loading. Even so the test duration was over 1 hour and if the 10 second load removal intervals had not been applied the expected friction coefficients would have been ~0.300-0.400. So even for this severe loading cycle the short 10

second unloading phase was clearly maintaining friction levels well below what would have been found after 1 hour of constant loading.

So once again this 3 minute loading/10 second unloading cycle, as expected, was shown to produce prolific increases in friction levels over the assessed 20 loading cycles. Test 5 was conducted in order to quantify the level of repeatability for a particular test and also ascertain the effect of reciprocating motion upon friction coefficients in relation to the standardly adopted stationary loading test procedure for the cyclic loading friction tests, Table 8-1. The 3 minute loading/10 second unloading cycle was chosen as it had already produced dramatic changes in friction coefficients (test 4A) and by using the same loading cycle adopted for a previous test (test 4A) a perception of reproducibility could also be obtained. Both repeatability within a test for the same cartilage specimen and reproducibility comparing results using different cartilage specimens for these cyclic load tests have been referred to in this discussion. It must be kept in mind however that only a relatively small number of cartilage plugs were tested and the quantification of repeatability and reproducibility was therefore restricted. Although, from the limited comparisons made, the levels of repeatability and reproducibility for these tests were sufficiently favourable to draw meaningful conclusions from.

8.5 General Comments and Conclusion

- Equilibrium friction coefficients were generally always higher than the initial first loading cycle friction measurements. Although sometimes this was considered to be within the level of friction variability for the particular cartilage specimen being tested, (e.g. Test 1A at the 5 seconds friction sampling time).
- A key factor increasing friction coefficients was a shorter unloading time relative to the loading time.
- Even for equal unloading/loading time cycles the friction coefficients were found to increase over the assessed number of loading cycles, (e.g. Test 1A at the 2 and 5 minute loading times and Tests 1B, 2C, 3C and 4C).
- Although the friction coefficients always tended to increase over the evaluated number of loading cycles it was recognised that the friction levels, as they reached equilibrium, were being maintained at surprisingly low values throughout tests 1-5, considering the test duration for each particular test, Table 8-1. It would appear that, ignoring effects of degradation and wear, for a given cartilage specimen and applied loading cycle a certain level of friction coefficient can be maintained indefinitely at a remarkably low level. For example steady state friction coefficients recorded at 30 seconds never rose above 0.030. For test 4B, in particular, these friction values remained at ~0.010 even after 90 minutes of testing while subject to a 3 minutes loading/90 seconds unloading cycle. The only exception was the 3 minutes loading/10 seconds unloading cycle (test 4A and test 5). However, even for this most severe load cycle investigated, the 10 second unloading phase clearly sustained friction levels well below those expected after an equal test duration of 1 hour but with the application of a constant load instead.

The overall conclusion from these tests was that during cyclic loading of articular cartilage, involving episodes of complete load removal, friction levels can be maintained indefinitely at remarkably low levels. Only the repeated subjection to extremely harsh loading cycles was sufficient to increase the friction coefficients

to values of >0.05, at which point, physiologically for prolonged periods of application substantial wear within a synovial joint may begin to occur.

The reason for increasing friction levels over a given number of applied loading cycles was attributed to the failure of the cartilage plugs to maintain full hydration, even for cycles with equal loading/unloading periods. Therefore, upon subjection to more and more loading cycles the cartilage lost more and more interstitial fluid. This fluid loss, in accordance with the biphasic lubrication hypothesis, thereby decreased the amount of load that the fluid phase could potentially support and subsequently increasing the friction levels for further loading cycles as more load had to be carried by the solid phase of the cartilage.

This deduction can be backed up by referring to observations made by Maroudas (1980). In a section concerning the amount of fluid loss during cyclic loading Maroudas stated that the rate of recovery (rehydration) was determined by the swelling pressure, which, for small strains (i.e. small loading periods) was far lower than the applied physiological load. Maroudas further postulated "Thus, recovery from small deformations is far slower than the initial rate of deformation produced by the physiological load. Accordingly, recovery in a cycle is likely to be incomplete unless the 'off-load' intervals are considerably longer than the 'on-load' periods. Hence, although the amount of water loss per cycle is extremely small, the total deformation after a period of walking may not always be negligible". This statement supports the build up effect of increasing water loss over a number of cycles even for equal loading/unloading times. The subsequent influence on friction levels has been discussed.

8.5.1 Comparison of In Vitro Testing and In Vivo Physiological Loading Regimes

The 1 minute loading/1 minute unloading cycle was the shortest practical loading cycle that could be used for these cyclic load friction tests. This was still a considerably longer duration load cycle than occurs in synovial joints during walking and running (Paul, 1967). However as mentioned in previous chapters

numerous common day-to-day activities such as washing the car, operating machinery, presenting a lecture etc., involve periods when the articular cartilage will be subjected to significant loads with little or no motion occurring. Any slight movement of the joint will, however, cause the load to be transferred to adjacent areas for the opposing cartilage surfaces. Considering these conditions the loading cycles that the cartilage plugs were subjected to were a reasonable analogue to those that any particular load bearing region of a synovial joint may expect to encounter *in vivo*. Furthermore, for a specified joint loading cycle the loading/unloading cycle for a particular load bearing region of the cartilage surface will vary according to its anatomical position. During walking and running the unloading phase time for a given cartilage area will probably last longer than that of the loading phase, ensuring little overall fluid loss for the opposing cartilage surfaces. Thus, while harsh loading cycles are probably experienced from time to time within a synovial joint their perpetual application upon given contact areas of opposing articular cartilage surfaces leading to eventually high friction coefficients is thought to be unlikely.

It should however be kept in mind that running and walking are not daily activities, per se, in the same vein as washing a car, ironing, operating machinery, gathering food. Therefore it is likely that the synovial joint has evolved various lubrication mechanisms in tandem to counter the more obvious threat of wear during high mobility activities such as walking and running, and of perhaps equal importance countering the potential high friction levels after prolonged periods of stationary loading or heavy duty high load duration/short load removal cyclic loading commonly associated with many day-to-day activities. To date workers have generally only discussed synovial joint lubrication within the context of walking and running (Unsworth, 1993). It is, of course, important that we understand lubrication mechanisms during walking and running. However, it is estimated that approximately 90% of the time synovial joints undergo little motion and during these periods significant loading is frequently encountered (Forster and Fisher, 1996). The lubrication mechanisms which continue to provide low friction

and minimum wear under these perceivingly harsh tribological conditions have within this study been addressed and appreciated for their essential role.

9. Conclusions

9.1 Surface Analysis

9.1.1 Surface/Boundary Layer(s)

The *'surface layer'* (first referred to as the *lamina splendens*) is presumed to be a distinct ~0.1-3 μm thick acellular, noncollagenous layer, composed of interfibrillar proteoglycan ground substance, which covers even the uppermost superficial collagen fibrils and chondrocytes in its native state. In this respect it could be considered as merely an extension of the interfibrillar cartilage solid matrix, see Figure 3-20. A further *'boundary' layer* (≤1 μm) may well also be present, resting upon the surface layer, having perhaps a more transient or ephemeral presence. Although the boundary layer was still considered to be intrinsic to the cartilage surface. The boundary layer constituents (phospholipids, glycoproteins) being derived from the synovial fluid and cartilage matrix cellular metabolites. The surface layer probably functions to prevent fibrillation of the superficial collagen fibrils while the boundary layer in turn minimises direct cartilage-cartilage asperity contact friction by boundary lubrication. The boundary and/or surface layer may also possess biphasic lubrication properties, being able to both expand and collapse rapidly, as a result of fluid flow, independently of the underlying cartilage matrix. This may occur over short time scales of less than a few seconds. It is worthwhile to point out this is still only a considered opinion. Further analysis, for instance, may well conclude that the surface and boundary layers are one and the same.

9.1.2 Articular Cartilage Surface Roughness

For general analyses of synovial joint lubrication, e.g. for the prediction of full fluid film or mixed regime lubrication, it is still probably appropriate to ascribe a quantitative value of roughness of 1-2 μm, as determined by R_a calculations at 0.8 mm cut-offs. However this 'roughness' was attributed to the general form or waviness of the cartilage over the adopted 0.8 mm sampling length. A more realistic value of inherent cartilage surface roughness of ~0.1-0.3 μm was

provided from line profiles filtered at 0.08 mm cut-offs. Cartilage lubrication theories involving microscopic analysis of surface roughness within a mixed lubrication regime would perhaps be better suited to use this R_a (0.08) value. The two profilometry techniques used may have been insensitive to the surface/boundary layer. This layer(s) would presumably cover surface features creating an even smoother surface. However due to its presumed transient and highly compliant nature, significant compression during joint loading creating sub-surface asperities to be prominent will probably still occur.

9.1.3 Wear due to Reciprocating Motion and Implications for Boundary Lubrication

An increase in roughness of ~ 1 μm was identified by laser profilometry due to reciprocating motion. This increased roughness may well have been partially caused by the removal of the 'surface coat', commonly quoted as being ~1 μm thick. Furthermore, the small increases in friction coefficient between the *initial* and *repeat* values, in Chapter 7, may have been caused by the removal of the 1 μm thick surface layer associated with boundary lubrication, during loaded reciprocating motion, and not simply due to the slight roughening of the compliant cartilage surface. Note that in these friction studies, for the *repeat* results the synovial fluid values were still statistically lower than those of the Ringer's solution. Therefore while substantial surface wear had occurred the boundary or boosted lubrication provided by the synovial fluid remained active.

It was difficult to speculate which of these two lubrication mechanisms the synovial fluid was providing. The removal of the uppermost surface, to a depth of ~1 μm, might indicate that the synovial fluid was supporting a boosted lubrication mechanism, dependent upon the HA constituent, in preference to boundary lubrication. This conclusion is however based on the assumption that boundary lubrication and boundary layer adherence was specific to the uppermost surface of the articular cartilage specimens, which may not necessarily have been the case.

From analysis of specimens previously used for stationary load friction testing no substantial wear was evident.

9.2 Friction Studies

The friction test protocol developed provided reliable friction data for cartilage/metal and cartilage/cartilage contacts within a mixed lubrication regime. From this consistency it was possible to publish the first detailed account seen in literature of cartilage/cartilage tribological behaviour (Forster and Fisher, 1996 - see appendix); for extracted 'small cartilage specimen' as opposed to entire joint friction studies.

9.2.1 Articular Cartilage's Loading Time vs. Friction Coefficient Relationship Biphasic Lubrication

Loading time, and subsequent load removal, has throughout the course of this project been shown to be a major determinant of friction for articular cartilage. The relationship between loading time and cartilage deformation/water content has long been recognised (Edwards, 1967). Therefore it has been reasonably argued that there exists a strong link between the fluid phase and friction coefficient of articular cartilage. The only theory put forward, to date, that could account for the relationship between water content and friction coefficient is McCutchen's weeping lubrication mechanism, (McCutchen, 1959 & 1962). However, the idea that fluid flows from the cartilage matrix and into the loaded contacting zones of the joints surfaces has been largely denounced, (Maroudas, 1980; Hou et al., 1992; Jin et al., 1992) and the general view is that fluid flow occurs away from the contact zones and loaded cartilage regions, and into the unloaded areas of the cartilage layers and joint capsule. However, even recent friction studies of synovial joint lubrication have, upon consideration of the time-dependent changes in cartilage exudation and friction, concluded that the weeping mechanism is a dominant feature in joint lubrication (Nickel and McLachlan, 1994; Ikeuchi et al., 1994).

The biphasic lubrication mechanism has been proposed as a rational explanation of the relationship between the fluid content and tribological behaviour of cartilage. This mechanism is in keeping with the broadly held opinion that fluid flow occurs

away from the contact zone and loaded areas of the tissue. It was first proposed by Forster et al. (1995), see appendix.

The most important consideration is how the load is carried or transferred in the biphasic cartilage. Upon initial loading a large amount of load is carried in the fluid phase of the cartilage (Ateshian et al., 1994) and it has been speculated that the proportion of load carried by the fluid phase contributes little to the total or aggregate friction force of the two phases. As the loading period is increased the load carried by the fluid phase decreases and that carried by the solid phase increases. Hence, the overall friction and friction coefficient increases. An expression has been derived by the author to account for this phenomenon, (see section 4.4.3, page 140).

It is very possible that even during walking and running in synovial joints of the lower limb asperity contacts may occur which are lubricated by the mechanisms referred to in section 9.2.1.1 below, of which biphasic lubrication is believed to be the major element.

9.2.1.1 Reappraisal of Mixed Lubrication Regime

Biphasic lubrication for well hydrated cartilage surfaces is considered to be of prime importance in maintaining low friction coefficients within a mixed lubrication regime. Biphasic lubrication was chiefly responsible for the low friction coefficients which were found when recording friction of cartilage specimens with a high water content. In synovial joints boundary lubrication is also active, as is possibly boosted[1] lubrication. However, without sufficient load carriage by the fluid phase, μ values were found to rise well above 0.3. Therefore, despite the presence of boundary (and boosted) lubrication, if the cartilage layer contacts become depleted of their high water content friction values can rise to what could be deemed catastrophic levels. In Figure 9-1 the lubrication mechanisms operating in a mixed lubrication regime have been depicted. This schematic diagram is intended to update the conventional view of cartilage lubrication in the mixed regime as illustrated in Figure 1-5. Compression of particularly prominent asperities may also occur and perhaps contribute to a localised weeping lubrication mechanism. Thus, micro-EHL may well have a role to play within a mixed lubrication regime, in combination with a 'micro-weeping' mechanism (although the net flow of fluid is still directed away from the load region).

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[1] The size alone of the HA macromolecule could be large enough to separate the cartilage surfaces within the fluid pockets without the need for a time consuming filter mechanism to be relied upon (Ogston and Stanier, 1951; Geborek and Wollheim, 1993).

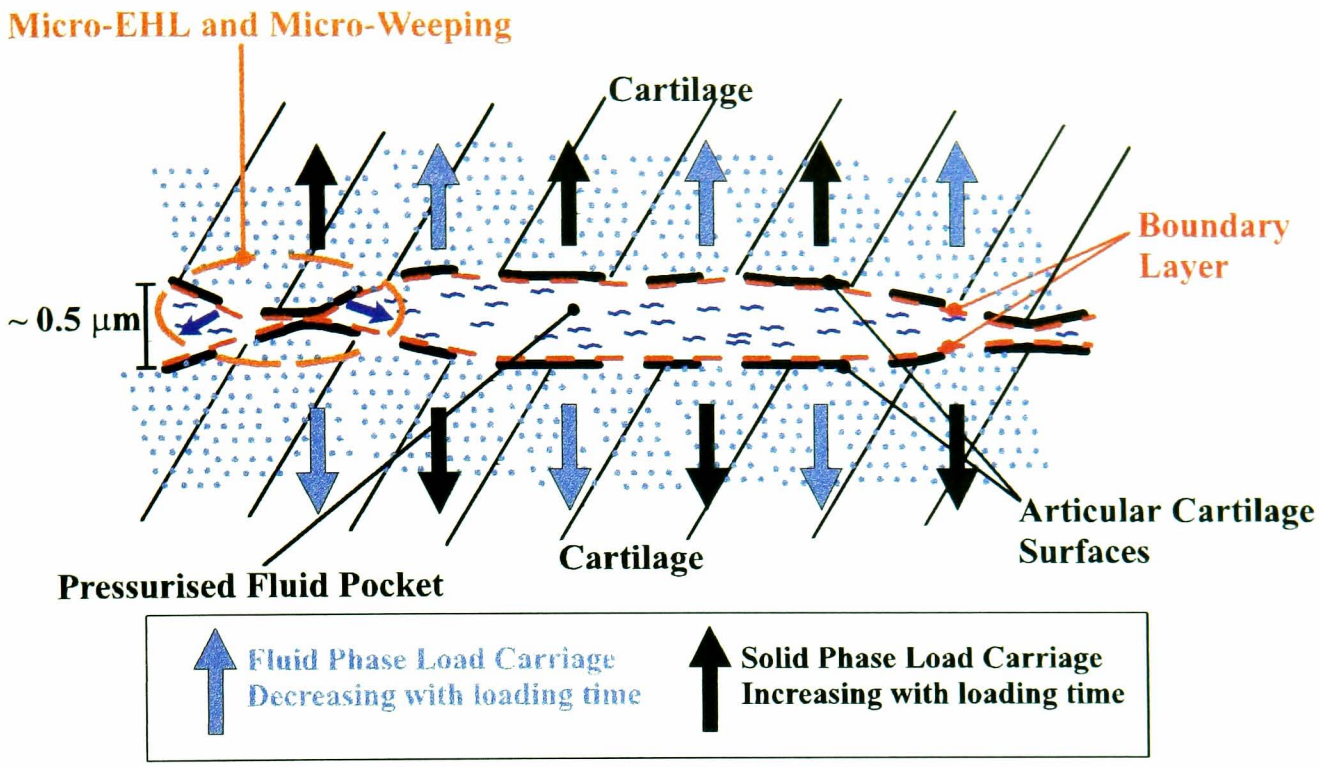

Figure 9-1 Biphasic lubrication operating within a mixed lubrication regime. Note that other lubrication mechanisms are also playing a role but ultimately if the opposing cartilage surfaces become substantially depleted of fluid then, irrespective of the other mechanisms, μ values will become very high.

The fluid phase load carriage mechanism in reducing friction could be specific to a certain region of the cartilage, e.g. the superficial tangential zone. In other words, it may not be a bulk tissue response. It is also quite possible that the surface or boundary layer acts independently from the cartilage matrix in affording biphasic lubrication to the cartilage surface, (Kobayashi et al., 1996).

9.2.2 Synovial Joint Lubrication, Wear and Osteoarthritis

It would appear from the friction experiments conducted that in a synovial joint there exist only two ways of producing high levels of friction. Firstly, by constant stationary loading, however prior to motion the joint surfaces may be momentarily separated to re-introduce a fluid film (e.g. after a period of standing shifting weight onto one leg while the other leg moves). Furthermore, upon motion high friction will only be experienced as instantaneous 'start-up' as the contact areas change on both the opposing cartilage surfaces, reducing friction as the contact moves onto hydrated cartilage layers. Secondly, the repeated application of harsh loading/unloading cycles on the same opposing contact areas of the joint could

take place, depleting the high water content of the cartilage layers and lead to high friction. However, although severe loading cycles, with only very short load removal periods, may be encountered from time to time their repeated application is thought to be unlikely.

The friction coefficient magnitude for permanent wear of synovial joints to commence *in vivo* is not known. Indeed for a healthy synovial joint irreparable wear due simply to high μ levels may well not occur *in vivo*. Even so from this study it would seem that sustained high friction levels in synovial joints are highly unlikely. This conclusion is primarily due to the fact that the region of load transmission continuously changes due to combined sliding and rolling, and the varying degrees of incongruence between the mating surfaces of synovial joints. The sliding and rolling articulation, thus, ensures that the hydrational state of the cartilage layers is well balanced and maintained at high levels throughout as any period of load application will be exceeded by the period of load removal. This statement is further confirmed by the fact that entire joint friction studies have all reported very low friction coefficients, ~ <0.03, while the small cartilage specimen friction studies varied from ~0.003 up to often as high as 0.30 and higher (see Tables 1-2 & 1-3). The latter results are believed to be due to the constant loading, depleting fluid content, which was applied to the cartilage specimens during friction measurement.

For permanent wear and the potential onset of osteoarthritis (OA) to result from catastrophically high friction levels one of the following factors must play a role.

i. Biochemical changes in the cartilage (e.g. proteoglycan loss diminishing the capability of the tissue to retain fluid).
ii. Prolonged subjection to unfamiliar loading patterns causing matrix fatigue and again diminishing the capability of the tissue to retain fluid under load.
iii. An episode of trauma leading to cartilage defects or alteration of the general joint physiological kinematics in such a way as to cause high loading on a particular cartilage area for extended periods.

The overall conclusion is that sufficiently high friction levels to cause any considerable wear in healthy synovial joints are not attained.

This finding is in disagreement with recent statements produced in the literature (Batchelor and Stachowiak, 1996). Batchelor and Stachowiak determined that substantial wear of synovial joints occurs *in vivo*. Metabolic feedback processes were consequently deemed to exist to control this wear. The authors went further to state that instabilities in the feedback processes were a basic cause of arthritis. However this statement was based on findings from their previous work (Stachowiak et al., 1994) which considered that lubrication in the mixed regime was largely provided by the boundary layer which quickly rubbed off during testing. The wear removal of the boundary layer was therefore accredited with increasing friction with loading time. This finding itself is believed to be misconceived, (Forster and Fisher, 1996). The requirement for a rigorous metabolic 'feedback system' (aside from the normal metabolic requirements of any living tissue) to continuously revitalise the cartilage surfaces is not therefore regarded as necessary in the first instance.

It is recognised that the aetiology of OA is a highly involved, complex and contentious issue. This section has hopefully provided some further incite.

9.2.3 Comparison of Friction Test Methodology with In Vivo Synovial Joint Physiological Function

The stationary load friction tests were harsh loading regimes in physiological terms. They did however show beyond doubt that the friction coefficient was strongly related to cartilage deformation/water content. The possibility of a wear component influencing friction levels was eliminated by adopting this test procedure. Entire joint friction studies could not have revealed this link between water content and friction due to the changing contact areas and thereby cyclic loading that the opposing cartilage surfaces are subjected to. Indeed it is this cyclic loading of the cartilage layers, both in joint friction studies and synovial

joints *in vivo*, which accounts for the maintenance of well hydrated cartilage surfaces and low friction levels.

The cyclic load friction experiments conducted in Chapter 8 were the closest approximation to loading cycles that articular cartilage may encounter *in vivo*. These tests illustrated the maintenance of low friction for adequate load application vs. load removal periods. The 1 minute loading/1 minute unloading cycle was the shortest practical loading cycle that could be used. This was still a considerably longer duration load cycle than occurs in synovial joints during walking and running (Paul, 1967). However numerous common day-to-day activities such as washing the car, operating machinery, presenting a lecture etc., involve periods when the articular cartilage will be subjected to significant loads with little or no motion occurring. Any slight movement of the joint will, however, cause the load to be transferred to adjacent areas for the opposing cartilage surfaces. Considering these conditions the loading cycles that the cartilage plugs were subjected to were a reasonable analogue to those that any particular load bearing region of a synovial joint may expect to encounter *in vivo*.

It should be kept in mind that running and walking are not daily activities, per se, in the same vein as washing a car, operating machinery, gathering food. Therefore it is likely that the synovial joint has evolved various lubrication mechanisms in tandem to counter the more obvious threat of wear during high mobility activities such as walking and running, and of perhaps equal importance countering the potential high friction levels after prolonged periods of stationary loading or heavy duty high load duration/short load removal cyclic loading commonly associated with many day-to-day activities. To date workers have generally only discussed synovial joint lubrication within the context of walking and running (Unsworth, 1993). It is, of course, important that we understand lubrication mechanisms during walking and running. However, it is estimated that approximately 90% of the time synovial joints undergo little motion and during these periods significant loading is frequently encountered (Forster and Fisher, 1996). The lubrication mechanisms which continue to provide low friction and

minimum wear under these perceivingly harsh tribological conditions have within this study been addressed and appreciated for their essential role.

In this study, as in many biomechanical studies, articular cartilage samples were extracted from the distal end of bovine femurs. It is believed that healthy human and bovine cartilage, of similar thickness, will behave in much the same way, having similar biphasic / biomechanical properties and composition (Athanasiou et al., 1991).

9.2.4 Articular Cartilage - The Bearing Material

Articular cartilage, as a bearing material, possesses some unique properties which sets it aside from traditional tribological engineering principles. It is important to reflect upon them as they have often been ignored or unacknowledged by previous workers. They, further, allow for a more considered interpretation of tribological studies investigating synovial joints.

i. The boundary lubricant is probably an inherent feature of the articular cartilage surface.

ii. In order to establish a boundary regime, whereby the expression μ=F/W remains constant, an equilibrium condition whereby $p_{applied} = p_{swelling} - p_{elastic}$ and $\Delta p = 0$, must first be achieved. If cartilage asperity contacts have been established and cartilage exudation is still occurring then the expression μ=F/W will change according to loading time and the adopted loading regime; a mixed lubrication regime therefore predominates. Hence, for a boundary regime to exist in synovial joints, whereby the expression μ=F/W remains constant, full 'wring out' of the opposing cartilage layers is required. Under physiological loading regimes, a true boundary regime, is therefore highly unlikely to exist in any circumstances (except instantaneously at the start-up of motion after a long period of stationary loading). Biphasic lubrication theory corrects the μ=F/W expression by adding a time dependency function.

iii. Friction coefficients of magnitudes of ≤ 0.03 may still be considered to occur within a mixed lubrication regime and not necessarily full fluid film. Previous workers (Little et al., 1969; Higaki et al., 1995), having initially recorded μ

values of ~0.010 in synovial joints, chemically removed the boundary layer. Subsequent measured values of μ, under identical conditions, rose to ~0.020-0.030. These studies would suggest that asperity contacts were occurring, to some extent, during recording of the initial readings. The μ values of ~0.020-0.030 were due to the loss of the boundary layer which protected the asperity contacts. Note that these friction coefficient values of ≤0.03 were still very low even after the removal of the boundary lubricant. Another lubrication mechanism was therefore still in operation and, from observations made in this work, was assumed to be biphasic lubrication.

9.2.5 Project Overview

The uniqueness and original approach of this project can be briefly summarised by the following points.

- The lubricating ability of synovial fluid was objectively assessed by comparison to a control Ringer's solution under known and widely varying tribological conditions. In the majority of these friction studies the test variables were always carefully controlled and at least seven specimens for every test condition was examined to ascertain the reproducibility of the results and allow for statistical comparison of the data.
- In the discussion and interpretation of this work detailed attention has been devoted to four important aspects concerning synovial joint boundary and mixed lubrication, namely, i. tribology; ii. biochemistry; iii. the articular cartilage surface; iv. biphasic properties of articular cartilage. Previous workers have always ignored at least one or more of these vital components in the understanding of articular cartilage lubrication.
- The need to remain as close as possible to synovial joint physiological conditions while, at the same time, trying to limit and control the number of test variables, unknowns and sources of error and optimise the reliability of the results were constantly of paramount importance when designing the friction experiments.

9.3 Summary

The major findings of this research are:

- The carriage of load by the fluid phase of articular cartilage is believed to be a key element reducing friction for mixed lubrication regimes. It has been named *"Biphasic Lubrication"*, in respect of work conducted by Mow and co-workers in modelling the biomechanics of this tissue. In view of the normal physiological kinematics of healthy diarthrodial joints, during day-to-day activities, which permit the cartilage layers to remain well hydrated, the occurrence of high, sustainable coefficients of friction, leading to cartilage wear and the potential onset of arthritis is deemed unlikely.

- The synovial fluid, under certain test conditions, was shown to provide superior lubrication than Ringer's solution. The synovial fluid is thought to replenish the cartilage's phospholipid/glycoprotein boundary layer and so aid *boundary lubrication*, and/or provide a *boosted lubrication* mechanism whereby the hyaluronic acid-protein macromolecules minimise further asperity contact between the opposing surfaces in a mixed lubrication regime.

- A distinct acellular, noncollagenous *surface/boundary layer* was observed (<1 μm thick) and, in consideration of the friction test results, was deemed to be an inherent feature of the articular cartilage surface. This layer has potential implications for cartilage nutrition and permeability, aside from it's role in synovial joint lubrication. Quantitative, accurate and reliable values of surface roughness for articular cartilage have been determined.

9.4 Future Work

In any follow-up studies undertaken the options listed below should be considered.

- The beneficial lubricating ability of synovial fluid has been systematically demonstrated in two contact configurations; namely the 9 mm cartilage plug on cartilage counterface (Chapter 6) and the 9 mm cartilage plug on metal counterface under reciprocating motion (Chapter 7). Using these contact configurations the lubrication mechanism (boundary or boosted) afforded by the synovial fluid, as well as the active boundary constituent of synovial fluid, might be discovered. This could be determined by isolation of the key components of the synovial fluid. The use of glycoprotein and phospholipid solutions as the lubricant would serve to promote the boundary mechanism, and the use of HA solutions would serve to promote the boosted mechanism. Note that the size alone of the HA macromolecule could be large enough to prevent widespread contact of the relatively smooth cartilage surfaces without the need for a time consuming filter mechanism to be relied upon (Ogston and Stanier, 1951; Balazs, 1982; Geborek and Wollheim, 1993). The relative success of these lubricants would help to further resolve the issue of synovial fluid function in joint lubrication. This work will provide important implications for the necessary constituents of clinical artificial joint lubricants which are administered directly into the joint capsule (Peyron, 1993).
- The effects of selective matrix degradation on the biomechanical / biphasic properties of articular cartilage have already been well characterised (Bader et al., 1992). It would be very interesting to establish any link between controlled and quantified proteoglycan release and the tribological behaviour of articular cartilage. Any such link could further support the biphasic lubrication hypothesis.
- Biphasic finite element analysis of articular cartilage under various loading regimes is now quite advanced, (Ateshian and Wang, 1995; Goldsmith et al., 1996). Further modelling in respect of the biphasic lubrication theory, paying particular attention to time-dependent fluid phase load carriage (and possibly

matching to experimental friction observations), would be a very interesting new dimension to explore.

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Appendix

Publications

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The effect of stationary loading on the friction and boundary lubrication of articular cartilage in the mixed lubrication regime

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Friction of cartilage/metal contacts, within a mixed lubrication regime, was recorded using synovial fluid, Ringer's solution or no lubricant present. The main test variable was the period of stationary loading prior to sliding which ranged from 5 seconds to 45 minutes. This was the amount of time that the cartilage spent loaded against the metal counterface prior to sliding the metal counterface horizontally and measuring friction. The calculated coefficient of friction rose with increasing stationary loading time, reaching an asymptote at ~0.270. Following the re-application of load, after short periods of load removal, friction was also found to drop sharply. The flow of liquid in the cartilage and load carriage by the fluid phase was highlighted as being an important factor influencing friction. At a 2 minute stationary loading time the synovial fluid beneficially lubricated the cartilage/metal contact compared to the Ringer's solution, by a mechanism which was no longer effective at 5 minute stationary loading times and beyond. The boundary lubricant was possibly an inherent part of the articular cartilage surface, rather than being absorbed from the synovial fluid, as all three lubricating conditions demonstrated equivalent μ readings for most tests.

1. Introduction

Human synovial joints have to withstand complex, varied and often harsh loading regimes, being subjected to both dynamic and impact loading cycles under conditions of sliding and rolling; often after considerable periods of stationary loading with little motion. Indeed it is estimated that for over 95% of the time our joints undergo very little movement, and these periods are commonly associated with significant loading. In order to explain the low friction and minimum wear of diarthrodial joints, throughout most people's lifetime, it is very probable that more than one lubrication mode is in operation (Dowson, 1967). It is generally accepted that some combination of fluid-film and boundary lubrication must co-exist (Mow and Soslowsky, 1991). This combination must be determined by the loads, kinematics, anatomy, and chemical and physical properties of the tissues and fluids within a specific joint at any given moment; that is the tribological conditions of the contact.

The two important elements providing lubrication of synovial joints are articular cartilage, covering the bone ends, and synovial fluid in the joint capsule. Articular cartilage is composed of a network of fine collagen fibrils enmeshed in a concentrated solution of proteoglycans. The collagen content of cartilage tissue ranges from 10 to 30 % by weight and for the proteoglycans 3-10 % by weight; the remaining 60 to 87 % being water, inorganic salts, and small amounts of other matrix proteins, glycoproteins, and lipids (Armstrong and Mow, 1982), collectively known as the interstitial fluid. The combination of its high water content, porosity {effective pore size of 2.0 to 6.5 nm (Dowson, 1990)} and compliance make articular cartilage an excellent bearing material. Collagen fibrils and proteoglycans are the structural components supporting the internal mechanical stresses that result from loads being applied to the cartilage. The structural components (the solid matrix) together with water determine

the biomechanical behaviour of this tissue (Mow et al., 1989). For this reason it is often modelled as a biphasic material (Spilker et al., 1992; Ateshian et al., 1994), or even triphasic with an additional ionic component (Mow et al., 1990). Under load the tissue expresses interstitial fluid into its unloaded areas and into the joint capsule. This is referred to as exudation. During the exudation process, the pressure differences produced in the fluid phase, that cause the flow, also carry varying amounts of the load, effectively reducing the load carriage and stresses in the solid matrix. When the load is removed the tissue recovers the lost water, due to its inherent swelling pressure. This is referred to as imbibition, rehydration or swelling.

Synovial fluid is a dialysate of blood plasma with the addition of a protein - polysaccharide, hyaluronic acid (HA), complex which is responsible for its high viscosity. At low shear rates (0.1 s^{-1}) the viscosity of synovial fluid is of the order of a few tens of Pas, whereas at much higher shear rates (1000 s^{-1}) it is only one thousandth of this value, ~0.02 Pas (Cooke et al, 1978), clearly portraying its non-Newtonian behaviour. Under physiological shear rates, being ~10^{5}-10^{6} s^{-1} , extrapolation of the data available shows that the viscosity of synovial fluid can be little more than twice that of water (Dowson, 1990).

Several mechanisms of fluid film lubrication in synovial joints have been proposed. As well as the more familiar elastohydrodynamic (EHL) and squeeze-film mechanisms (Unsworth, 1993), other theories relate specifically to the characteristics of articular cartilage and synovial fluid. These include micro-EHL (Dowson and Jin, 1986), weeping (McCutchen, 1962), boosted/ultrafiltration mechanism (Maroudas, 1967; Walker et al., 1970) and a self-generating interstitial fluid film mechanism (Mow et al., 1989).

When solid-to-solid contacts occur, a boundary lubricant upon the cartilage surface helps to sustain the minimum friction and wear associated with a healthy synovial joint. The boundary lubricant was originally thought to be a large macromolecular hyaluronic acid (HA) - protein complex adsorbed from the synovial fluid (McCutchen, 1966; Walker et al., 1970). More recently both glycoproteins (Davis et al., 1979; Swann et al., 1981; Jay, 1992) and phospholipids (Hills, 1989; Williams et al., 1993; Kirk et al., 1993) have been implicated. Whether the boundary layer is adsorbed from the synovial fluid or is an inherent part of the articular cartilage surface is not known. In general the fact remains that "The relative importance of synovial fluid and cartilage in terms of lubrication [for both boundary and fluid film] is still unclear" as stated by Stachowiak et al. (1994).

There are several advantages and disadvantages associated with measuring the frictional behaviour of articular cartilage for both entire joint (Table 1) and small cartilage specimen (Table 2) experiments. In brief, entire joint experiments allow preservation, as far as possible, of physiological conditions. The use of small cartilage specimens on a flat synthetic counterface, on the other hand, allows friction to be recorded directly and also a better understanding and control of the tribological and general experimental conditions to be achieved.

Throughout these types of studies widely differing methodologies have been adopted. These methodologies result in different lubrication conditions in the contact, from full fluid film to mixed regimes with increasing amounts of boundary contact. It is, therefore, not surprising that the values for the coefficient of friction found in Tables 1 and 2 vary by two orders of magnitude and worthwhile comparisons of friction measurements can generally only be made within a carefully controlled study (Clift et al., 1989).

In order to investigate the effect of different boundary lubricants on the friction of cartilage, contacts have to be established in the mixed lubrication regime. This can be achieved by applying a constant stationary load to the contact for extended periods of time, so producing very thin lubricating films due to squeeze film action. By subsequently using very low sliding velocities the contribution made by elastrohydrodynamic entraining action, to the film thickness, is very small and the mixed lubrication regime, which results from the constant stationary load, remains (see Appendix). Under these conditions the

Reference	Joint	μ
Jones (1934)	Horse stifle	0.02
Charnley (1959)	Human ankle	0.014 , 0.024
Barnett & Cobbold (1962)	Canine ankle	0.018-0.03
Linn (1968)	Canine ankle	0.0044
Little et al. (1969)	Human hip	0.003-0.015
Unsworth et al. (1975)	Human hip	0.02-0.042
Clarke et al. (1975)	Human hip	0.001-0.030
O'Kelly et al. (1978)	Human hip	0.01-0.08
*Roberts et al. (1982)	Human hip	0.04
Mabuchi et al. (1994)	Canine hip	0.007 (±0.004)

Table 1 Coefficient of friction, μ, measured in entire synovial joints. *Friction measured by hip function simulator machine.

Reference	Contacts	Lubricant	μ
McCutchen (1962)	cartilage/glass	SF	0.003-0.1
Dowson et al. (1968)	cartilage/glass	SF	0.1-0.9
Walker et al. (1970)	cartilage/glass	SF	0.0014-0.07
Chappuis et al. (1983)	cartilage/glass	SF	0.01-0.1
Clift et al. (1989)	cartilage/glass	None	0.1-0.2
Stachowiak et al. (1994)	cartilage/metal	SF	0.02-0.20

Table 2 Coefficient of friction, μ, measured for small cartilage specimens. (SF - Synovial fluid.)

contacts enter the mixed lubrication regime, with increasing amounts of boundary contact for increasing loading time prior to sliding, as the squeeze film is depleted in the contact. In the biphasic cartilage material, extended periods of constant loading produce a further effect in the exudation of water out of the cartilage, which is subject to the external stress. The time constant involved in the exudation of water from the cartilage, which can be up to 60 minutes (Edwards, 1967), is much longer than the time required to deplete the squeeze film in the contact, (Unsworth, 1993), to a level of 0.1 micrometres which is typically less than 5 seconds (see Appendix). Hence by carrying out comparative friction experiments where the period of loading, prior to sliding, is carefully controlled, the tribological conditions in the contact and the condition of the cartilage can be independently varied.

The aim of this study was to investigate boundary lubrication of articular cartilage by measuring friction, within a mixed lubrication regime, after periods of stationary loading varying from 5 seconds to 45 minutes, upon a smooth metal counterface. Both Ringer's solution and synovial fluid were studied in order to assess whether or not the constituents of the synovial fluid, not present in the Ringer's solution, reduced friction by providing a boundary lubricating action.

2. MATERIALS

Articular cartilage was collected from bovine femoral condyles and from the patello-femoral articulating surfaces. The subchondral bone and underlying cancellous bone was retained, in the form of 9 millimetre diameter bone plugs, to facilitate handling and mounting of the cartilage on the friction apparatus, and also to ensure cartilage integrity (Figure 1). Three millimetre diameter cartilage disc specimens were produced, centrally located upon the 9 millimetre diameter bone plugs, for the friction experiments in order to allow application the of physiological loads and to ensure a reasonably flat cartilage surface in the contact areas. As it was important to maintain

cartilage surface integrity and obtain reproducible specimens, a special rig was developed to machine and drill the specimens. A universal drilling machine was used along with customised drill bits. During joint sectioning and manufacture of these specimens, the cartilage was kept hydrated by frequent spraying of Ringer's solution. This method allowed the 3 mm cartilage discs to be prepared without damage to the articulating surface (Figure 1).

Synovial fluid was collected from bovine ankle joints within a period of 24-36 hours after slaughter. Synovial fluid was exposed by sectioning across intact, cleaned and skinned, ankle joints and extracted by a syringe through a 2 mm diameter flexible plastic tube. This technique was found to provide easy and rapid collection. The ankle joints provided a plentiful and fresh supply of synovial fluid. The fluid was filtered through a 500 micron sieve to remove cartilage debris.

Articular cartilage and synovial fluid samples were stored frozen at -20°C. The cartilage specimens were kept frozen in Ringer's solution. Prior to the tests the cartilage specimens were always defrosted and immersed in the appropriate lubricant for at least 1 hour before use. Additional tests were also carried out on fresh articular cartilage and synovial fluid samples that had not been frozen, in order to check that the freezing of specimens did not influence the friction readings.

3. METHODS

Friction was measured on a sliding-friction machine adapted from a sledge microtome (Caravia et al., 1993), using both synovial fluid and Ringer's solution as lubricants. A flat metal counterface, with a surface roughness of Ra≅0.01 μm, was fixed in position in a bath of lubricant which was driven by a motor to slide in one direction at a constant speed. The cartilage specimen was loaded onto the counterface while secured to one end of a balanced loading arm, pivoted at a fulcrum by an air bearing. The other end of the loading arm was restrained by a piezoelectric force transducer and thereby served to record the frictional force (N). Displacement of the metal counterface was monitored by a linear variable differential transducer. The signals from both transducers were transferred to a computer via an analogue-to-digital converter. The sliding speed and the coefficient of friction, μ, were recorded using the Unkelscope data acquisition and manipulation software package. The software used previously determined calibration data to convert the transducer signals into values of friction force and sliding distance, which were then plotted against time. The coefficient of friction readings had a sensitivity of ±0.001. The calibration for friction was conducted by attaching a series of loads, from 0 to 500 grammes (up to equivalent frictional forces of 4.9 Newtons), to the specimen end of the loading arm, such as to mimic friction forces during testing. The Unkelscope software package, using an analogue-to-digital converter, recorded the voltages for each applied load and a linear least squares fitting procedure was used for the calibration graph for the transducer and data acquisition system.

A normal load of 30 N was applied to the 3 mm diameter cartilage discs providing a nominal contact pressure of approximately 4 MNm^{-2}, which was representative of physiological loading. A low sliding speed of 4 mm s^{-1} was adopted to minimise elastrohydrodynamic lubrication, which could reduce the amount of boundary contact (see Appendix). The main test variable, the period of loading prior to sliding, was varied between 5 seconds and 45 minutes. This was the amount of time that the cartilage spent loaded against the metal counterface prior to sliding the metal counterface horizontally and measuring friction. When the cartilage was loaded onto the metal counterface a digital timer was simultaneously started and just prior to the end of the stationary loading period the piezoelectric force transducer was switched on. Once the loading time was reached, as indicated by the digital timer, the motor of the friction apparatus was started and, at the same time, data processing by the computer was initiated. A 5 second loading period, prior to sliding, was the shortest time that could practically be adopted for this routine. At this time the predicted squeeze film thickness for both

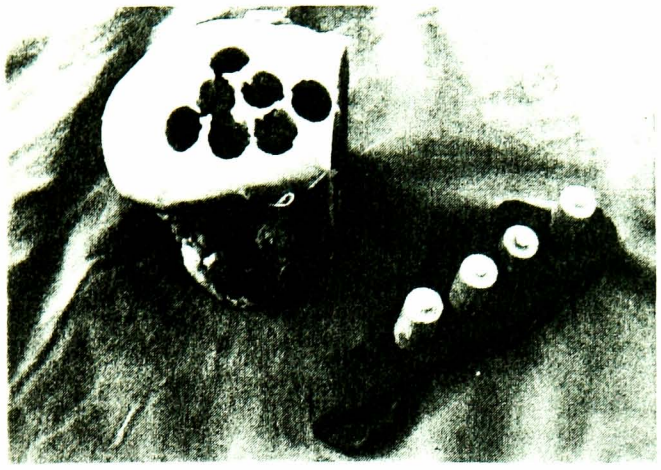

Fig 1 Small cartilage specimens taken from a bovine femoral condyle

lubricants was less than 0.1 μm, indicating that the contacts had entered the mixed lubrication regime (see Appendix).

The stationary loading times adopted were 5 seconds, 2 minutes, 5 minutes and 45 minutes. Friction values after 5 seconds loading were always recorded and for each loading period (2, 5 and 45 minute) seven cartilage specimens were used to measure friction. Occasional specimens with μ values above 0.015 after 5 seconds loading were rejected, being indicative of fibrillation or handling damage. This procedure served to ensure specimen repeatability. For each cartilage specimen seven readings were taken after 5 seconds loading and then seven readings at either 2 or 5 minutes loading. Between each test run for the respective 2 and 5 minute stationary loading periods the specimen was unloaded, while immersed in the lubricant, for either 2 or 5 minutes. Thus the cartilage specimens were allowed to rehydrate for a period equal to the amount of time spent under load. This was carried out for both Ringer's solution and synovial fluid as the lubricant. For the 45 minute loading times only one reading per specimen was taken. The reason being that this was an extreme loading condition and subsequent runs would not have necessarily provided repeatable results. Possible damage to the cartilage could have been caused by disruption of the collagen-proteoglycan matrix and leaching of the proteoglycans from the solid matrix, causing softening and increased permeability (Mow et al., 1989). In addition, with a large amount of boundary contact after 45 minutes loading, successive runs could have damaged the boundary lubricating layer (Stachowiak et al., 1994).

In addition to using Ringer's solution and synovial fluid, cartilage specimens were also tested with no lubricant for the 45 minute loading tests. Furthermore, after the 45 minute loading period, having measured friction, it was again measured after a 1 second or 1 minute period of load removal. After the load removal period the cartilage specimen was again loaded against the metal counterface for 5 seconds prior to sliding and then friction was recorded once more. The load removal time of 1 second was selected to allow the fluid to fully re-enter the contact, while allowing minimum cartilage rehydration. So when the load was reapplied the normal approach squeeze film action occurred. The 1 minute load removal period allowed some rehydration of the cartilage, in addition to re-introducing the squeeze film action.

Between each set of tests the metal counterface was cleaned with hot water and dried using alcohol. Before the metal counterface was secured into the lubricant bath it was always wiped with tissues and then blasted by air to remove dust and loose particles.

For each set of conditions the mean and standard deviation of the friction coefficient was calculated. The statistical significance of the difference in the means was tested using the Student's t-Test on a Microsoft Excel spreadsheet application software package. Statistical significance was tested at the 95% confidence level ($p<0.05$).

4. RESULTS

Upon the initiation of sliding, following a given period of stationary loading, the friction rose instantaneously to a certain value and then generally remained at this value while the cartilage was sliding across the metal counterface. Hence the start-up[1] and steady state values were similar and both were influenced primarily by the duration of stationary loading prior to sliding.

[1]Occasionally the start-up friction exhibited a small peak in relation to the steady state friction.

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The steady state coefficient of friction values for the cartilage specimens sliding on a metal counterface are shown in Figures 2 and 3, for the loading times of 5 seconds, 2, 5 and 45 minutes. These graphs represent readings taken using Ringer's solution and synovial fluid respectively. Figures 2 and 3 clearly display the rise in friction with stationary loading time for the cartilage/metal contacts. Stationary loading time was thus shown to be a critical variable for the coefficient of friction for both lubricants. On the whole the results for Ringer's solution and synovial fluid were similar with the coefficient of friction at 5 seconds being ~0.010 and at 45 minutes ~0.270. However the 2 minute stationary loading period results, Figure 4, revealed that synovial fluid was beneficially lubricating the cartilage/metal contacts compared to the Ringer's solution ($p<0.001$). For loading times longer than two minutes prior to sliding, the friction rose rapidly for both lubricants, with similar values of friction for the five minute and forty five minute loading tests (Figures 2 & 3). At 45 minutes the coefficient of friction, being ~0.270, appeared to be reaching an asymptote for both lubricants. This asymptotic behaviour was highlighted by studying the variation in friction at 10, 15 and 30 minute stationary loading periods also, on four individual specimens (Figure 5).

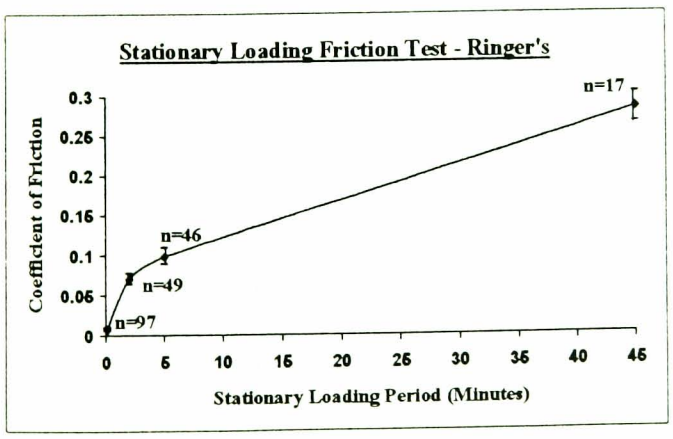

Fig 2 Means and standard deviations of friction for articular cartilage specimens sliding against a metal counterface, as measured after 5 second, 2 minute, 5 minute and 45 minute stationary loading periods. The number of readings (n) taken for each loading period is shown. The lubricant was Ringer's solution.

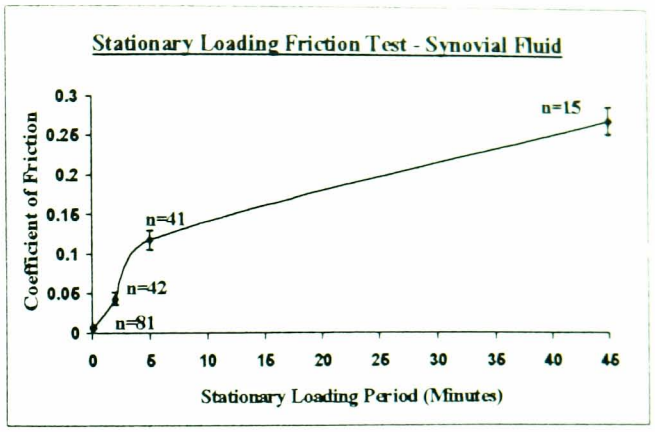

Fig 3 Means and standard deviations of friction for articular cartilage specimens sliding against a metal counterface, as measured after 5 second, 2 minute, 5 minute and 45 minute stationary loading periods. The number of readings (n) taken for each loading period is shown. The lubricant was synovial fluid.

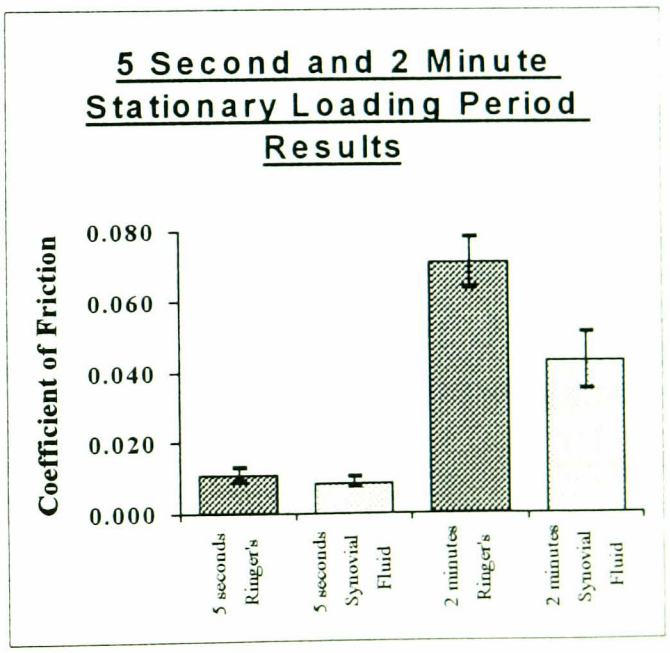

Fig 4 Means and standard deviations of friction for articular cartilage specimens sliding against a metal counterface, as measured after 5 second and 2 minute stationary loading periods. The number of readings taken for each column was approximately 49. For the 2 minute loading period the difference in friction between the synovial fluid and Ringer's solution group was statistically significant ($p<0.001$).

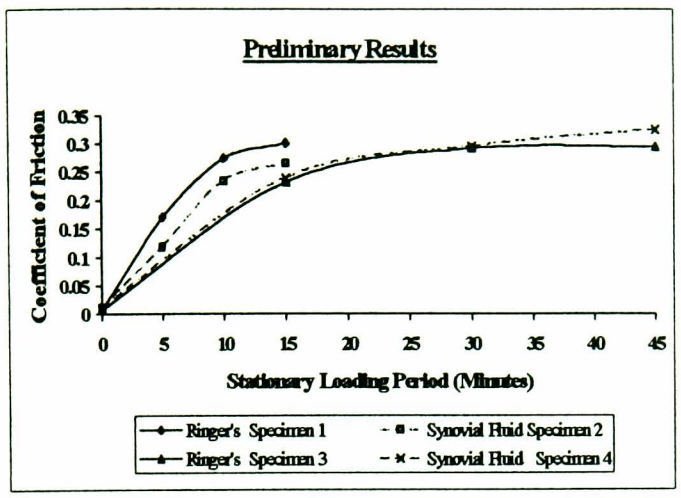

Fig 5 Coefficient of friction plotted against stationary loading period for four cartilage specimens. Each point represents an average of 3 readings. Differences of the μ values, between the two lubricants, generally appeared less for longer loading periods. At 45 minutes it appeared that an asymptote was being approached.

For the friction measurements made after 45 minutes loading, (Figures 6 and 7) no statistical difference was found between the three lubricating conditions (i.e. synovial fluid, Ringer's solution and no lubricant). Figures 6 and 7 also show the effect of the removal of the load for 1 second and 1 minute, respectively, following the period of 45 minutes loading. Figure 6 shows that for each of the three lubricating conditions, the removal of the load for 1 second, after the 45 minute loading period, immediately followed by a period of 5 seconds loading, only reduced the high level of friction found after 45 minutes by a small amount. This reduction, using a paired Student t-Test, was found to be significant for Ringer's solution ($p<0.003$) and synovial fluid ($p<0.001$) but not for the no lubricant condition. The μ values were still far higher than after the initial 5 second loading period. Following this 1 second load removal, the high level of friction found for the synovial fluid was significantly less than for the no lubricant condition ($p<0.004$) but not significantly less compared to the Ringer's solution, $p=0.06$.

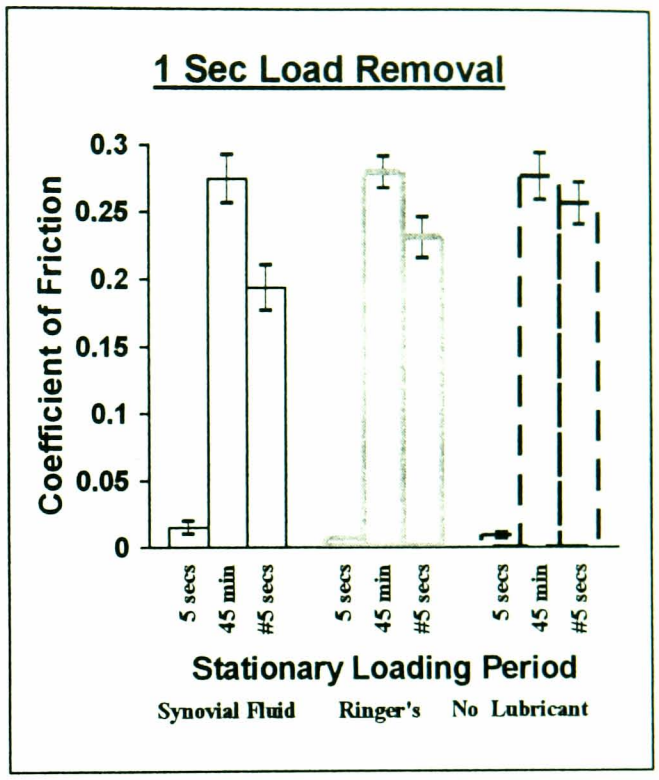

Fig 6 Coefficient of friction plotted against stationary loading period for the 5 second and 45 minute loading period tests. Tests were conducted using either synovial fluid or Ringer's solution as the lubricant, and also with no lubricant. #Immediately after the 45 minute loading period test the cartilage specimen was lifted away from the metal counterface for 1 second and then placed back onto the counterface for another 5 second loading period test.

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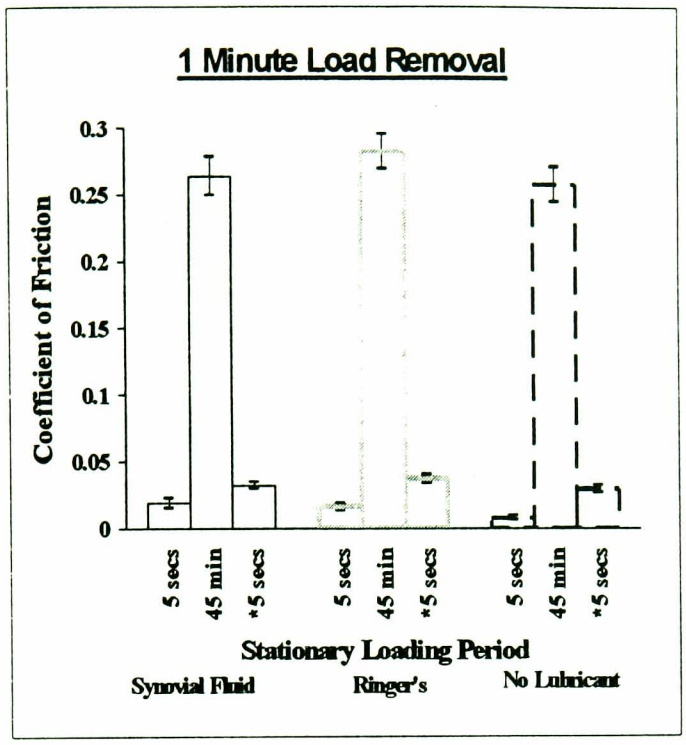

Fig 7 Coefficient of friction plotted against stationary loading period for the 5 second and 45 minute loading period tests. Tests were conducted using either synovial fluid or Ringer's solution as the lubricant, and also with no lubricant. *Immediately after the 45 minute loading period test the cartilage specimen was lifted away from the metal counterface for 1 minute and then placed back onto the counterface for another 5 second loading period test.

For a one minute load removal after 45 minutes loading, Figure 7, the coefficient of friction dropped considerably compared to the 45 minute loading level, and the values were only slightly, though significantly, higher than for the initial 5 second loading period; using a paired Student t-Test for synovial fluid, Ringer's solution and no lubricant respectively $p<0.05$, $p<0.004$ and $p<0.004$. For the 5 second loading test with a 1 minute load removal after 45 minutes loading (Figure 7), no significant difference in friction was found between the lubricating conditions.

5. DISCUSSION

The results have indicated a satisfactory level of repeatability of the friction values, as can be seen by the standard deviations for Figures 2 and 3. This was very encouraging as previous studies investigating boundary lubrication of synovial joints have often tended to use synthetic materials as their contacts (McCutchen, 1966; Wilkins, 1968; Reimann et al., 1975; Davis et al., 1979; Jay, 1992 and Williams et al., 1993), glass/rubber for example, as experiments utilising cartilage specimens were found to be time consuming and lacking reproducibility (Davis et al., 1979; Hills and Butler, 1984 and Jay, 1993). Cartilage friction experiments are essential and these difficulties were overcome through conscientious methodology. The relevance of any other type of friction testing would appear at the very least ambiguous.

The squeeze film model (see Appendix) demonstrated that some boundary contact had already occurred at the 5 second loading period. The entraining action model (see Appendix), for the experimental conditions, showed that upon sliding, full fluid film lubrication did not occur and boundary contacts remained. Both the start-up and steady state friction values were similar, in all experiments.

The results compare well with the literature (Tables 1 and 2). For the loading periods 5 seconds, 2 minutes, 5 minutes and 45 minutes, μ values were approximately in the range of 0.005-0.015, 0.040-0.070, 0.100-0.120 and 0.250-0.290 respectively. Most of the results in the literature, especially the entire joint experiments, Table 1, approximate to the 5 second loading values. Many of these studies proposed a full fluid film lubricating mechanism but at the 5 second loading time, for this study, contact between asperities was predicted to have occurred (see Appendix) and the coefficient of friction was still low. Therefore it would appear that low friction can be maintained, for short loading times at least, with boundary contacts, for well hydrated articular cartilage. For this to occur a good boundary lubricating mechanism must be in operation, or alternatively a high proportion of the load is carried by the fluid, within a mixed lubrication regime. It is recognised that in the biphasic theory of cartilage deformation a large proportion of the load is carried by the fluid phase at the onset of loading (Ateshian et al., 1994) and this would be consistent with the very

low values of friction after 5 seconds loading, for both lubricants.

At the 2 minute stationary loading period, however, synovial fluid was found to provide lower friction than the Ringer's solution ($p<0.001$), Figure 4. While at the 5 minute loading period and beyond the μ values were not showing any distinction between the two lubricants, Figures 2 and 3. It is thought that at 2 minutes the high viscosity of synovial fluid could, conceivably, have been substantially reducing the amount of boundary contact in comparison to the Ringer's solution by a boosted/ultrafiltration mechanism (Maroudas, 1967; Walker et al., 1970) whereby synovial fluid, in trapped pools upon the cartilage surface, increases its macromolecular concentration and also its resultant viscosity. Thus, in a mixed lubrication regime, for stationary loading periods of up to 2 minutes at least, the synovial fluid was better able to support the load in these trapped pools, before subsequently being squeezed out and increasing the amount of boundary contacts. At periods of 5 minutes and above, this mechanism may have been no longer effective and so a more equal amount of boundary contact, for the two lubricants, would have occurred. Another possible explanation is that there could have been an equal amount of boundary contact occurring for both lubricants at 2 minutes but the synovial fluid provided a better boundary lubricating action to the cartilage surface, in comparison to the Ringer's solution. Following this hypothesis, it is assumed that this boundary lubrication, imparted by the synovial fluid, was no longer effective for stationary loading periods of 5 minutes and above.

The most dominant factor influencing friction in this study was the period of loading. Above the 2 minute loading time the increase in friction, with loading time prior to sliding, was the same for both lubricants and therefore was not influenced by the viscosity of the lubricant or the elements of the fluid, such as glycoproteins or phospholipids in the synovial fluid, which are believed to provide boundary lubrication (Jay, 1992; Williams et al., 1993). Superficially the increase in friction can be linked to the increase in boundary contact between the asperities. In conventional, non-porous engineering materials this would be linked to depletion of the squeeze film and the squeezing out of fluid between the surface asperities. However the depletion of the squeeze film happens very rapidly and contact occurs in less than 5 seconds, with film thicknesses of less than 0.1 micrometres (see Appendix). Furthermore, it has been shown that with non-porous elastomers, with a similar test configuration, this leads to very high values of maximum friction being reached in approximately 200 seconds (Caravia et al., 1993). Therefore it is extremely unlikely that the more gradual rise in friction that occurs, in the high water content cartilage over a period of 45 minutes, is purely associated with the squeezing out of fluid between the surfaces. It is more likely that the rise in friction, over a 45 minute period, is primarily controlled by the flow of water (exudation) in the biphasic cartilage itself. Edwards (1967) and Mow et al. (1989) both showed long exudation times before reaching equilibrium, of the order of 45 to 60 minutes for articular cartilage. It is interesting to speculate about how the mechanics of biphasic deformation and flow of water in articular cartilage may influence surface friction. The authors do not fully support McCutchen's weeping lubrication theory (1962), as analyses have shown a net flow of fluid into the cartilage (Jin et al., 1992; Hou et al., 1992), and not out of the cartilage into the surface contacts. The most important consideration is how the load is carried or transferred in the biphasic cartilage. Upon initial loading a large amount of load is carried in the fluid phase of the cartilage (Ateshian et al., 1994) and it is interesting to speculate that the proportion of load carried by the fluid phase contributes little to the total or aggregate friction force of the two phases. As the stationary loading period is increased the load carried by the solid phase increases and that carried by the fluid phase decreases. Hence, the overall friction and friction coefficient increases. This is simply expressed as

$$\mu(t) = \frac{F(t)}{W} = \mu_s \left(\frac{W_s(t)}{W} \right) + \mu_f \left(\frac{W_f(t)}{W} \right)$$

where

$\mu_{(t)}$ overall or aggregate friction coefficient

$F_{(t)}$ overall friction force

W	total load
$W_{s(t)}$	load carried by the solid phase
$W_{f(t)}$	load carried by the fluid phase
μ_s	effective coefficient of friction attributed to the solid phase
μ_f	effective coefficient of friction attributed to the fluid phase

Following this relationship, for $\mu_s \gg \mu_f$, the coefficient of friction, $\mu_{(t)}$, increases as the proportion of the load carried by the solid phase increases with time.

A further explanation of the rise in friction can be attributed to the flow of fluid out of the surface asperities of the cartilage. Although there is no net overall flow of fluid out of the cartilage surface, as individual asperities are deformed fluid may well be released locally into the contact, helping to lubricate the asperities in contact.

The results of unloading the cartilage for 1 second and 1 minute periods, after 45 minutes loading, then reloading (Figures 6 and 7) provide further evidence that the water flow and content of cartilage is a critical determinant of friction. A 1 second load removal, which allowed fluid to cover the surface but not enter into the cartilage, only produced a small reduction in friction. For the 1 minute load removal period, where the fluid had an opportunity to re-enter the exuded cartilage, friction was dramatically reduced.

Furthermore the results in Figures 6 and 7 for the unlubricated contacts, which closely matched the results of the two lubricants, provides additional indication that the fluid flow in the cartilage itself, and not the lubricant applied to the surface, is the critical factor in determining friction in a mixed or boundary regime.

Stachowiak et al. (1994) blamed wear, particularly of the boundary layer, for rising friction with time for constant load, reciprocating motion studies, thus ignoring exudation effects. This present study has shown that even for a stationary constant load, upon the initiation of sliding, friction will increase with time. It is the authors' belief that both superficial boundary wear (possibly recoverable as transient in nature (Ghadially, 1983)) and cartilage exudation had a role to play in the rising friction with time characteristic for Stachowiak's et al. (1994) studies.

This study has vividly portrayed the importance of fully hydrated cartilage and load carriage by the fluid phase for low friction levels to be achieved in a mixed lubrication regime. The degeneration and disease of articular cartilage which can be associated with the loss of proteoglycans[2] and thus the fluid retaining properties, is therefore likely to cause increased levels of friction.

6. CONCLUSIONS

- There was good repeatability in the measurement of friction and the corresponding determination of the coefficient of friction values using cartilage specimens under varying tribological conditions.

- For a 5 second loading period the coefficients of friction were very low, 0.005-0.015, for all lubricating conditions. This was within a mixed, rather than a full fluid film, lubrication regime, with a substantial amount of load carried by the fluid and a good boundary lubricant in operation.

- The friction rose sharply with stationary loading time, and then levelled off to an asymptote, for loading times from 5 seconds to 45 minutes. The carriage of load by the fluid phase of the biphasic cartilage is thought to be an important factor influencing friction, as well as the amount of boundary contact taking place and any boundary lubrication action.

- There was generally no difference in friction between the different lubricants used, within a mixed lubrication regime. It is therefore quite probable that the boundary lubricant is a tenacious layer bonded to the articular cartilage surface under all the experimental conditions

[2] proteoglycans play a key role in retaining fluid within the collagen matrix.

considered. The synovial fluid may be serving to maintain or replenish this layer during periods when the cartilage surfaces are separated by the synovial fluid film.

- At the 2 minute stationary loading time synovial fluid beneficially lubricated the cartilage/metal contacts due to either a boosted/ultrafiltration mechanism or boundary lubricating mechanism. This mechanism was no longer effective at periods of 5 minutes and beyond.

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APPENDIX

Squeeze film - for stationary loading periods

To model the normal approach of the cartilage discs towards the metal counterface and predict squeeze film thicknesses while under stationary load the following equation, for perfectly smooth surfaces, was used after Dowson et al. (1991),

$$h = a^2 \sqrt{\frac{3\pi\eta}{4Wt}}$$

η - viscosity (Pa.s)
W - load (N)
t - squeeze film time (s)
h - squeeze film thickness (m)
a - contact radius (m)

For this study $a = 0.0015$ m and $W = 30$ N. For Ringer's solution η was 0.001 Pa.s and for synovial fluid η was assumed to be 0.1 Pa.s.

For a stationary loading time of 5 seconds the predicted squeeze film thickness for both Ringer's solution and synovial fluid was calculated to be 0.01 and 0.09 μm respectively. As articular cartilage surface roughness is of the order of 1 μm, then clearly, at 5 seconds loading and beyond, a mixed lubrication regime was predicted for both lubricants.

Entraining action - during sliding

The minimum film thickness, h_{min}, for entraining action during sliding of the cartilage specimens can be estimated by assuming that the flat-ended cylindrical cartilage disc could be approximated to a square pad such that the length of the square, L, was

$$L^2 = \pi r^2$$

where r - radius of cartilage specimen (m) (Halling, 1978) such that,

$$h_{min} = \sqrt{0.071\left(\frac{\eta u L}{W}\right)} \cdot L$$

where W - load (N)
η - viscosity (Pa.s)
μ - sliding velocity (m/s)
$L = \sqrt{\pi . r}$ (m)

Accounting for some degree of shear thinning (Cooke et al., 1978) the viscosity of synovial fluid was taken to be 0.01 Pa.s during sliding. For Ringer's solution and synovial fluid the predicted film thicknesses for entraining action were 0.01 μm and 0.04 μm respectively. Both of these predictions were equal to or less than the squeeze film calculations at 5 seconds and thereby still well within a mixed lubrication regime.

The influence of loading time and lubricant on the friction of articular cartilage

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Friction of cartilage on metal, metal on cartilage and cartilage on cartilage contact configurations, within a mixed lubrication regime, was measured using synovial fluid, Ringer's solution or with no lubricant present. The main test variable was the period of stationary loading which ranged from 5 s to 45 min, prior to sliding and consequently measuring friction. The coefficient of friction rose gradually with increasing stationary loading time, up to a value of approximately 0.3 at 45 min for all the contact configurations. Following the re-application of load, after short periods of load removal, friction was also found to drop sharply. The flow of liquid in the biphasic cartilage and load carriage by the fluid phase was highlighted as being an important factor in reducing friction within the mixed or boundary lubrication regime. Movement of the contact zone over the cartilage counterface ensured very low friction as the slider moved over fully hydrated cartilage. For the cartilage–cartilage contacts synovial fluid significantly reduced friction compared to Ringer's solution. This was attributed to an effective boundary lubrication action, which was not as effective for the cartilage–metal contacts. contacts.

Key words: friction, articular cartilage, lubricant, loading time

1 INTRODUCTION

Human synovial joints have to withstand complex, varied and often harsh loading regimes, being subjected to both dynamic and static loads under conditions of sliding and rolling; often after considerable periods of stationary loading with little motion. Indeed it is estimated that for over 90 per cent of the time joints undergo very little movement, and these periods are commonly associated with significant loading. In order to explain the low friction and minimum wear of diarthrodial joints throughout the lifetime of most people, it is very probable that more than one lubrication mode is in operation (**1**). It is generally accepted that combinations of fluid-film and boundary lubrication must co-exist (**2**). This combination must be determined by the loads, kinematics, anatomy and chemical and physical properties of the tissues and fluids within a specific joint at any given moment; that is the tribological conditions of the contact.

The two important elements providing lubrication of synovial joints are articular cartilage, covering the bone ends, and synovial fluid in the joint capsule. Articular cartilage is composed of a network of fine collagen fibrils and within this network hydrophilic proteoglycan aggregate molecules are immobilized and restrained. The collagen content of cartilage tissue ranges from 10–30 per cent by weight and for the proteoglycans 3–10 per cent by weight; the remaining 60–87 per cent being water, inorganic salts and small amounts of other matrix proteins, glycoproteins and lipids (**3**), collectively known as the interstitial fluid. The combination of its high water content, porosity [effective pore size of 2.0–6.5 nm (**4**)] and compliance make articular cartilage an excellent bearing material. Collagen fibrils and proteoglycans are the structural components transmitting the internal mechanical stresses that result from loads being applied to the cartilage. The structural components (the solid matrix) together with water determine the biomechanical behaviour of this tissue (**5**). For this reason it is often modelled as a biphasic material (**6**, **7**), or even triphasic with an additional ionic component (**8**). Under load the tissue expresses interstitial fluid into its unloaded areas and into the joint capsule. This is referred to as exudation. During the exudation process, the pressure differences produced in the fluid phase that cause the flow also carry varying amounts of the load, effectively reducing the load carriage and stresses in the solid matrix. When the load is removed the tissue recovers the lost water, due to its inherent swelling pressure. This is referred to as imbibition, rehydration or swelling.

Synovial fluid is a dialysate of blood plasma with the addition of a protein–polysaccharide, hyaluronic acid (HA), complex which is responsible for its high viscosity. At low shear rates (0.1/s) the viscosity of synovial fluid is of the order of twenty Pa s, whereas at much higher shear rates (1000/s) it is only one thousandth of this value, ~0.02 Pa s (**9**), clearly portraying its non-Newtonian behaviour. Under physiological shear rates, being $\sim 10^5$–10^6/s, extrapolation of the data available shows that the viscosity of synovial fluid can be little more than twice that of water (**4**).

Several mechanisms of fluid film lubrication in synovial joints have been proposed. As well as the more familiar elastohydrodynamic (EHL) and squeeze-film mechanisms (**10**), other theories relate specifically to the characteristics of articular cartilage and synovial fluid. These include micro-EHL (**11**), weeping (**12**), boosted ultrafiltration mechanism (**13**, **14**) and a self-generating interstitial fluid film mechanism (**5**).

When solid-to-solid contacts occur, a boundary lubricant upon the cartilage surface can help to sustain the low level of friction and wear associated with a healthy synovial joint. The boundary lubricant was originally thought to be a large macromolecular HA–protein complex adsorbed from the synovial fluid (**14**, **15**). More recently both glycoproteins (**16**–**18**) and phospholipids (**19**–**21**) have been implicated, with current work suggesting both may combine to form a complex boundary

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Table 1 Coefficient of friction μ measured in entire synovial joints

Reference	Joint	μ
Jones (**23**)	Horse stifle	0.02
Charnley (**24**)	Human ankle	0.014, 0.024
Barnett and Cobbold (**25**)	Canine ankle	0.018–0.03
Linn (**26**)	Canine ankle	0.0044
Little *et al.* (**27**)	Human hip	0.003–0.015
Unsworth *et al.* (**28**)	Human hip	0.02–0.042
Clarke *et al.* (**29**)	Human hip	0.001–0.030
O'Kelly *et al.* (**30**)	Human hip	0.01–0.08
Roberts *et al.** (**31**)	Human hip	0.04
Mabuchi *et al.* (**32**)	Canine hip	0.007 ($\pm$0.004)
Higaki *et al.* (**33**)	Porcine shoulder joint	0.003–0.023

* Friction measured by hip function simulator machine.

layer. Whether the boundary layer is adsorbed from the synovial fluid or is an inherent part of the articular cartilage surface is not known. In general the fact remains that 'The relative importance of synovial fluid and cartilage in terms of lubrication [for both boundary and fluid film] is still unclear' as stated by Stachowiak *et al.* (**22**).

There are several advantages and disadvantages associated with measuring the frictional behaviour of articular cartilage for both entire joint (Table 1) and small cartilage specimen (Table 2) experiments. In brief, entire joint experiments allow preservation, as far as possible, of physiological conditions. The use of small cartilage specimens on a flat synthetic counterface, on the other hand, allows friction to be recorded directly; also a better understanding and control of the tribological and general experimental conditions can be achieved.

Widely differing methodologies have been adopted for the measurement of friction. These methodologies result in different lubrication conditions in the contact, from full fluid film to mixed regimes with increasing amounts of boundary contact. It is, therefore, not surprising that the values for the coefficient of friction found in Tables 1 and 2 vary by two orders of magnitude and worthwhile comparisons of friction measurements can generally only be made within a carefully controlled study (**36**).

In order to investigate the effect of different boundary lubricants on the friction of cartilage, contacts have to be established in the mixed lubrication regime. This can be achieved by applying a constant stationary load to the contact for a period of time, so producing very thin lubricating films due to squeeze-film action. On subsequently using very low sliding velocities the contribution made by elastohydrodynamic entraining action to the film thickness is very small and the mixed lubrication regime, which results from the constant stationary load, remains (see Appendix). Under these conditions the contacts enter the mixed lubrication regime, with increasing amounts of boundary contact for increasing loading time prior to sliding. In the biphasic cartilage material, extended periods of constant loading produce a further effect in the exudation of water out of the cartilage, within the regions subject to the external compressive stress. The time taken for fluid flow away from the loaded cartilage regions to cease, which can be up to 60 min (**39**), is much longer than the time required to deplete the squeeze film in the contact (**10**) to a level of 0.1 μm which is typically less than 5 s (see Appendix). Hence by carrying out comparative friction experiments where the period of loading, prior to sliding, is carefully controlled, the tribological conditions in the contact and the condition of the cartilage can be independently varied.

The aim of this study was to investigate the behaviour of articular cartilage within mixed/boundary lubrication regimes, by measuring friction after periods of stationary loading varying from 5 s to 45 min, upon a smooth metal counterface. Both Ringer's solution and synovial fluid were studied in order to assess whether or not the constituents of the synovial fluid, not present in the Ringer's solution, reduced friction by providing a boundary lubricating action.

2 MATERIALS AND METHODOLOGY

2.1 Materials

Articular cartilage was collected from bovine femoral condyles and from the femoral–patella articulating surfaces. The subchondral bone and underlying cancellous bone was retained, in the form of 9 mm diameter bone plugs, to facilitate handling and mounting of the cartilage, and to ensure cartilage integrity. In addition to 9 mm cartilage plugs, 3 mm diameter cartilage disc specimens were cut, being centrally located upon the 9 mm

Table 2 Coefficient of friction μ measured for small cartilage specimens (SF = synovial fluid)

Reference	Contacts	Lubricant	μ
McCutchen (**12**)	Cartilage–glass	SF	0.003—0.1
Dowson *et al.* (**34**)	Cartilage–glass	SF	0.1–0.9
Walker *et al.* (**14**)	Cartilage–glass	SF	0.0014–0.07
Chappuis *et al.* (**35**)	Cartilage–glass	SF	0.01–0.1
Clift *et al.* (**36**)	Cartilage–glass	None	0.1–0.2
Ikeuchi *et al.* (**37**)	Cartilage–PMMA	Saline	~ 0–0.28
	Cartilage–cartilage	Saline	0.016–0.028
Stachowiak *et al.* (**22**)	Cartilage–metal	SF	0.02—0.20
Forster *et al.* (**38**)	Cartilage–metal	SF	0.003–0.35

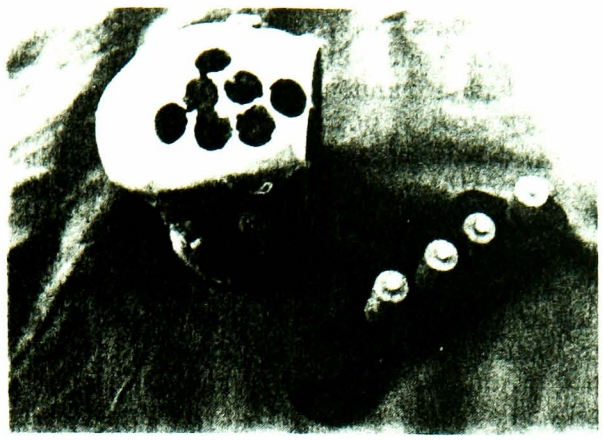

Fig. 1 Small cartilage specimens taken from a bovine femoral condyle

diameter bone plugs (Fig. 1). The curvature of the cartilage surface means that the area of contact does not always extend over the 9 mm diameter face. The 3 mm diameter discs were cut to ensure the contact area was the same on all specimens. As it was important to preserve the cartilage surface integrity, a special rig was developed to carefully extract the specimens from the femoral components. This comprised a drilling machine with customized drill bits. During joint sectioning and manufacture of these specimens, the cartilage was kept hydrated by frequent spraying of Ringer's solution.

Synovial fluid was collected from bovine ankle joints within a period of 36 h following slaughter. Synovial fluid was exposed by sectioning across intact, cleaned and skinned ankle joints and extracted by a syringe through a 2 mm diameter flexible plastic tube. This technique was found to provide easy and rapid collection. The ankle joints provided a plentiful and fresh supply of synovial fluid. The fluid was filtered through a 500 μm sieve to remove cartilage debris.

Articular cartilage and synovial fluid samples were stored frozen at −20 °C. The cartilage specimens were kept frozen in Ringer's solution. Prior to the tests the cartilage specimens were defrosted and immersed in the appropriate lubricant for at least 1 h before use. Additional tests were also carried out on fresh articular cartilage and synovial fluid samples that had not been frozen, in order to check that the freezing of specimens did not influence the friction readings.

2.2 Methods

2.2.1 Friction measurements

Friction was measured on a sliding-friction machine (**40**), using both synovial fluid and Ringer's solution as a lubricant. A flat metal counterface, with a surface roughness of $Ra \cong 0.01$ μm, was fixed in position in a bath of lubricant which was driven by a motor to slide a distance of 50 mm in one direction at a constant speed. The cartilage specimen was loaded on to the counterface while secured to one end of a balanced loading arm, pivoted at a fulcrum on an air bearing. The other end of the loading arm was restrained by a piezoelectric force transducer and thereby served to record the frictional force (N). Although piezoelectric force transducers are essentially used for measuring dynamic loads, during these experiments they measured forces over a period of up to 12 s. During the calibration of the transducer, forces were applied for periods of approximately 12 s without any variation in the data obtained. Displacement of the metal counterface was monitored by a linear variable differential transducer. The signals from both transducers were transferred to a computer via an analogue-to-digital converter. The sliding speed and the coefficient of friction, μ, were recorded using the Unkelscope data acquisition software package. The software used previously determined calibration factors to convert the transducer signals into values of friction force and sliding distance, which were then plotted against time. The coefficient of friction readings had a sensitivity of ±0.001. The calibration for friction was conducted by attaching a series of loads, from 0–500 g (up to equivalent frictional forces of 4.9 N), to the specimen end of the loading arm, such as to mimic friction forces during testing. The Unkelscope software package, using an analogue-to-digital converter, recorded the voltages for each applied load and a linear least-squares fitting procedure was used for the calibration graph for the transducer and data acquisition system.

A normal load of 30 N was applied to the specimens which produced stress levels in the range of 0.5–4 MN/m^2 depending on the contact area and was representative of physiological loading. A low sliding speed of 4 mm/s was adopted to ensure the contact operated in the mixed or boundary lubrication. The Appendix shows that the theoretically predicted elastohydrodynamic film thickness was less than 0.05 μm which is one twentieth of the roughness of the cartilage surface. The main test variable, the period of loading prior to sliding, was varied between 5 s and 45 min (Fig. 2). This was the amount of time that the cartilage spent loaded against the metal counterface prior to sliding the metal counterface horizontally and measuring friction. When the cartilage was loaded on to the metal counterface a digital timer was simultaneously started and just prior to the

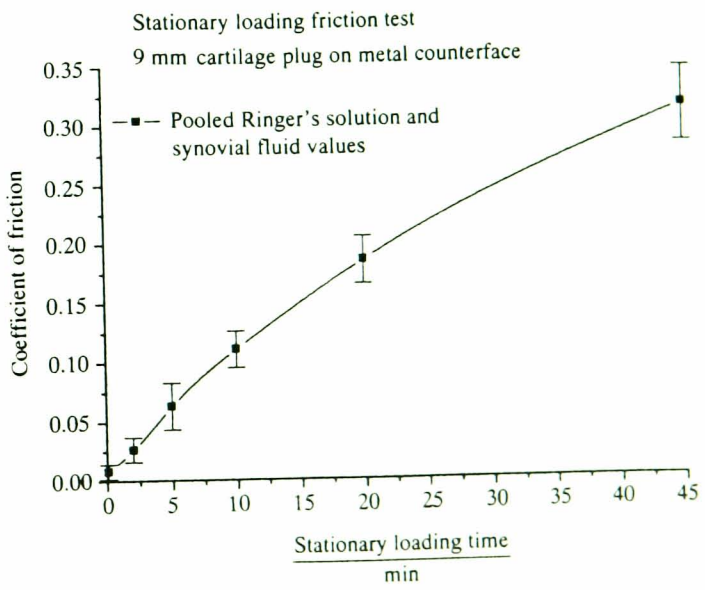

Fig. 2 Coefficient of friction versus loading time plot for the 9 mm diameter cartilage plug on metal counterface contact configuration for both synovial fluid and Ringer's solution. Means and standard deviations of the coefficient of friction are provided for articular cartilage specimens sliding against a metal counterface, as measured after 5 s, 2 min, 5 min, 10 min, 20 min and 45 min stationary loading periods ($n = 14$)

end of the stationary loading period the piezoelectric force transducer was switched on. Once the loading time was reached, as indicated by the digital timer, the motor of the friction apparatus was started and, at the same time, data processing by the computer was initiated. A 5 s loading period, prior to sliding, was the shortest time that could be practically adopted for this routine. At this time the predicted squeeze-film thickness for both lubricants was less than 0.1 μm for the 3 mm cartilage on metal configuration (see Appendix), indicating that the contacts had entered the mixed lubrication regime.

For each set of conditions the mean and standard deviation of the friction coefficient between the specimens were calculated. The statistical significance of the differences in the mean was tested using the Student's *t*-tests on a Microsoft Excel spreadsheet application software package. Statistical significance was tested at the 95 per cent confidence level, that is the probability that the differences in the means could have occurred purely by chance was less than 5 per cent.

2.2.2 9 mm diameter cartilage plug on metal counterface

For this contact configuration friction readings were recorded at 5 s, 2 min, 5 min, 10 min, 20 min and 45 min stationary loading times. For each loading time one reading was taken per specimen, eight specimens being used for Ringer's solution and a further eight for synovial fluid. Following the 45 min test the specimens were immersed in the lubricant with no load for another 45 min and then the same friction readings were repeated.

Between each test run, for the respective stationary loading periods, the specimen was unloaded, while immersed in the lubricant, for the same length of time. Thus the cartilage specimens were allowed to rehydrate for a period equal to the amount of time spent under load. The metal counterface was cleaned with hot water and dried using alcohol between each set of tests for any particular cartilage specimen. Before the metal counterface was secured into the lubricant bath it was always wiped with tissues and then blasted by air to remove dust and loose particles. These procedures were carried out for all contact configurations.

2.2.3 3 mm diameter cartilage disc on metal counterface

For the 3 mm cartilage disc on metal counterface tests the stationary loading times adopted were 5 s, 2 min, 5 min and 45 min. Friction values after 5 s loading were always recorded and for each loading period (2, 5 and 45 min) seven cartilage specimens were used to measure friction. Occasional specimens with μ values above 0.015 after 5 s loading were rejected, being indicative of fibrillation or handling damage. This procedure served to ensure specimen reproducibility. For each cartilage specimen a reading was taken after 5 s loading and then a reading at either 2, 5 or 45 min loading.

2.2.4 Metal slider on cartilage counterface

Cartilage counterfaces were sectioned from bovine humeral and femoral heads to dimensions similar to those of the metal counterface, approximately 30 × 70 mm and having a thickness of 10–20 mm (largely consisting of the underlying subchondral and cancellous bone), in such a manner as to provide as even a cartilage surface area as possible. Six cartilage counterfaces were tested with either or both of two metal plugs (or sliders). It was deemed necessary to use more than one cartilage counterface to prevent any deterioration of the cartilage during extended periods of testing and also to ensure that the collected cartilage samples were of good quality and produced reproducible results. Both metal plugs 1 and 2 had surface roughnesses, *Ra* values, of less than 0.15 μm, for a 0.25 mm sampling cut-off. Metal plug 1 had a radius of curvature of 66 mm. Metal plug 2 had a radius of curvature of 137 mm. The respective diameters of metal plugs 1 and 2 were 25 and 11 mm. Seven sets of friction data were taken using Ringer's solution as the lubricant and eight sets of data were taken using synovial fluid as the lubricant. Each data set consisted of two readings at 5 s, 30 s, 2 min, 5 min and 20 min with a single reading taken after 45 min stationary loading. For this contact configuration the initial friction at the start-up of motion was recorded, instead of steady state friction, as steady state friction was not found to be a function of loading time, simply because only the cartilage at start-up was subjected to any load.

2.2.5 Cartilage plug on cartilage counterface contact configuration

For these tests six cartilage counterfaces were used, again extracted from bovine femoral or humeral heads and sixteen 9 mm diameter cartilage plugs were used. Eight cartilage plugs were tested with synovial fluid as the lubricant and eight with Ringer's solution as the lubricant. Apart from having cartilage plugs these tests were carried out in exactly the same way as for the metal plug on cartilage counterface contact configuration tests in Subsection 2.2.4.

2.2.6 Indentation test

In order to investigate how the cartilage responded to loading and to obtain an idea of how the fluid was expressed out of the cartilage while under load, an indentation test was performed. Figure 3 shows how the linear displacement of an indentor changes with time under load. After the initial elastic deformation, the increase in displacement with time is associated with fluid flow out of the cartilage under load, thereby making this a useful and popular way to characterize the biomechanical properties of this tissue and also quantitate fluid expression while under load (**39**). A load of 6 N was applied via a hemispherical indentor to a section of articular cartilage (femoral cartilage counterface) while immersed in Ringer's solution. The radius of the indentor was 3.2 mm. The indentor was lowered on to the cartilage surface while data sampling for both vertical displacement and load, with the Unkelscope data acquisition software package, was initiated every 0.5 ms so as to record accurately the point of contact between the indentor and the cartilage counterface (**41**). This point of contact was identified by a sharp rise in the measured load. Measurements of linear displacement were then taken up to 60 min after

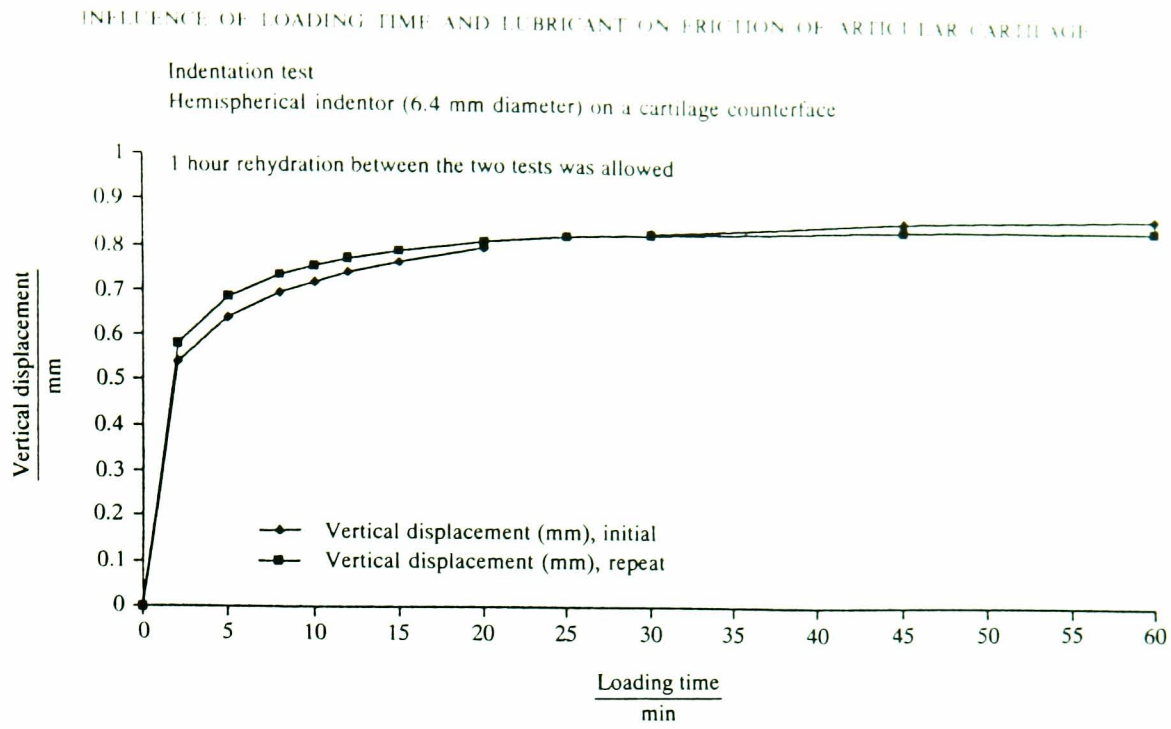

Fig. 3 Indentation test on a cartilage counterface—linear displacement versus loading time plot

the initial load was applied. This procedure was then repeated, upon exactly the same cartilage area, after the cartilage had been allowed to rehydrate while unloaded and immersed in Ringer's solution for 1 h.

2.2.7 Load removal tests with and without lubricant

Additional tests were carried out on the 3 mm cartilage disc on metal counterface configuration using Ringer's solution, synovial fluid, and no lubricant for a loading time of 45 min. After the 45 min loading period friction measurement, friction was again recorded after a 1 s or 1 min period of load removal. After the load removal period the cartilage specimen was again loaded against the metal counterface for 5 s prior to sliding and then friction was recorded once more. The load removal time of 1 s was selected to allow the fluid to fully re-enter the contact, while allowing minimum cartilage rehydration. When the load was reapplied, the normal approach squeeze-film action occurred. The 1 min load removal period allowed some rehydration of the cartilage, in addition to re-introducing the squeeze-film action at the surface.

3 RESULTS

3.1 Cartilage on a metal counterface

3.1.1 Influence of loading time

Upon the initiation of sliding, following a given period of stationary loading, the friction rose instantaneously to a certain value and then generally remained at this value while the cartilage was sliding across the metal counterface, during the 10 s data sampling period (or 40 mm sliding distance). Hence the friction at start-up* and during steady state sliding were similar and both were influenced primarily by the duration of stationary loading prior to sliding. Figure 2 plainly demonstrates the rise in the coefficient of friction for articular cartilage when subject to increasingly higher periods of stationary loading. For this 9 mm cartilage on metal contact configuration the values for synovial fluid and Ringer's solution were not significantly different and are both therefore presented as one data set. This marked rise in the coefficient of friction with loading time may be linked to the expression of fluid, and more specifically load carriage by the interstitial fluid flow, from the cartilage matrix under load. The results of the indentation tests (Fig. 3) showed continued displacement of the indentor, hence flow of fluid through the cartilage matrix, over a period of 60 min.

3.1.2 Comparison of lubricants

For the 3 mm cartilage on metal test, Fig. 4, on the whole the results for Ringer's solution and synovial

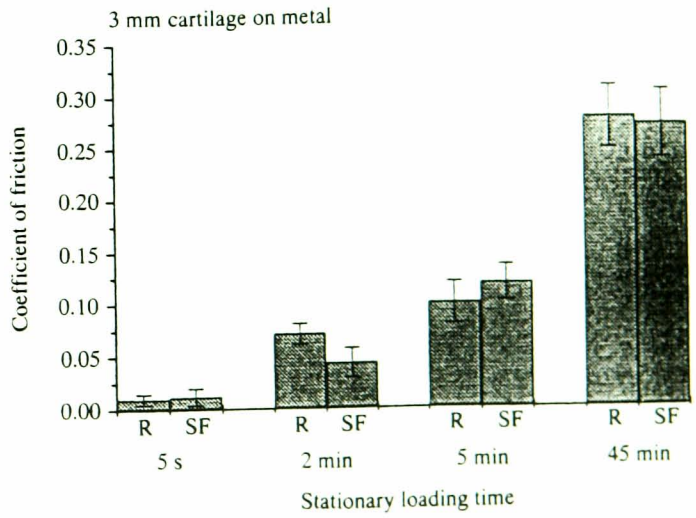

Fig. 4 Means and standard deviations of the coefficient of friction values for the 3 mm diameter cartilage disc on metal counterface contact configuration; as measured after 5 s, 2 min, 5 min and 45 min stationary loading periods, using both Ringer's solution (R) and synovial fluid (SF) as the lubricant

* Occasionally the start-up friction exhibited a small peak in relation to the steady state friction.

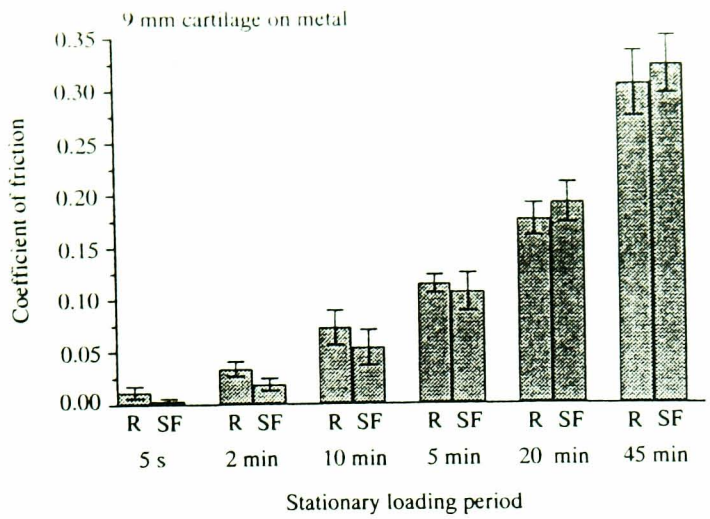

Fig. 5 Means and standard deviations of the coefficient of friction values for the 9 mm diameter cartilage disc on metal counterface contact configuration; as measured after 5 s, 2 min, 5 min, 10 min, 20 min and 45 min stationary loading periods, using both Ringer's solution (R) and synovial fluid (SF) as the lubricant

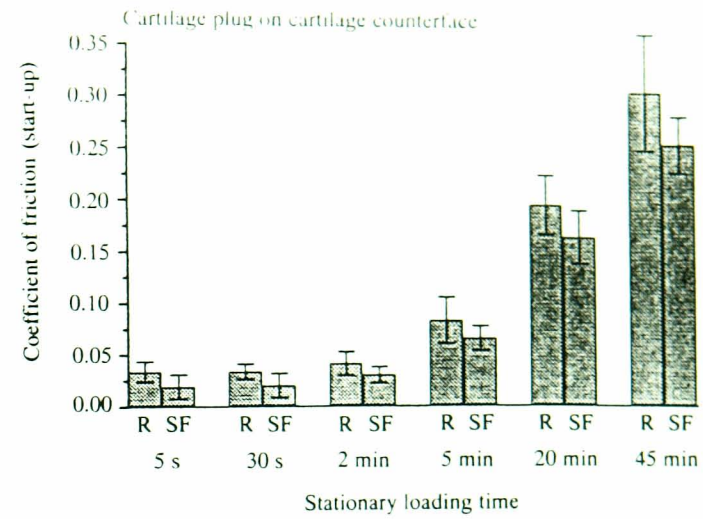

Fig. 7 Means and standard deviations of the start-up coefficient of friction values for the 9 mm diameter cartilage plug on cartilage counterface contact configuration; as measured after 5 s, 30 s, 2 min, 5 min, 20 min and 45 min stationary loading periods, using both Ringer's solution (R) and synovial fluid (SF) as the lubricant

fluid were similar with the coefficient of friction at 5 s being ~0.010 and at 45 min ~0.270. However, the 2 min stationary loading period results revealed that synovial fluid was beneficially lubricating the cartilage–metal contacts compared to the Ringer's solution ($p < 0.005$). Figure 5 contains the same results as in Fig. 2 for the 9 mm cartilage specimens, this time displayed as a histogram and plotting the results from the two lubricants separately. As already mentioned there were no statistically significant differences in coefficient of friction values for the two lubricants tested with this configuration. However, it is interesting to note that for the shorter periods of loading the synovial fluid gave lower values of coefficient of friction than the Ringer's solution.

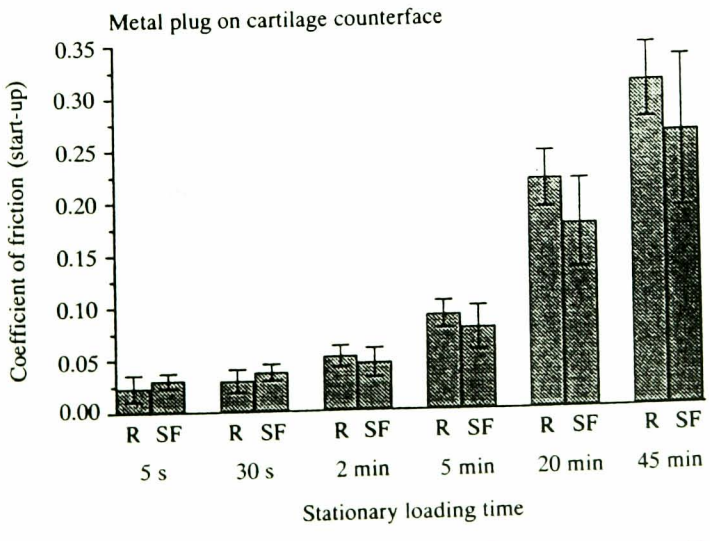

Fig. 6 Means and standard deviations of the start-up coefficient of friction values for the metal plug on cartilage counterface contact configuration; as measured after 5 s, 30 s, 2 min, 5 min, 20 min and 45 min stationary loading periods, using both Ringer's solution (R) and synovial fluid (SF) as the lubricant

3.2 Cartilage counterfaces (9 mm diameter cartilage plug and metal plug on cartilage counterface)

For the tests conducted on the cartilage counterface only the start-up friction was recorded. The reason for this was two-fold, firstly the cartilage counterfaces were not sufficiently even (both in the direction of sliding and perpendicular to the direction of sliding) to record reliable results and secondly the steady state coefficient of friction was not deemed to be dependent on loading time, as after the immediate initiation of the sliding that is the start-up, the cartilage or metal plug was then

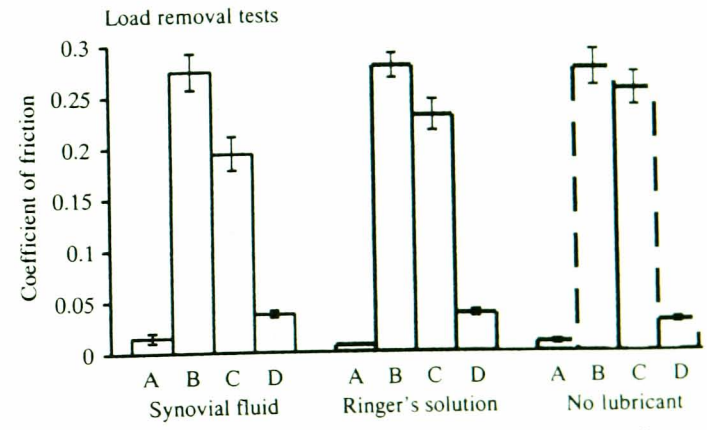

Fig. 8 Coefficient of friction plotted against stationary loading period for the 3 mm diameter cartilage disc on metal counterface contact configuration. Coefficient of friction is shown for four loading conditions, for each of the lubricants (synovial fluid, Ringer's solution and no lubricant). The loading conditions were: A—5 s of loading; B—45 min of loading; C—immediately after the 45 min loading period test the cartilage specimen was lifted away from the metal counterface for 1 s and then placed back on to the counterface for another 5 s loading period test; D—immediately after the 45 min loading period test the cartilage specimen was lifted away from the metal counterface for 1 min and then placed back on to the counterface for another 5 s loading period test

sliding upon previously unloaded and thereby a fully hydrated part of the cartilage counterface.

The start-up coefficient of friction for both the metal and cartilage plug on cartilage counterface was shown to increase with stationary loading time, Figs 6 and 7, in much the same manner as for the steady state coefficient of friction for the two previous metal counterface tests, Figs 4 and 5. For both contact configurations Student's *t*-tests were used to compare the means between the two lubricants at each of the loading periods tested. It was interesting to find that for the cartilage on cartilage contacts, Fig. 7, the synovial fluid was significantly reducing the coefficient of friction at all loading times compared to the Ringer's solution.

3.3 Load removal tests with and without lubricant

The general pattern of the results in Fig. 8 is similar for all three lubricating conditions, that is Ringer's solution, synovial fluid and no lubricant. This indicates that the lubricant was not the important factor controlling the friction coefficient. The loading condition was shown to have a marked effect on friction. The friction for condition A, 5 s of loading, was low and rose to a high value after 45 min of loading (condition B). Load removal for 1 s after 45 min loading (condition C) which allowed the fluid back into the surface of the contact, did not reduce the coefficient of friction considerably. However, condition D with load removal for 1 min after the 45 min loading, allowed the cartilage to rehydrate and the friction level was markedly reduced back to a value close to the original value seen in condition A, for all three lubricant conditions.

4 DISCUSSION

4.1 Comparison with other work

For all the four tested contact configurations the results have indicated a satisfactory level of reproducibility of the friction values, as can be seen by the standard deviations for Figs 2 and 4 to 8. This was very encouraging as previous studies investigating boundary lubrication of synovial joints have often tended to use synthetic materials as their contacts (**15**, **42**, **43**, **16**, **18**, **20**), glass/rubber for example, as experiments utilizing cartilage specimens were found to be time consuming and lacking reproducibility (**16**, **44**, G. D. Jay, 1993, personal communication). As the friction in the cartilage contact is controlled by the biphasic nature of the cartilage and the chemistry of the boundary layers interacting with the cartilage the relevance of any other type of friction testing would appear to cause concern.

The purpose of this study was to investigate friction when the contact was in the mixed or boundary lubrication regime. For the 3 mm cartilage on metal configuration the squeeze-film model (see Appendix) predicted that some boundary contact had already occurred at the 5 s loading period. The entraining action model (see Appendix), for the experimental conditions, predicted that upon sliding, full fluid film lubrication did not occur and boundary contacts remained. The experimental results showed that both the start-up and steady state friction values were similar for the metal counterface tests. This similarity tended to confirm that the entraining action produced by the low-velocity sliding was not sufficient to produce fluid film lubrication and reduce friction by elastohydrodynamic effects and hence the design of the experiment was considered to be successful in ensuring that the contact remained in the mixed or boundary lubrication regime.

The results compare well with the literature (Tables 1 and 2). Most of the results in the literature, especially the entire joint experiments, Table 1 (for example Unsworth *et al.* **31**), approximate to the 5 s loading values for the 3 mm and 9 mm cartilage on metal configurations. Many of these studies proposed a full fluid film lubricating mechanism but at the 5 s loading time, for this study, contact between asperities was predicted to have occurred (see Appendix) and the coefficient of friction was still low. Dowson and Jin (**11**) have proposed from theoretical studies that cartilage asperities are flattened by micro-elastohydrodynamic effects, and hence thinner lubricating films can be predicted to be maintained compared to the prediction presented in the Appendix. Therefore micro-elastohydrodynamic lubrication could be offered as an explanation for the low friction of around 0.01 found with cartilage plugs on metal counterfaces after 5 s of loading, Figs 4 and 5. However, the experimental results presented in Fig. 8 help to offer a fuller interpretation of the low values of friction. Firstly, the no surface lubricant condition also produced extremely low values of friction after 5 s of loading, indicating that bulk fluid on the surface was not required to obtain this low value of friction. Secondly, after 45 min of loading the coefficient of friction was extremely high (>0.25), and the entraining action did not reduce friction. Unloading the contact for 1 s and allowing a lubricant film on the surface only reduced the friction by a small amount from this high value. It was necessary to unload the contact for 1 min and allow the fluid to rehydrate the cartilage in order to reduce the friction towards the original low value found after the initial 5 s of loading. These results question the importance of the micro-elastohydrodynamic lubrication action in reducing friction in the contacts studied after 5 s of load application. Assuming that the low levels of friction after 5 s loading were achieved in the mixed or boundary regime, either an excellent boundary lubricant was present on the surface of the cartilage, or the biphasic nature of the cartilage (7) ensures extremely low friction in the fully hydrated condition. The authors favour the second explanation, as on initial loading of the biphasic cartilage, a large proportion of the load is carried by the pressure difference in the fluid phase of the cartilage and this may produce a very low coefficient of friction for the synovial fluid, for the Ringer's solution and for the no surface lubricant conditions. For the cartilage counterface configurations the start-up friction readings taken after a 5 s loading time were slightly higher due to the compliant nature of the cartilage counterface resulting in a degree of ploughing friction (**5**) when sliding was initiated for both the cartilage and metal plugs.

4.2 Influence of loading time

The overall results in this study showed that the friction was primarily controlled by the duration of loading

placed on the cartilage (Fig. 2), and therefore by the amount of water content and the biphasic fluid flow in the cartilage at the contact.

The most important consideration is how the load is carried or transferred in the biphasic cartilage. Upon initial loading a large amount of load is carried in the fluid phase of the cartilage (7) and it is interesting to speculate that the proportion of load carried by the fluid phase contributes little to the total or aggregate friction force of the two phases. As the stationary loading period is increased the load carried by the solid phase increases and that carried by the fluid phase decreases. Hence, the overall friction and friction coefficient increases. This is simply expressed as

$$F_T(t) = \mu_T(t)W$$

or

$$F_T(t) = \mu_s W_s(t) + \mu_f W_f(t)$$

$$\mu_s \gg \mu_f$$

therefore

$$F_T(t) = \mu_s W_s(t)$$

where $W_s(t) = W - W_f)(t)$

$\mu_T(t)$ = overall or aggregate friction coefficient
$F_T(t)$ = overall friction force
W = total load
$W_s(t)$ = load carried by the solid phase
$W_f(t)$ = load carried by the fluid phase
μ_s = effective coefficient of friction attributed to the solid phase
μ_f = effective coefficient of friction attributed to the fluid phase
(t) = indicates time function dependency

Following this relationship, for $\mu_s \gg \mu_f$, the coefficient of friction, $\mu_T(t)$, increases as the proportion of the load carried by the solid phase increases with time.

The indentation response of the cartilage to load with time (Fig. 3) can be used as an indication of the flow of fluid in the cartilage and the load carriage by the fluid phase. An initial instantaneous elastic deformation in response to load is primarily due to change in shape of the cartilage (not change in volume). At this point there are maximum pressure differences in the fluid phase, and maximum fluid flow and the largest proportion of the load is carried by the fluid phase, as the loading time progresses the slope of the displacement–time curve reduces. As the rate of flow of fluid in the cartilage declines in response to reducing pressure difference there is a subsequent reduction in the load carried by the fluid phase and hence increasing load carried by the solid phase. At 45–60 min the contact is close to equilibrium, little further deformation or fluid flow is occurring and the pressure difference and load carried by the fluid are close to zero. This is consistent with the long exudation times found by Edwards (**39**). The rise in friction in Fig. 2 with loading time is thought to reflect the relative proportion of load carried by the solid phase of the cartilage, $W_s(t)$. Upon removal of load fluid content is rapidly recovered (**39**), explaining the greatly reduced friction levels which were recorded subsequent to 1 min load removal after 45 min loading in Fig. 8. Following an equal period of load removal to that of previous loading, the indentation test produced a similar displacement time curve when subsequently reloaded which was indicative of a full recovery of fluid content, and this equal loading and load removal procedure was subsequently used in all friction tests.

For the cartilage counterface tests where most of the counterface was not loaded steady state friction was not affected by the loading time. This is in agreement with previous studies (**40**) whereby water-swollen polymers, hydrogels containing approximately 50 per cent water by volume, were friction tested as counterfaces in much the same manner. This would indicate that, in the natural joint, for two opposing cartilage surfaces translating across each other, even if one layer has a low water content, as long as the opposing cartilage surface is relatively well hydrated then friction would still remain low, as a large proportion of load would be carried by the fluid phase.

The above statement highlights an interesting contrast of results between the cartilage counterface work and the metal counterface work. For the 3 mm cartilage disc and 9 mm cartilage plug loaded against the metal counterface the friction instantaneously rose to a certain value (start-up friction) and then remained at that value while the cartilage slid along the flat metal counterface (steady state friction). Thus start-up and steady state friction were equal for all the tests and steady state friction was dependent on the loading time. As mentioned above, the difference would appear to be that, for these cartilage plugs on metal configurations, after start-up the cartilage was *not* moving on to a hydrated and porous surface (for example cartilage or hydrogel counterface) and thereby steady state friction was similar in magnitude to start-up friction and continued to increase as the loading time increased, as the proportion of load carried by the solid phase increased.

The 3 mm cartilage disc on metal configuration exhibited the sharpest rise in friction with time due to the higher contact stresses involved, being approximately 4 MPa. For the 9 mm cartilage plugs and metal plugs the contact stresses would have been roughly 0.5 MPa, depending on contact area which would have increased with load duration. It could be argued that for the 9 mm cartilage plugs and metal plugs friction was rising simply due to the increasing nominal contact area. This was not the case as for the 3 mm cartilage disc tests the nominal contact area was constant and friction, nonetheless, rose sharply with time. For the 3 mm cartilage disc the exudation of fluid from the matrix under load may have been exacerbated by the exposed surrounding areas of the matrix resulting from sectioning.

4.3 Influence of lubricants

For the 3 mm cartilage plugs on metal counterface configuration the differences between the two lubricants was variable, with the synovial fluid only consistently reducing frictions at the 2 min loading time, possibly due to the effect of its increased viscosity. The difference between the two lubricants was most evident for the cartilage–cartilage contact, with synovial fluid producing significantly lower ($p < 0.05$) coefficients of friction at all loading times. For the metal on cartilage contacts

a significantly lower value for friction was found for a few loading periods. Therefore under the experimental conditions considered in these studies the boundary (**18**, **20**) or boosted (**14**) lubrication mechanisms provided by the synovial fluid would only appear to be effective for opposing cartilage surfaces and not for cartilage–synthetic contacts.

4.4 General comments

There has been much discussion over the last 30 years about the role of different lubrication mechanisms in cartilage–cartilage contacts in natural synovial joints. It is clear that under the wide range of tribological conditions that occur *in vivo* a number of different mechanisms can play important roles at different times. This study was designed to consider the case of mixed and boundary lubrication, when elastohydrodynamic effects were less important. The differences found between the two lubricants were generally small even for the cartilage–cartilage contacts where it was statistically significant. So for this study the beneficial lubrication of the synovial fluid and in particular the macromolecules that have been extensively described elsewhere, can only be considered of secondary importance to the biphasic properties of the cartilage and lubrication afforded through load carriage by the fluid phase. The fact that after 2 min continuous loading the coefficient of friction levels are generally just less than 0.05, and after 5 min continuous loading these levels are just below 0.1, illustrates the importance of the biphasic lubrication mechanism during adverse lubrication conditions, such as continuous standing *in vivo*. Stachowiak *et al.* (**22**) has recently cited wear and removal of boundary lubricating layers for the rise in friction with time under a constant load and unidirectional continuous motion. The present study would indicate that the exudation of fluid from the cartilage and rise in the load carried by the solid phase is a more likely cause of increased friction.

It is a considerable period of time since McCutchen (**12**) proposed the weeping lubrication theory for synovial joints, whereby under load fluid from the cartilage was thought to enter the contact. More recently theoretical predictions by Hou *et al.* (**45**) and Jin *et al.* (**46**) have indicated that there is instead a net flow of fluid from the fluid film into the cartilage under normal approach. Nevertheless, the experimental results in this study confirm that the condition of the fluid phase of the cartilage plays a very important role in determining friction in the mixed or boundary lubrication regime. Consideration of the pressure differences and load carriage by the fluid phase of the biphasic cartilage can provide a clearer interpretation of the changes in the coefficient of friction with loading time.

5 CONCLUSIONS

1. There was good reproducibility in the measurement of friction and the corresponding determination of the coefficient of friction values using cartilage specimens under varying tribological conditions.
2. For a 5 s loading period the coefficients of friction were very low, 0.005–0.015, for all lubricating conditions. This was considered to be within a mixed, rather than a full fluid film, lubrication regime, with a substantial amount of load carried by the fluid phase of the biphasic cartilage and effective boundary lubrication in operation.
3. The friction rose sharply with stationary loading time, and then levelled off to an asymptote, for loading times from 5 s to 45 min. The carriage of load by the fluid phase of the biphasic cartilage is thought to be an important factor influencing friction.
4. Translation of metal or cartilage plugs over fully hydrated cartilage (in the case of the cartilage counterface) ensured very low values for the coefficient of friction, irrespective of the plugs' prior stationary loading times.
5. For cartilage–cartilage contacts synovial fluid significantly reduced friction compared to Ringer's solution. This was attributed to an effective boundary lubrication action, which was not as effective in cartilage–metal contacts.

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APPENDIX

Theoretical predictions of film thickness for the 3 mm cartilage disc on metal counterface configuration

(a) *Squeeze film (for stationary loading periods)*

To model the normal approach of the cartilage discs towards the metal counterface and predict squeeze-film thicknesses while under stationary load the following equation, for perfectly smooth surfaces, was used after Dowson *et al.* (**47**):

$$h = a^2\sqrt{\left(\frac{3\pi\eta}{4Wt}\right)}$$

η = viscosity (Pa s)

W = load (N)

t = squeeze film time (s)

h = squeeze film thickness (m)

a = contact radius (m)

For this study $a = 0.0015$ m and $W = 30$ N. For Ringer's solution η was 0.001 Pa s and for synovial fluid η was assumed to be 0.1 Pa s.

For a stationary loading time of 5 s the predicted squeeze-film thickness for both Ringer's solution and synovial fluid was calculated to be 0.01 and 0.09 μm respectively. As articular cartilage surface roughness is of the order of 1 μm, then clearly at 5 s loading and beyond a mixed lubrication regime was predicted for both lubricants.

(b) *Entraining action (during sliding)*

The minimum film thickness h_{min} for entraining action during sliding of the cartilage specimens can be estimated by assuming that the flat-ended cylindrical cartilage disc could be approximated to a square pad such that the length of the square, L, was $L^2 = \pi r^2$ where r = radius of cartilage specimen (m) (**48**) such that

$$h_{min} = \sqrt{\left\{0.071\left(\frac{\eta u L}{W}\right)\right\}}L$$

where

W = load (N)

η = viscosity (Pa s)

μ = sliding velocity (m/s)

$L = \sqrt{(\pi r)}$(m)

Accounting for some degree of shear thinning (9) the viscosity of synovial fluid was taken to be 0.01 Pa s during sliding. For Ringer's solution and synovial fluid the predicted film thicknesses for entraining action were 0.01 and 0.04 μm respectively. Both of these predictions were equal to or less than the squeeze film calculations at 5 s and thereby still well within a mixed lubrication regime.

Friction Apparatus

In Figure i and Figure ii below plan and side schematic drawings of the friction apparatus are displayed. These schematics are supplied in addition to the photographs in section 2.4, page 49, of the friction apparatus.

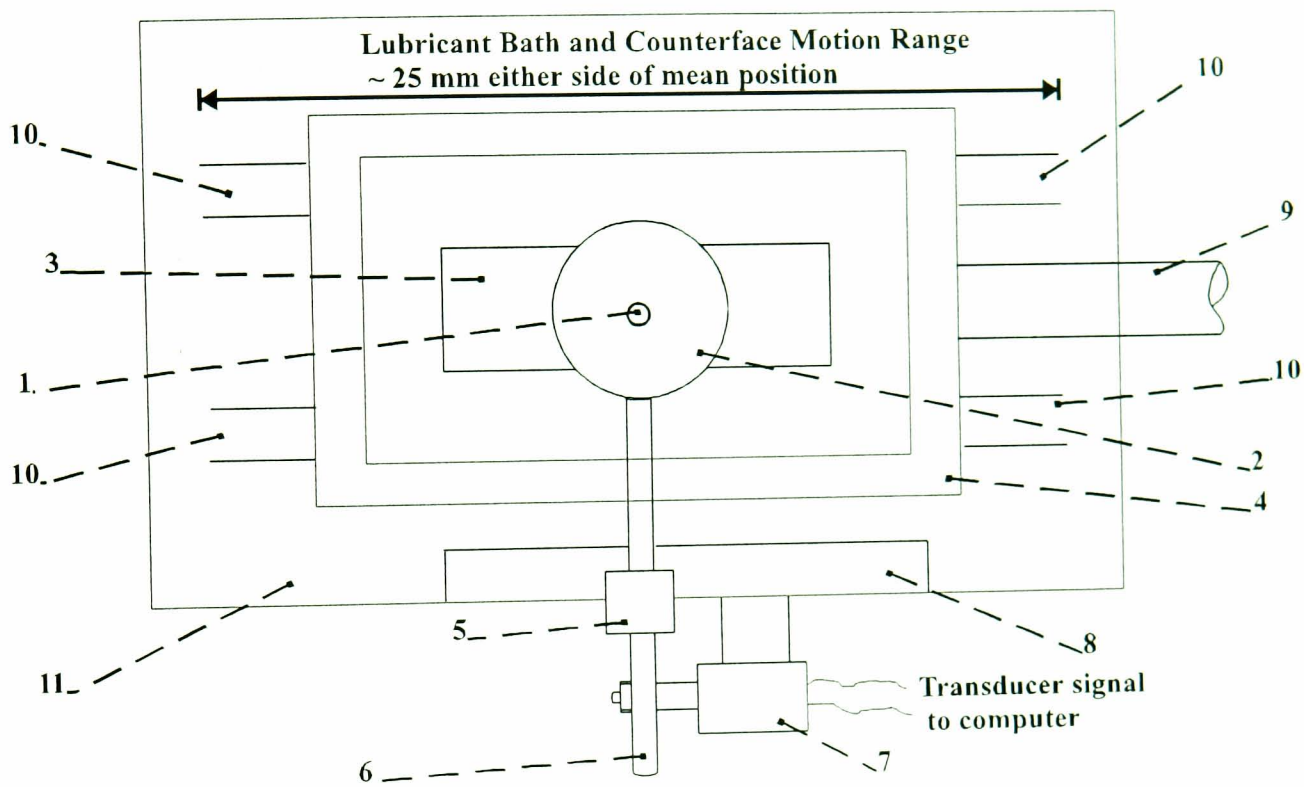


Figure i Schematic drawing of the principal components of the friction apparatus. The diagram is a plan view of the rig.

Key for Figures i and ii:

1 Specimen holder
2 30 N Load (dead weight)
3 Counterface (Metal counterface shown ~30x70 mm^2 and 15 mm in depth)
4 Lubricant bath
5 Air bearing
6 Loading Arm
7 Friction force transducer
8 Elevated fixed frame supporting loading arm
9 Screw thread to motor
10 Runners supporting lubricant bath
11 Fixed frame base
12 Plug (9 mm diameter cartilage plug shown ~15 mm in length largely consisting of subchondral and cancellous bone with a 1-2 mm layer of cartilage)
13 Lubricant (Ringer's solution or synovial fluid)

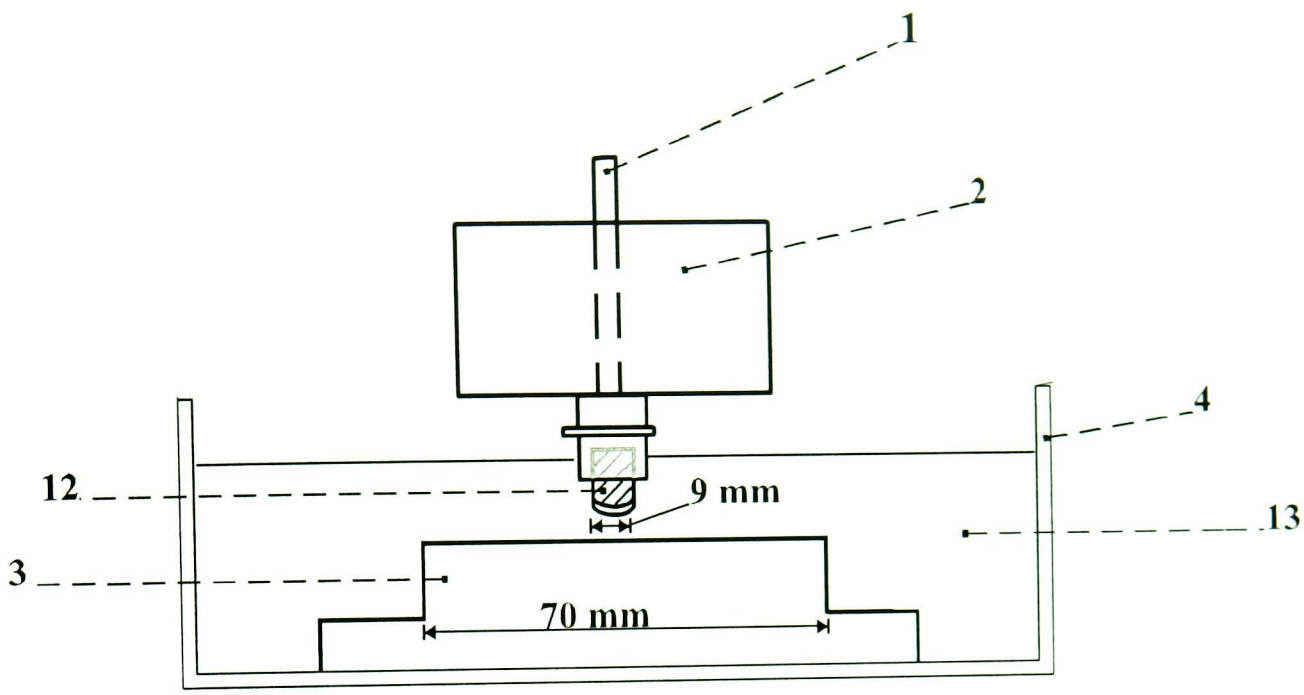


Figure ii This schematic diagram is a side view detailing how the plugs were loaded against the counterfaces.

Also in section 2.4 the calibration procedure for friction force was explained, page 50. Figure iii details the obtained calibration fit between actual and recorded forces.

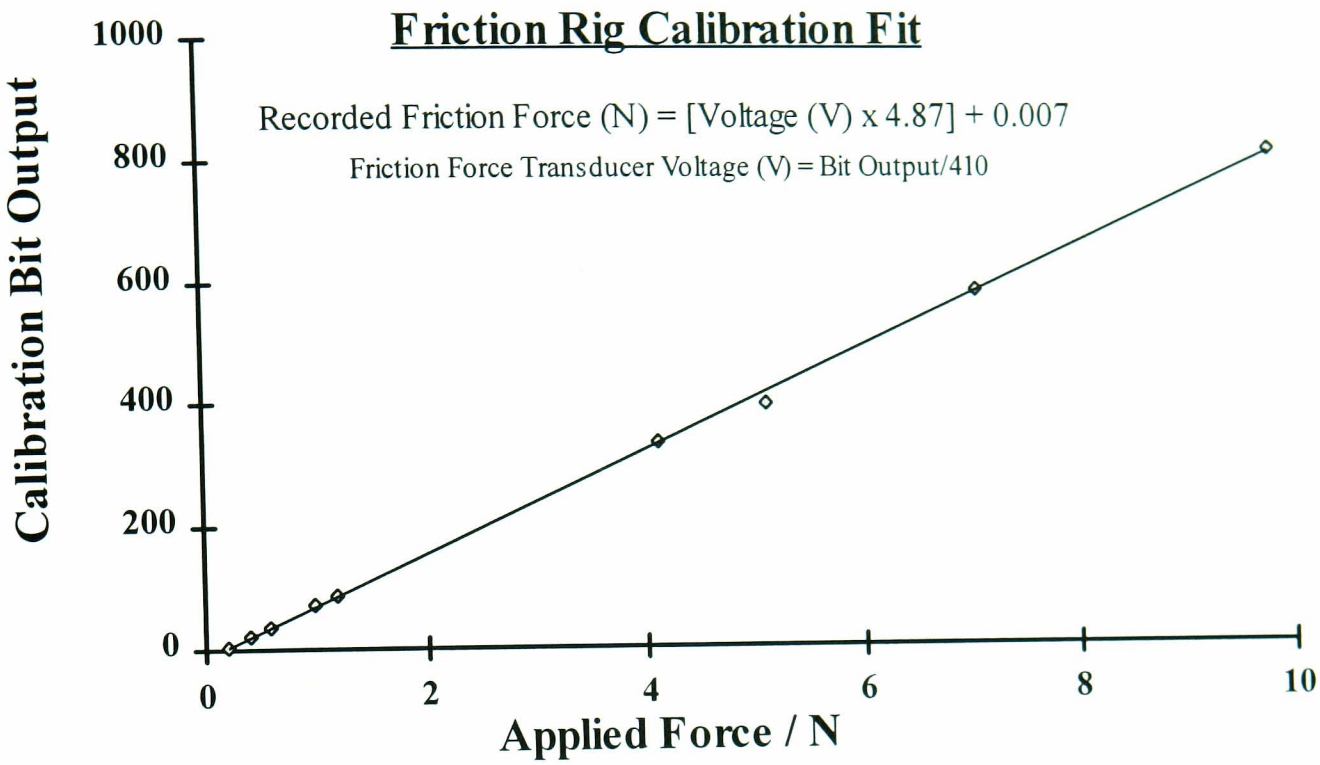
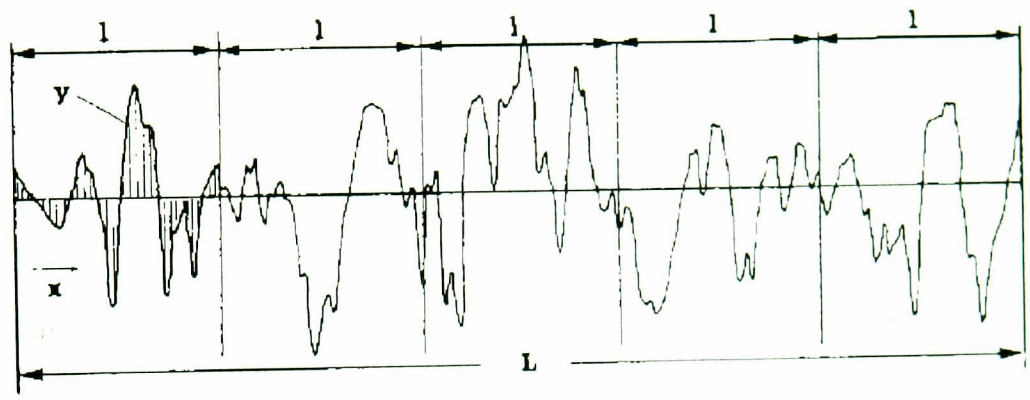


Figure iii Calibration graph from linear least squares fit between applied and computer recorded forces.

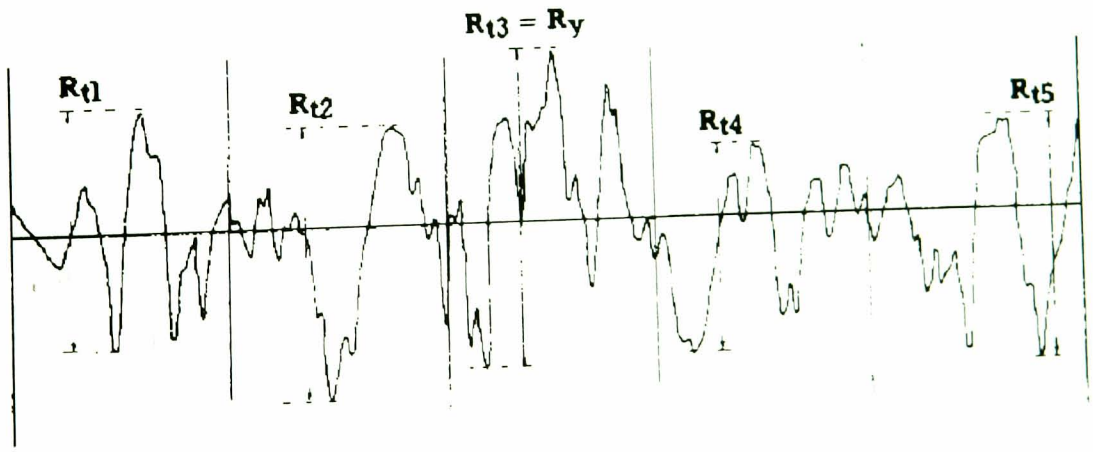
Profilometry

The diagram immediately below represents a profilometry trace of the surface of a material. The profile has been divided into equal sample lengths l, which are long enough to include a statistically reliable amount of roughness, yet short enough to exclude any waviness from the roughness measurement. The assessment length L is defined as the length of the profile used for the measurement of surface roughness parameters.



Surface roughness, R_a, (defined on page 67) can be mathematically expressed as

$$R_a = \frac{1}{L}\int_0^L |y(x)|\,dx$$



R_{ti} is the maximum peak-to-valley height of the profile in one sampling length, as depicted in the above diagram. R_{tm} is the mean of all the R_{ti} values obtained within the assessment length such that

$$R_{tm} = \frac{1}{n}\sum_{i=0}^{i=n} R_{ti}$$