INTRODUCTION

Most modern cementless joints depend on a good press-fit primary fixation, which stabilises the component in the early weeks. This allows bony ingrowth or ongrowth to occur which in turn provides durable long-term fixation. Increased bearing friction in early weeks and months after implantation may lead to micromotion and has the potential to prevent effective bony ingrowth from occurring. Therefore friction in the early postoperative period can be critical to the long-term success of the fixation. This has been one of the concerns raised in a recent clinicoradiological study, of the metal-metal (MoM) bearing with closely controlled (100 µm) clearance [1]. A progressive radiolucent line around the socket component, seen in some of these cases at follow up raises the possibility that increased friction is affecting component fixation [1]. This phenomenon was not observed in the devices with higher clearances. Press-fitting the cup into the acetabulum also results in non-uniform compressive stresses on the cup and consequently causes non-uniform cup deformation. That in turn may result in equatorial contact, high frictional torque and femoral head seizure. The aim of this study was to investigate the effects of cup deformation on friction between the articulating surfaces in MoM bearings with various clearances using clotted blood as lubricant.

The engineering issues surrounding optimal metal on metal bearings have been the centre of much debate and research in the past. Ongoing research into the in vitro wear performance of these bearings as a function of macrogometry (bearing diameter, clearance, and component thickness) and microgeometry (roundness and surface finish) are done in hip motion simulators with lubricants that are believed to simulate the natural fluid in terms of viscosity. However these lubricants have the limitation of being unable to simulate the friction effects of macromolecules.

MATERIALS AND METHODS

Six MoM hip resurfacing devices with a nominal diameter of 50mm each and a range of diametral clearances (80, 135, 175, 200, 243, and 306µm) were used in this study. Frictional measurements of all the joints were carried out on a Prosim Hip Friction Simulator (Simsol Simulation Solutions, Stockport, UK). The acetabular cup is positioned in a low-friction carriage below and the femoral head in a moving-frame above. The carriage sits on pressurized hydrostatic bearings generating negligible friction compared to that generated between the articulating surfaces. A pneumatic mechanism controlled by a microprocessor generates a dynamic loading cycle and the load is measured by a piezo-electric force transducer. Friction measurements were made in the ‘stable’ part of the cycle. The loading cycle was set at maximum and minimum load of 2000N and 100N respectively. In the flexion plane, an oscillatory motion of amplitude ±24° was applied to the femoral head with a frequency of 1 Hz. The angular displacement, frictional torque (T) and load (L) were recorded through each cycle. The frictional torque was then converted into friction factor (f) using the equation $f = T/rL$, where r is the femoral head radius. An average of three test was taken.

Initially the test was conducted with clotted blood as the lubricant for each joint. Then each cup was subjected to unidirectional plastic deformation of between 25-35 µm. The rim of the cup was loaded using a two-point pinching action. The load applied was measured using a load cell, and the deformation was measured using a coordinate measuring machine (CMM). The friction tests were then repeated on the deflected cups.

RESULTS AND DISCUSSION

A force of 80 N applied at the rim of the cup can elastically deform it by approximately 30 µm (Figure 1). Up to 100 µm of deformation can be observed with 400 N load applied. The non-uniform loading which occurs in vivo, due to press fitting, is sufficient to cause elastic deformation, therefore when the components are retrieved, there is no deformation observed. Smaller clearance components show an increased friction coefficient compared to larger clearance components when clotted blood is used as lubricant. The 80 µm clearance had a friction coefficient of 0.19, compared to 0.12 for the 250 µm clearance component (Figure 2). The effects of cup deformation on friction is varied across the clearances. The larger clearance components showed no significant change in friction with deformation. Both the 250µm and 300µm components showed that they were able to accommodate 25-35µm of deformation. However the lower clearance components (80µm) showed a significant increase in friction with deformation. The friction coefficient of the 80 µm clearance component showed friction coefficient as high as 0.22. It appears that the lower pre-deformed clearance bearings do not have sufficient reserve to accommodate the deformation caused by external pressure on the cup.

CONCLUSION

Low clearance components generate high friction when clotted blood is used as the lubricant, which is the case early post operative period. Deformation of the components by 25-35 µm leads to further increase in friction in the low clearance components. However the larger clearance components can accommodate the deformation with no significant change in friction. Increased friction in low clearance bearings may produce micromotion and may hamper bony ingrowth which is essential for long-term for survival.

REFERENCES