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Motion of the bipolar hip prosthesis components

Friction studied in cadavers

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It has been postulated that most hip motion occurs at the inner bearing in bipolar hip prosthesis. However, radiography and cineradiography analyses have shown that the inner bearing is not always the primary articulation. We studied the frictional behavior of bipolar prosthesis and Austin-Moore prostheses and the motion of the bipolar prosthesis using a pendulum apparatus.

The primary articulation was altered according to the amount of loading. When a load of 10 kg was applied to the joint, motion occurred at both bearings with friction coefficients 0.061 at the inner bearing and 0.026 at the outer bearing. With loads of over 20 kg, the outer bearing was the primary articulation. The inner bearing was the dominant articulation only when the acetabular cartilage had been removed.

Our results suggest that the motion of bipolar prostheses occurs mainly at the outer bearing during normal walking, and that this prosthesis cannot be expected to reduce wear of articular cartilage.

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In place of the Austin-Moore femoral head prosthesis, the bipolar prostheses, developed by Bateman (1974) and Gilberty (1974), were introduced to reduce acetabular wear in the hope that movement would occur mainly at the inner bearing.

However, observations of movements by videoradiography or cineradiography are not consistent. Some have reported that the inner bearing was the dominant articulation (Chen et al. 1980, Mess and Barmada 1990) and others said that the outer bearing dominated (Drinker and Murray 1979, Verbaerne 1983, Phillips 1987). We report an experimental analysis of motion between the inner and outer bearing surfaces of the bipolar hip prosthesis.

Material and methods

4 hip joints were obtained at autopsy. Absence of lesions was previously confirmed by radiography, and the friction was measured in each joint by a pendulum machine, without opening the joint capsule. The friction coefficient at a load of 60 kg was between 0.003–0.007, proving that there were no abnormalities in the lubricating condition of the joints used in the experiments. The diameters of the femoral heads ranged from 48.5–52.0 mm. For the hip prostheses, we used the Austin-Moore prosthesis and Bateman UPF II prosthesis with a stem head of 22 mm in diameter equipped with a suitable head size for each test acetabulum.

The prostheses were mounted with acrylic resin (ostron 100) filled in the metal frame which was attached firmly to the pendulum. The acetabulum was also firmly fixed to the pendulum's support arm. Hyaluronic acid solution (1%) was used as lubricating fluid for both bearings.

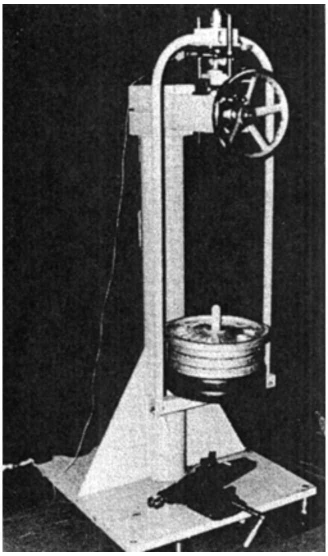
To measure the friction coefficient, a pendulum machine (developed by us) was used (Figure 1). The machine enables one to give a swing and simultaneously apply a load to the pendulum from zero to an arbitrary level. Experiments were always conducted by suddenly applying a load in which the joint was placed at the fulcrum of the pendulum. The initial amplitude was selected at 0.1 radian. The swing was initiated in the direction of extension-flexion. The amplitude decay was determined by a device which measured the angle of inclination (inclinometer), and which was fixed to the frame of the pendulum. The electrical signals from the device were recorded with a pen recorder.

The following formula was used to calculate friction coefficients (see Appendix):

$$f = l \cdot (\Theta_n - \Theta_{n+1}) / (4 \cdot r) = l \cdot \Delta\Theta / (4 \cdot r)$$

where:

f friction coefficient;
Θ_n amplitude of n'th swing cycle;
Θ_{n+1} amplitude of n+1'th swing cycle;
$\Delta\Theta$ decay in amplitude per cycle; $\Delta\Theta = \Theta_n - \Theta_{n+1}$
l distance between the center of gravity and the center of the fulcrum of the pendulum;
r prosthesis head radius.



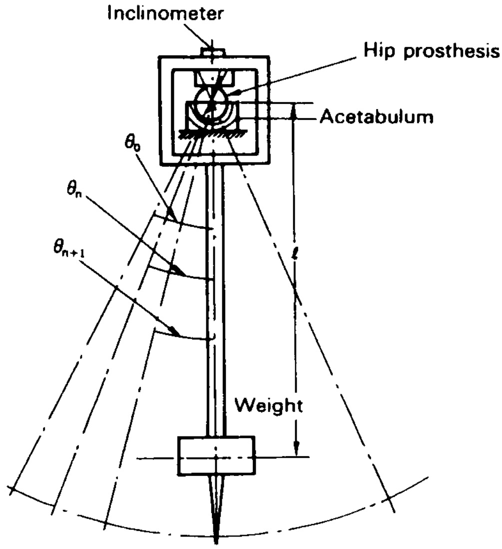


Figure 1. The pendulum used for measuring the frictional resistance. Θ amplitude of a swing cycle and l distance between the center of gravity and the center of the fulcrum of the pendulum

Table 1. Friction coefficient between acetabular cartilage and Austin-Moore prostheses. Mean *SD*

Load (kg)	Friction coefficient	
10	0.035	*0.003*
20	0.026	*0.006*
40	0.021	*0.005*
60	0.020	*0.003*

Table 2. Motion (+/–) and friction coefficient (mean *SD*) of inner bearing and outer bearing at 20-kg load

Time (min)	Stage	Outer bearing			Inner bearing		
		Motion	Frict. coeff.		Motion	Frict. coeff.	
0	1st	+	0.022	*0.004*	–		
2.2	2nd	+	0.026	*0.004*	+	0.060	*0.006*
3.0	3rd	–			+	0.071	*0.005*

10, 20, 40, and 60 kg loads were applied in three series of experiments using an Austin-Moore prosthesis and a bipolar prostheses, respectively; the experiments with the bipolar prostheses being repeated after removal of acetabular cartilage by reaming. Experiments were performed twice for each load.

The frictional torque was calculated using the following formula (Charnley 1979, Krein and Chao 1984).

$$T = F \cdot r = f \cdot L \cdot r$$

where:

T frictional torque;
F frictional force;
f friction coefficient;
L load;
r prosthesis head radius.

Results

For the Austin-Moore prostheses (acetabulae with intact articular cartilage), the mean friction coefficient immediately after 10 kg loading was 0.035 (*SD 0.003*). The coefficient gradually decreased with increases in load (Table 1).

In the bipolar prosthesis, complex movements were observed. With a load of 10 kg, movement of the prosthesis occurred both at the inner and outer bearings and continued until the pendulum stopped. The friction coefficient was 0.061 (*SD 0.007*) at the inner bearing and 0.026 (*SD 0.006*) at the outer bearing.

With loads of 20 kg or more, movement of the bipolar prostheses always occurred at the outer bearing first (1st stage), and then at the inner bearing after a time interval. That is, motion occurred at both bearings (2nd stage), and then movement of the outer bearing stopped while only the inner bearing moved (3rd stage). These results are shown in Table 2.

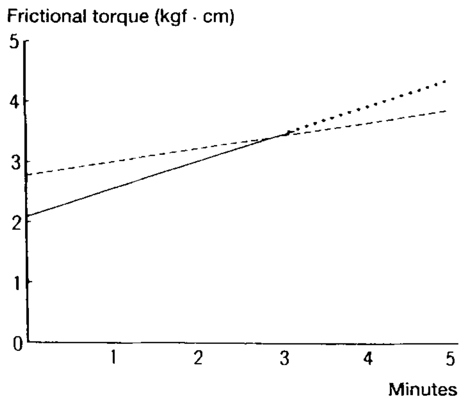


Figure 2. Change of frictional torque at the inner bearing (- - - -) and outer bearing (——) at 60-kg load. ···· surmiced line.

Table 3. Motion (+/–) and friction coefficient (mean *SD*) of inner bearing and outer bearing at 60-kg load

Time (min)	Stage	Outer bearing		Inner bearing	
		Motion	Frict. coeff.	Motion	Frict. coeff.
0	1st	+	0.014 *0.005*	–	
2.2	2nd	+	0.022 *0.004*	+	0.052 *0.005*
3.5	3rd	–		+	0.054 *0.007*
4.0		–		+	0.057 *0.007*
5.0		–		+	0.059 *0.003*

With a load of 60 kg, the friction coefficient was 0.014 (*SD 0.005*) at the 1st stage when only the outer bearing moved. 2 min 30 sec later, it shifted to the 2nd stage where both the bearings moved. Approximately 1 minute later, it shifted to the 3rd stage, where only the inner bearing moved, with a friction coefficient of 0.054 (*SD 0.007*) (Table 3).

After the acetabular cartilage was removed, the outer bearing did not move with any load; only the inner bearing moved. The friction coefficient shortly after a swing was given was 0.068 (*SD 0.004*) at a load of 10 kg, 0.070 (*SD 0.007*) at 20 kg, 0.059 (*SD 0.005*) at 40 kg, and 0.050 (*SD 0.003*) at 60 kg.

The magnitude of the difference of frictional torques governed the occurrence of motion in each bearing (Figure 2 and Table 3).

Discussion

In this study, we wished to clarify frictional behavior between articular cartilage and a metallic outer head, and between a polyethylene bearing insert and a metallic head.

The friction coefficient was not constant between the two bearings, and it changed with load magnitude. If a load, as used in our experiment, is suddenly applied, the friction coefficient is reduced with a greater load. This phenomenon means that the lubricating condition at the two bearings is not boundary lubrication. Moreover, the friction coefficient was lower than that in pure boundary lubrication (Tsukamoto et al. 1983). Judging from the value of the friction coefficient, it is unlikely that complete fluid film lubrication was established. Accordingly, mixed lubrication with partial fluid film and partial boundary contact is assumed as Unsworth et al. (1975) postulated concerning the lubrication mechanism of total hip prostheses. If fluid film lubrication is involved even partially, then the elasticity of articular cartilage and polyethylene may be significant.

With loads beyond 20 kg, movement initially occurred in the outer bearing, then both bearings moved, and later only the inner bearing moved. This observation can be interpreted as the area of fluid film being reduced with time because the fluid was squeezed out, so that the friction coefficient gradually increased, leading to the occurrence of reversed frictional torque between the two bearings.

The maximum load tested in our study was 60 kg, i.e., far less than that occurring during normal walking. However, the friction coefficient at loads of more than 60 kg may not differ markedly from that obtained at 60 kg. The film thickness of lubricant at a load around 160 kg (set as maximum load during walking) was almost the same as at a load around 60 kg in a numerical analysis of lubrication of a total hip prosthesis (Mabuchi and Sasada 1990). Moreover, Mabuchi et al. (1984) showed that damping curves of an oscillating pendulum obtained by theoretical analyses were in good accordance with those in the real pendulum experiments.

The developer of the bipolar hip prosthesis stated that since the friction in the inner bearing is low and, furthermore, the stem head is always smaller than the outer head, the inner bearing is the primary articulation, and the outer bearing begins to move only when the stem neck is impinged at the bearing insert; this impingement occurs in any direction of 50° movement of the inner head. Ordinarily, movement of the hip joint is within 50°, for instance, in normal walking.

Thus, this prosthesis is alleged to protect the acetabular cartilage. However, there is no experimental support for this hypothesis. Moreover, there are very few studies available on friction or lubrication of artificial joints or between joint cartilage and metallic femoral heads.

Krein and Chao (1984) performed mechanical analyses of the bipolar prosthesis. In their study, the distribution of inner and outer head motion was evaluated theoretically. They stated that if the friction coefficient of the inner bearing is equal to twice that of the outer bearing, it would be possible to achieve inner-bearing motion when the head diameter was 22 mm. However, they did not measure the friction coefficient of the prosthesis.

In our experiments, as the load is increased, only the outer bearing begins to move, and the friction coefficient is reduced. We assume this is because, as the load is increased, then fluid-film lubrication becomes more important in the outer bearing with presence of richly elastic articular cartilage than with polyethylene at the inner bearing.

The fact that reports on movement of prostheses to date are variable is explained to some extent by the results of this experiment. In the supine position, almost all reports agree that both the inner bearing and the outer bearing move, corresponding to our case with 10 kg loading in the experiment. In case of weight bearing, there are two different results with the inner bearing (Eiskjær et al. 1989, Mess and Barmada 1990) or the outer bearing (Drinker and Murray 1979, Verberne 1983, Philips 1987) supporting primary articulation. The results are tentatively interpreted as contradictory because either bearing can support primary articulation, depending on the conditions affecting movement.

If the patient moves his hip joint after standing for a certain time, the increase in frictional torque caused by the diminishing area of fluid film lubrication may affect the outer bearing more than the inner bearing. This may possibly lead the inner bearing to support primary articulation, as seen in our experiment where movement with over 20 kg loading was transferred from the outer bearing to the inner bearing after a short time. On the other hand, if the joint begins to move with sufficient lubricant, the outer bearing appears to support primary articulation.

Eiskjær et al. (1989) reported that the function of the Hastings bipolar prosthesis was not correlated to the amount of intraprosthetic motion with weight bearing and only a minor part of the motion in this prosthesis takes place at the inner bearing. In a comparison of clinical results between a bipolar prosthesis (Batemann UPF II: 15 joints) and a unipolar prosthesis (Austin-Moore prosthesis: 28 joints) in our series of 43 joints followed for an average of 4 years, there were no significant differences (Tsukamoto et al. 1988), confirming other reports (Drinker and Murray 1979, Long and Knight 1980) although some authors stated that the results of bipolar prosthesis were superior to those of unipolar prosthesis (Schildhaus 1980, Devas and Hinves 1983, Lestrange 1990).

If it is true that the inner bearing, as the primary articulation, protects against wear of the articular cartilage, efforts to minimize the frictional resistance at the inner bearing should be pursued.

Appendix

The equation for calculation of the friction coefficient is based on the following analysis. (ΔE): The decrease in potential energy within one cycle is given by

$$\Delta E = m\, g\, l\, \{\cos(\Theta - \Delta\Theta) - \cos\Theta\} \qquad \text{(A-1)}$$

ΔE should be used up by friction. The energy lost by friction within one cycle of oscillation is represented by the following equation

$$\Delta E = 4\, r\, m\, g\, f\, \Theta \qquad \text{(A-2)}$$

if r is constant. From equation A-1 and equation A-2 we can state that

$$f = l\, \{\cos(\Theta - \Delta\Theta) - \cos\Theta\}/(4r\, \Theta) \qquad \text{(A-3)}$$

Here, under the condition of $\Delta\Theta << \Theta$, $\Theta << 1$, we can obtain

$$\begin{aligned} &\cos(\Theta - \Delta\Theta) - \cos\Theta \\ &= \cos\Theta \cos\Delta\Theta + \sin\Theta \sin\Delta\Theta - \cos\Theta \\ &\cong \Theta \cdot \Delta\Theta \qquad \text{(A-4)} \end{aligned}$$

From equation A-3 and A-4, we can arrive at equation

$$f = l\, \Delta\Theta\, /(4\, r)$$

where:

$\Delta\Theta$ decay in amplitude per cycle;
Θ amplitude;
r femoral head radius or prosthesis head radius;
m total mass of the pendulum and the weight;
g gravitational acceleration;
l distance between the center of gravity and the center of the fulcrum of the pendulum;
f friction coefficient.

Acknowledgements

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