Lubrication Property of Total Knee Prostheses with PVA Hydrogel Layer as Artificial Cartilage*

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To solve the tribological problems such as wear and loosening in existing knee prostheses composed of ultrahigh molecular weight polyethylene (UHMWPE) and metal or ceramics, a new design was attempted to realize adaptive multimode lubrication in the artificial joint. For this purpose, artificial articular cartilage is required, which has a similar structure and function to natural articular cartilage. In this study, PVA (polyvinylalcohol) hydrogel was used as artificial cartilage in the total knee prostheses. A PVA hydrogel layer was placed on an articulating surface of the tibial component, and their lubrication property during walking was evaluated by the knee joint simulator test. Silicone rubber (another compliant material) and UHMWPE were also tested and their frictional behaviors were compared with that of PVA hydrogel. As a result, PVA hydrogel showed suitable frictional behavior as artificial cartilage.

Key Words: Lubrication, Friction, Biotribology, PVA Hydrogel, Artificial Cartilage, Total Knee Prostheses, Knee Joint Simulator

1. Introduction

An ultrahigh molecular weight polyethylene (UHMWPE) is currently used as the rubbing material against harder materials, such as anticorrosive metals or ceramics, in total replacement joints. UHMWPE is harder than articular cartilages which comprise articulating surfaces of natural synovial joints. Most of artificial joints which consist of such hard materials, except some artificial hip joint⁽¹⁾, are unlikely to preserve an full fluid film between articulating surfaces and they appear to operate in mixed lubrication regime, because little EHL (elastohydrodynamic lubrication) effect is expected. This results in wear of the softer material, i.e. UHMWPE. Wear debris of UHMWPE causes histological changes of the surrounding living tissue and induces the loosening of

joint prostheses^{(2),(3)}.

The application of an artificial cartilage to the surface of total replacement joints is being investigated in several facilities^{(4)–(8)} to solve tribological problems described above. The artificial cartilage has a low elastic modulus and makes easy to form a fluid film between surfaces by soft EHL effect. However, the solid contact is likely to occur at some phases during daily activities even though the artificial cartilage is used, because synovial joints operate under various severe conditions. So the friction and wear of the artificial cartilage must be low even in the mixed lubrication regime.

For these purpose, Murakami et al⁽⁶⁾ suggested that the design concept of the adaptive multimode lubrication should be applied to artificial knee joints. The adaptive multimode lubrication⁽⁹⁾ is the lubrication mechanism of natural synovial joints, and in this lubrication mechanism, EHL enhanced by the elastic deformation of articular cartilages and viscous effect of the synovial fluid is predominant, but when the fluid film is broken down, several supplemental lubrication mechanisms, such as the weeping and boundary lubrication by constituents of the synovia, function

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synergistically. To establish these lubrication mechanisms, a soft, porous nature and high water contents of articular cartilages and a complex composition of synovial fluid are required. So an artificial cartilage material should have similar mechanical and chemical properties and structures to an articular cartilage.

In this study, PVA (polyvinylalcohol) hydrogel was used as an artificial cartilage. PVA hydrogel is a soft, porous material and it contains a lot of water in itself. Conductive silicone rubber was also used as another candidate material for an artificial cartilage. The cylindrical knee prostheses which have an artificial cartilage layer on their tibial component was constructed, and their lubrication properties during walking was evaluated by the knee joint simulator test. The knee prosthesis with conventional UH-MWPE tibial surface was tested too, and their frictional behavior was compared with those of artificial cartilages. And, the importance of geometrical design parameters of total knee prostheses with PVA hydrogel layer was discussed.

2. Experimental Methods

2.1 Knee joint simulator test

Experiments were carried out using the knee joint simulator (Fig. 1). This simulator has two hydraulic actuators controlled by the personal computer. One of them produces the flexion-extension motion of the femoral component (Fig. 2) through the rack-and-pinion system, and another applies the variable axial load to the tibial component. Walking period is 1 s in all tests.

Several kinds of total knee prostheses were tested by this knee joint simulator, and their frictional behavior and fluid film formation during walking were

Fig. 1 Knee joint simulator

examined. The frictional behavior was evaluated by the variation in the torque of the femoral driving shaft which was measured by strain gauges attached to the surface of the shaft. The fluid film formation was examined by the electric resistance method since both PVA hydrogel and silicone rubber used in this study were conductive. The electric circuit used for the measurement is shown in Fig. 3. In this method, degree of separation is defined as the ratio of the measured voltage to the applied voltage. But the degree of separation may be changed with the loading condition during the simulator test because the electric resistance of these artificial cartilage materials are varied with their stress and strain. Therefore, the definition of the degree of separation was modified. Firstly, the simulator test was carried out without lubricant, and the voltage measured in this test was defined as contact voltage. Then the degree of separation was defined as follows; (measured voltage - contact voltage) / (applied voltage - contact voltage).

2.2 Total knee prostheses

The schematics of the total knee prostheses

Fig. 2 Walking condition in knee joint simulator test

Fig. 3 Electric circuit for evaluation of fluid film formation

Fig. 4 Total knee prostheses with artificial cartilage layer

which have an artificial cartilage layer in their articulating surface are shown in Fig. 4. PVA hydrogel and conductive silicone rubber were used as artificial cartilages. PVA hydrogel is the hopeful candidate biomaterial, and it has a few advantages over UHMWPE. First, PVA hydrogel is softer than UHMWPE and an EHL effect is enhanced. Second, the structural feature of PVA hydrogel (porous and high water contents) is similar to that of an articular cartilage, so the supplemental lubrication mechanism, such as a weeping lubrication, may function. And next, its biocompatibility is excellently good. PVA used in this study has an average degree of polymerization of 2 000 and an average degree of saponification of 98.5~99.4 mol%. It was dissolved in a distilled water and gelled by the repeated freezing-thawing method^{(10),(11)}. Silicone rubber is another candidate material for an artificial cartilage because it has a similar elastic property to an articular cartilage. To avoid damages of the soft artificial cartilage in the mixed lubrication regime, the mean contact pressure must be lowered. Therefore the cylindrical design with a good geometric congruity was chosen. Femoral component was made of stainless steel (SUS 316) and the artificial cartilage layer was placed on the surface of the tibial component which was made of UHMWPE. UHMWPE tibial component without artificial cartilage layer was also tested to compare its frictional behavior.

Two design parameters, i.e. the equivalent radius R ($1/R = 1/R_1 - 1/R_2$, R_1 : Femoral radius, R_2 : Tibial radius) and the layer thickness of the artificial cartilage t_n , should be considered when PVA hydrogel is used as an artificial cartilage. The femoral radius was fixed at 30 mm and equivalent radius was altered by changing the tibial geometry. In this study, equivalent radius was adjusted to 30 mm (flat tibial geometry),

Table 1 Mechanical properties

	Elastic modulus MPa	EWC %
PVA hydrogel	0.5	79
Silicone rubber	6	
UHMWPE	800	

*EWC = Equilibrium Water Content

Table 2 Viscosity of lubricants

	Viscosity(16°C) Pa·s
Glycerin 61 wt.% solution	0.011
Glycerin 86 wt.% solution	0.11
Silicone oil S100	0.13
Silicone oil S1000	1.2

180 mm (cup tibial with radial clearance of 6 mm) and 1 830 mm (radial clearance of 0.5 mm). Layer thickness for each case was 3 mm, and for $R=1\ 830$ mm, layer thickness of 6 mm was also tested. Material properties of tibial articulating surfaces were shown in Table 1. For artificial cartilages, the indentation test with the sphere (diameter of 9.5 mm) was performed and elastic modulus was determined by using Hertzian theory. A Poisson ratio of 0.5 was assumed.

2.3 Lubricants

A distilled water and glycerin solutions were used as lubricants. Silicone oils (S 100, S 1000, Number means a kinematic viscosity mm^2/s at 25°C) were also used in several tests. Glycerin solutions are Newtonian and their viscosity could be adjusted by changing the concentration. In this study, glycerin 61 wt.% solution and 86 wt.% solution were used because their viscosity were similar to the viscosity of the secondary synovial fluid⁽¹²⁾ (0.005~0.1 Pa·s) which is a lubricant for a replacement joint in the human body. All tests were carried out at the room temperature ($15 \pm 1^\circ\text{C}$) and the viscosity of these lubricants were shown in Table 2.

3. Theoretical Prediction of Minimum Film Thickness

3.1 Notation

- b : dry contact length
- B : bearing length
- D : bearing inclination
- E : elastic modulus
- F' : load per unit width
- h : minimum film thickness
- h_c : central film thickness
- $h_{\min}$: minimum film thickness under steady state (constant load and constant speed)

l : contact width
 p : fluid pressure
 R : equivalent radius
 t : time
 t_h : layer thickness
 T : torque
 u : entrainment velocity
 U : sliding velocity
 η : viscosity

3.2 PI model

The transient minimum fluid film thickness during the knee joint simulator test was predicted theoretically to compare with the measured fluid film formation. The plane inclined surface model (PI model) proposed by Medley et al⁽¹³⁾ was used in theoretical treatment. This model simplifies the EHL theory for the line contact, and can easily predict the variation of the fluid film thickness under transient conditions.

The variation of the minimum film thickness under the walking condition used in this simulator test (Fig. 2) was calculated for a walking period by using the calculation procedure applied by Gladstone and Medley⁽¹⁴⁾ to thin layered line contact problem. This model assumed that the elastohydrodynamic surface profile was inclined plane and the maximum fluid pressure would occur at the centre of the dry contact length (Fig. 5). Equation (1) is derived from one dimensional Reynolds equation for the plane inclined surface.

$$\frac{dh}{dt} = -\frac{D^3 F' (2h + D)}{12\eta B^3 \left\{ 2D - (2h + D) \ln \left(1 + \frac{D}{h} \right) \right\}} + \frac{DU}{2B} \quad (1)$$

And the maximum fluid pressure occurs at

$$x = \frac{-hB}{2h + D} \quad (2)$$

Since the maximum pressure occurs at the centre of the dry contact length,

$$B = b \left(1 + \frac{D}{2h} \right) \quad (3)$$

where the dry contact width was calculated from the joint geometry (R and t_h) and the elastic modulus which derived from measured contact area at the static loading (2 kN) by using the expressions of Meijers⁽¹⁵⁾.

Fig. 5 Schematic of PI model

Then B and D were decided from the steady state formula

$$F' = \frac{6\eta UB^2}{D^2} \left\{ \ln \left(1 + \frac{D}{h_{\min}} \right) - \frac{2D}{2h_{\min} + D} \right\} \quad (4)$$

and the minimum film thickness under steady state $h_{\min}$ was calculated by Eq.(5) which is taken by Gladstone et al from the numerical analysis of Hooke⁽¹⁶⁾.

$$\bar{h}_{\min} = 2.5 - 5.0g_3^{0.8} + 6.630 \left(\frac{b}{2t_h} \right)^{-0.1225} g_3^{0.8189} \quad (5)$$

where

$$h_{\min} = \frac{h_{\min} F'}{\eta u R} \quad (6)$$

$$g_3 = \sqrt{\frac{0.375}{\eta u E R}} F' \quad (7)$$

Then Eq.(1) was solved by Runge-Kutta method to obtain the minimum film thickness h . Where the time step of an integration was 0.004 s which was the same as the data sampling interval in the simulator test. When the sliding speed U was nearly zero, Eq.(1) was reduced to

$$\frac{dh}{dt} = -\frac{F' h^3}{\eta B^3} \quad (8)$$

From the shear stress relation for a Newtonian fluid, frictional torque was assumed that

$$T = \eta b l R \frac{U}{h_c} \quad (9)$$

where

$$h_c = h + \frac{Db}{2B} \quad (10)$$

4. Results and Discussion

4.1 Fluid film formation

For the total knee prostheses which had a PVA hydrogel as an artificial cartilage, a few results of the measured fluid film formation and the variation of the minimum film thickness calculated from the PI model were shown in Fig. 6. Since PI model is a two-dimensional model, effects of side leakage are neglected. Therefore the actual film thickness appears to be thinner than the calculated value.

When the equivalent radius was 30 mm and the lubricant was S 100 (Silicone oil), the degree of separation decreased continuously during the stance phase, and reached minimum at 50% of cycle. It means that the fluid film was squeezed out and its thickness became the thinnest at the end of the stance phase. This experimental result agreed with theoretical analysis, but the slight drop of the separation at 15% of cycle could not be predicted. This decrease in the separation might be due to the complicated surface deformation caused by the squeeze action, because it occurred at the stroke end of the femoral motion and the peak of the tibial load. PI model assumed that the surface profile was plane, so this decline could not

Fig. 6 Comparison between fluid film formation in knee joint simulator test and minimum film thickness calculated from PI model

be predicted.

The increase in the equivalent radius reduces the mean contact pressure and enhances the squeeze effect. Therefore the decrease in the fluid film thickness during stance phase is suppressed in theoretical analysis. The experimental result is consistent with this theoretical prediction, the separation drop during stance phase is suppressed and a fluid film formation is improved through a cycle. With increase in the viscosity of the lubricant, the degree of separation was improved well, but the separation drops at 15% and 50% of cycle were still remained.

These results of experiments and theoretical analysis give the following indication. The lubrication film thickness was minimized and the lubrication condition was getting severe at the end of the stance phase under the walking condition. So, the frictional behavior at this phase is the most important to improve the frictional property of the total knee prostheses. Since considerable agreement was shown

Fig. 7 Viscous effect in datum torque

between the variation of the degree of separation measured experimentally and the variation of the theoretical minimum film thickness, the degree of separation could be a measure of the fluid film formation. And in these experiments mentioned above, full fluid film was almost formed.

4.2 Frictional behavior

4.2.1 Relative torque A torque was measured by strain gauges at the surface of the femoral driving shaft. The measured torque was caused by not only the friction between femoral and tibial components but also the inertia, the bending and eccentric load. So, the measured value of the previous test, in which the tibial articulating surface of a silicone rubber was lubricated with S100 and the full fluid film formation between articulating surfaces was confirmed by the degree of separation, was selected as a datum. This datum included the frictional torque caused by the viscous resistance of the lubrication film, but its value was expected to be very small from PI model (Fig. 7). Then, the relative value of measured torque to the datum was defined as relative torque. This corresponds to the frictional torque induced by solid contacts. In this study, the variation of the relative torque was used to evaluate the frictional behavior of knee prostheses during walking.

4.2.2 The effect of an artificial cartilage A comparison of the frictional behavior among three articulating surface materials, i.e., PVA hydrogel, UHMWPE and silicone rubber, is shown in Fig. 8. All types have equivalent radius of 30 mm, and distilled water and glycerin 61 wt.% solution are used as lubricants. For silicone rubber, degree of separation is also revealed. In Fig. 9, coefficient of friction for three tibial components at near the end of stance phase (from 35 to 45% of cycle), where the rubbing condition is the severest, are compared in Stribeck curves for two lubricants. Here, load in the abscissa means the load per unit width.

For the glycerin 61 wt.% solution, PVA hydrogel and silicone rubber exhibit a lower friction than UHMWPE, and silicone rubber shows a fluid film formation to some extent. This means that a compliant artificial cartilage makes easy to preserve a fluid film between articulating surfaces and improves the

Fig. 8 Comparison of frictional behavior among three materials

Fig. 9 Comparison of frictional behavior at end of stance phase

frictional behavior of the total knee prostheses by enhancing an EHL effect. For distilled water with extremely low viscosity, however, silicone rubber could not form a fluid film at stance phase and its frictional behavior became worse rapidly. As shown by these result, silicone rubber can improve a lubrication mode of the total knee prostheses when it is lubricated with viscous liquid, but once the solid contact occurs with decreasing viscosity, it exhibits the unacceptable high friction and a surface damage or a dislocation of a soft artificial cartilage may occur

Fig. 10 Effect of equivalent radius on theoretical minimum film thickness (PVA, Viscosity of lubricant : 0.11Pa·s)

under walking conditions. So silicone rubber is not suitable for an artificial cartilage. On the other hand, PVA hydrogel shows a better frictional behavior than UHMWPE even if distilled water is used as lubricant. Therefore, it appears that PVA hydrogel can be a good artificial cartilage.

4.2.3 The effect of geometric design parameters

The cylindrical knee prosthesis which has a PVA hydrogel layer as an artificial cartilage has two important geometric design parameters, i.e., equivalent radius and layer thickness. The equivalent radius is a prime design parameter because it controls the contact area and the contact pressure. PI model predicted that the larger equivalent radius produces the thicker fluid film (Fig. 10). Moreover, the strength of PVA hydrogel is so low that the equivalent radius must be large to lower the contact pressure and prevent damages of the PVA hydrogel layer. The layer thickness also has an influence upon the contact condition and affects the equivalent modulus and a stress distribution in the artificial cartilage layer.

The effects of these design parameters on the frictional behavior is shown in Fig. 11 for two different lubricants. When equivalent radius increases from 30 mm to 180 mm, relative torque decreases for both lubricants. But if it increases to 1830 mm, frictional behavior of layered cylindrical knee joint becomes worse slightly. There is a possible explanation for this result. PVA hydrogel is a very soft material, so the contact width becomes large excessively when the equivalent radius becomes too large, and the wedge effect of EHL loses its function. Too large layer thickness also increases a relative torque. When the layer thickness is increased, the equivalent modulus is getting smaller and a strain of the artificial cartilage layer increases. This results in the considerable hysteresis loss in the soft layer, and frictional behavior becomes worse. PVA hydrogel used in this study does not have a sufficient mechanical strength for an artificial cartilage material yet. So design parameters should be chosen carefully to avoid surface and bulk damages of PVA hydrogel when it is used in knee prostheses. The equivalent radius and

Fig. 11 Effect of geometric design parameters on frictional behavior of PVA hydrogel

layer thickness should be large sufficiently to lower the contact stresses and strains. However, the experimental results show that the excessive large equivalent radius and layer thickness have a harmful effect upon the frictional behavior. Therefore, the optimum design considering both the durability and lubrication must be achieved when PVA hydrogel is used as an artificial cartilage practically.

4.2.4 Comparison with theory For three types of total knee prostheses with artificial cartilages, each coefficient of friction at 45% of cycle, where lubrication condition becomes severe during walking, is compared with the minimum film thickness predicted from the theoretical analysis in Fig. 12. The relationship between the minimum film thickness and coefficient of friction shows the advantage of PVA hydrogel over silicone rubber clearly.

Silicone rubber increases its coefficient of friction steeply when the minimum film thickness is decreased below $1\mu\text{m}$. Then a film parameter⁽¹⁷⁾ becomes less than one third, where the r.m.s. roughness of silicone rubber and stainless femoral component are $3.3\mu\text{m}$ and $0.02\mu\text{m}$, respectively. The roughness of silicone rubber was measured by the replica method. For a soft material like silicone rubber, surface asperities can be deformed easily by a fluid pressure and the

Fig. 12 Relation between coefficient of friction and minimum film thickness calculated by PI model

Fig. 13 Surface image of PVA hydrogel taken by low vacuum scanning electron microscope

practical surface roughness may become smaller. Moreover micro EHL effect⁽¹⁸⁾ may function too. But some direct contacts must have happened in this condition. On the contrary, the increase in coefficient of friction for PVA hydrogel is gentle even if the film thickness is decreased below $1\mu\text{m}$. For PVA hydrogel, a r.m.s. surface roughness could not be determined by the replica method and ambient water will swell its surface structure. But a fibrillar structure with a diameter of a few μm could be observed on its surface by the low vacuum scanning electron microscope (Fig. 13). So, like silicone rubber, the solid contact may occur, too. Once solid contacts occurs at surface asperities, a water may be exuded from PVA hydrogel by the pressure gradient near the surface, and adhesion between surfaces can be prevented⁽¹⁹⁾, by so called weeping lubrication. In the previous paper⁽²⁰⁾, the authors have confirmed the weeping effect that water exuded from PVA hydrogel covered a conjunction and friction was lowered. This supplemental lubrication mechanism is likely to contribute the advantage of PVA hydrogel over silicone rubber. But, other factors, such as a chemical property and a microstructure, may also affect the frictional behavior, so detailed examination must be carried out.

PVA hydrogel has shown the suitability for an artificial cartilage, but coefficient of friction is raised to about 0.05 when fluid film thickness decreases to

Fig. 14 Effect of sodium hyaluronate contained in PVA hydrogel

less than $1\text{ }\mu\text{m}$. A normal natural synovial joint does not show such an increase of friction under walking conditions. So the frictional behavior of PVA hydrogel must be improved further to survive as long as a natural synovial joint. For this purpose, the authors have tried to give adaptive multimode lubrication mechanism to the artificial cartilage. If an additional lubricant, such as a boundary lubricant, is contained in PVA hydrogel, it may be exuded with water when surfaces come into contacts, and it protects contacting surfaces^{(4),(21)}. In this study, sodium hyaluronate (molecular weight $=1.02 \times 10^6$) was added to PVA hydrogel (0.1 wt.%), and its effects on the frictional behavior was examined. The results at the stance phase are shown in Fig. 14. Relative torque can be reduced slightly by sodium hyaluronate but the frictional behavior at the 45% of cycle, where the tribological condition is the severest, is not improved. Like this result, sodium hyaluronate can not form a firm boundary lubrication layer on the surface of PVA hydrogel. In natural synovial joints, it is known that hyaluronic acid contributes to the viscous nature of synovial fluid⁽²²⁾ and serum proteins⁽²³⁾ and phospholipids⁽²⁴⁾ can play a role of boundary lubricants. So the effect of these constituents of synovia on the frictional behavior of PVA hydrogel must be investigated in detail. And further works are required to realize adaptive multimode lubrication in a total knee prostheses.

5. Conclusions

Some conclusions from this study are as follows.

The fluid film formation measured experimentally corresponds to the variation of the minimum film thickness predicted from PI model qualitatively. But, the PI model can't predict a slight drop of the measured film formation at the stroke end of the swinging motion because the detailed surface deformation is neglected in this model. For the low viscous (0.01 Pa·s) lubricant, such as secondary synovial fluid, an artificial cartilage makes easy to preserve fluid film between articulating surfaces. And it can improve the lubrication mode of a total knee prostheses.

Silicone rubber shows the unacceptable high fric-

tion in the mixed lubrication regime. On the contrary, PVA hydrogel keeps low friction even if some asperity contacts occur. Therefore, PVA hydrogel is more suitable for an artificial cartilage material.

In total knee prostheses with PVA hydrogel layer, a large equivalent radius and considerable layer thickness are required to get the sufficient durability but the excessive equivalent radius and layer thickness have a harmful effect upon the frictional behavior. Therefore the optimum design for both the durability and lubrication must be achieved to use PVA hydrogel as an artificial cartilage practically.

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