Physical Properties of Ultra-High Molecular Weight Polyethylene in Retrieved Artificial Knee Joint

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(Received October 31, 2000; Accepted November 14, 2000)

Abstract: Physical properties of ultra-high molecular weight polyethylene (UHMWPE) in retrieved artificial knee joints were investigated. Lamella thickness, crystallinity, and mechanical chain scission appeared to increase during service. Crystallinity-depth profile was related to the conformity of joint and the thickness of tibial component. For the joint of low conformity, stress, crystallinity, and lamellae alignment were highest at subsurface 1-2 mm beneath the surface, where crack propagation and delamination eventually occurred. It was suggested that the thickness and conformity of UHMWPE part should be increased for better wear-resistant tibial component of knee joint prosthesis.

Keywords: Ultra-high molecular weight polyethylene (UHMWPE), retrieved artificial knee joints, crystallinity-depth profile, conformity, thickness

1. Introduction

The average span of life of artificial knee joint is between 5 and 15 years, and the considerable portion undergoes re-operation in less than 10 years [1]. There are several reasons for the failure of prosthesis; infection, instability, fracture, pain, loosening, and wear of substitute cartilage. The biggest problem is the excessive wear of cartilage material, for which ultra-high molecular weight polyethylene (UHMWPE) is currently used. It is therefore desired to minimize the wear of UHMWPE in order to develop prosthesis with prolonged span of life. In the present study, physical properties of the UHMWPE tibial parts retrieved in the re-operation of failed artificial knee joint were examined. In the course, the reasons for the failure were to be elucidated, and suggestion for better wear-resistant prosthesis would be submitted.

2. Experimental

The characteristics of the retrieved tibial components examined were listed in Table 1. The thickness was measured at the condyle part. For MGs and PCAs, which are of metal-backed design, the thickness was measured with the metal tray on. Two types of raw UHMWPE were used. One was the medical grade UHMWPE powder (Hizex 240 M

<table>
<thead>
<tr>
<th>Sample</th>
<th>Conformity</th>
<th>LOS† (month)</th>
<th>Thickness‡ (mm)</th>
<th>Melting Point (°C)</th>
</tr>
</thead>
<tbody>
<tr>
<td>MG1</td>
<td>flat on flat</td>
<td>60</td>
<td>8.5</td>
<td>138</td>
</tr>
<tr>
<td>MG2</td>
<td>flat on flat</td>
<td>58</td>
<td>11</td>
<td>137</td>
</tr>
<tr>
<td>PCA1</td>
<td>flat on flat</td>
<td>90</td>
<td>7</td>
<td>137</td>
</tr>
<tr>
<td>PCA2</td>
<td>flat on flat</td>
<td>108</td>
<td>9</td>
<td>136</td>
</tr>
<tr>
<td>Bps</td>
<td>full</td>
<td>50</td>
<td>15</td>
<td>137</td>
</tr>
<tr>
<td>Pre-implantation</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>134</td>
</tr>
<tr>
<td>Rod</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>134</td>
</tr>
<tr>
<td>Molded plate</td>
<td>-</td>
<td>-</td>
<td>-</td>
<td>133</td>
</tr>
</tbody>
</table>

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‡ Length of service, which means the period implanted in vivo
§ Thickness of tibial component
supplied by Mitsui Petrochemical Industrial Co.), which was compression molded to plates at 220 °C and 6000 psi for 20 min. Bulk UHMWPE was supplied in the form of a rod by The Hospital for Special Surgery/Poly Hi Solidur, which was ram-extruded GUR415 resin from Hoechst-Celanese.

The samples were cut to 0.5-mm thick pieces and the physical properties were investigated. Thermal properties were measured using a Perkin-Elmer DSC-4. Thermograms were obtained during heating ca. 5 mg of samples up to 170 °C at 10 °C/min. Heat of fusion of UHMWPE was calculated based on the heat of fusion of 100% crystalline polyethylene, which is 292 J/g [2]. Crystal structure was examined using a MacScience M18XHF-SRA diffractometer. Diffractograms were obtained in the range of 5° ≤ 2θ ≤ 40° using Cu Kα radiation at 50 kV and 100 mA (λ = 1.5045 Å). Gel content was determined by measuring the weight of gel after boiling the specimen in p-xylene (containing 0.5 wt% 2,6-di-tert-butyl-4-methyl phenol as an antioxidant) for 72 h, extracting p-xylene in acetone for 24 h, and dried at 60 °C for 24 h.

Thin sections containing subsurface cracks were observed using a Leitz Laborlux optical microscope. One side, which is supposed to be perpendicular to the plane of crack propagation, of the sample was ground, polished, and glued onto a slide glass using epoxy adhesive. The other side of the sample was ground to give a thin section with a thickness of ca. 200 μm and polished. Fracture surface was etched with KMnO4 and examined with a JEOl JSM-840A scanning electron microscope. The condition for the etching suggested by Olley and Bassett was followed to minimize the artifact by the experiment [3].

### 3. Results and Discussion

A total of 48 artificial knee joints retrieved in re-operation practice were examined. Only a few representatives were listed in Table 1. The relation between design and mode of failure will be discussed elsewhere, and only the physical properties are discussed in the present report. Melting temperature of UHMWPE was higher for the retrieved joints than for the unused or raw material by a few degrees. The increase in melting temperature was thought to be the result of lamella thickening during the service in the human body. As it would become apparent below, the thickening was considered to be due to the annealing effect of applied stress.

Crystallinity-depth profiles of the UHMWPE were shown in Figure 1. Degree of crystallinity was higher for the UHMWPE in the retrieved components than for those from unused UHMWPE. It was evident that the increase in crystallinity occurred during the period of service in the human body, since the crystallinity of UHMWPE in the pre-implantation components was not much different from that in the virgin materials, as shown in Figure 1(b). Pre-implantation part is UHMWPE that were processed, machined and sterilized, and packaged for practice.

The profile appeared to be related to conformity of femoral and tibial part. For MGs and PCAs, which are of low conformity, the crystallinity showed its maximum at the depth between 1 and 2 mm beneath the surface. The crystallinity profile did not show a peak for IBPs, which is an all-polymerimplant.

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J. Ind. Eng. Chem., Vol. 6, No. 6, 2000
with high conformity, although the crystallinity was high down to 2-mm depth. The results agreed with the previous observations; the density profile showed a peak in the used tibial components of knee implants, while density is highest at the surface of acetabulum component of hip joint implant, where conformity is much higher than knee joint [4-6].

The above results could also be explained by the extent and state of stress experienced by prosthesis. It was found that, with the same applied load, the contact stress on the surface of low conformity was always higher than that of high conformity, which could be understood by the smaller contact area for high conformity contact [7]. Bartel and coworkers investigated the state and distribution of stress in prosthesis using finite element method [8,9]. Maximum principal stress was highest at the surface for both of femoral part of knee joint and acetabulum of hip joint prostheses. Maximum shear stress, however, was highest at a sub-surface, 1-2 mm from surface, for femoral part of knee joint, while it was highest at surface for acetabulum. The peak in crystallinity profiles for MGs and PCAs and the highest crystallinity at the surface for 1Bps in the present study could be explained by the distribution of maximum shear stress, which resulted in the distribution of deformation and crystallinity.

Density profile was related to the thickness of tibial part also. As shown in Figure 2, for the thinner tibial components, the crystallinity was higher and the location of maximum crystallinity was farther from the surface. It has been observed that the maximum stress increased with decreasing thickness of UHMWPE on metal tray, and the increment became much larger for the thickness lower than 6 mm [9,10]. Through a study with retrieved knee implants, Wright and coworkers suggested the optimum thickness for tibia as 6-8 mm [11]. It was thought that, as the polymer part became thinner, the stress experienced by the component increased. As a result crystallinity became higher, and the location of maximum crystallinity moved to a deeper subsurface.

Figure 3 showed the crack propagating at subsurface and the fracture surface of subsurface crack, which eventually developed to delamination. The depth of maximum crystallinity and that of delamination crack were approximately the same. Konvopoulos and coworkers have shown, through a transmission electron microscopy work, that lamellae aligned to the direction parallel to surface at the location where delamination occurred [12]. It was therefore thought that the lamellae of UHMWPE are aligned to the direction of maximum shear stress at subsurface. The accumulated deformation and alignment might eventually cause the crack and delami-
nation at the same location.

Ohta and coworkers reported, through the wear experiments, that (200) plane of orthogonal unit cell structure of UHMWPE was oriented to the direction of the wear surface [13]. Butler and coworkers also found that, when polyethylene yielded, (100)[001] chain slip and (200) plane orientation to the direction of stress occurred [14]. Figure 4 shows the relative intensities of (200) plane peaks measured at surface, subsurface, and bottom of a PCA-type UHMWPE tibial component. It was observed that the intensity was highest at the subsurface, and it coincided with the relative crystallinity. It was therefore affirmed that stress, crystallinity, and the alignment of lamellae were the highest at subsurface, and the chance of failure was also the highest at subsurface.

For lamella thickening to occur during the service, the chains in the amorphous or interfacial region should newly crystallize. Since the very large number of entanglements in UHMWPE might not allow the chain to participate in the crystallization without scissoring of chains. Chemical degradation [16] by gamma ray sterilization is also possible in addition to mechanical chain scission during the service [17]. Since the radicals formed by scission would cause crosslinking of chains, gel content was measured. For the retrieved joints, gel content in the condyle (more stressed) region was lower, than that in the intercondyle (less or not stressed) region, and the difference became larger as the length of service increased. There was no such difference in gel content for the preimplantation samples. It was therefore considered that, in the early stage of service, crosslinking due to the sterilization and chemical degradation might occur. As the period of service increased, mechanical scission by applied stress became dominant, resulting in the decrease in gel content.

The results of the present study suggested an approach for developing a better wear-resistant prosthesis. To prevent the accumulation of shearing, deformation, and alignment of lamellae in a narrow region at subsurface, the thickness of polyethylene part should be thick and conforms well to the femoral part. It was further desired that the tibial part is free of metal tray, which would enlarge the stress and confine the thickness of polymer part.

4. Conclusions

By examining the physical properties of UHMWPE in the retrieved artificial knee joint and comparing them with those of raw UHMWPE and preimplantation samples, lamella thickness, crystallinity, and mechanical chain scission were determined to increase by applied stress during the service. Crystallinity-depth profile was related to the conformity of joint and the thickness of tibial component. For the joint with low conformity, stress, crystallinity, and lamellae alignment was highest at a layer at subsurface, where crack propagation and delamination eventually occurred. It was suggested that the thickness and conformity of UHMWPE part should be increased for better wear-resistant tibial component of artificial knee joint.

Acknowledgment

The authors wish to acknowledge the financial support of the Korea Research Foundation made in the program year of 1997.

References