MECHANICAL PROPERTIES OF ARTICULAR CARTILAGE IN KNEES WITH UNICOMPARTMENTAL OSTEOARTHRITIS

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We studied the mechanical properties of cartilage from the apparently unaffected compartment of knees with unicompartamental osteoarthritis (OA). Plugs of cartilage and subchondral bone, 8 mm in diameter, were obtained from the tibial plateau of seven patients treated by total knee replacement. Control specimens were obtained from eight cadaver knees of similar age. Specimens were loaded in a plane-ended indentor in a hydraulic materials testing machine; we measured thickness, 'softness', rate of creep, and compressive strength of the articular cartilage.

We found that the 'unaffected' cartilage from OA knees was significantly thinner and softer than control cartilage, and it was slightly, although not significantly, weaker. We conclude that the apparently unaffected cartilage in knees with unicompartamental OA is mechanically inferior to normal cartilage, even although clinically, radiologically and morphologically it appears to be sound.

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In knee surgery for unicompartamental osteoarthritis (OA) one compartment is sometimes replaced on the assumption that the other is normal. This assessment of integrity is based on clinical and radiological judgement and on visual appraisal of the articular surface. Corpe and Engh (1990) consider that changes in limited areas of articular cartilage are acceptable in this context, but this has not been supported by biochemical or biomechanical evidence. Armstrong and Mow (1982), however, concluded that the visual and histological appearance of cartilage is a poor indicator of its ability to function in an intact joint. Because of this uncertainty about the quality of the 'unaffected' cartilage we studied its mechanical properties, using a range of tests to assess its ability to withstand the high forces present in a functioning joint.

MATERIALS AND METHODS

We studied eight patients diagnosed as having unicompartamental osteoarthritis of the knee on clinical assessment, supported by radiography. The knees were considered to be suitable for unicompartamental replacement, but had total replacement as part of a randomised prospective trial. At operation we were able to collect the entire tibial plateau with its subchondral bone. In each case it was confirmed that the cartilage on the other 'unaffected' compartment appeared to be normal.

One of the eight specimens was damaged during preparation and was therefore discarded, leaving seven for the tests, four from female and three from male knees. The average age of the patients was 63 years (47 to 79; Table I). Eight control specimens were collected from the normal knees of patients who had died from unrelated causes; their average age was 66 years (38 to 82; Table I). Specimens were stored at −20°C in sealed plastic bags.

Table I. Source of specimens obtained at surgery and the control knees

<table>
<thead>
<tr>
<th>Sex</th>
<th>Age (yr)</th>
<th>Knee</th>
<th>Affected compartment</th>
</tr>
</thead>
<tbody>
<tr>
<td>Surgery</td>
<td>M 61</td>
<td>Right</td>
<td>Lateral</td>
</tr>
<tr>
<td>M 47</td>
<td>Left</td>
<td>Medial</td>
<td></td>
</tr>
<tr>
<td>F 79</td>
<td>Right</td>
<td>Medial</td>
<td></td>
</tr>
<tr>
<td>M 55</td>
<td>Right</td>
<td>Medial</td>
<td></td>
</tr>
<tr>
<td>F 73</td>
<td>Right</td>
<td>Medial</td>
<td></td>
</tr>
<tr>
<td>F 65</td>
<td>Left</td>
<td>Medial</td>
<td></td>
</tr>
<tr>
<td>F 65</td>
<td>Right</td>
<td>Medial</td>
<td></td>
</tr>
</tbody>
</table>

| Control | M 67 | Right and left |
|---------| M 80 | Right and left |
| M 82 | Right and left |
| M 38 | Right and left |

For mechanical studies, plugs of cartilage and attached subchondral bone were cut from the 'unaffected' tibial plateau while still frozen, using an 8 mm diameter trephine. Approximately 1 to 2 mm of subchondral bone was retained, the remainder being removed with a fine saw. Two plugs were taken from each compartment, one from the weight-bearing area at the centre of the plateau.

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and the other from a ‘non-weight-bearing’ area in the posteromedial or posterolateral part of the plateau (Maquet 1984). A total of 60 plugs was obtained. The thicknesses of the cartilage and of the subchondral bone were then measured to an accuracy of 0.1 mm, using a Vernier caliper. Three readings were taken from different sites on each plug, and the mean thickness calculated.

Each plug was thawed at room temperature for five minutes in a bath of physiological saline. It was carefully blotted dry to remove any excess moisture and mounted in a metal cup containing dental plaster (Fig. 1). Tapping the edge of the cup assisted plaster penetration into the subchondral bone and enhanced fixation. Towards the end of the five-minute setting period, a 5 mm diameter plane-ended metal indentor was used to apply a 5 to 10 N load to the specimen to ensure that the surface of the cartilage was exactly parallel to the indentor when the plaster had finally set. The surface of the plaster was at the bone-cartilage junction, so that the cartilage was not restrained laterally. This is an important detail because the lateral restraint has a marked effect on compressive properties (Mizrahi et al 1986; McPherson et al 1990).

Each plug was tested on a Dartec computer-controlled hydraulic materials testing machine, being immersed in a shallow bath of physiological saline at 21°C to keep the cartilage moist and prevent tissue dehydration (Elmore et al 1963). Three mechanical properties were assessed; the experimental methods are described and justified in the Appendix.

**Softness.** A linear ‘ramp’ load was applied up to a maximum stress of about 7 MPa and down again, in 1.0 sec, at an approximate displacement rate of 0.5 to 1.0 mm/sec. Compressive force and vertical displacement were recorded at a frequency of 500 Hz and stored on a microcomputer. Cartilage softness was then calculated from the gradient of the graph (see Appendix).

**Creep.** A constant compressive stress of 2 MPa was applied in 1.0 sec and retained for 300 sec. Loss of specimen height was recorded continuously at a frequency of 1.7 Hz.

**Compressive strength.** Five minutes were allowed for the specimen to recover from the creep test, and the plug was then compressed to failure, using a displacement rate of 1 mm/sec. The load-displacement curve was plotted in real time and the test was stopped when this curve showed a marked reduction in resistance to compression.

When testing had been completed, the cartilage was cut away from a number of specimens so that the underlying bone could be examined for macroscopic damage.

**RESULTS**

Typical force-deformation curves for a cartilage-on-bone plug loaded in compression (Fig. 2) show a marked increase in stiffness (gradient) at high load indicating that the material is non-linear. Cartilage ‘softness’ was calculated from the gradient of the curve at a force of approximately 140 N, which corresponds to a stress of 7 MPa. This is a measure of compressive deformability per unit thickness of cartilage (see Appendix). Preliminary

![Diagram of the apparatus used to test the plugs of cartilage-on-bone.](image)

**Fig. 1**

Typical force-deformation curve for a plug of cartilage-on-bone loaded in compression. The arrows indicate the direction of loading. The second loading cycle was more severe than the first, but at low loads, successive loading curves are closely reproducible. The gradient indicates specimen stiffness. The specimen was from the weight-bearing area of the medial condyle.

**Fig. 2**
tests had shown that the deformation of the apparatus and the plaster-reinforced subchondral bone accounted for less than 10% of overall deformation; this contribution was therefore ignored.

During 'creep' tests, cartilage plugs lost height rapidly (Fig. 3); again, preliminary tests had shown negligible creep in the subchondral bone and apparatus. All the height loss was attributed to the cartilage and expressed as a percentage of the original thickness.

The 'compressive strength' of cartilage was recorded as the compressive stress at which damage was first detected (Fig. 4). The nature of this damage was not ascertained, but the subchondral bone was grossly intact in all specimens from which the cartilage had been cut at the end of the test.

The results for the control cartilage specimens are summarised in Table II. In these, cartilage from weight-bearing areas was stronger than that from 'non-weight-bearing' areas, and lost less height during the creep test. These differences were less in the lateral compartment. Medial compartment cartilage was thinner and softer than lateral cartilage, and tended to creep less.

Table III compares 'unaffected' cartilage from OA knees with the control cartilage. Data have been pooled for weight-bearing and non-weight-bearing specimens, but we have excluded medial compartment control data because the 'unaffected' compartment was the lateral one in six of our seven cases. We found that unaffected cartilage was on average 22% thinner and 71% softer than control cartilage and these were statistically significant differences. It was also 12% weaker on average, but this difference was not significant.

We calculated the age-dependence of the mechanical properties because there was a slight discrepancy between the ages of 'unaffected' and control material. Linear regression analysis showed that compressive strength

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**Table II.** Comparison of medial and lateral compartment cartilage from normal knees. Values are mean ± sd. The significance of differences was established by a 1-tailed t-test

<table>
<thead>
<tr>
<th>Compartments</th>
<th>Area*</th>
<th>Medial (n=8)</th>
<th>Lateral (n=8)</th>
<th>p value of difference</th>
</tr>
</thead>
<tbody>
<tr>
<td>Thickness (mm)</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>WB</td>
<td>2.40±0.49</td>
<td>3.29±0.78</td>
<td>0.004</td>
<td></td>
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<tr>
<td>NWB</td>
<td>2.16±0.35</td>
<td>3.16±0.49</td>
<td>0.001</td>
<td></td>
</tr>
<tr>
<td>Softness (1/N)</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>WB</td>
<td>0.39±0.14</td>
<td>0.31±0.15</td>
<td>&gt;0.1</td>
<td></td>
</tr>
<tr>
<td>NWB</td>
<td>0.47±0.22</td>
<td>0.31±0.11</td>
<td>0.040</td>
<td></td>
</tr>
<tr>
<td>Creep (per cent)</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>WB</td>
<td>9.9±4.5</td>
<td>12.4±5.0</td>
<td>0.006</td>
<td></td>
</tr>
<tr>
<td>NWB</td>
<td>14.2±6.7</td>
<td>15.1±6.2</td>
<td>&gt;0.1</td>
<td></td>
</tr>
<tr>
<td>Strength (MPa)</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>WB</td>
<td>14.8±4.1</td>
<td>15.3±3.7</td>
<td>&gt;0.1</td>
<td></td>
</tr>
<tr>
<td>NWB</td>
<td>10.8±4.8</td>
<td>13.7±2.7</td>
<td>0.057</td>
<td></td>
</tr>
</tbody>
</table>

* WB, weight-bearing; NWB, non-weight-bearing

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**Table III.** Comparison of apparently unaffected cartilage from OA knees with cartilage from the lateral compartment of control knees. The data are pooled for weight-bearing and non-weight-bearing areas, values are mean ± sd, and the significance of differences was established by a 1-tailed t-test

<table>
<thead>
<tr>
<th></th>
<th>Unaffected (n=14)</th>
<th>Control (n=16)</th>
<th>p value of difference</th>
</tr>
</thead>
<tbody>
<tr>
<td>Thickness (mm)</td>
<td>2.51±1.04</td>
<td>3.23±0.63</td>
<td>0.019</td>
</tr>
<tr>
<td>Softness (1/N)</td>
<td>0.53±0.35</td>
<td>0.31±0.13</td>
<td>0.020</td>
</tr>
<tr>
<td>Creep (per cent)</td>
<td>12.5±6.5</td>
<td>13.8±5.6</td>
<td>0.29</td>
</tr>
<tr>
<td>Strength (MPa)</td>
<td>12.7±6.8</td>
<td>14.5±3.3</td>
<td>0.18</td>
</tr>
</tbody>
</table>

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Fig. 3

Creep curve for a plug of cartilage-on-bone loaded in compression for five minutes. After this time, the specimen has lost 0.4 mm, about 13% of its original thickness.

Fig. 4

We defined the compressive strength of a plug of cartilage-on-bone as the stress at which damage first occurs; it is marked by a sudden reduction in gradient, and a displacement to the right of the subsequent loading curve. In this example the strength was 291 N divided by the cross-sectional area of the indenter.

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decreased slightly with increasing age ($R^2 = 0.19$, $p = 0.11$, $n = 15$) but none of the other properties showed any apparent age-dependence. The slight age difference is therefore unlikely to be of any consequence. Gender did not significantly affect any of the mechanical properties.

**DISCUSSION**

Any method of measuring the mechanical properties of articular cartilage introduces some degree of artifact; this is discussed in the Appendix. In our experiment, the use of a plane-ended indenter must lead to high stress gradients around its edge. Even when intact joint surfaces are pressed together, however, articular cartilage has only a limited ability to equalise compressive stress, and it is not unusual to find an area of uniformly high stress surrounded by steep stress gradients (Afoke, Byers and Hutton 1987). The stress distributions acting on the cartilage in our experiments, therefore, would not have been grossly unphysiological. It seems probable that we underestimated the compressive strength of articular cartilage, but this is difficult to assess because little has been reported about true compressive strength.

Our mechanical tests were designed to measure the ability of cartilage to function normally under high compressive loads. Cartilage *thickness* has a strong influence on its ability to equalise stresses between opposing bone surfaces, and marked thinning is one of the features of early OA. *Creep* is a measure of how rapidly water is expelled from the cartilage under sustained loading. Excessive water expulsion would prevent cartilage from behaving as a hydrostatic cushion (Broom and Oloyede 1993). Excessive *softness* impairs the ability of cartilage to equalise stress at high loads; this occurs very early in animal models of joint degeneration (Lane, Chisena and Black 1979; Hoch et al 1983) and is a feature of human OA (Roberts et al 1986). *Strength* (stress at failure) indicates the vulnerability of cartilage to damage at high loads. Thus the mechanical properties that we measured are important for the normal function of articular cartilage, and the inferior properties shown by ‘unaffected’ cartilage from the OA knees may well have clinical implications.

Unicompartamental knee replacement in selected cases has been justified previously by the work of Brocklehurst et al (1984). They analysed the water content, fixed charge density, and sulphate incorporation rate of cartilage from arthritic human knees, and concluded that OA changes are focal. They did not perform mechanical tests, however, to confirm that apparently normal areas of cartilage could function normally at high load levels. In some cases, there is failure of the contralateral compartment after unicompartamental knee replacement (Laskin 1978; Insall and Aglietti 1980), although this may be due in part to polyethylene debris from the unicompartamental prosthesis. Our study suggests that the visually normal cartilage of tibial plateau in the ‘unaffected’ compartment may not be mechanically sound, and that in some cases unicompartamental replacement may not be appropriate.

One limitation of our study is that no histological tests were done to confirm the visual normality of control and ‘unaffected’ cartilage specimens. Brocklehurst et al (1984), however, found a good correlation between the results of histology and the visual appearance of human knee articular cartilage, so that this omission is unlikely to be important.

In the control knees, we found that medial compartment cartilage was inferior to that in the lateral compartment (Table II). OA usually affects the medial compartment much more (Ahlbäck 1968; Hernborg and Nilsson 1977; White, Ludkowski and Goodfellow 1991; Hodge and Chandler 1992), and it is therefore possible that our control specimens were showing preclinical changes in the medial compartment which in some cases might have developed into OA.

**APPENDIX - Justification of experimental methods**

There is no ideal way of assessing the mechanical properties of articular cartilage. Cartilage is a structure rather than a material (Jeffery et al 1991) which makes it impossible to remove small samples for testing without disrupting its structural integrity. On a larger scale, mechanical experiments on whole joints are difficult to interpret because the properties of cartilage vary with location on the joint surface, and they may be masked by the properties of subchondral bone. We tried to steer a middle course: we tested large cartilage-on-bone plugs, aiming to assess the local properties without too much disruption.

**Size of the plugs.** The larger the plug, the less disruption of the collagen network, but plugs larger than about 8 mm in diameter do not have an acceptably flat surface and should not be used with a plane-ended indenter (see below). Removal of a plug allowed us to trim the subchondral bone so that its mechanical properties would not obscure those of the cartilage. This is not possible when an indenter is applied to an intact joint surface.

**Size and shape of the indenter.** We chose a plane-ended indenter, in order to apply even compressive stress. The plane end gives artificial stress gradients around its periphery, but the relative importance of these decrease as the diameter of the indenter increases, and so we used the largest possible indenter. The cartilage plugs were 8 mm in diameter, and the use of a 5 mm indenter ensured that the loaded cartilage had some natural lateral constraint from the peripheral 1.5 mm unloaded rim of cartilage. Without some lateral constraint, cartilage develops a characteristic (and artificial) ‘barrel’ deformation under compression. We used an impervious indenter because under physiological conditions, fluid cannot flow out of one cartilage surface and into the opposing cartilage: this would violate symmetry. Tissue fluid must flow tangentially, or into spaces between the opposed surfaces.

**Magnitude of applied loads.** In a normal synovial joint, articular cartilage is subject to high and rapidly fluctuating compressive loads; its functional ability must therefore be assessed under similar circumstances. For this reason, we performed the tests on a powerful hydraulic materials testing machine; stress levels in the stiffness and creep tests were similar to those measured on the human hip during simulated walking (Afoke et al 1987).

**‘Softness’ rather than stiffness.** Softness was calculated from the formula:

$$\text{softness} = \frac{1000}{\text{thickness} \times \text{stiffness}}$$

This is the inverse of stiffness, normalised for the thickness of the cartilage, while the multiplication factor of 1000 gives convenient numbers. The inverse of stiffness is a measure of deformability and ‘softness’ is therefore...
deformability per unit thickness of cartilage. We prefer to use ‘softness’ rather than stiffness because it is less sensitive to variations in the mechanical properties of the subchondral bone.

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REFERENCES


