MECHANICAL TESTS ON THE TIBIAL COMPONENTS OF NON-HINGED KNEE PROSTHESES


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Cadaveric knees replaced with the Geomedic, ICLH, Marmor and Total Condylar prostheses were tested in axial compression, in rotation and in hyperextension in order to observe the strength of fixation of the tibial components. In axial compression the strengths at failure varied widely, both with any one prosthesis and between prostheses. This is attributed largely to the strength of the cancellous bone of the tibia, which was measured in each case and also varied widely. Three natural knees failed at loads of 7300, 7600 and 8300 newtons respectively, whereas the strengths of replaced knees ranged from 3000 to 15 750 newtons. At least one example of each design failed at less than 7300 newtons, suggesting little or no reserve of strength. The strength of fixation was greater when the tibial prosthesis was large enough to rest on the whole cross-section of the tibia. In rotation the three prostheses embodying rollers in troughs were stiffer than the Marmor which had a nearly flat tibial-bearing surface. The presence or absence of the cruciate ligaments had a negligible effect on torsional stiffness. In hyperextension, knees replaced with the ICLH, Marmor and Total Condylar prostheses failed by rupture of the posterior capsule at moments of about 60 newton-metres, compared with about 100 for natural knees. With the Marmor prosthesis the anterior cruciate ligament was avulsed at about 20 newton-metres compared with about 75 in natural knees, suggesting that in this respect the retention of the cruciate ligaments contributes little. None of the four knees tested after inserting a Geomedic prosthesis showed strengths as high as those replaced with the other three designs.

Knee prostheses consisting essentially of hinges have been used for about twenty-five years, but more recently the use of non-hinged prostheses has increased rapidly. It is therefore desirable to know something of their mechanical properties as a guide to the selection of a design.

At the time of writing, a limited amount of mechanical testing has been reported. Walker, Ranawat and Insall (1976) tested plastic tibial plateaux in compression after cementing them to cadaveric tibiae, and found that they bent elastically when implanted on soft bone but to an extent that was not thought significant in terms of fixation. They found also that a metallic tibial plateau, when loaded by a vertical compressive force of 840 newtons applied towards the front or the back, tilted so as to open a gap of up to 0.5 millimetre at the back or front respectively; this gap did not open when holes were made into the cancellous bone to improve the grip of the cement. Nogi et al. (1976) tested three Geomedic and three Polycentric knee prostheses, implanted in amputated knees, in compression and torsion. They found that no failures occurred in the femoral components, and that fixation of the tibial components failed at vertical forces ranging from 637 to 2720 newtons with the Polycentric prostheses and from 1700 to 3230 newtons with the Geomedic, all of which loads were within the physiological range. Some of the failures occurred when rotation had been imposed under compressive load, but the turning moments were apparently not measured.

The purpose of the work described in this paper was to examine types representative of a wider range of non-hinged prostheses in three modes of loading—compression, torsion and hyperextension—and to compare the strengths of the tibial components implanted in cadaveric knees with the loads applied in life or the corresponding performance of natural cadaveric knees. Attention was confined to the tibial components because clinically these loosen more frequently than the femoral components in the types of replacement considered. In the compression tests loads were applied direct to the tibial component, but for torsion and hyperextension tests the entire prosthesis was implanted and tested.
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Fig. 1

The four types of prosthesis tested. Above, from left to right, the ICLH, with original smaller tibial component, the Geomedic and the Total Condylar. Below, the later, larger tibial component of the ICLH, and the Marmor.

MATERIALS

Prostheses. Four types were used—ICLH, Geomedic, Total Condylar and Marmor (Fig. 1). These were selected from the large number on the market because they are all used in considerable numbers and because they present a wide range of design. The ICLH and the Geomedic each have conforming articular surfaces, but whereas the Geomedic is intended to be inserted with retention of the cruciate ligaments, the ICLH requires these to be removed. The Total Condylar and the Marmor each have non-conforming articular surfaces, the Marmor having almost flat tibial surfaces; the Marmor allows the cruciate ligaments to be retained but the Total Condylar does not. The Marmor has two separate femoral components and two separate tibial components, whereas the other three each have one. The Geomedic femoral and tibial components engage only part of the cross-sections of the femur and tibia, but the ICLH and Total Condylar components, having no gaps to accommodate the cruciate ligaments, present larger areas of contact to the bones. The Total Condylar tibial component has a short intramedullary stem. The ICLH knee was tested with both the original smaller tibial component and with the later, larger one (see Fig. 1). Between them these four designs contain most of the significant variations found in other types of non-hinged knee prostheses.

Cadaveric knee joints. These were obtained when the cause of death did not suggest abnormality of the bone or soft tissues; all were normal to the naked eye. About 200 millimetres each of the tibial and femoral shafts were retained; the skin was removed but all other soft tissues were present. The specimens were stored at −18 degrees Celsius and thawed to room temperature on the day of testing. Altogether, thirty-six specimens, from subjects aged from twenty-two to eighty-three years, were used.

METHODS

Implantation of prostheses. Each type was implanted as described by the originator; the implanting surgeon was able to satisfy himself that the alignments achieved would have been acceptable clinically. All the preparations that included a total prosthesis were checked for laxity by the application of a distracting force of 90 newtons with the joint in extension; none showed any visible distraction.

Indentation tests on bone. Because the strength of cancellous bone varies widely, indentation tests were performed on most of the tibiae after they had been sectioned to receive the prosthesis. This provided
some indication of the strength of the particular tibia at the level at which load would be transmitted to it.

The test used a Rockwell conical indentor made of steel (Fig. 2), but a simpler method than the usual Rockwell procedure was used. The indentor was pressed into the bone at a constant rate until a force of 98 newtons was reached after a few seconds, the force and depth of indentation being continuously recorded and the average compressive stress under the indentor calculated.

With the Geomedic and Marmor prosthesis four indentations were made, two medially and two laterally, each 10 millimetres in from the cortex. With the ICLH and Total Condylar prostheses, two further indentations were made centrally. A few knees were rejected because their indentation strengths were unusually high, quite unlike that to be expected in rheumatoid or osteoarthritic knees. One tibia was used solely for indentation testing at different levels in order to gain some idea of the variation of strength with increasing distance from the articular surface.

Compression tests. Tibial preparations alone were used. The shaft was held in a suitable clamp so that the articular surface of the component was horizontal, and an acrylic cement pad was cast *in situ* to fit it. Force was applied through this pad in the testing machine, the axis of application passing through the centre of the total area of each prosthesis. From a continuous record the instants of first and subsequent failures were noted.

Torsion tests. The tibial and femoral shafts were cast into metallic pots using plaster of Paris. With the joint in extension and neutral rotation, these pots were then placed in a torsion testing rig which was in turn placed in a compression-testing machine (Fig. 3). Thus turning moments could be applied simultaneously with compressive forces. Rotation of the femur relative to the tibia and vertical displacement of the femur relative to the tibia were both measured as the turning moment was increased from zero to 24 newton-metres while the vertical compressive force was held constant at one, two or three times the body weight.

RESULTS AND COMMENTS

The various findings are set out in Tables I to VIII.

The indentation strength of the tibia. As expected, there was a wide variation from 1.1 to 5.2 meganewtons per square metre for the average result on any one tibia (Tables I and II). This reflects the differences in strength of the cancellous bone upon which the tibial prostheses tested were placed. The more proximal the level of section, the stronger was the cancellous bone.

The lateral tibial condyles were noticeably weaker than the medial, and the central portions even more so. This suggests that although a prosthesis may engage a larger area of the bone, the increase in strength should not be expected to be proportional if the extra bone at the interface is central. Furthermore, any stud on the undersurface of the tibial component is best placed centrally, as in the Total Condylar prosthesis, though in this position it is relatively inefficient with regard to the transmission of twisting moments.

Compressive strengths. In an attempt to eliminate the effect of bone strength, the compressive strengths achieved with various prostheses were recorded in five different tibiae which happened to have approximately

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**Fig. 3**
A line diagram of the apparatus used for torsion testing.

**Fig. 4**
A line diagram of the four-point bending rig used for hyperextension tests.
the same indentation strengths at the relevant level of section (Table III). It will be seen that the large-area ICLH was the strongest, which might have been predicted from a consideration of the surface areas and the fact that it rests upon all the tibial cortex. The Total Condylar prosthesis was the second strongest; although it is smaller, it transmits some load to the cortex of the proximal shaft through the cemented intramedullary stem. The small-area ICLH has an interface of the same area as the Total Condylar and is about as strong; it rests only on the posterior tibial cortex, except in very small knees in which it was not tested here and for which alone it is now used clinically. The Marmor prosthesis, although having the smallest area of all, was not the weakest because the level of section is very close to the articular surface. The Geomedic was the weakest in compression, partly because it has a relatively small surface area and partly because it is placed more distally than the Marmor.

Table IV presents the ranges of compressive strengths measured with each design. These ranges overlap considerably, but the general conclusion can be drawn that tibial components which engage the largest practicable area of bone, including all the cortices, and which make use of the stronger bone available proximally, are more likely to be strong in compression.

Morrison (1970) has shown that, in the normal knee, axial compressive loads of up to three times the body weight are transmitted during normal level walking. If it may be assumed that fatigue strength is one third of static strength, then the replaced joint, which will be cyclically loaded, should be capable of withstanding a load of about nine times the body weight, equivalent to a force of about 6300 newtons. Tables III and IV show that with the weakest tibiae all of the designs tested failed to attain this strength, and that the Geomedic attained it only on an unusually strong tibia. It is interesting to note that the three natural knees tested gave compressive strengths about halfway up the ranges found with all the prostheses except the Geomedic, and comfortably above the 6300 newtons suggested above as a notional level giving freedom from fatigue failure.

Table I. Indentation stresses at the level of the prosthesis used in twenty-five tibiae

<table>
<thead>
<tr>
<th>Area indented</th>
<th>Number of indentations</th>
<th>Indentation stress (MN/m²)</th>
<th>Average</th>
<th>Range</th>
</tr>
</thead>
<tbody>
<tr>
<td>Medial</td>
<td>49</td>
<td>3.3</td>
<td>0.8 to 9.3</td>
<td></td>
</tr>
<tr>
<td>Central</td>
<td>31</td>
<td>1.43</td>
<td>0.7 to 3.6</td>
<td></td>
</tr>
<tr>
<td>Lateral</td>
<td>50</td>
<td>2.46</td>
<td>0.7 to 5.4</td>
<td></td>
</tr>
<tr>
<td>Average result for any one tibia</td>
<td>4 to 6</td>
<td>2.51</td>
<td>1.1 to 5.2</td>
<td></td>
</tr>
</tbody>
</table>

Table II. Indentation stresses at four levels of section in one tibia

<table>
<thead>
<tr>
<th>Area indented</th>
<th>Number of indentations</th>
<th>Average indentation stress (MN/m²) at given depth (mm) from articular surface</th>
</tr>
</thead>
<tbody>
<tr>
<td>Medial</td>
<td>6 total</td>
<td>4.4 3.3 3.5 1.6</td>
</tr>
<tr>
<td>Central</td>
<td>6 total</td>
<td>3.1 2.5 2.4 -</td>
</tr>
<tr>
<td>Lateral</td>
<td>7 total</td>
<td>4.2 4.3 2.4 1.6</td>
</tr>
<tr>
<td>Average for level of section</td>
<td>3.9 3.4 2.7 1.6</td>
<td></td>
</tr>
</tbody>
</table>

Table III. Compressive strengths of five designs of prosthesis in selected tibiae of similar indentation strengths

<table>
<thead>
<tr>
<th>Prosthesis</th>
<th>Area of interface (mm²)</th>
<th>Average indentation stress of tibia (MN/m²)</th>
<th>Compressive strength of tibial preparation (N)</th>
</tr>
</thead>
<tbody>
<tr>
<td>ICLH (large)</td>
<td>3100</td>
<td>2.65</td>
<td>11 000</td>
</tr>
<tr>
<td>Total Condylar</td>
<td>2220</td>
<td>2.1</td>
<td>7 000</td>
</tr>
<tr>
<td>ICLH (small)</td>
<td>2200</td>
<td>2.55</td>
<td>6 500</td>
</tr>
<tr>
<td>Marmor</td>
<td>1040</td>
<td>2.6</td>
<td>6 500</td>
</tr>
<tr>
<td>Geomedic</td>
<td>1710</td>
<td>2.65</td>
<td>3 000</td>
</tr>
</tbody>
</table>

Table IV. Compressive strengths of five designs of prosthesis and of natural knees

<table>
<thead>
<tr>
<th>Prosthesis</th>
<th>Area of interface (mm²)</th>
<th>Average indentation stress of tibia (MN/m²)</th>
<th>Compressive strength of tibial preparation (N)</th>
</tr>
</thead>
<tbody>
<tr>
<td>ICLH (large)</td>
<td>3100</td>
<td>-</td>
<td>15 400</td>
</tr>
<tr>
<td>Total Condylar</td>
<td>2220</td>
<td>2.65</td>
<td>11 000</td>
</tr>
<tr>
<td>ICLH (small)</td>
<td>2200</td>
<td>3.65</td>
<td>12 000</td>
</tr>
<tr>
<td>Marmor</td>
<td>1040</td>
<td>3.3</td>
<td>12 000</td>
</tr>
<tr>
<td>Geomedic</td>
<td>1710</td>
<td>2.1</td>
<td>7 000</td>
</tr>
<tr>
<td>Natural knees</td>
<td>-</td>
<td>2.2</td>
<td>9 800</td>
</tr>
<tr>
<td></td>
<td>-</td>
<td>3.2</td>
<td>7 100</td>
</tr>
<tr>
<td></td>
<td>-</td>
<td>2.6</td>
<td>6 500</td>
</tr>
</tbody>
</table>
Table V. Rotational stiffness of knees replaced with various prostheses and loaded in axial compression at 1 x body weight combined with a turning moment of 5.7 newton-metres

<table>
<thead>
<tr>
<th>Prosthesis</th>
<th>Number of knees tested</th>
<th>Average rotation (degrees)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td>Internal</td>
</tr>
<tr>
<td>Geomedic</td>
<td>2</td>
<td>1.5</td>
</tr>
<tr>
<td>Total Condylar</td>
<td>1</td>
<td>3.0</td>
</tr>
<tr>
<td>ICLH (small)</td>
<td>2</td>
<td>3.0</td>
</tr>
<tr>
<td>ICLH (large)</td>
<td>3</td>
<td>6.0</td>
</tr>
<tr>
<td>Marmor</td>
<td>1</td>
<td>14.0</td>
</tr>
<tr>
<td>Natural knee</td>
<td>1</td>
<td>3.0</td>
</tr>
</tbody>
</table>

Table VI. Strengths of natural and replaced knees in torsion compared with some turning moments transmitted in life

<table>
<thead>
<tr>
<th>Turning moment (newton-metres)</th>
<th>Observation</th>
</tr>
</thead>
<tbody>
<tr>
<td>17</td>
<td>Natural knee, with only cruciate ligaments and bone present, rotated uncontrollably under a vertical load about twice body weight</td>
</tr>
<tr>
<td>17 to 28</td>
<td>Femoral components of prosthetic knees twisted out of tibial components under a vertical load of body weight</td>
</tr>
<tr>
<td>15</td>
<td>Foot slipped on floor under body weight</td>
</tr>
<tr>
<td>20</td>
<td>Pain in knee of weight-bearing leg of a man aged fifty-nine</td>
</tr>
<tr>
<td>60</td>
<td>Pain in knee of weight-bearing leg of a man aged twenty-three</td>
</tr>
</tbody>
</table>

Table VII. Rotational stiffness of one knee replaced with the ICLH prosthesis and loaded in axial compression at 1, 2 and 3 x body weight combined with a turning moment of 5.7 newton-metres

<table>
<thead>
<tr>
<th>Compressive load (multiples of body weight)</th>
<th>State of capsule</th>
<th>Rotation (degrees)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>Present</td>
<td>4.5 6.0 10.5</td>
</tr>
<tr>
<td>2</td>
<td>Present</td>
<td>1.5 2.0 3.5</td>
</tr>
<tr>
<td>3</td>
<td>Present</td>
<td>0.5 1.0 1.5</td>
</tr>
<tr>
<td>1</td>
<td>Excised</td>
<td>6.5 7.5 14.0</td>
</tr>
<tr>
<td>2</td>
<td>Excised</td>
<td>2.0 1.0 3.0</td>
</tr>
<tr>
<td>3</td>
<td>Excised</td>
<td>0.5 1.0 1.5</td>
</tr>
</tbody>
</table>

Strength and stiffness in rotation about a vertical axis. In prostheses that have a roller-in-trough geometry, as have all the prostheses in this study other than the Marmor, rotation of the femoral component tends to lift it away from the tibia as the roller climbs up the trough, and thereby to tense the soft tissues connecting the femur and the tibia. If the prosthesis has been implanted with these tissues tense in neutral rotation, as all prostheses were in this study, the soft tissues will resist any tendency for the tibia to rotate on the femur. Reference to Table V shows that the Geomedic was the stiffest of the prostheses. The Marmor prosthesis was markedly less stiff, rotating four times more at the turning moment of 5.7 newton-metres than did the normal knee. Thus the presence or absence of the cruciate ligaments had very little effect on the stiffness of the replaced knee, as both these prostheses retained the cruciate ligaments. The factor that had the greatest effect upon rotational stiffness was the shape of the trough. In this respect the Geomedic, the Total Condylar and the ICLH all had stiffness close to that of the natural knee.

Table V is concerned only with comparisons between the different prostheses at one turning moment. All the tests were continued until the femoral component twisted out of the tibial component, which occurred at turning moments of from 17 to 28 newton-metres with a vertical load equal to body weight. In Table VI these values are listed, together with the results of other torsional tests which are self-explanatory. These results suggest that the four designs tested have a reasonable margin of strength in torsion in relation to the turning moments that can be transmitted across their articulating surfaces, and also in relation to the turning moments that can be transmitted through a shoe or cause pain in the knee of an elderly person. The moment sustained by a young adult male is included for interest, but none of the prostheses tested could transmit one so high.

The effect of varying the axial compressive load upon the rotational stiffness of a prosthesis having a roller-in-trough geometry is shown in Table VII. At higher compressive loads, less rotation occurs and what does occur depends less on the constraint provided by the capsule.

Table VIII. The strength in hyperextension of knees replaced with various prostheses

<table>
<thead>
<tr>
<th>Prosthesis</th>
<th>Number tested</th>
<th>Mean bending moment at failure (newton-metres)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td>Anterior cruciate avulsion</td>
</tr>
<tr>
<td>Geomedic</td>
<td>4</td>
<td>31.7</td>
</tr>
<tr>
<td>ICLH (large)</td>
<td>3</td>
<td>—</td>
</tr>
<tr>
<td>Marmor</td>
<td>2</td>
<td>21.9</td>
</tr>
<tr>
<td>Total Condylar</td>
<td>2</td>
<td>—</td>
</tr>
<tr>
<td>Natural knees</td>
<td>4</td>
<td>75.3</td>
</tr>
</tbody>
</table>

Strength in hyperextension. When the cruciate ligaments were retained, as in the Geomedic and the Marmor, hyperextension resulted first in avulsion of the bony attachment of the anterior cruciate ligament at loads...
about one-third as great as that required to produce a
similar avulsion in the natural knee (Table VIII). This
may be because the resection of bone around the
intercondylar eminence of the tibia significantly
weakens the attachment. After this avulsion, all knees
depended upon tension in the posterior capsule to
prevent hyperextension.

Table IX. Strengths in hyperextension compared with bending
moments transmitted in life

<table>
<thead>
<tr>
<th>Bending moment (newton · metres)</th>
<th>Observation</th>
</tr>
</thead>
<tbody>
<tr>
<td>27.5</td>
<td>Caused failure, by avulsion of collateral ligament, of knee with ICLH prosthesis implanted but with posterior capsule removed</td>
</tr>
<tr>
<td>33.8</td>
<td>Caused failure, by avulsion of cruciate ligaments from femur, of natural knee with only cruciate ligaments present</td>
</tr>
<tr>
<td>41.2</td>
<td>Was supported, when applied by means of a force on the front of the knee, by forty-year-old males either passively with discomfort in posterior region of knee or by muscular effort</td>
</tr>
</tbody>
</table>

The results given in Table IX show that in replaced knees in which the cruciate ligaments are retained the anterior cruciate attachment is avulsed at bending moments less than those which provoke posterior discomfort in the knees of healthy forty-year-old males. Elderly patients with one or more replaced knees are unlikely to subject their legs to such hyperextending moments, but even in such patients the practical contribution of the anterior cruciate ligament to strength in hyperextension seems doubtful.

CLINICAL IMPLICATIONS

These results show that one major variable outside the surgeon's control affects the compressive strength of the replaced knee, namely the strength of the tibia itself. It therefore seems reasonable to adjust the patient's postoperative activity in the light of the quality of the bone found at operation. This having been said, it is also obviously desirable to confer the maximum possible strength upon the replaced tibia, and the results show that this can best be done by employing a tibial component of the largest possible area, enough to ensure that it rests on both cancellous and cortical bone, and by aligning the knee so as to ensure that the resultant of the loads passing through it traverses the centre. If all these considerations could be met reliably in every surgical procedure, there would be no advantage from a short intramedullary stem such as that of the Total Condylar. Such a stem, however, does provide a more reliable resistance to eccentrically applied loads than does the very much smaller stud on the ICLH prosthesis.

The present results show that there is no particular advantage to be gained by the retention of the cruciate ligaments. An advantage of the roller-in-trough geometry employed by most condylar replacements is demonstrated by the finding that the more heavily such a prosthesis is loaded, the stiffer it becomes in torsion. In these respects the effect of geometry on rotational stiffness is similar to that in the natural knee, as described by Walker, Wang and Masse (1975).

REFERENCES


