Effect of femoral component designs on the contact and tracking characteristics of the unresurfaced patella in total knee arthroplasty

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Objectives: To determine the effect of 5 different femoral components used in total knee arthroplasty (TKA) on the contact area and tracking characteristics of the nonresurfaced patella and to identify any design features that might adversely affect these characteristics. Design: An in-vitro study. Setting: The biomechanics laboratory, Department of Mechanical Engineering, McGill University, Montreal. Specimens: Six fresh-frozen cadaveric knee-joint specimens. Interventions: An unconstrained quadriceps simulator was used to apply the conditions of static lifting to the specimens first in their normal state and then sequentially implanted with femoral and tibial components of various designs (Miller/Galante II, Anatomic Modular Knee [AMK] System, Whiteside Ortholoc Modular, press-fit condylar and Insall–Burstein II). Outcome measures: Patellar 3-dimensional tracking characteristics, determined by using a 6 degrees-of-freedom electromechanical goniometer attached directly to the patella, and patellar contact pressure measurements, obtained using low-range Fuji Prescale film. Results: Articulation of the normal patella on a prosthetic femoral component resulted in alterations in the normal patellofemoral contact and tracking characteristics. The exact departure depended on the design of the prosthetic trochlea. Although all of the selected prostheses demonstrated satisfactory contact characteristics near extension, marked alterations occurred at higher flexion angles. With 90° or more of flexion, there was incompatibility between the geometries of the prosthetic notch of 2 femoral designs (AMK and PFC) and the normal knee. Conclusion: The design of the prosthetic femoral component must be taken into account when determining whether or not to resurface the patella at the time of TKA.
contact patellofémoral normal et du mouvement. L’écart exact dépendait de la forme de la trochilée de la prothèse. Même si toutes les prothèses choisies ont montré des caractéristiques de contact satisfaisantes en quasi-extension, on a constaté des altérations marquées à des angles de flexion plus élevés. Avec une flexion de 90 degrés ou plus, on a constaté une incompatibilité entre les caractéristiques géométriques de l’échancrure de la prothèse de deux pièces fémorales (AMK et PFC) et le genou normal. Conclusion : Il faut tenir compte de la forme de la pièce fémorale lorsqu’il s’agit de déterminer s’il faut resurfercer la rotule au moment de l’ATG.

Early designs of prostheses used for total knee arthroplasty (TKA) did not allow for patellar resurfacing and were associated with a high rate of persistent patellofemoral pain postoperatively. As a result, modern prostheses have been designed to incorporate a resurfaced patella. Although patellar resurfacing decreased the frequency of patellofemoral pain, particularly in patients with rheumatoid arthritis, patellar complications have now become the second leading cause, after infection, for revision TKA. Patellar complications include maltracking, pain, loosening, wear, mechanical failure, soft-tissue impingement and patellar fracture.

The alarming number of patellofemoral complications have influenced many orthopedic surgeons to selectively determine which knees require resurfacing or to abandon patellar resurfacing altogether at the time of TKA. Unfortunately, it is now apparent that up to 29% of patients with TKAs without patellar resurfacing may suffer anterior knee pain, 10% may require subsequent surgery for patellar resurfacing and one-third have difficulty climbing stairs. Picetti and colleagues evaluated 100 TKAs without patellar resurfacing at a mean of 4.5 years postoperatively: 29% of the knees continued to cause anterior knee pain, 10% may require subsequent surgery for patellar resurfacing and one-third have difficulty climbing stairs. Barrack and associates found that at only 30 months’ follow-up, 6 (10%) of the 60 knees that had not undergone patellar resurfacing required it subsequently because of anterior knee pain. In a study comparing 27 knees not subjected to patellar resurfacing with 100 knees with patellar resurfacing, Soudry and colleagues found that patients without a patellar button had more difficulty using the knee for stair climbing, and one-third avoided this activity altogether.

Because it is not apparent why some patients have patellofemoral symptoms after TKA without patellar resurfacing and others do not, the orthopedic surgeon cannot determine preoperatively with any certainty whether a patient will suffer patellofemoral symptoms after TKA without patellar resurfacing.

It has been speculated that anterior knee pain from a patella that is not resurfaced is secondary to altered patellofemoral biomechanics. The femoral components of current TKAs are designed to articulate with a corresponding patellar prosthesis. Articulation of the normal patella with the prosthetic femoral component poses a potential risk in terms of abnormal contact and tracking characteristics. The degree by which patellofemoral biomechanics will depart from normal depends on the design features of the femoral component. Therefore, we undertook a study to evaluate the effects of different femoral component designs on the contact and tracking characteristics of the unresurfaced patella and to identify any design features that may adversely affect these characteristics.

Five designs were considered: Miller/Galante II (MG II) prosthesis (Zimmer, Warsaw, Ind.), Anatomic Modular Knee (AMK) System (DePuy, Warsaw, Ind.), Whiteside Ortholoc Modular (Ortholoc) prosthesis (Wright Medical Technology, Arlington, Tenn.), press-fit condylar (PFC) prosthesis (Johnson & Johnson, Warsaw, Ind.) and Insall-Burstein II (IB II) prosthesis (Zimmer). To allow direct comparisons of the effects of each design, corresponding to simulated loading conditions, the foregoing characteristics were determined in knee joint cadaveric specimens first in their normal state and then sequentially implanted with the femoral and tibial components of each design.

Material and methods

Six fresh-frozen cadaveric knee joint specimens from 4 males and 2 females were used. The age of the donors ranged from 25 to 61 years (mean 38 years). To minimize prosthetic component inventory, only left knees with an anterior–posterior (A–P) femoral dimension of approximately 64 mm were selected. The specimens were verified to be free of abnormalities, patellar maltracking, arthritic changes and moderate to severe chondromalacia on visual and radiographic inspection. For mounting in the load simulator, the femurs and tibias were transected about 25 cm from the joint line and all skin, subcutaneous tissues and muscles removed. The knee capsule, ligaments and tendons were left intact, but a lateral retinacular release was done in all cases. No further patellar realignment procedures were carried out on any of the knees.

With respect to implantation of the prosthetic components in the frontal plane, the tibia and femur were cut perpendicular to their respective mechanical axes. In keeping with contemporary surgical practice, the femoral components were externally rotated by 3°. Although the thickness of the components varied from one design to another, we attempted in all cases to maintain the level of the joint line. With sequential prosthetic implantation, cuts were redone or custom-made shims inserted as required to maintain the
original joint line and preserve soft-
tissue balance. Restoration of the
joint line was confirmed by postoper-
ative radiographs. The femoral and
tibial components were all placed
centrally in the medial–lateral (M–L)
direction. In the A–P direction, the
tibial components were placed up to
the posterior margin of the cut prox-
imal tibia and aligned anatomically.
A single posterior slope of 5° was
adopted for the tibial cut, being the
average slope recommended by the
prosthetic component manufactur-
ers. The specifications of the pro-
thetic components are listed in Table
1. In view of the fixed A–P dimen-
sion of the intact distal femurs
selected, only a single size of each
femoral component design was re-
quired, allowing direct comparisons
of the contact area and tracking mea-
surements among specimens without
the need for scaling. With the excep-
tion of the Ortholoc prosthesis, the
size of each femoral component de-
sign was such that its A–P dimen-
sion was the closest available to that of
the intact femur while not exceeding
it. However, only a large size was
available for the Ortholoc prosthesis.
Although its A–P dimension of 65
mm was sufficiently close to that of
the intact femurs to retain it in our
study, its accompanying M–L dimen-
sion of 79 mm (5 mm more than the
next biggest femoral component)
was excessive in 2 of the 6 speci-
mens, resulting in collateral ligament
impingement and flexion contractu-
ture. Therefore, in theses 2 cases, test
measurements were not performed.

For the loading conditions, an un-
constrained quadriceps simulator was
used to apply the conditions of "sta-
tic lifting," an activity similar in na-
ture to that of rising from a chair
without the aid of an arm rest.22 In
this simulation, a foot–floor reaction
force of 334 N (half the body weight
of a person weighing 68 kg) was
applied to the tibia with its line of
action being adjusted at each consid-
ered flexion angle in a manner com-
mensurate with the activity. To
counteract the flexional moment of
the foot–floor reaction, 3 extensor
muscle group forces — FL, FC, and
FM — were applied to the patella
with their combined magnitude be-
ing correspondingly adjusted at each
flexion angle. FL represented the
effects of the vastus lateralis, FC re-
presented those of the rectus femoris,
the vastus intermedius and the vastus
medialis longus and FM represented
those of the vastus medialis oblique.
The 3 forces were applied with steel
cables through a brass cap fitted
snugly on the outer surface of the
patella. Their lines of action were
based on previous work by Ahmed
and associates.22 On the basis of mus-
cle cross-sectional measurements in
that report, the total extensor load
was apportioned in a ratio of
1.75:2:1, respectively, for FL, FC,
and FM. The flexion angles consid-
ered in the present study were 0°,
15°, 30°, 60°, 90° and 105°. Corre-
spondingly over this flexion range,
the total extensor load increased
from about 90 N to 950 N.

Contact pressure measurements
were taken with use of low-range
Fuji Prescale film (Fuji Photo Film,
Tokyo, Japan). The film sheets were
enclosed in thin plastic packets (total
thickness less than 0.3 mm) to pro-
tect them from the surrounding
moisture and inserted superiorly be-
tween the patella and femur. To reg-
ister the location of the contacting
areas relative to the patella, land-
marks were inscribed on each side of
the patella in the mid-region and
transcribed to each test-condition
film packet using a dull point. Load-
ing was gradually applied over 30
seconds and held for 60 seconds, fol-
lowed by a quick unload. To deter-
mine the pressure distribution, the
film was calibrated under the same
load and unload conditions by ex-
posing given areas of it to com-
pressed air at given pressure levels
against a flat ground surface. Both
the test-condition and calibration
traces were then digitized into gray
level bitmaps (256 gray levels) using
a flat-bed scanner (HP 1C ScanJet
et; H ewlett Packard, Palo Alto, Calif.)
and analyzed using an image analysis
software program (Image Pro Plus;
M edia Cybernetics, Baltimore). The
overall accuracy for the pressure and
spatial measurements was ±0.05 MPa
and ±0.1 mm respectively. The
threshold and saturation levels of the
film were approximately 1 MPa and
5.5 MPa respectively.

The patellar 3-dimensional track-
ing characteristics were measured us-
ing a 6 degrees-of-freedom electro-
mechanical goniometer attached
directly to the patella. The tracking
characteristics were described as 3
Euler rotations — flexion, tilt and
spin — and 3 Cartesian translations
— medial–lateral, anterior–posterior
and proximal–distal. The rotation
axes were based on the patella’s
anatomically defined axes, whereas
the translation axes were based on the
anatomical axis of the femur and a
line parallel to the posterior
condyles. The initial position from
which the displacements were mea-
sured in each specimen corresponded

### Table 1

<table>
<thead>
<tr>
<th>Prosthesis design</th>
<th>Femoral size (A–P dimension, mm)</th>
<th>Tibial surface type</th>
<th>PCL status</th>
</tr>
</thead>
<tbody>
<tr>
<td>MG II</td>
<td>5 (64)</td>
<td>Flat (unconstrained)</td>
<td>Intact</td>
</tr>
<tr>
<td>AMK</td>
<td>3 (64)</td>
<td>Flat (unconstrained)</td>
<td>Intact</td>
</tr>
<tr>
<td>Ortholoc</td>
<td>L (65)</td>
<td>Flat (unconstrained)</td>
<td>Intact</td>
</tr>
<tr>
<td>PFC</td>
<td>3 (61)</td>
<td>Curved (constrained)</td>
<td>Intact</td>
</tr>
<tr>
<td>IB II</td>
<td>64</td>
<td>Curved (constrained)</td>
<td>Sacrificed</td>
</tr>
</tbody>
</table>

MG II = Miller/Galante II prosthesis (Zimmer, Warsaw, Ind.), AMK = Anatomic Modular Knee System (DePuy, Warsaw, Ind.),
Ortholoc = Whiteside Ortholoc Modular prosthesis (Wright Medical Technology, Arlington, Tenn.), PFC = press-fit condylar
prosthesis (Johnson & Johnson, Warsaw, Ind.) and IB II = Insall-Burstein II (Zimmer), A–P = anteroposterior, PCL = posterior
cruciate ligament.
to that of the patella in the intact case at 15° of knee flexion. The overall accuracies for the rotations and translations were determined to be ±0.5° and ±0.5 mm respectively.

Statistical comparisons, between the cadaveric specimens in their normal state and sequentially implanted with the femoral and tibial components of each design, were carried out using a repeated one-way analysis of variance at each flexion angle for the following parameters: total contact area, percentage of the total contact area subjected to high pressures (more than 5 M Pa) and patellar tracking. The Student–Newman–Keuls test was then used to further evaluate differences among the individual implant designs. A p value of less than 0.05 was considered significant.

Results

Patellar contact characteristics (total area, location and area subjected to high pressures) varied from implant to implant as well as with knee flexion. Between 0 and 60° of knee flexion, there was no significant difference in the total patellar contact area between the implanted specimens and the normal knee (p = 0.7 at 30°, p = 0.09 at 60°) (Table 2). On average, the implanted specimens had 97% and 79% of the patellar contact area of the normal knee at 30° and 60° of flexion respectively.

As seen in the normal knee at 0 to 60° of flexion, the patellar contact area tended to be distributed across a broad transverse region of the patella in the implanted specimens (Fig. 1). However, in deep knee flexion (90° and 105°), there was a pronounced decrease in the patellar contact area in the implanted cases (Table 2). The average contact area was 69% and 65% of the normal knee at 90° and 105° of knee flexion respectively (p = 0.002 at 90°, p = 0.004 at 105°). All of the 5 implants tested had significantly less patellar contact area than the normal knee at these higher flexion angles (p < 0.05).

Furthermore, the implanted knees tended to demonstrate severely altered patellar contact distributions at 90° and 105° of knee flexion. Unlike

![FIG. 1. Comparison of the contact pressure distributions measured in the intact articulation (first row) with those measured when the intact patella articulates with the femoral components of 5 different component designs used in total knee arthroplasty, for 4 different flexion angles (30°, 60°, 90° and 105°). MG II = Miller Galante II prosthesis (Zimmer, Warsaw, Ind.), AMK = Anatomic Modular Knee System (DePuy, Warsaw, Ind.), Ortholoc = Whiteside Ortholoc Modular prosthesis (Wright Medical Technology, Arlington, Tenn.), PFC = press-fit condylar prosthesis (Johnson & Johnson, Warsaw, Ind.) and IB II = Insall–Burstein II prosthesis (Zimmer).](image)
implants had significantly greater
planted cases. At 90° of flexion all
was most pronounced in the im-
90° and 105° of flexion, this effect
lar contact pressures occurred with
articulating with the normal knee
sures greater than 5 MPa than when
contact area being subjected to pres-
greater proportion of the patellar
nents resulted in a significantly
PFC and Ortholoc femoral compo-
faced patella with the AMK, IB II,
contact pressures greater than 5 M Pa at 30° of knee flexion
(p = 0.2). Furthermore, there were
no significant differences among any
of the various implants. At 60° of
flexion, articulation of the unresur-
faced patella with the AMK, IB II,
PFC and Ortholoc femoral compo-
nents resulted in a significantly
greater proportion of the patellar
contact area being subjected to pres-
sures greater than 5 M Pa than when
articulating with the normal knee
(p < 0.05). Although increased patel-
lar contact pressures occurred with
90° and 105° of flexion, this effect
was most pronounced in the im-
planted cases. At 90° of flexion all
implants had significantly greater
proportions of their patellar contact
areas subjected to pressures greater
than 5 M Pa compared with normal
knees (p < 0.05). At 105° of flexion,
the IB II, MG II and PFC prostheses
continued to have a greater propor-
tion of their contact areas subjected
to these high pressures than normal
knees (p < 0.05).

Incompatibility between the
geometries of the AMK and PFC
femoral intercondylar notches and
the normal patella occurred at 90°
and 105° of flexion. At these high
flexion angles, the apex of the central
ridge of the retropatellar surface im-
pinged on the prosthetic intercondy-
lar notch. The impingement either
created a marked indentation into
the apex of the patellar central ridge
or resulted in frank displacement of
the patella into the notch. This effect
is illustrated in Fig. 1 where the pres-
sure distributions are seen to un-
dergo abrupt changes, outlining the
shape of the notches. As a result of
the patellar groove’s quick transition
into the notch and its sharp border,
indentation of the retropatellar sur-
face was particularly severe with the
AMK design.

In-plane tracking characteristics
did not differ between the implanted
specimens and the normal knees.
There was no significant difference in
patellar flexion, proximal-distal
translation and A–P translation
among any of the implanted knees,
or between the implanted knees
and the normal knees.

The out-of-plane tracking charac-
teristics were altered to varying de-
gress when the normal patella was
required to articulate with a pro-
thetic femoral prosthesis. From 15°
to 30° of knee flexion, significant lat-
eral tracking of the patella occurred
at 15° with the MG II prosthesis (p <
0.05) and at 30° with the AMK, MG
II and PFC prostheses (p < 0.05).
On average, implantation of these
prostheses resulted in patellar transla-
tions that were approximately 3 to 5
mm more lateral than that seen in
the normal knee (p < 0.001). This
lateral translation appeared to occur
as a result of an intentional design
feature of the femoral components to
reduce maltracking and was not asso-
ciated with clinically obvious patellar
subluxation. From 60° to 105° of
flexion, the implanted specimens
demonstrated M–L patellar transla-
tion similar to that of the normal
knee. Patellar spin similar to that in
the normal knee was seen in all im-
plants between 30° and 105° of knee
flexion. Only from 0 to 15°, did
patellar spin vary significantly among
the normal knee and the IB II, PFC
and Ortholoc prostheses (p < 0.05).
Various implants demonstrated aber-
rant patellar tilt. Compared with the
normal knee, increased medial tilt
of the patella occurred with the
Ortholoc prosthesis from 15° to
105° of flexion (p < 0.05), with the
IB II prosthesis from 60° to 105° (p
< 0.05) and with the PFC prosthesis
at 90° and 105°.

Discussion

Alterations in normal patello-
femoral contact and tracking char-
acteristics can be expected when the
unresurfaced patella is brought to
articulate with a prosthetic femoral
component. The exact departures
will depend on the design of the
prosthetic trochlea. At low flexion
angles, patellar contact area, location
and pressures were similar between
implanted specimens and normal
knees. However, with increasing
flexion there was a profound alter-
ation in patellar contact distribution,
a progressive decrease in the patellar
contact area and, as a result, a signif-
icient increase in patellofemoral con-
tact pressures. At 90° and 105° of
knee flexion, the implanted speci-
mens tended to have only discrete
isolated zones of contact on each
facet of the patella, and these areas
were subjected to higher than nor-
mal pressures. These findings may
help explain why patients can expe-
rience difficulties carrying out activi-
ties that require loading the

\( \text{Table 3} \)

**Percentage of the Patellar Contact Area Subjected to Contact Pressures Greater Than 5 MPa for Normal Knees and Unresurfaced Patellae Articulating With Each Femoral Component Design**

<table>
<thead>
<tr>
<th>Prosthesis design</th>
<th>Knee flexion angle</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>30°</td>
</tr>
<tr>
<td>Normal</td>
<td>0.3</td>
</tr>
<tr>
<td>MG II</td>
<td>0.5</td>
</tr>
<tr>
<td>AMK</td>
<td>2.7</td>
</tr>
<tr>
<td>Ortholoc</td>
<td>0.5</td>
</tr>
<tr>
<td>PFC</td>
<td>5.0</td>
</tr>
<tr>
<td>IB II</td>
<td>0.3</td>
</tr>
</tbody>
</table>

MG II = Miller-Galante II prosthesis (Zimmer, Warsaw, Ind.), AMK = Anatomic Modular Knee System (DePuy, Warsaw, Ind.), Ortholoc = Whiteside Ortholoc Modular prosthesis (Wright Medical Technology, Arlington, Tenn.), PFC = preflex condylar prosthesis (Johnson & Johnson, Warsaw, Ind.) and IB II = Imal-Burstein II prosthesis (Zimmer).
patellofemoral joint at high degrees of knee flexion, such as stair climbing.\textsuperscript{21} Even patients who can ascend stairs normally when tested, may not do so in everyday life. Levitsky and associates\textsuperscript{18} found that although 53 (80\%) of their 66 patients whose patella was not resurfaced were able to ascend stairs reciprocally when tested postoperatively in the office, only 24\% climbed stairs reciprocally in their routine day-to-day activities.

At 90° and 105° of knee flexion, articulation of the unresurfaced patella with 2 of the femoral component designs resulted in impingement of the patella on the intercondylar notch. At these high flexion angles, the intercondylar notch design of both the AMK and PFC implants did not adequately support the patella, thereby allowing it either to be indented by or actually to displace into the notch. This lack of patellar support was due to the high, wide femoral component that subse-

References


