The Influence of Total Knee-Replacement Design on Walking and Stair-Climbing

BY THOMAS P. ANDRIACCHI, PH.D.,† JORGE O. GALANTE, M.D.,Ž AND REX W. FERMER, B.S.,† CHICAGO, ILLINOIS

From the Department of Orthopedic Surgery, Rush-Presbyterian-St. Luke’s Medical Center, Chicago

ABSTRACT: The relationship between gait and prosthetic design was studied during level walking and stair-climbing for twenty-six asymptomatic patients after total knee replacement. An age-matched group of fourteen control subjects was also studied. Five designs of total knee replacement — Geomedic, Gunston, total condylar, dupatellar, and Cloutier — were used.

Differences in gait could be identified on the basis of prosthetic design. The more stressful stair-climbing test produced more clearly differentiated function among the different designs. Patients who were treated with the least-constrained cruciate-retaining (Cloutier) design of prosthesis were the only group that had a normal range of motion during climbing up and down stairs. Two groups of patients with semiconstrained (total condylar and Geomedic) designs had a lower than normal range of knee flexion while descending stairs. Patients with the other designs of prosthesis had a normal range of knee motion on stair-climbing.

Kinematic and anatomical differences among the five designs did not have as great an influence on function during level walking as they did during stair-climbing. The results of this study indicate that after total knee replacement even asymptomatic patients with excellent clinical results have an abnormality of gait. The features of the abnormality were common to most of the patients in the series, and consisted of a shorter than normal stride length, reduced mid-stance knee flexion, and abnormal patterns of external flexion-extension moment of the knee. Although an explanation of these abnormalities of gait is not completely possible at this time, they appear to be related to the interaction of the kinematics of the knee and surrounding soft tissues.

CLINICAL RELEVANCE: It appears that patients with less constrained cruciate-retaining designs of total knee replacement have a more normal gait during stair-climbing than patients with more constrained cruciate-sacrificing designs. During level walking, patients with five quite different designs all had abnormalities of gait in spite of a successful clinical result.

There is currently a great deal of controversy regarding which type of total knee prosthesis provides better gait. An improved understanding of gait and the variables associated with total knee designs is essential in addressing this controversy. Quantitative studies of gait during activities of daily living are needed to generate this information, and will be useful for the evaluation of total knee-replacement devices and for providing understanding of the loading patterns that may occur during normal activity.

Several studies have evaluated gait in patients with knee disease. These investigations included kinematic analyses 1, 6, 7, 9, 13, 21, time-distance measurements, and force-plate measurements. There have also been several kinetic and force-analysis studies of function in normal subjects and in patients after treatment for knee disabilities. The common finding of these studies was that patients who appear to be clinically asymptomatic after joint replacement have abnormal gait patterns. Currently, little is known about the nature of the gait abnormality in patients after total knee replacement or its relationship to total knee-replacement design.

The purpose of this study was to evaluate the relationship between gait and total knee-replacement design. The prosthetic knees that were selected for this study were considered to be representative of cruciate-sacrificing and sparing designs with varying amounts of constraint. The parameters of gait that we observed included time-distance patterns and motion and moments of the knee joint. The gait of patients who had received one of five different designs of total knee replacement was evaluated and compared with that of control subjects.

Materials and Methods

Twenty-six patients, in five experimental groups, were studied during level walking and stair-climbing. Patients were grouped according to which of five total knee designs they had received. The five implants selected for this study were the Geomedic, Gunston, total condylar, dupatellar, and Cloutier designs.

The five designs of prosthesis were selected to represent varying shapes of the articular surfaces and the retention of one, both, or neither cruciate ligament. The Geomedic prosthesis has fairly congruous articular surfaces; requires removal of the anterior cruciate ligament, permits retention of the posterior cruciate ligament, and does not include a patellar flange or resurfacing. The Gunston prosthesis consists of two separate semicircular runners that articulate with two independent tibial components, permits retention of both cruciate ligaments, and does not include patellar resurfacing or a patellar flange. The total condylar design requires the sacrifice of both cruciate ligaments, with anterior-posterior stability provided by the conformity of the tibial articularating surfaces; all patients with this design had patellar resurfacing. The dupatellar prosthesis permits retention of the posterior cruciate ligament, includes a patellar flange, and allows patellar resurfacing, which was performed in all of the patients whom we examined. The Cloutier prosthesis allows retention of both cruciate ligaments and the femoral condyles are asymmetrical, diverge, and have varying radii of curvature. The tibial component of the Cloutier device consists of flat articular surfaces supported on a metal retainer, and the design has a patellar flange, but patellar resurfacing was not performed in our patients.

The patients selected for this study were matched according to postoperative pain, function, passive range of motion, and joint stability. A point system based on The Hospital for Special Surgery knee-rating system was used to quantitate
postoperative status. All patients were evaluated at least one year postoperatively, and to qualify for this study the result had to score 85 points or more on the basis of this system. All patients had an excellent clinical result, were able to walk without aids, had little or no pain, and were able to climb stairs in a reciprocal manner.

No attempt was made to match the patients in terms of diagnosis, involvement of other joints, sex distribution, age, or presence of bilateral total knee replacement. However, patients with moderate or severe involvement of other joints that was associated with pain were not included in the study, since the result would not have achieved the clinical rating that was required as a prerequisite for inclusion.

The resulting population for the experimental study consisted of thirty-six knees in twenty-six patients (Table I). In eighteen patients with twenty-three involved knees rheumatoid arthritis was diagnosed, while the remaining eight patients with thirteen involved knees had osteoarthritis. Six patients had had a previous operation: four on the foot (three Gunston and one Geomedic prosthesis), one total hip replacement (duopatellar prosthesis), and one osteotomy of the hip (Geomedic prosthesis). No patient had a passive knee flexion contracture of more than 5 degrees. The posterior cruciate ligament was retained in the patients with the appropriate designs of prosthesis. Four of the eight knees with the Cloutier prosthesis still had the anterior cruciate ligament, while no documentation was available on the presence of the anterior cruciate ligament in patients with the Gunston design.

The control population consisted of fourteen healthy adults (seven men and seven women) with an average age of 62.4 years (standard deviation, 6.3). Both the control group and the study group were tested using the same protocol. The gait of each subject was measured during twelve stride cycles, each of which occurred midway during a separate walking trial on a ten-meter walkway. Each subject was instructed to walk at three nominal speeds: slow, normal, and fast. Patients were never requested to walk faster or slower than would normally be comfortable. All measurements of gait were analyzed in relation to walking speed for each subject. Each subject was also observed while ascending and descending a three-step staircase.

The experimental observations were based on an idealization that considered the lower extremities as a three-dimensional linkage with movable joints at the hip, knee, and ankle. The joints were assumed to have fixed axes of motion. The approach was similar to that reported by others.5-13 Motion of the lower limbs was monitored by observing the spatial position of six light-emitting diodes that were placed at the anterior superior iliac spine, at the center of the greater trochanter, over the midpoint of the lateral joint line of the knee, on the lateral aspect of the malleolus at the ankle, at the base of the calcaneus, and at the base of the fifth metatarsal. The positions of the joint centers at the hip, knee, and ankle on the sagittal plane were located relative to the position of the light-emitting diodes at the greater trochanter, lateral joint line of the knee, and lateral malleolus. The position of the center of the joint of the knee joint on the frontal plane was located by identifying the mid-point of a line between the peripheral margins of the medial and lateral plates of the level of the joint surfaces. The hip joint was located 1.5 centimeters distal to the mid-point of a line from the anterior superior iliac spine to the pubic symphysis. The ankle joint was estimated to be at the mid-point of a line from the tip of the lateral malleolus to the tip of the medial malleolus. The joint centers were located and marked on each subject prior to the observation of gait.

In the three-dimensional position of each light-emitting diode was sampled seventy-five times per second using an optoelectronic system consisting of two optical digitizers and electronic signal-conditioning. Ground-reaction force measurements were acquired simultaneously with the measurements of limb position using a piezoelectric force platform. The force platform provided the three components of ground-reaction force, vertical twisting moments, and location of resultant forces at the foot.

To calculate the moments, each segment of the limb (thigh, shank, and foot) was idealized as a rigid body with a coordinate system chosen to coincide with anatomical axes. Angular velocity and acceleration about the longitudinal axis of the limb segment were assumed to be negligible. The inertial properties of the limb segments were approximated from data derived from the literature.15 It was assumed that the flexion-extension axis remained perpendicular to the plate of progression, that the abduction-adduction and internal-external rotation axes at the hip joint moved with the thigh segment, that the axes of the knee joint moved with the shank segment, and that the ankle joint moved with the foot. The components of the moment vector were resolved into directions such that moments producing flexion-extension, abduction-adduction, and internal-external rotation at each joint could be identified. In addition, time-distance measurements and parameters of limb motion were derived for these measurements. For the purpose of this study, only data about the knee joint will be described.

The moments were analyzed for both patterns and magnitudes. The patterns were identified by timing the occurrence of the relative maximum and sign reversal. The sensitivity of the moment patterns and magnitudes to shifts in the position of the fixed joint center (for normal gait) were studied by theoretically moving the joint center a total of thirty millimeters in five-millimeter increments (fifteen millimeters on either side of the nominal position) in three orthogonal directions.

Differences between the averages for the experimental and control groups were tested for significance using the Student t test with a level of p < 0.05. A least-square curve was used to fit first, second, and third-order polynomials to the relationships of the various measurements of gait and walking speed. An analysis of the multiple correlation coefficient was used to determine which order of polynomial was used. Differences in regression coefficients were tested independently among patients and control subjects, assuming a Student t distribution.

Results

Level Walking

The range of walking speeds of the patients with total knee replacement was 0.5 to 1.92 meters per second, while the range of walking speeds for the normal individuals was 0.61 to 2.05 meters per second. The average self-selected walking speeds were placed in three categories (slow, normal, and fast) on the basis of the instructions given at the time of testing. The patients' average slow and normal speeds were both 79 per cent of those of the normal group (Fig. 1). The range of self-selected speeds tended to overlap those of the three nominal-speed categories, producing measurements of gait that were distributed over a range of walking speeds. Each of the other measurements of gait was considered with walking speed as an independent variable in order to separate differences in gait that were due to factors other than changes in speed.16

The stride length of the patients with total knee replacement was shorter than that of the control group over the range of walking speeds (Fig. 2). The relationship be-
between stride length and walking speed for the control subjects and the patients was best fit by a first-order least-square polynomial. The intercept of the curve for the groups of patients was found to be less (p < 0.05) than that for the control group, while the slopes of the two curves (24 per cent of height per meter per second) were not different.

The majority (75 per cent) of the patients with knee replacement had abnormal patterns of flexion-extension moments during stance phase. In response to an assumed error of fifteen millimeters in the location of all three axes of rotation, the largest change in the magnitude of the moment (16.5 per cent) was for the moment about the flexion-extension axis. However, the normal biphasic pattern of the flexion-extension moment during gait was not altered by displacing the position of the joint center by as much as fifteen millimeters on either side of the nominal position of the joint. These results are similar to those reported by other investigators. There was no significant difference between the regression coefficients in the relationship of stride length to walking speed among the patients in the five groups.

The change in the angle of knee flexion during stance, defined as the difference between the angle of knee flexion at heel-strike and the relative maximum flexion occurring during mid-stance (Fig. 3), was also less in the patients than in the control subjects over the range of walking speeds. The relationship between changes in knee flexion...
during stance and walking speed was found to be similar to
the relationship between stride length and walking speed
(Figs. 2 and 3). An increase in walking speed was asso-
ciated with an increased change in knee flexion during
stance. A first-order polynomial was found to best fit the
change in the relationship of knee flexion to walking speed
during stance for both the control subjects and the patients.
The intercept of the curve in the patients with a total knee
replacement was found to be less ($p < 0.05$) than that for
control subjects, while the slopes were not different.
Again, there were no differences between the patients in
terms of the different designs of prosthesis. A patient who
was walking at the same speed as a control subject tended
to change the angle of knee flexion substantially less dur-
ing stance than did the control subject. The reduced stride
length in the patients appeared to be coupled with a re-
duced change in the amount of knee flexion during the
middle portion of the stance phase.

A third characteristic of abnormal gait in the patients
with knee replacement was the pattern of external flexion-
extension moment about the nominal fixed center of the
knee joint. The moments were compared by selecting the
test conducted at a walking speed that was closest to one
meter per second for each subject (Fig. 1). The average
trial speed that was used when comparing the moments of
the controls ($1.02 \pm 0.14$ meter per second) and of the pa-
tients ($0.97 \pm 0.09$ meter per second) was not different.
The variations in the patterns were too large to be ex-
plained by systematic errors in the location of the nominal
joint center. In all of the patients the moment patterns in
the directions of abduction-adduction and internal-external
rotation were normal.

The normal curve of flexion-extension moment was
characterized by a biphasic pattern, with two relative
maximums tending to flex the joint and two relative
maximums tending to extend the joint (Fig. 4). The times
when the relative maximums occurred as well as the
number of occurrences of crossing the zero axis were used
to classify three distinct patterns. The pattern charac-
terized as *normal* was found to occur in 80 per cent of the
control subjects (Table II). The two abnormal moment pat-
terns (Fig. 4) tended to maintain an extrinsic moment pre-
dominantly tending to either flex the joint (flexional
moment pattern) or extend the joint (extensional moment pat-
tern) throughout stance phase. The abnormality in the flex-
ional and extensional moment patterns occurred during the
middle portion of stance phase. The distribution of the pat-
terns for each of the groups of patients and for the control
subjects is shown in Table II.

Differences in the maximum magnitude of the
moments tending to flex and extend the knee joint were
consistent with the distribution (Tables III and IV) of the
moment patterns. The two groups of patients (those with
total condylar and those with Geomedic prostheses) with a
majority of extensional moment patterns had a signif-
ically lower than normal moment tending to flex the
knee. The only other moment amplitude that was different

<table>
<thead>
<tr>
<th>Type of Prosthesis</th>
<th>Normal</th>
<th>Flexional</th>
<th>Extensional</th>
</tr>
</thead>
<tbody>
<tr>
<td>Total condylar</td>
<td>0</td>
<td>3</td>
<td>5</td>
</tr>
<tr>
<td>Geomedic</td>
<td>1</td>
<td>1</td>
<td>5</td>
</tr>
<tr>
<td>Duopatellar</td>
<td>1</td>
<td>3</td>
<td>2</td>
</tr>
<tr>
<td>Gunston</td>
<td>3</td>
<td>3</td>
<td>1</td>
</tr>
<tr>
<td>Cloutier</td>
<td>3</td>
<td>2</td>
<td>3</td>
</tr>
<tr>
<td>Total</td>
<td>8</td>
<td>11</td>
<td>16</td>
</tr>
<tr>
<td>Control group</td>
<td>11</td>
<td>3</td>
<td>0</td>
</tr>
</tbody>
</table>

* Walking speed of one meter per second.
from normal occurred in the patients with the Gunston prosthesis, who had an abnormally higher moment tending to abduct the knee joint. All of the other measurements of moments showed no difference between the patients and the control subjects. In addition, when the patients were grouped according to moment patterns, the patients with the flexional moment pattern had a significantly larger angle of knee flexion at heel-strike (10.6 ± 5.7 degrees) than did patients with the extensional moment pattern (4.24 ± 3.1 degrees). Both groups tended to have the smaller amount of change in knee flexion during stance, as described earlier.

Stair-Climbing

All patients were able to climb up and down stairs in a reciprocal manner. The speeds of ascending and descending stairs were not different for the patients and the control subjects (0.32 ± 0.09 meter per second). For the patients

<table>
<thead>
<tr>
<th>Type of Prosthesis</th>
<th>Extension</th>
<th>Flexion</th>
<th>Adduction</th>
<th>Abduction</th>
<th>External</th>
<th>Internal</th>
</tr>
</thead>
<tbody>
<tr>
<td>Total condylar</td>
<td>2.72</td>
<td>1.86†</td>
<td>3.05</td>
<td>0.41</td>
<td>0.45</td>
<td>0.39</td>
</tr>
<tr>
<td>(1.42)</td>
<td>(0.96)</td>
<td>(0.93)</td>
<td>(0.39)</td>
<td>(0.14)</td>
<td>(0.30)</td>
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</tr>
<tr>
<td>Geomedic</td>
<td>2.00</td>
<td>1.80†</td>
<td>4.00</td>
<td>0.40</td>
<td>0.60</td>
<td>0.40</td>
</tr>
<tr>
<td>(1.00)</td>
<td>(0.80)</td>
<td>(0.70)</td>
<td>(0.02)</td>
<td>(0.20)</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Duopatellar</td>
<td>2.07</td>
<td>2.78</td>
<td>3.78</td>
<td>0.22</td>
<td>0.61</td>
<td>0.44</td>
</tr>
<tr>
<td>(1.26)</td>
<td>(1.70)</td>
<td>(1.57)</td>
<td>(0.11)</td>
<td>(0.31)</td>
<td>(0.15)</td>
<td></td>
</tr>
<tr>
<td>Gunston</td>
<td>2.40</td>
<td>2.30</td>
<td>4.20</td>
<td>0.70†</td>
<td>0.80</td>
<td>0.40</td>
</tr>
<tr>
<td>(1.99)</td>
<td>(1.71)</td>
<td>(0.24)</td>
<td>(0.08)</td>
<td>(0.60)</td>
<td>(0.30)</td>
<td></td>
</tr>
<tr>
<td>Cloutier</td>
<td>2.49</td>
<td>2.67</td>
<td>3.70</td>
<td>0.41</td>
<td>0.56</td>
<td>0.50</td>
</tr>
<tr>
<td>(1.07)</td>
<td>(1.87)</td>
<td>(1.43)</td>
<td>(0.07)</td>
<td>(0.24)</td>
<td>(0.23)</td>
<td></td>
</tr>
<tr>
<td>Control group</td>
<td>2.29</td>
<td>2.74</td>
<td>3.57</td>
<td>0.25</td>
<td>0.54</td>
<td>0.41</td>
</tr>
<tr>
<td>(1.10)</td>
<td>(0.98)</td>
<td>(0.90)</td>
<td>(0.25)</td>
<td>(0.14)</td>
<td>(0.19)</td>
<td></td>
</tr>
</tbody>
</table>

* Walking speed of one meter per second.
† Standard deviations are in parentheses.
§ Significant difference from the control value.

An illustration of the characteristics of the three patterns of flexion-extension moments about a fixed point at the knee joint. The average times t1-t4 were characteristic of the so-called normal pattern, and the average times t5-t1 were characteristic of the patients with total knee replacement.

The differences in total range of motion of the knee while climbing up stairs among the five groups of patients and the control subjects. The patients with the Cloutier prosthesis were the only group whose range of motion did not differ from that of the control group.
in all five groups, with the exception of those with the total condylar prosthesis, the speed while descending stairs did not differ from that of the controls (0.49 ± 0.07 meter per second). The patients with the total condylar prosthesis descended stairs at a speed of 0.31 ± 0.10 meter per second.

The patients with the Cloutier prosthesis had the same range of motion of the knee while climbing up stairs as the control subjects (Fig. 5). The patients in the other groups, however, used a significantly smaller range of knee motion than the control group while climbing up stairs. Patients with the dupopatellar, Gunston, and Cloutier prostheses were not different than the control subjects during descending stairs, while patients with the total condylar and Geomedic prostheses had significantly smaller ranges of knee motion (Fig. 6).

Two distinctive patterns of flexion-extension moments of the knee while ascending stairs were identified (Fig. 7). Pattern 1 was characterized by a waveform that reached a maximum in a flexion direction during mid-support phase, while Pattern 2 was characterized by a sign reversal from flexion to extension at 44 per cent of stance phase. The variations of these patterns could not be explained by systematic differences in the location of the nominal center of the knee joint of as much as one and one-half centimeters of displacement from the fixed center. All control subjects had Pattern-1 flexion-extension moment of the knee during stair-climbing. Five patients with the total condylar, two with the Geomedic, and one with the dupopatellar prosthesis had Pattern-2 moment characteristics at the knee, while all of the patients with a Gunston or Cloutier prosthesis had Pattern-1 moments. The maximum magnitude of the moment components during stair-climbing did not differ among the five groups of patients. All patients had a flexion moment that was less than that of the control subjects when descending stairs.
(Table IV). In the patients who climbed up stairs with the Pattern-2 flexion-extension moment, the moment tending to extend the knee (1.24 ± 6.3 per cent body weight × height) was higher than that for the control subjects.

**Discussion**

The results of this study indicate that differences in gait can be identified on the basis of prosthetic design. In addition to measurements of gait during level walking, more stressful tests such as stair-climbing are useful in determining differences. The patients who were treated with the least-constrained cruciate-retaining (Cloutier) design of prosthesis were the only ones who had a normal range of motion during ascending and descending stairs. With the Cloutier design both cruciate ligaments are retained, and the prosthesis has flat tibial surfaces that minimally constrain anterior-posterior and rotatory displacement. In assessing the reasons for the more normal gait of these patients during stair-climbing, some aspects of the kinematic behavior of the prosthesis appear to be factors. However, the range of motion of the knee alone does not provide the entire explanation, since all of the groups of patients had the same passive range of knee flexion. Because differences were observed during activity, the possibility exists that the interaction of the kinematics of the knee joint with the surrounding ligaments and muscles produces functional differences between designs. As the knee flexes from zero to 90 degrees, the contact area between the femur and tibia will move posteriorly increasing the mechanical advantage of the quadriceps muscles significantly by increasing their moment arm about the condylar contact point. This explanation is consistent with the finding that the myoelectric activity of the quadriceps muscles under constant moment is reduced by nearly a factor of two as the knee moves from full extension to 40 degrees of flexion. The mechanical advantage of this posterior movement may not be possible to the same degree in a more constrained design, and thus gait during stair-climbing is compromised.

The presence and function of the posterior cruciate ligament may also be a factor influencing the patient's ability to descend stairs. The patients with the total condylar (posterior-cruciate-sacrificing) prosthesis had a reduced descent velocity as well as a reduced range of knee motion. As the knee flexes while the patient descends stairs, the posterior cruciate ligament is in a position to resist the forward thrust of the femur on the tibia. In the absence of the ligament, the joint must be stabilized primarily by constraint built into the prosthesis, since neither the muscular nor the secondary ligamentous restraints are in a mechanically advantageous position to provide an efficient substitute.

The differences in design among the five total knee replacements did not influence gait during level walking as much as during stair-climbing. The fact that common characteristics of gait were present in a rather inhomogeneous mixture of subjects tends to indicate that there may be factors other than clinical differences or differences in prosthetic design that influence the way that people walk after total knee replacement.

One explanation for these abnormal characteristics of gait could be that after total knee replacement the patients continued to walk with a pattern that they had learned prior to treatment. Since we did not obtain preoperative measurements for these patients, this explanation could not be evaluated in our study. However, there is some indication from other studies that the gait characteristics of patients with untreated arthritis are different from those of normal subjects.

Another possible cause for these abnormalities in walking is abnormal muscle function. Several characteristics of the gait in these patients seem consistent with this explanation. The abnormality in gait occurred during the middle portion of the stance phase, when normally the quadriceps muscle is eccentrically contracting to balance gravitational and inertial forces at the knee. In the process, the quadriceps allows the knee to flex to approximately 20 degrees during mid-stance in a controlled manner. Patients with total knee replacement do not flex the knee in this manner. The abnormal patterns of flexion-extension moments seen in this study are also consistent with the explanation that the muscles are functioning abnormally, since the flexor and extensor muscle groups of the knee are the primary internal structures that are capable of equilibrating the extrinsic flexion-extension moments.

Abnormal muscle function could be caused by a partial loss of proprioceptive control or by a reduction or imbalance in muscle capacity after joint replacement. Both proprioceptive control and muscle capacity can be influenced by the interaction of the joint kinematics with the surrounding soft tissues and muscles. It is possible that different prosthetic designs impose different strains on the soft tissues (where afferent receptors are located) and thus alter the feedback on joint position, which depends on a combination of skin, muscle, and joint receptors. Similarly, a loss of quadriceps-muscle capacity or mechanical advantage could inhibit its use during gait and produce the abnormality of an extensional moment pattern.

As indicated earlier, the different kinematics associated with the various prosthetic designs may significantly alter the mechanical advantage of the quadriceps mechanism. The total explanation of the abnormality in gait may include loss of both proprioception and muscle capacity.

In assessing the results of this study, it is important to consider the limitations of the methods and assumptions used in acquiring the data. The moments reported in this study were calculated about a fixed point at the geometric center of the knee joint. Thus, for every subject, external moments were calculated about the same relative point at the knee joint. The sensitivity of both the magnitude and patterns of moments in the position of the fixed nominal joint center has been tested. Both in this study and in another study the normal patterns of the flexion-extension
moment of the knee were not altered by relatively large changes (± 1.5 centimeter) from the nominal position of the joint center. The abnormality of gait in these patients was sufficiently large so that the characteristics of the patterns that we have identified would not be altered by errors in the position of the joint center.

In comparing the characteristics of gait among the groups of patients and the control subjects, walking speed was used as an independent variable. Many aspects of gait change with changes in walking speed. Therefore, in order to differentiate between the effects of differing walking speeds and actual abnormalities of gait, gait measurements were compared in terms of their relationship to walking speed. In spite of the fact that the patients’ so-called normal walking speed was slower than that of the control group, we were able to identify differences between the mechanisms of walking in the two groups. Thus, this approach allowed interpretation of results when comparing data between two groups walking at different speeds. For example, if two control subjects were observed, one walking at the average speed for the control group and the second walking at the average speed for the knee-replacement patients, the stride length would be expected to be shorter for the latter subject due to the fact that he or she was walking at a slower speed. However, without collecting measurements over the range of walking speeds, we would not know if a subject was walking with a normal relationship between stride length and walking speed or with the abnormal relationship seen in the knee-replacement patients. The normalization of moments by the product of the subject’s height and weight was assumed to reduce variability, but no statistical assessment was performed.

It is not completely possible at this time to explain the abnormalities in gait observed in this study. However, their existence raises some new questions regarding the designs of total knee-replacement prostheses and the rehabilitation methods used after operation. We still do not know the effects of these abnormalities of gait on energy consumption and joint forces, nor do we know the effects of these forces on bone, implant, and cement interfaces. At this stage in the evaluation of total knee-replacement design, functional testing of patients treated with these devices constitutes an important source of new information.

References