Micromotion Measurements With Hip Center and Modular Neck Length Alterations

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Hip center relocation often is necessary because of acetabular deformity or in revision surgery. Superolateral relocation of the acetabular component increases the hip joint reaction forces and has been associated with early femoral implant loosening. In addition, relocation can necessitate the use of extended femoral neck lengths. The purpose of this study was to compare the initial stability (micromotion) of an anatomically placed femoral component with that of a superolaterally relocated component and with a component having an extended neck length. A six-degree of freedom device was constructed to measure three-dimensional micromotion at the proximal and distal regions of the femoral component. The instrumented femur was loaded using a unique loading device that included musculature necessary to simulate stairclimbing. Results showed that superolateral relocation of the hip center (25 mm) only moderately increased femoral component micromotion (13%). However, it was found that extending the neck length 12.5 mm produced a dramatic increase in micromotion (38%). Clinically this suggests that hip center lateralization and the use of long modular neck lengths should be avoided.

When un cemented acetabular implants are used in total hip arthroplasty, relocation of the hip center from its anatomic position often becomes necessary during revision surgery or because of pelvic deformity. Historically, cemented acetabular components or bulk grafting were used in these cases to maintain or restore the anatomic hip center location. However, because allograft bone has not shown adequate ingrowth characteristics, bulk allografts often are contraindicated for the uncemented acetabulum. In addition, high rates of aseptic loosening have been associated with allograft and cemented acetabular components, especially in cases involving revision or deficient acetabular bone stock. This has led to increased use of uncemented, porous coated acetabular components in primary and revision surgery.

To promote bone ingrowth and provide adequate long term fixation, the uncemented
acetabular component requires a stable implant-bone interface. Because of the problems with cement and allograft in acetabular replacement, hip center relocation has become the primary alternative for maximizing bone and implant contact. However, relocation alters the normal kinematics of the hip joint and possibly the long term stability of the implant.9 Experimental and theoretical studies have found that superolateral relocation of the hip center increases the hip joint reaction force.11,12,19 Clinical studies have associated hip center relocation, particularly superolateral, with an increased failure rate of the femoral component.7,20,31 In contrast, other studies have reported no increase in failure rate for components placed superiorly (not laterally).24 For instance, Schutzer and Harris26 reported no increase in failure rate with superior relocation as great as 29 mm. These findings indicate that mechanical factors associated with superolateral hip center relocation may affect the femoral component failure rate. Currently, few biomechanical data are available to compare with the clinical results.

Superolateral relocation of the hip center also may require the surgeon to increase the neck length of a modular prosthesis to maintain muscle tension, avoid impingement, and prevent dislocation.23 Modular implants allow the surgeon to customize the neck length intraoperatively by changing the femoral head of the prosthesis. Unfortunately, increasing the neck length of the femoral component also will increase the moments acting about the axis of the implant, possibly affecting implant stability. This nonsagittal moment or torque has been implicated as a major factor associated with implant failure.6,13 In addition, there recently has been much concern about the use of modular implants from a mechanical and a materials standpoint.4

Micromotion, defined as small relative motion between the implant and the bone, has been investigated in numerous in vitro studies.3,5,14,25,29 These studies rely on the premise that early stability of the bone-implant interface is essential for bony ingrowth and for a stable, long lived uncemented hip replacement. In a canine study, Pilliar et al.22 found that micromotions of greater than 150 μm can inhibit bone ingrowth. Long term clinical studies by Jasty et al.17 showed that micromotion continued to affect implant performance years after the initial postoperative period. In addition, femoral autopsy analysis of implant surfaces with poor bone ingrowth or a fibrous interface have shown evidence of stem micromotion and subsequently were found to be less stable in mechanical tests.30

Several investigators have assessed the direction and magnitude of the hip joint forces that cause micromotion. Gebauer et al.14 reported that micromotion takes place with a combination of axial, anteroposterior, and rotational forces. In addition, in vitro experiments with simulated walking forces produced no appreciable increase in micromotion compared with those in single leg stance. A study using a canine model showed that torsional loading produced the greatest degree of micromotion in cemented implants.28 Nonwalking activities, such as climbing stairs or rising from a chair, produce these rotational forces in vivo.

Few experimental studies have examined how increases in the length of the femoral neck or relocation of the hip center affect femoral micromotion. Davey et al.10 studied the effect of a similar variable, femoral component offset, which is defined as the perpendicular distance from the center of the femoral head to the long axis of the prosthesis. They found that increasing the offset increased the nonsagittal moment about the implant axis, resulting in increased micromotion of the prosthesis. However, this does not directly address the problem of neck length change because lengthening of the femoral neck also increases the moments in the sagittal plane.

Hip center relocation and neck length changes are becoming more common in total hip arthroplasty. Thus, this study was per-
formed to quantify independently the effects of hip center relocation and neck length increase on the micromotion of uncemented femoral components using an advanced hip joint loading simulator and micromotion measuring device.

MATERIALS AND METHODS

Specimen Preparation
Using a standard operating procedure, six fresh frozen human anatomic specimen femurs (left) were templated to determine proper implant size, thawed, and implanted by an orthopaedic surgeon (DD) with an uncemented femoral prosthesis (HG Multiloc, Zimmer Inc, Warsaw, IN). Two of the femurs were from male donors 36 and 44 years of age, whereas the other four were from female donors 29, 52, 64, and 71 years of age, respectively. All femurs were radiographed and determined to be free of abnormality. The implant used was a distally fluted, collared, T1AIV component with a fully circumferential proximal porous coating. For this study, the component was inserted line to line, that is, the femoral canal was reamed to the exact size of the implant stem (no distal locking was used). Immediately after the component was implanted, a torque wrench device (Zimmer Inc) was used to quantify stability of the implant. Readings from the torque wrench showed that all specimens were implanted with a consistently tight fitting femoral component. No femoral cracking was observed. Implant sizes ranged from 11 to 14 mm diameter, and care was taken to seat the collar properly during implantation.

Micromotion Measurement
Micromotion of the prosthesis was determined by measuring the movement of two sets of target balls attached in triad formation to the implant (proximal and distal). One triad was located in a transverse plane 4 mm inferior to the implant collar (proximal cross section), and the other was located 5 mm superior to the distal implant tip. The six balls were attached to the implant through window holes (7/16 inch) drilled in the bone, as shown in Figure 1. The implant was drilled and tapped before implantation with 3.2-mm (7/64 inch) holes using a specially designed drill press guide to ensure accurate hole placement.

Fig 1. Linear variable differential transducers are mounted to the bone, and target balls are mounted to the implant. Six-degree of freedom motion is measured independently at proximal and distal cross sections.

From each triad, the six-degree of freedom motion of the implant was calculated as follows. First, digital calipers were used to measure the positions of the target balls relative to the center of the implant. These measurements were necessary to establish the position of the target balls relative to the implant. The motion of each of the target balls was measured using two linear variable differential transducers (TransTek Inc, Ellington, CT) mounted on a bracket fixed to the bone (Fig 1). Small magnets fixed to the transducer cores were used to maintain contact with the target balls. This greatly simplified construction of the brackets and eliminated the possible errors attributable to forces produced by spring loaded transducers. The small magnets had no detectable effect on transducer calibration. Transducer readings were obtained continuously during loading with a computerized data acquisition system. Each transducer was factory calibrated to within 1 μm accuracy. In addition, a calibration stand incorporating a micrometer was constructed to perform two-dimensional calibrations for each bracket. Each bracket assembly was calibrated independently to a resolution of 3 μm and a range of 750 μm.
Two types of micromotion have been defined in the literature: (1) recoverable, when the implant returns to its original position after loading, and (2) subsidence type motion, when the implant does not return to its original position (settling or seating of the implant). Preconditioning of the specimen was performed to seat the implant into the bone, reducing the effect of subsidence type micromotion. In this experiment, total subsidence from the onset of loading could not be measured because of the limited range (750 μm) of the transducers. Occasionally during preloading the implant would subside more than 750 μm. This required rezeroing of the transducers, preventing direct measurement of total subsidence in this study. Thus, micromotions reported in this study are recoverable micromotions. After preloading, subsidence during cyclic loading tests was minimal.

**Hip Joint Simulator and Testing Protocol**

After the micromotion measurement apparatus was attached to the femur, the entire assembly was placed into a hip joint simulator (Fig 2). This device was designed previously to simulate loading in a physiologic stairclimbing position. The simulator (Fig 3) incorporates all three major muscle groups active in stairclimbing (extensors, abductors, and adductors). The unique feature of this simulator is a ball caster foot piece that allows the femur position to be set solely by contracting or lengthening the muscle groups. No

![Fig 2. The hip stairclimbing simulator loads the femur physiologically. Included are the abductor, adductor, and extensor muscle groups. A ball caster footpiece provides a vertical reaction force, simulating the knee.](image)

![Fig 3. Schematic of hip stairclimbing simulator. Load is applied to the top plate with a materials testing machine. Femoral position is maintained solely by the three muscle groups.](image)
distal clamping or potting is required. In addition, the ball caster provided the capability to adjust for the proper physiologically offset center of gravity. In this study, femoral position and the physiologic center of gravity were chosen based on the work of Andriacchi et al., to represent a standard stairclimbing position (34° flexion, 15° adduction) with an applied flexion moment of 28 N-m.

This stairclimbing loading position was chosen for two reasons; it produces a high hip joint reaction force, and it is used commonly in the literature, allowing comparison of results with previous studies. A vertical compressive cyclic load of 0 to 445 N (0–100 lb) was applied to the fixture with a materials testing machine. This applied load and moment resulted in a hip joint reaction force of approximately 3.5 to 4 times body weight, which is similar to forces found in telemetric studies. Loading was performed quasi-statically with a period of approximately 20 seconds per cycle (0.05 Hz). Additional details regarding the loading device are detailed in a previous study.27

The loading protocol for each specimen was as follows: (1) 15 cycles preconditioning to reduce the effect of subsidence motion, (2) 15 cycles with a normal hip center and short femoral neck length, (3) 15 cycles with a normal hip center and long plus extended head (12.5 mm longer than the short neck), (4) 15 cycles with a 25 mm superolaterally relocated hip center and short neck length, and (5) 10 cycles normal position (repeat of normal hip center with short neck).

Superolateral reorientation of the femur was achieved by shifting the position of the femur relative to the pelvic insertion sites and physiologic center of gravity while maintaining a constant femoral angle. In two specimens combined superolateral relocation with neck length extension was attempted; however, this produced high abductor muscle forces that resulted in fracture of the greater trochanter in two consecutive specimens. Thus, this study does not report data for this position.

Repeat loading in the normal position (Step 5) was performed to verify that the bone was not damaged permanently during previous steps. In all five specimens reported, the micromotion measured in Step 5 was similar to that of Step 2. This shows that recoverable micromotion was not affected by order of testing. A sixth specimen underwent part of the experiment but showed evidence of cracking on the proximal medial surface during Step 4. In addition, two specimens fractured at the greater trochanter. Data from these three specimens were discarded, bringing to five the total number of specimens reported in the current study.

Data Collection and Analysis

During cyclic loading, analog readings of micromotion from the transducers and hip joint reaction force from strain gauges were collected simultaneously using a computerized data acquisition system. Data underwent digital filtering before storage to eliminate all frequencies greater than 10 Hz. This significantly reduced noise and increased the accuracy of measurement. After the digital filtering, data were stored on the computer at 1 Hz.

A comprehensive affine transformation algorithm using Euler Z-Y-X angles was developed to convert the transducer measurements to motion at the implant-bone interface. A coordinate system was defined with the X axis directed medially, the Y axis directed posteriorly, and the Z axis directed inferiorly along the femoral axis (Fig 1). The conversion algorithm, similar to those of previous studies,21 assumed rigidity of the bone and the implant. This assumption was tested in one specimen by fixing one of the steel target balls directly to the cortical bone using the method described by Callaghan et al.5 The transducer brackets were mounted, and pseudomicromotion measurements were obtained during preloading with the steel ball and the bracket fixed to the cortical bone. If the bone and bracket were perfectly rigid, the result of this test would be no micromotion. However, the maximum motion obtained from this test was 6 μm, indicating that the cortical bone deformed slightly on loading.

The conversion algorithm was designed to convert the six-degree-of-freedom motion of the steel balls to micromotion of any point on the surface of the implant. This surface motion was reduced to three components: anteroposterior (AP), mediolateral, and axial micromotion. For the results that follow, transverse micromotion was defined as the vector sum of the AP and mediolateral components. The same coordinate system and conversion algorithm were used for the proximal and distal measurement transformations.
Data were divided into three groups corresponding to the loading protocol: a normal group, a long plus group, and a superolateral group (Fig 4). For each specimen, the amplitudes of the last 10 cycles of motion were averaged to obtain a value of recoverable micromotion. A paired, two-tailed t test was used to compare the superolateral and long plus group measurements with those of the normal group.

RESULTS

In all specimens tested, implant micromotion was highly recoverable, with only a small amount of subsidence (< 50 μm). Recoverable micromotion tended to decrease slightly during the first 10 preloading cycles and then stabilize. Figure 5 shows a plot of the transverse and axial micromotions for the proximal and distal measurements. For this plot, the data are computed in terms of micromotion of a point P fixed to the center of the implant cross section (not interface motion). Recoverable micromotion, rotational and transverse, was sensitive to the hip joint force, as can be seen in the figure. The patterns of mediolateral motion of point P were

**Fig 4.** The three configurations tested in this study were: (1) normal anatomic position, (2) superolateral relocation, and (3) normal hip center position with a long plus modular head.

**Fig 5.** Raw data from a typical specimen showing translational and rotational micromotion of the proximal cross section.
similar to the hip joint force, whereas AP and axial motions reached critical values that were less dependent on force. Transverse micromotion was much larger than axial micromotion in all specimens. In addition, the largest motion measured was AP translation, comprising approximately 65% of the total micromotion.

Figure 6 shows the transverse micromotion (magnified × 100) of the proximal and distal implant cross sections of a typical implant. Transverse micromotion was greatest on the proximal medial implant surface, where the rotational and translational components were additive. The proximal implant tended to translate posteromedially while simultaneously pivoting around the posterolateral surface. The average maximum proximal micromotion detected was 75 μm on the posteromedial surface. Distal micromotion was less than proximal motion for all loading conditions. Patterns of distal micromotion also were distinctly different from those of proximal motion. For example, in Figure 6B the distal implant translates anterolaterally, exactly opposite of the direction for the proximal implant. This shows that the implant was toggling within the femur. The average maximum micromotion of the distal implant was 45 μm on the anterolateral surface.

Figure 6 also shows the increases in micromotion resulting from hip center relocation and neck length extension. Relocation of the hip center tended to increase the posterior motion of the proximal implant. Increasing the neck length caused the implant to move in a medial direction. At the distal measurement site, the effects of both changes were similar to the proximal site but in opposite directions.

Compared with the normal group, micromotion in the superolaterally relocated group was significantly greater (p < 0.05) only on the proximal medial implant surface (Fig 7). Although the difference was not significant, strong trends showed hip center relocation produced increases in micromotion. Relocation produced micromotion increases of 13%, 9%, and 10% on the proximal medial, anterior, and posterior surfaces, respectively. Hip center relocation did not significantly increase distal micromotion in any of the specimens tested.

Larger increases in micromotion were found when comparing the long plus neck group with the normal group, with significant increases of 32%, 28%, and 23% at the proximal medial, anterior, and posterior implant surfaces, respectively. Distal micromotion also was affected significantly by neck length extension, with increases of 18% and

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**Fig 6.** Micromotion of the proximal and distal cross sections of a typical implant. Micromotion is magnified 100 times.
15% on the anterior and lateral surfaces. Neither neck length changes nor hip center relocation had a significant effect on axial micromotion.

**DISCUSSION**

The purpose of this study was to determine the effects of superolateral hip center location and neck lengthening on the initial stability of the implant. On the basis of previous studies of hip joint reaction force, it initially was hypothesized that superolateral relocation would produce significant micromotion increases. Neck length changes were added to the protocol to emulate surgical procedures, but at the time it was assumed that there would be little effect on micromotion. On the contrary, experimental results showed a significant effect of neck length changes on femoral micromotion, much greater than for superolateral relocation. This possibly was because of increased moment about the axis of the implant that results from neck length extension. Extending the neck length directly increases sagittal and nonsagittal moments acting on the implant, regardless of the changes in hip joint reaction force. Superolateral relocation only indirectly increases the hip joint force via reorientation of the associated musculature.

A neck length extension of 12.5 mm in this study corresponds to a 6-mm increase in the femoral component offset, defined by Davey et al as the perpendicular distance from the hip center to the implant long axis. An increase in offset directly increases the nonsagittal moment about the implant axis, producing a corresponding increase in micromotion. In general the result of the current study concurs with that of Davey et al, who found that increasing the offset produced increases in femoral micromotion and cortical bone strain. The current study’s micromotion increases were larger than those in the study of Davey et al, probably because of the higher joint reaction forces used in this study. In addition, results showed that increasing the neck length primarily increased the mediolateral translation of the proximal implant, having a lesser effect on implant axial rotation. This could be because the implant used in this study was designed specifically with distal flutes to reduce axial rotation. Differences in implantation technique also may have resulted in a tighter fit than those described in previous studies.

Many other factors also can confound comparison with previous studies. Because of the
difficulty of measuring micromotion, many different loading configurations and measurement techniques have been developed. For example, Fischer et al.\textsuperscript{13} used a different implant, loading configuration, and micromotion measurement sensor. As expected, the current results also were much different. Fischer et al found axial micromotion to be greater than transverse motion, exactly the opposite of the findings of the current study. This discrepancy could be because of differences in measurement locations and the assumptions of bone and implant rigidity. In the current study, six-degree-of freedom micromotion measurement at proximal and distal implant locations was considered necessary to reduce possible errors because of bone deformation. This study and most previous studies rely on the assumption that the bone and the implant can be treated as rigid bodies. However, the rigid body assumption recently has been in question. During preliminary investigations for this study, it was found that micromotion measurements were valid only at or near the location of measurement, necessitating measurement at proximal and distal implant locations.

Considering the factors mentioned, trends of proximal micromotion of the normally implanted specimens compared reasonably well with those of previous studies, notably with those of Gilbert et al.\textsuperscript{15} However, distal micromotion tended to be lower than that of previous results. A unique aspect of the current experimental design was the inclusion of musculature, which may explain partially the lower distal micromotion findings. Previous micromotion studies all have used some kind of distal fixation (clamping or potting) to maintain position of the femur. The current study does not use distal clamping; three muscle groups (extensor, abductor, adductor) maintain the position of the femur in a physiologic fashion, eliminating the artificial bending moment that results from distal fixation. These results imply that deformation may have an important effect on micromotion measurements.

This study has shown that superolateral hip center relocation and neck length changes affect femoral micromotion. A previous study showed that superior relocation without lateralization had little or no effect on hip joint reaction force.\textsuperscript{12} If there is no change in forces in the implant, little change in micromotion will result. The results of the current study support clinical findings that associate superolateral hip center relocation with increased failure rates of the femoral component. Little is known about the long term affects of micromotion, and it is possible that initial increases could have significant long term consequences. Thus, when the uncemented femoral component is used, the clinical indication is to reduce lateralization of the component by increasing neck length, although superior relocation may be acceptable.

More significantly, the results study show that the use of excessively long neck lengths should be avoided because they can produce significant increases in micromotion and possibly have a negative effect on bone ingrowth. Unlike relocation, neck length extension also increased distal implant micromotion, which has been associated with increased patient thigh pain. Although the absolute magnitudes of the micromotion increases with long neck extensions were small, they were large percent increases compared with the data of short neck lengths offset acetabular liners may offer an acceptable alternative to increasing neck length.

Data from this study indicate that femoral micromotion, especially mediolateral and axial rotation, may be affected more by increases in neck length than by alterations of the hip center. This emphasizes the importance of careful implant design on the initial stability and long term performance of total hip arthroplasty.

References

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