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Original Article

Initial Stability of Cemented vs Cementless Tibial Components Under Cyclic Load

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ABSTRACT

Background: Cement fixation of total knee components remains the gold standard despite resurgence in cementless fixation with the goal of long-term durable fixation. Initial stability is paramount to achieve bony ingrowth of cementless components.
Methods: Twelve cemented and cementless tibial baseplates were implanted into sawbones and tested using a physiological medial-lateral load distribution for 10,000 cycles to represent 8 weeks of in vivo function. Micromotion was measured at 5 locations around the baseplate during loading.
Results: Cycling had a significant effect on the change in micromotion between maximum and minimum loads at the anterior, medial, lateral, posteromedial, and posterolateral tray edge locations. A significant effect of fixation technique was detected for the anterior ($P < .001$), medial ($P = .002$), and lateral ($P = .0056$) locations but not for the posteromedial ($P = .36$) or posterolateral ($P = .82$) locations. Differences in micromotion between cemented and cementless components did not exceed 150 μm at any tested location.
Conclusion: The micromotion experienced by cementless tibial components in the present study may indicate a lower initial mechanical stability than the cemented group. However, this difference in initial stability may be subclinical because the differences between average cemented and cementless micromotion were <150 μm at all measured locations under the loading regime implemented.

Cement fixation of total knee arthroplasty (TKA) components has been accepted as the gold standard with more durable fixation and better survivorship when compared with cementless fixation [1]. Thus, it is still the preferred choice among knee surgeons [2]. Despite this success, there remain concerns regarding the use of cement as it relates to longevity [3–6] as well as the additional time required during the surgical operation. Integrity of the cement is also a concern as cement strains increase with a higher body mass index [7]. Cementing is an art, and previous literature has suggested that outcomes are technique dependent [8,9]. With increasing numbers of younger patients requiring TKA [10], more reliable methods of long-term durable fixation such as those potentially offered by cementless fixation are attractive.

This interest in long-term fixation combined with ease of component insertion has induced a steady change in total hip arthroplasty fixation. The cementless acetabular component became accepted first because of clinical results, which were clearly better than those of cemented components. Cementless femoral components subsequently gained in popularity as early failure rates declined to that of cemented components, and cementless components became easier and faster to implant. Similarly, in TKA, cementless fixation on the femoral side (and to a lesser degree, patellar components) has shown similar results to cemented fixation for many years [11]. The weakness of cementless tibial baseplates in TKA has always been tibial fixation, with insufficient short-term stability to encourage long-term ingrowth. The cause for this insufficient stability is likely multifactorial, including patient, surgeon, and implant design–related factors. Specifically, lack of medial cortical support of the tibial baseplate has been suggested to lead to stress shielding and, ultimately, resorption of

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the proximal tibia, which may lead to enough micromotion to prevent long-term ingrowth [12].

Initial mechanical stability is critical to the success of cementless components [13]. Failure of early designs of cementless TKAs has been demonstrated [14,15]. Migration of the tibial component because of micromotion at the implant-bone interface has been a common mode of failure of cementless fixation [16]. It has been shown that micromotion >150 μm at the implant-bone interface can inhibit bony ingrowth [15,17,18]. Results in the literature are mixed as to which form of fixation has more tibial component migration [19–21]. However, migration is strongly related to initial micromotion [22–24], which dominantly influences cementless fixation. With material advances in biological fixation surfaces such as porous metal [25–28] and hydroxyapatite coating [18,29–31], there has been a resurgence of cementless TKA as a potential solution to reliable long-term fixation.

Recently, Bhimji and Meneghini [32] evaluated the micromotion of 2 different cementless tibial component designs with compressive, shear, and torsional loads applied through a femoral component articulating on an insert. The first was a monoblock design made of porous tantalum with two 16-mm-long hexagonal pegs, and the second was a modular design made of titanium with a keel and four 9-mm-long cruciform pegs with porous features. They found that the 2 peg design experienced more micromotion concluding that the design with a keel and 4 pegs will have better initial stability. Given the renewed interest in cementless TKA, we set out to evaluate the initial stability of a cementless tibial component with a keel and 4-peg design to that of the gold standard, which is a cemented tibial component.

Methods

Specimen Selection

A priori power analyses indicated that a sample size of 12 for each of 2 implant designs would be sufficient, while maintaining a power of 80%, to detect differences comparable with those seen between other designs in prior studies. Thus, 12 each of cementless and cemented tibial baseplates (Triathlon Knee System, size 4, Triathlon Tritanium; Stryker, Mahwah, NJ) were used for this study. The cementless implant is a titanium design with porous features, a keel, and four 9-mm-long cruciform pegs. The cemented implant is titanium with a keel but without porous features or cruciform pegs.

A synthetic bone model of a cut tibial plateau (Model 1522-912, Tibial plateau profile #3401; Pacific Research Laboratories, Vashon, WA) was used with each implant for testing in this study as previously described [33]. The model replicates the cut profile that would exist at a 9-mm-depth resection plane of medium-sized tibia and is composed of dual-density polyurethane foam. The inner core is filled with a lighter density 12.5-pcf (pounds per cubic foot) open cellular foam to represent cancellous bone, while the outer 2.5-mm-wide rim is made of a higher density 40-pcf closed cellular foam to represent cortical bone.

Specimen Preparation

The tibial bone model was prepared in a standard fashion for both the cementless and cemented tibial components according to the manufacturer's surgical technique guide. The tibial components were positioned so that there was cortical rim support anterior and posterolateral with cancellous support posteromedial. For the cementless tibial component, the bone model was prepared using the keel punch corresponding to the size 4 tibial component followed by drilling 1/8″ diameter holes for the 4 cruciform pegs. This preparation was designed for press-fit final components. For the cemented tibial component, the corresponding mill and keel punch were used to prepare the bone model. The components were cemented into place using 20 gm of polymethylmethacrylate bone cement (Surgical Simplex P, Stryker, NJ). The cement was vacuum mixed at −20 to 22 in Hg using Stryker Revolution Cement Mixing System for 90 s to obtain a doughy consistency. Cement was then applied to the entire undersurface of the tibial component and to the prepared tibial bone model. The cement was finger-packed to ensure adequate penetration of cement into the sawbone material. The tibial component was then compressed into place, and all excess cement was removed. Cement was allowed to cure for at least 30 min before further preparation.

Similar to other studies [32,33], micromotion of the baseplate was measured throughout loading. Small brass balls (3/8″ diameter, 8-32 threads, #60250; Jandorf, Northfield, OH) were permanently applied using cyanoacrylate (Duro #81742; Henkel Corporation, Rocky Hill, CT) to threaded steel rods (1/8″ × 8-32) mounted in modified pronged tee nuts (8-32 threads, #23431; Midwest Fastener Corporation, Kalamazoo, MI). These were then attached via epoxy putty (#48526; Ace Hardware Corporation, Oak Brook, IL) to the anterior, lateral, posterior, and medial rims of each tibial baseplate. After hardening of the epoxy putty, measurements were taken in the transverse plane of brass ball positions by digital calipers (Model 500-171; Mitutoyo, Aurora, IL) and digital photographs (Samsung S7 model SM-G930V; Samsung) for later use in post-test analyses.

Testing Protocol

Immediately before testing, a size 4 posterior stabilized polyethylene insert was placed in the tibial baseplate and rigidly clamped via custom fixtures to a biaxial servohydraulic testing machine (Model 1321; Instron Corporation, Canton, MA) retrofitted with MTS TestStar II digital controller (MTS Corporation, Eden Prairie, MN). The construct orientation was adjusted to properly align the posterior stabilized with a size 4 femoral component mounted to the axial actuator via a custom fixture designed to load the medial side more than the lateral. The design targeted a 33%-67% lateral-medial load distribution to fall within 30%-70% to 40%-60% range reported in vivo conditions [34] and confirmed before and after cycling with Fujifilm (Low Prescale; Fujifilm Corporation, Japan). Four linear variable differential transformer (LVDT) bodies were then mounted via custom fixtures to the rigid base on which the construct was clamped and adjusted, so that the measuring rod of each LVDT rested on the center of each of the anterior, posterior, medial, and lateral brass balls to measure vertical micromotion in the sagittal and frontal planes during testing (Fig. 1).

Each tibial baseplate was then loaded axially with the femoral component in a ramp fashion between 20 and 2000 N to mimic the gait cycle at a rate of 1 Hz for 10,000 cycles. Micromotion at each LVDT was recorded throughout testing, along with actuator position and load. Ten thousand cycles was chosen to mimic 8 weeks of in vivo use, that is, the amount of time to initial biologic fixation [35–37].

Post-Test Processing and Statistical Analysis

Using NIH ImageJ freeware (National Institutes of Health, Bethesda, MD), digital caliper measurements of brass ball spacing were used to calibrate digital photographs from which brass ball center positions (x-y coordinates) in the transverse plane were extracted. These x-y coordinates were combined with LVDT micromotion measurements (z coordinate) to represent 3D positions of the brass balls after the 1st, 5000th, and 10,000th cycles of