Characterization of Femoral Component Initial Stability and Cortical Strain in a Reduced Stem Length Design

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Background: Short stemmed femoral components facilitate reduced exposure surgical techniques while preserving native bone. A clinically successful stem should ideally reduce risk for stress shielding while maintaining adequate primary stability for biological fixation. We asked (1) how stem length changes cortical strain distribution in the proximal femur in a fit-and-fill

geometry, and (2) if short stemmed components exhibit primary stability on par with clinically successful designs.

Methods: Cortical strain was assessed via digital image correlation in composite femurs implanted with long, medium, and short metaphyseal fit-and-fill stem designs in a single-leg stance loading model. Strain was compared to a loaded, unimplanted femur. Bone-implant micromotion then compared with reduced lateral shoulder short stem, and short tapered-wedge designs in cyclic axial and torsional testing.

Results: Femurs implanted with short-stemmed components exhibited cortical strain response most closely matching that of the intact femur model, theoretically reducing the potential for proximal stress shielding. In micromotion testing, no difference in primary stability was observed as function of reduced stem length within the same component design.

Conclusions: Our findings demonstrate that within this fit-and-fill stem design, reduction in stem length improved proximal cortical strain distribution and maintained axial and torsional stability on par with other stem designs in a composite femur model. Short stemmed implants may accommodate less invasive surgical techniques while facilitating more physiological femoral loading without sacrificing primary implant stability.

Keywords: primary stability, micromotion, THA, short stem, digital image correlation

1 **Introduction**

2 Renewed interest in minimally invasive surgical techniques has led to the increased
3 utilization of the direct anterior approach in total hip arthroplasty (THA) [1]. While THA
4 is a highly successful surgical procedure across a variety of implant designs and surgical
5 approaches [2-7], the direct anterior approach has increased in popularity over the past
6 decade as several studies have reported advantages in decreased blood loss, early
7 functional recovery, decreased dislocation rates and shorter hospital stays [8]. Femoral
8 preparation in this approach, however, can be difficult as a result of restricted visibility
9 and exposure of the proximal femur. Consequently, shorter stemmed femoral components
10 continue to be developed in an effort to accommodate less invasive surgical techniques.

11 Adaptive bone remodeling as a result of stress shielding is a well-documented product of
12 altered load distribution and bone-implant stiffness mismatch [9,10]. In addition to
13 accommodating surgical approaches with smaller exposures, shorter stemmed implants
14 may additionally enable advantageous loading distribution within the femur. To enable
15 reduced stress shielding effects, stem designs must minimize the change in load
16 distribution between the intact and implanted femur. The design rationale for short
17 stemmed femoral components focuses on minimizing the risk of distal locking and
18 reducing the distal shunting of joint loads into the diaphysis. Subsequently, reduced stem
19 length implants have been clinically shown to reduce periprosthetic bone resorption when
20 compared to autopsy retrieval studies of standard components [11-13], resulting in the
21 preservation of proximal bone stock. Short stemmed designs have also shown promising
22 early survivorship with lower rates of intraoperative complications [14]. In order to
23 ensure long term clinical success, uncemented femoral components must maximize primary

stability to allow for adequate bone ingrowth, particularly when missing the distal support of a traditional length stem. Current literature lacks a paired analysis of both strain and primary stability as a result of shortening a clinically established metaphyseal fit-and-fill femoral component. Therefore, the purpose of the current study was twofold: 1) to quantify the altered mechanical load distribution in the femur as a result of reduced stem length in a clinically established fit-and-fill stem design, and 2) to evaluate and compare axial and rotational primary stability between varied stem lengths and component designs via micromotion analysis.

Materials and Methods

In order to characterize the mechanical response to loading as a function of stem design, this study was divided into two distinct phases: cortical strain and micromotion analysis. In both phases, femoral components were implanted into fourth generation composite femur specimens (Model 3403, Pacific Research Laboratories, Vashon, WA) by a board certified orthopaedic surgeon using incremental broaching and manual impaction.

Cortical Strain Analysis

Cortical strain analysis was designed to isolate reduced stem length as the sole variable in examination of cortical strain in a clinically established, uncemented fit-and-fill femoral component design. Femur specimens were implanted with one of three primary stem designs: short (96 mm from shoulder to distal tip) (JISRF Short Stem Classification 3B) [McTighe], medium (118 mm), and long/standard (151 mm) variations of a Size 11 mm, reduced proximal profile, forged titanium fit-and-fill stem (Echo® Bi-metric®, Zimmer Biomet, Warsaw, IN) (Figure 1). A fourth experimental group serving as a baseline

control consisted of intact, unimplanted composite femur models, with six ($n=6$) specimens per experimental group. Following femoral component implantation, all specimens were distally potted to a depth of 80 mm in a fast-curing polyester resin in a cylindrical aluminum mold while using digital inclinometers to insure repeatable positioning within the potted base. Following potting, specimens were prepared for digital image correlation (DIC) strain analysis by painting a black and white speckle pattern around the proximal half of the exposed femoral cortex.

Femoral strain testing was conducted using an electrodynamic load frame (ElectroPuls E10,000, Instron, Norwood, MA) incorporating a custom-designed femoral loading jig allowing for a compressive joint reaction force at the femoral head paired with a 15° abductor force applied through an abductor plate on the greater trochanter [15,16] (Figure 2). Potted femurs were aligned at 12° of valgus consistent with the scenario of single-legged stance [17]. A joint reaction force of 2060 N, approximately 3x assumed body weight of 70 kg, was applied to the femoral head paired with a 1375 N, 2x body weight, abductor force. Joint reaction force was measured via a compressive subminiature button load cell (LBS-1K, Interface, Scottsdale, AZ), while abductor force was measured via a subminiature threaded load cell (LCFD-1K, Omega Engineering, Stamford, CT). Both load cells were live monitored on a multi-channel high speed data acquisition system (System 7000, Vishay Micro-measurements, Raleigh, NC) to verify proper load distribution during each trial. These paired forces were generated through the custom loading apparatus as a result of a compressive force applied by the load frame at a rate of 60 N/s to a peak load of 685 N. All specimens were loaded with three preconditioning cycles prior to initial data collection. Because DIC is a line-of-sight measurement

technique, trials were repeated with cameras recording medial, lateral, anterior, and posterior views. Four trials were repeated for each specimen in each view.

Strain measurements were collected across the surface of the proximal femur using a speckle-based digital image correlation (DIC) system (ARAMIS 5M, GOM, Inc. Braunschweig, Germany). Full-field von Mises strain maps were collected for each femur in medial, lateral, anterior, and posterior views for each specimen. Additionally, key points of interest within each of the four strain views were selected at 2 cm increments along the long axis of the femur for quantitative strain comparison between experimental groups (Figure 3).

Micromotion Analysis

This phase of the study was designed to examine the role of stem length and component design on axial and rotational primary stability. Five experimental groups (n=6 specimens per group) consisted of short (96 mm), medium (118 mm), long/standard (151 mm) variations of the forged titanium fit-and-sill Echo® Bi-Metric® stem, a 104 mm short variation with reduced lateral shoulder (RLS) (Echo® Bi-Metric® Microplasty™, Zimmer Biomet, Warsaw, IN), and a commercially available short tapered-wedge stem (107.5 mm) (Taperloc® Microplasty™, Zimmer Biomet, Warsaw, IN) (JISRF Short Stem Classification 3A) [McTighe]0. All specimens were prepared and implanted in the same manner as in cortical strain analysis. To enable DIC micromotion analysis, three measurement viewing windows, 8 mm in diameter, were drilled through the anterior aspect of the femur to align with the proximal aspect of the component ingrowth surface, and 42 mm distally to align near the distal border of the ingrowth surface in the fit-and-

fill stems (Figure 4). A fine black and white speckle pattern was applied to the exposed surface of the implanted component through the viewing window to allow for DIC tracking of the implanted specimen at those positions. A preliminary evaluation of point-to-point DIC micromotion repeatability was carried out by measuring quantifying the displacement of a metal rod past a rigid base by means of the electrodynamic load frame. Through this method, repeatability of the current camera setup was quantified at ± 0.0034 mm. Separate validity testing was conducted in order to assess the degree of instability added to the system by introducing viewing windows in the micromotion analysis methodology.

Specimens were tested in separate axial and torsional setups to independently determine stability in each of two loading scenarios. To test torsional stability, femurs were mounted in 12° valgus with the taper head-neck junction inserted into a customized loading fixture allowing for free medial-lateral bending and translation while restraining rotation [16]. An axial load of 1675 N was held constant while a cyclic torque of +5 to - 20 Nm was applied at a rate of 1 Hz for 550 cycles [18]. Specimens were then testing for axial stability by means of a 100 N to 1675 N cyclic axial load applied to the stem again at a rate of 1 Hz for 550 cycles. All specimens underwent axial preconditioning loads to ensure full component seating and preliminary analysis which showed no significant change in micromotion within this model after 500 axial or torsional cycles. Implant stability was assessed through the quantification of the relative femoral component micromotion between the implanted stem and surrounding bone.

Statistical analysis for both strain and stability tests were performed utilizing Minitab 17 statistical software (Minitab, Inc., State College, PA). Analysis of variance (ANOVA)

techniques were used to determine the influence of stem design on strain and micromotion response at each individual measurement region. Statistically significant differences in strain or micromotion response were indicated at $p < 0.05$ in all analyses.

Results

Cortical Strain

As can be observed qualitatively through the full-field DIC von Mises strain maps (Figure 5), as well as through quantitative DIC point strain analysis, a repeatable pattern of distally shifting load distribution following stem length was observed. Full field analysis demonstrates this clear trend of increased distal strain within the long-stemmed implanted femurs, with increased proximal loading in the short-stemmed and unimplanted femur specimens. Strain results at each measurement point along the proximal femoral cortex are presented in Figure 6, with statistical analysis presented in Table 1. Quantitatively, overall strain response was greatest in the medial and lateral aspects of the femur as a result of bending under single-legged stance loading. These regions of the bone are also the areas in which the stem length of the implanted prosthesis most greatly affected bone mechanics. The unimplanted intact femur, serving as a model for an undisrupted, healthy bone, exhibited higher proximal strain response to loading than any of the implanted specimens. Following the unimplanted femur, the specimens implanted with short-stemmed prostheses generally exhibited their highest strain response near the proximal end of the femur and toward the distal end of the stem. Specimens implanted with medium-length stems exhibited higher strains relative to the other implants at the end of that stem, while the specimens implanted with long-length stems exhibited highest strains at the location of the distal end of that stem.

Laterally, the short-stemmed specimens exhibited a strain response most closely matching that of the intact femur, with strain ranging from within -5% ($p=0.119$) to +7% ($p=.032$) across all measurement regions. Comparatively, medium and long-stemmed specimens ranged from within -21% ($p<0.001$) to +4% ($p=.220$), and 27% ($p<0.001$) to +6% ($p=.064$) of the intact femur, respectively. Medially, strain in the short-stemmed specimens ranged within -43% ($p<0.001$) to +6% ($p=0.064$) of the unimplanted femur response while medium and long-stemmed specimens differed from the intact femur by -52% ($p<0.001$) to +13% ($p<0.001$) and -62% ($p<0.001$) to +6% ($p=0.043$), respectively.

Femoral Component Micromotion

In femoral component micromotion analysis, similar degrees of primary stability were observed across the majority of stem lengths and designs. Axial stability was assessed by means of micromotion data was collected at 125, 250, 375 and 500 cycles of axial loading. Because these specimens were preconditioned to ensure full implant seating, no statistically significant difference in micromotion values were observed between subsequent cycle measurements ($p=0.10$ to 0.97). Peak micromotion due to axial loading at the final measurement cycle is presented in Table 2. Three-dimensional axial micromotion ranged from 0.047 to 0.091 mm at the most proximal measurement point. There was no statistically significant difference between the short, medium, long, and RLS fit-and-fill stems, however, the tapered-wedge stem exhibited 48% less micromotion than the RLS stem at that measurement point ($p=0.014$). At the distal measurement point corresponding with the distal boundary of the fit-and-fill ingrowth surface, axial micromotion ranged from 0.052 to 0.075 mm with no statistically significant difference between any component designs ($p=0.153$). When comparing stem length exclusively, no significant difference between the short,

medium, and long stemmed versions of otherwise identical implants was observed at either border of the proximal ingrowth surface ($p>0.063$) as a result of axial loading.

Similar to axial testing, no statistically significant difference in micromotion values were observed between subsequent cycles measured in torsional loading. Peak micromotion due to torsional loading are also reported in Table 2. Torsional loading resulted in greater micromotion response across all stems versus axial loading ($p<0.001$). Torsional micromotion ranged from 0.348 to 0.409 mm at the proximal border of the ingrowth surface with the tapered-wedge stem demonstrating the lowest torsional micromotion. However, no difference instability was observed between any of the stem lengths or designs at this point ($p=0.508$). At the distal border of the ingrowth surface, peak torsional micromotion ranged between 0.213 to 0.275 mm with the tapered-wedge stem again exhibiting the greatest torsional stability. This difference was again insignificant as no difference in torsional stability was observed between femoral component stem designs at this point ($p=0.176$).

Validity measurements quantifying the effect of the viewing windows on implant stability were performed in both axial and torsional testing. At 500 cycles, the presence of view windows resulted in 0.006 ± 0.006 μm greater micromotion in axial loading and 0.011 ± 0.005 μm greater micromotion in torsional loading at the exposed surface of the implant when compared to femurs without viewing windows.

Discussion

Short stemmed femoral components more readily allow for the use of minimally invasive surgical techniques, particularly in the direct anterior surgical approach [7,14].

Furthermore, improved mechanical load distribution in the proximal femur resulting in decreased risk for stress-shielding and bone resorption are a key component of the design rationale for short-stemmed THA femoral components. To ensure clinical viability of a short stemmed femoral component, the primary stability of shortened stem must not be compromised if biological fixation and long term clinical success are to be achieved. In the current biomechanical model, a shortened variation of a clinically established fit-and-fill femoral stem demonstrated improved proximal load distribution when compared to longer stem lengths, while maintaining a comparable degree of resistance to micromotion at the bone-implant interface.

Standard length femoral stems have a solid clinical history of high survivorship at long term follow-up [2-7]. The first generation of the standard length fit-and-fill component investigated in this study has a well-established clinical history with reported survivorship of 100% at a mean of 12 years [5], and 100% survivorship at a mean of 15.2 years [6] within two independent cohort studies. The second generation version of this design, tested in this study, incorporates minor geometry changes which have resulted in 4 to 6 times decreased risk of intraoperative fracture when compared to the original stem geometry [19]. Additionally tested in this study is the short stemmed version of a standard length tapered-wedge stem with reported 99.1% survivorship at 11 years [7], and 97% at mean 16 year follow-up [20]. Molli *et al.* [14] recently reported early clinical results from the short-stemmed version of the tapered-wedge design including a significant reduction from an intraoperative fracture rate 3.1% to 0.4%, but no significant difference between the short and standard length versions in patient function, pain, or

number of reoperations. The authors concluded that the reduced stem length in the tapered-wedge design impairs neither fixation nor short-term clinical results.

Several studies have investigated femoral component design factors in an effort to promote improved loading distribution following THA. Gillies *et al.* [21] compared multiple femoral component designs of the same stem length and again showed minimal effect of those design changes on strain. Vail *et al.* [16] compared three geometrically identical femoral stems with varied modulus, 110 GPa, 52 GPa and 1.9 GPa, and still observed significant changes in bony loading and reduced proximal strain when compared to native femora. The authors concluded that stem design changes may be a more economical means to achieve higher proximal femur strain.

In the current study, the phenomenon of decreased proximal strain in longer stems, with corresponding increased distal strain in close proximity to the distal stem tip demonstrates the load shunting properties of the implanted stems. This deviation from the physiological load distribution of the native proximal femur is a mechanical process through which stress shielding is initiated. DIC strain analysis exhibited that specimens implanted with reduced length prostheses most closely matched the cortical strain distribution patterns of the unimplanted specimens and thus, within this model, would be less likely to initiate a proximal stress shielding response. In a recent study, Arno *et al.* [22] examined cortical strain in cadaveric femur specimens within short, ultra-short, and stemless variations of a design featuring a lateral flare for improved proximal fill. Parallel to the results of the current study, Arno *et al.* reported increased proximal loading, most closely matching the intact femur, in the shortest stemmed components. Several other studies have presented similar findings, with short stemmed components improving bone strain distributions and

lowering risk for stress shielding and subsequent bone resorption compared with longer stems across a variety of stem designs [23-27]. Similarly, Decking *et al.* [25] reported significant improvement in strain distribution when comparing a stemless femoral neck prosthesis to two long stemmed prostheses. However, this implant subsequently demonstrated poor clinical performance with high rates of early and mid-term aseptic loosening [28]. While improved strain distribution may decreased the risk for bone resorption and aseptic loosening, the failure of some bone-conserving designs demonstrates the need for component design focused not only on bone conservation and improved proximal loading, but additionally coupled with robust primary stability enabling biological fixation and long-term clinical success.

Laboratory studies have shown that primary stability is essential for biological fixation and bone ingrowth in cementless constructs. Recurrent bone-metal relative micromotion greater than 150 μm has been correlated with the development of a fibrous layer at the interface [29]. Conversely, post-operative micromotion less than 40 μm is adequately stable for bone ingrowth and stable fixation [30].

Investigations in the primary stability of THA femoral components have utilized bench testing in cadaveric [23, 24, 25, 31, 32] and composite [18, 33- 38] femur specimens, as well as finite element analysis [23, 24, 34, 39, 40] in reporting a wide range of stability data, dependent on measurement techniques, loading profiles, and setup design. Prior *in vitro* THA micromotion studies have primarily used linear variable differential transformer (LVDT) or differential variable reluctance transducer (DVRT) sensors to quantify implant stability: instrumentation which typically require complex instrumentation to be rigidly fixed to the specimens and allow for displacement

measurements at a single point in space [31, 33-37, 40-43, 48]. In contrast, DIC micromotion analysis is a non-contact measurement which allows for three dimensional relative motion to be captured without the need for secondary fixation of a transducer onto the bone specimen. Viewing windows are needed for internal micromotion capture, but those windows have minimal effect on resultant micromotion, accounting for a small percentage of the overall instability of the implanted stem system.

True values of micromotion are difficult to make interstudy comparisons as most studies have selected varying loading profiles, test mediums, experimental setups, and measurement techniques. For example, a recent finite element model of short and long stem THA designs reports large micromotions up to 1.017 mm for short stem and 0.727 mm for long stem, with 40-94% greater micromotion in the short compared to long stem design [27]. Other studies have reported a wide range in micromotion from less than 0.032 mm to up to 0.526 mm cadaveric micromotion studies in various stem designs during bench tests in cadaveric and composite femur specimens [23, 24, 26, 31, 33-37, 44]. Because of these differences, the most notable conclusions from THA primary stability investigations are those made within the individual studies comparing the effects of design changes on the relative changes in component stability. These comparisons are further strengthened when new iterations device designs are compared, within the same experimental protocol, to clinically successful stems.

In an early finite element analysis, Biegler *et al.* [45] observed that torsional loading, associated with activities such as stairclimbing, results in a greater bone-implant interface micromotion than purely axial loading. These findings match well to the current study where mean axial micromotion ranged between 0.047 and 0.091 mm, while mean

torsional micromotion ranged between 0.219 mm and 0.438 mm across all implant designs. Similar to the current results, in a later study cadaveric study, Bieger *et al.* [24] also observed that within a curved, triple-tapered stem design, a short stem did not negatively influence primary stability when compared to standard length straight stemmed, ribbed prosthesis. Additionally, the authors observed a trend to greater rotation stability in other short stemmed femoral components [23]. Interestingly, we also observed slight trends in increased rotational stability in shorter stemmed prostheses in the current, however, this difference also statistically non-significant.

We acknowledge several limitations in the methodology in this study. First, we used a manufactured composite femur model as a substitute for native tissue. Having used these manufactured specimens in several studies, it is our experience that composite long bones are stiffer than their cadaveric counterparts and thus generate lower cortical strains. Composite femurs have been validated against cadaveric specimens and demonstrate improved interspecimen uniformity. These specimens provide extremely uniform test media allowing for comparative analysis between strain and micromotion responses as a function of varied device design which might otherwise be eclipsed by interspecimen variability within native tissue. The authors acknowledge that these specimens do not generate strain and micromotion responses equivalent to *in vivo* clinical or *in intro* cadaveric models. For example, torsional micromotion in all stems tested in the composite femurs demonstrated movement greater than would allow for clinical ingrowth in a physiological specimen, even stem designs well supported in the literature by clinical survivorship and positive patient outcomes. We therefore find it important to draw comparisons of strain and micromotion response relative to clinically established designs

in which biological ingrowth and adequate clinical performance has been well documented. Secondly, because DIC is a line-of-sight measurement technique, micromotion analysis of the implanted component at the ingrowth surfaces required viewing windows compromising contact at those locations. These viewing windows were created at the smallest possible size to allow for visualization of the speckled component while minimizing the effect on component stability. As reported in validity testing, the presence of viewing windows did increase micromotion by 0.005 to 0.011 mm, however this represents a small contribution in error relative to the overall micromotion measured in these specimens. Third, this model does not account for the time-dependent bone remodeling response, and thus represents an early post-operative model. Fourth, both strain and micromotion setups were simplified representations of a complex dynamic gait. The model did not fully account for the numerous soft tissue interactions which actively contribute to bony loading and joint stability. Finally, the testing regimen incorporated loads associated with a 70 kg patient, while patient weight varies greatly in clinical practice. As a result, reported strain and micromotion values in this loading model may be lower than in a higher BMI patient model; however, relative comparisons and observed trends should remain consistent.

In conclusion, the findings of this laboratory study suggest that short stemmed metaphyseal fit and fill implants may reduce the risk for stress shielding by eliciting a femoral loading response similar to that of the intact femur, without sacrificing primary implant stability in a composite femur model. While improving strain distribution,

322 shortening the stem length of this component design demonstrated no significant
323 degradative effects on the axial or torsional primary stability of the implanted stem.

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Figure Legend

Figure 1: Left to right: long, medium, short, and reduced lateral shoulder variations of the fit-and-fill stem design, with the short-stemmed tapered-wedge stem at the far right. The first three were utilized in strain testing, while all five were included in micromotion analysis.

Figure 2: Black and white speckled femur within the customized loading apparatus [15,16] incorporating a distal universal joint along with proximal loading via a compressive joint reaction force and a tensile abductor force.

Figure 3: Key points of interest selected for quantitative strain comparison between experimental groups. Point A1 is 2 cm distal to the anterior tubercle of the greater trochanter and P2 was placed at the lesser trochanter. All other points are placed at 2 cm increments along the long axis of the femoral shaft.

Figure 4: Left to right: long, medium, short, and reduced lateral shoulder variations of the fit-and-fill stem design, with the short-stemmed tapered-wedge stem at the far right. Micromotion measurement points are placed on each component corresponding with the proximal and distal ingrowth boundaries of the fit-and-fill stem.

Figure 5: Representative digital image correlation full-field von Mises strain analysis in each of the four fields of view.

Figure 6: Von Mises strain response to single-legged stance loading in (A) anterior, (B) posterior, (C) medial, and (D) lateral measurement regions.

Load
Application

15°

Abductor
Plate

12°

Anterior

Medial

Posterior

Lateral

Table 1: Strain Comparison at All Measurement Points, P-Values

	A1	A2	A3	A4	A5	A6	A7
Long to Medium	<0.001	<0.001	<0.001	0.018	0.491	0.076	0.002
Long to Short	<0.001	<0.001	0.012	0.564	0.635	0.181	0.007
Long to Unimplanted	0.001	0.354	0.027	0.050	0.111	0.643	<0.001
Medium to Short	0.737	0.516	0.076	0.054	0.727	0.629	0.510
Medium to Unimplanted	0.001	<0.001	<0.001	<0.001	0.001	<0.001	0.033
Short to Unimplanted	0.001	<0.001	<0.001	0.004	0.001	0.010	0.149

	P1	P2	P3	P4	P5	P6	P7	P8	P9
Long to Medium	<0.001	<0.001	0.014	0.438	0.264	0.357	0.052	0.039	0.069
Long to Short	<0.001	<0.001	<0.001	0.045	0.860	0.189	0.036	0.827	0.062
Long to Unimplanted	<0.001	<0.001	0.347	0.595	0.906	0.582	0.910	0.874	<0.001
Medium to Short	<0.001	0.840	<0.001	0.089	0.188	0.406	0.938	0.004	0.975
Medium to Unimplanted	<0.001	0.138	0.630	0.061	0.033	0.348	0.002	0.007	0.052
Short to Unimplanted	<0.001	<0.001	0.001	0.005	0.910	0.072	<0.001	0.948	0.054

	M1	M2	M3	M4	M5	M6	M7
Long to Medium	<0.001	<0.001	<0.001	<0.001	0.058	0.174	0.257
Long to Short	<0.001	<0.001	<0.001	0.014	0.605	0.215	0.516
Long to Unimplanted	<0.001	<0.001	<0.001	<0.001	0.155	0.043	0.020
Medium to Short	<0.001	0.003	0.610	0.012	0.207	0.916	0.056
Medium to Unimplanted	<0.001	<0.001	0.064	<0.001	<0.001	0.646	0.001
Short to Unimplanted	<0.001	0.799	0.363	0.310	0.064	0.568	0.018

	L1	L2	L3	L4	L5	L6	L7	L8
Long to Medium	0.221	0.479	<0.001	<0.001	0.006	0.970	0.005	0.145
Long to Short	0.016	<0.001	<0.001	<0.001	0.078	0.950	0.741	0.082
Long to Unimplanted	0.043	<0.001	<0.001	<0.001	0.101	0.457	0.007	0.131
Medium to Short	0.117	<0.001	<0.001	0.603	0.268	0.906	0.020	0.002
Medium to Unimplanted	0.178	<0.001	<0.001	0.527	0.220	0.408	0.470	0.006
Short to Unimplanted	0.946	0.119	0.085	0.151	<0.001	0.314	0.032	0.920

Table 2: Peak Axial and Torsional Micromotion at Proximal and Distal Measurement Locations after 500 Cycles

	Axial Micromotion (mm)		Torsional Micromotion (mm)	
	Proximal	Distal	Proximal	Distal
Long	0.082 ± 0.029	0.061 ± 0.017	0.395 ± 0.035	0.275 ± 0.070
Medium	0.078 ± 0.010	0.052 ± 0.013	0.409 ± 0.038	0.269 ± 0.042
Short	0.073 ± 0.018	0.067 ± 0.009	0.375 ± 0.088	0.240 ± 0.025
RLS	0.091 ± 0.020	0.075 ± 0.008	0.371 ± 0.035	0.219 ± 0.023
Tapered-Wedge	0.047 ± 0.026	0.057 ± 0.013	0.438 ± 0.053	0.213 ± 0.031
<i>Anova</i>	<i>p=0.038</i>	<i>p=0.153</i>	<i>p=0.508</i>	<i>p=0.176</i>

* RLS vs. Tapered-Wedge axial micromotion at proximal point ($p=0.014$); all other comparisons ($p>0.05$)