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Interface Micromotion of Uncemented Femoral Components from Postmortem Retrieved Total Hip Replacements

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Abstract

Axial torsional loads representative of gait and stair climbing conditions were applied to transverse sections of 8 uncemented postmortem retrievals and a high-resolution imaging system with digital image correlation was used to measure local micromotion along the bone-implant interface. For seven components that were radiographically stable, there was limited micromotion for gait loading ($1.42 \pm 1.33 \mu\text{m}$) that increased significantly ($p=0.0032$) for stair climb loading ($7.32 \pm 9.96 \mu\text{m}$). A radiographically loose component had motions on the order of 2.3 mm with gait loading. There was a strong inverse relationship between the amount of bone-implant contact (contact fraction) ($p=0.001$) and micromotion. The uncemented components had greater contact fraction ($41.8 \pm 14.4\%$ vs. $11.5 \pm 10.2\%$, $p=0.0033$) and less median micromotion ($0.81 \pm 0.79 \mu\text{m}$ vs. $28.8 \pm 51.1 \mu\text{m}$) compared to a previously reported study of cemented retrievals.

Introduction

Uncemented and cemented femoral implant components have been used successfully in total hip replacements (THR) with many reports of long term, stable fixation [1–5]. The method by which uncemented and cemented components achieve fixation is somewhat different. Uncemented components rely on initial stability via rigid press-fit macro-lock to provide conditions conducive for bony apposition and ingrowth to the metallic stem surface (micro-lock). In contrast, cemented constructs achieve interlock by flow of cement into trabecular and lacunar spaces in cortical bone. In both instances, there is a biological response of the bone to the implantation, and this can affect the fixation of the component. Recent work with postmortem retrieved cemented hip replacements has shown that there is small, but measurable micromotion at the cement-bone interface [6]. It also appears that this interface micromotion increases from the initial ‘freshly cemented’ condition due to changes in the morphology of the cement-bone interface [7]. Further, the magnitude of interface micromotion is inversely proportional to the amount of bone in contact with the cement.

There have been several reports of stability measurements of postmortem retrieved, uncemented femoral components [8, 9], but these rely on indirect measures of micromotion

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at the interface between implant and bone. The direct measurement of micromotion between implant and bone of well-fixed uncemented components from postmortem retrievals has not been reported. We used a torsional loading apparatus [10] to apply 'functional' loading to postmortem retrieved uncemented components and directly measured the relative motion at the implant-bone interface. We asked four research questions: (1) how large is the inducible micromotion at the implant-bone interface for uncemented THR subjected to gait and stair climbing loading conditions?; (2) is there a relationship between bone-implant apposition and micromotion?; (3) does the amount of micromotion depend on axial position along the stem; and (4) how does micromotion of uncemented THR compare to cemented THR?

Methods

Eight post-mortem retrieved uncemented implants were obtained from the SUNY Upstate Anatomical Gift Program (Table 1). Donor gender, age, body mass index, and time in service were collected. The implant type and manufacturer were determined from markings on implant and implant shape. Plane radiographs were evaluated to document implants that were well-fixed or loose by our orthopaedic surgeon (TI). Each implant/femur construct was sectioned transversely using a water irrigated silicon carbide blade at 10mm intervals, starting at the lesser trochanter and moving distally. Specimens at the lesser trochanter level (0 level), 10 mm distal from the lesser trochanter (10 mm level), and 20 mm distal from the lesser trochanter (20 mm level) were chosen for testing because they represent regions of metaphyseal load transfer from stem to bone and had implant coatings over these levels. The surfaces of the 10 mm thick transverse sections were polished to 1200 grit using a water irrigated polisher. High-resolution white light images (1.8 $\mu\text{m}/\text{pixel}$) were captured of each specimen surface to document morphology.

A torsion loading device with provision to sample images of the transverse stem-bone surface during loading was used (Fig 1a). Details on the system are described elsewhere [10]; a brief summary of the approach is included here for completeness. The periosteal surfaces of the specimens were bonded to cut-outs in acrylic blocks with epoxy and mounted in a pure torsion loading device. An overspray of black and white acrylic paint was applied to the stem-bone surface to provide texture for digital image correlation (DIC) analysis. The loading device was immersed in a calcium buffered saline bath at 37°C and an imaging system was used to capture images of the transverse section during loading. The relative motion between the implant and bone (micromotion) was quantified using eight pairs of sampling points (Fig 1b), spatially located at 45° intervals (Fig 1b) around the interface. If there was no bone in the vicinity of the interface (shown as open squares in Fig 1b), a measurement was not made at that location.

Morphology Measurements

A stereology method (Fig 2) was used to determine the *contact fraction* between stem and bone. One hundred lines were projected orthogonally across the stem-bone interface and the fraction of lines that crossed the interface where there was stem-bone apposition was calculated. The *texture fraction* was calculated as the portion of the stem section contour with a roughened (not smooth) surface. The *perimeter length* of the stem section was determined by tracing the outer contour of the stem. The *aspect ratio* was calculated as the major to minor aspect of the stem section.

Gait Loading Conditions

Gait loading was simulated using torques of 0.73 N-m in retroversion followed by 0.22 N-m of ante-version [10]. Specimens were loaded for three full preconditioning cycles (Fig 1c) and images were captured at 4 Hz during the fourth cycle of loading. Because motions were

very small, very high resolution images (3 $\mu\text{m}/\text{pixel}$) were captured at eight locations around the stem (Fig 1b) during the loading and unloading cycle (Fig 1c). At this resolution, the RMS error of the DIC measurements was calculated to be 0.1 pixel or 0.3 μm [10]. The loading sequence was completed for each sampling location and relative normal, shear, and total motion was calculated between the peak retroversion and ante-version torques. The global response of the sections was characterized (Fig 1c) in terms of global torsional stiffness and torsional span. Gait loading was applied to the '10 mm level' sections. The results from the gait loading condition for the uncemented postmortem retrievals were compared to previously reported results for cemented postmortem retrievals using the same loading protocol and DIC measurement system [6].

Stair Climbing Loading Conditions

To simulate stair climbing conditions, the retroversion torque magnitude was increased to 2.6 N-m; this was based on torques measured during stair climbing for an instrumented hip [11]. A 40 Nm torque uniformly applied over a 150mm stem length would be equivalent to 2.6Nm applied to our 10mm sections. Following one preconditioning cycle, images were captured of the complete implant-bone interface using a high resolution image (20 $\mu\text{m}/\text{pixel}$). At this resolution, the RMS errors are calculated to be 2.0 μm . Total micromotion at eight sampling locations was calculated for excursions between peak retroversion (2.6 N-m) and ante-version (0.22 N-m). Stair climbing loading was applied to the 0, 10, and 20 mm level sections.

To gain a better understanding of the local displacement field across the bone-implant interface for a well-fixed component (donor bone E) with stair climbing loads, an 8 $\times$ stereo-microscope system (0.76 $\mu\text{m}/\text{pixel}$) was used with the DIC technique. With this approach, a static load with 2.6 N-m torque was applied to the construct and a digital image was captured before loading and after static load application. This approach was repeated for a cemented retrieval section used in a previous study [6].

Statistical Analysis

Regression analysis was used to relate the independent variables of morphology (such as contact fraction), loading condition, specimen type, specimen level and the main dependent variable of bone-implant micromotion. Because of the large and non-normal distribution of contact fraction and micromotion, these were log transformed before statistical analysis.

Results

The contact fraction between stem and bone ranged widely from 0% for one case with a radiographically loose prosthesis to 72% for a radiographically well-fixed component (Table 1). Examples from sections near the lesser trochanter illustrates a case with bony ingrowth into a beaded surface (donor bone H, Fig 3a), apposition between stem and bone for a plasma spray/HA coated stem (donor bone F, Fig 3b), and a loose satin finish stem with organized fibrous tissue between stem and bone (donor bone B, Fig 3c). One stem was radiographically loose, two had some indications that the stems were not fully integrated with the bone, and the remaining five were radiographically well-fixed.

For the seven components that were radiographically stable, there was limited micromotion between stem and bone for gait loading ($1.42 \pm 1.33 \mu\text{m}$, range: 0.39–4.01 μm) that increased significantly ($p=0.0032$, paired t-test) for stair climb loading ($7.32 \pm 9.96 \mu\text{m}$, range: 1.26–28.87 μm) (Fig 4). The radiographically loose component had excessive motions (on the order of 2.3 mm) with gait loading, and was therefore not subjected to stair climb loading. Using a regression model ($r^2=0.92$, $p<0.0001$) with micromotion as the dependent variable,

we found a strong inverse relationship with independent variables contact fraction ($p=0.001$) and load type (gait or stair) ($p=0.0008$). We also found that the increase in micromotion due to changing from gait to stair climb loading was inversely proportional to contact fraction ($p=0.03$). That is, specimens with less contact fraction had greater increases in micromotion when switching from gait to stair climb loading.

For the investigation of micromotion with stair climb loads applied to different section levels (Fig 5), two of the 20mm level sections were lost during specimen preparation; one due to mechanical damage to the interface during preparation for a case where there was very low contact fraction between cement and bone, and one due to damage to the stem component during a drilling operation to attach the torsion loading device. This left a total of 19 specimens (7- 0mm level, 7 – 10mm level, and 5–20mm level) for analysis from well-fixed components. Using a multiple regression model ($r^2=0.76$, $p<0.0001$), there was a significant inverse contribution from contact fraction ($p<0.0001$) and a significant positive contribution from section level (0, 10, or 20mm, $p=0.044$) in predicting interface micromotion. These results suggest that increasing contact fraction reduced interface micromotion, while moving distally from slice 0 to 20mm increased interface micromotion. The contact fraction did not change much moving from the 0 to 20mm section ($49\pm13\%$, $42\pm14\%$, and $46\pm19\%$, for 0, 10, and 20mm sections) and there was a wide range of responses for the same section level (see Fig 5). As such, there must be other factors that were not captured in the regression model to explain why there would be more motion moving from the 0 to 20mm section. It should be noted that the other morphology parameters (texture fraction, perimeter length, aspect ratio) did not contribute significantly to the regression models. Increasing the number of test specimens at each level could identify additional factors contributing to interface stability.

The uncemented components had significantly greater contact fraction ($p=0.0033$) when compared to 10 cemented retrievals [6] that were radiographically stable (Table 2). In terms of global response of the constructs, the torsional stiffness of the constructs was approximately twice as large for the uncemented components ($p=0.0194$) and the torsional span was two orders of magnitude lower ($p=0.0048$) compared to previously tested cemented components. The shear, normal, and total micromotions were more than an order of magnitude smaller for the uncemented components. There was a strong power-law relationship ($R^2=0.88$, $p=0.0001$) between contact fraction and micromotion (Fig 6) for the combined uncemented and cemented implant data sets. The two radiographically loose components, one uncemented and one cemented, are included in the graph for completeness. Overall, it is evident that there is limited overlap between the uncemented and cemented data sets for the components that were radiographically stable.

Close-up digital image correlation analysis of sections subjected to stair climbing loads showed that a well-fixed uncemented construct (Fig 7a) exhibits no slip across the interface. In this case, the micromotion across the interface would be zero, providing evidence that the bone is chemically or mechanically bonded to the metal. In contrast, a typical cemented construct (Fig 7b) exhibited a displacement jump across the interface. The cement was typically separated from the bone by a very small gap that opened or closed under load.

Discussion

The results of this study suggest that there is minimal micromotion at the implant-bone interface for uncemented total hip replacements that are radiographically well-fixed. In regions of bony ingrowth or apposition, the bone-implant interface had micromotions on the order of one micrometer for gait loading conditions. However, a radiographically loose implant was surrounded by extensive fibrous tissue with no bone-implant contact; this

implant moved independently from the bone. Increasing loading to stair climbing levels caused an increase in micromotion as would be expected, but the increase in motion was greater of constructs with less contact between cement and bone. The magnitude of micromotion was found to depend on section level, with more distal sections having more micromotion.

Overall, the well-fixed, uncemented retrievals had less micromotion and greater contact fraction at the implant-bone interface compared to well-fixed cemented retrievals [6]. There was some overlap between the implant types; micromotions from the best-fixed cemented implant were similar to the less-well fixed uncemented components. It is interesting to note that previous work [7] with cemented stem constructs prepared in the laboratory, had contact fractions (58%) and interface micromotions (1.2 μm) similar to the uncemented components studied here. The reduced contact fraction and increased micromotion for the previously reported cemented postmortem retrievals [6] could be due to differences in cementing technique, with the laboratory prepared constructs more aggressively pressurized than the postmortem retrievals. Or, there could have been bony remodeling changes at the cement-bone interface in the retrieval cases, resulting in less cement-bone contact and more interface micromotion.

There have been a number of different approaches used to measure motion of implant components relative to bone following mechanical loading of bone implant systems. The simplest approach is use of external extensometers attached between the metal component and the bone [12]. While this is useful for measuring motion of the stem relative to the bone in the region of the calcar, there is not a direct measure of interface micromotion, nor is there any information provided as to motion over the distal portions of the construct. Full 6 degree-of-freedom systems have been used to document global migration and inducible displacement of stems relative to bone [13–16]. These systems incorporate a triad of spheres attached to the stem through a drill hole in the bone and 6 LVDT's to measure motion of the spheres relative to the bone. These systems are particularly useful for fatigue studies to quantify migration and inducible motion of the stem, but are not capable of measuring motion at the interface. Drilling through the bone at different locations along the construct and attaching extensometers or LVDT's provides more information regarding spatial distribution of micromotion [17–20]. However the measurements assume that the bone is a rigid body relative to the moving implant.

Burke and coworkers [17] used an extensometer based system to measure micromotion between the cortical bone shell and stem for laboratory prepared constructs that had either cemented or uncemented components. These constructs would be representative of conditions that would exist in the immediate post-operative state. In the Burke study, micromotions were similar for cemented and uncemented implants under single legged stance loading conditions (mean total motion of 12 μm cemented, 13 μm uncemented). However, increasing the loads to simulate stair climbing resulted in increased micromotion more for uncemented (104 μm) compared to cemented (27 μm) implants. The micromotion magnitudes for the uncemented stems in the Burke study were much larger than those measured in the present study. This is likely due to the fact that bony ingrowth was not present in the Burke study because implants were placed postmortem. In contrast, the micromotion magnitudes for the cemented stems in the Burke study were less than those presented here. Previous work in our laboratory has shown that laboratory prepared cemented constructs have much less micromotion when compared to postmortem retrievals [7].

Engh et al [8] reported on a series of 14 postmortem retrievals with porous coated anatomic medullary locking prostheses (AML) using extensometers that measured motion between

the stem and the bone for simulated gait and stair climbing loads. The maximum relative motion between implants and bones in regions of porous coating was 40 μm for 13 cases that were well fixed. In all of these cases, the motion was said to be elastic, suggesting that when load was removed, there was no permanent micromotion at the interface. For one case that did not have bony ingrowth, micromotion increased to 150 μm . The results of Engh et al (1992) are generally consistent with those of the present study, in that large motions were not found for the uncemented components.

There has been long standing interest in the relationship between micromotion and osseointegration of the implant to bone. Cylindrical porous coated implants placed in the distal femoral metaphysis in dogs [21] were found to have bony ingrowth and continuity between the ingrown bone and surrounding bone when micromotion was 0 or 20 μm . When micromotion was increased to 40 or 150 μm the implant surfaces were often surrounded by fibrocartilage or fibrous tissue. Liu and Niebur [22] predicted that 20 μm of transverse micromotion at a porous coated surface would result in soft tissue formation at the bone-implant interface using a mechano-regulatory tissue differentiation algorithm. When micromotion was reduced to 10 or 5 μm , bone was predicted to fill the porous coating surface. The current investigation is not able to determine the amount of micromotion that could be allowed and still achieve fixation, but is able to document the amount of micromotion present after *in vivo* service. Even for a very short term implant (donor bone G, 0.25 years in service), interface motions were less than 10 μm with gait loading, suggesting that more complete integration of the implant would occur over time. Indeed, the contralateral limb from this donor (donor bone F, 1 year in service) shows less interface micromotion and greater contact fraction between implant and bone.

There are several limitations to this study. First, loading was only in pure torsion. Coupled loading with axial compression, torsion, and bending would occur *in vivo*. Further, we assumed that torques would be applied uniformly along the length of the stem. However, since we applied the same torques to all sections, we could still compare response with the independent variables (such as contact fraction) and implant type (uncemented or cemented). There are too few samples to make meaningful conclusions about how the type of stem, surface treatment, donor age, and time in service influence fixation. A larger sample population would be desirable to determine conditions that result in the most stable fixation.

In summary, by direct measurement of the interface micromotion between implant and adjacent bone, we show that the uncemented implant-bone interfaces tested in this study can behave as a bonded interface (as illustrated in Fig 7) with negligible micromotion due to loads consistent with activities of daily living. In contrast, the cemented implants tested previously in our lab [6] have small, but measurable micromotion and are not as well integrated to the femoral canal. The differences in interface compliance between these two implant types would likely affect the way load is transferred between implant and bone. In addition, for high demand patients with cemented implant, large micromotions between cement and bone could result in abrasive wear [23] and cyclic pressurization of the interface [24], resulting in progressive loss of fixation. For these reasons, the senior author restricts cemented femoral component fixation to older and low demand patients.

The debate over choice of fixation method for the femoral component of total hip replacement continues [25]. In the United States, the use of cemented femoral components has diminished to a small fraction of all procedures, while in parts of Europe, particularly Scandinavia, the use of cemented fixation is dominant. Interestingly, new results from the Swedish registry [1] suggest that the relative risk of revision (0.4) is lower for uncemented stems compared to cemented stems. It is important to note that the comparisons made here between uncemented and cemented femoral implants from hip replacement were from a

small sample (19 postmortem retrievals total) and that this sample may or may not represent the full population of both fixation types. However, the finding that increased contact fraction reduced interface micromotion, likely is relevant to all implant systems.

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Figure 1.

The mechanical test setup (A) is shown without the front cover on the saline bath environmental chamber. Load application is applied to a lever arm to provide a pure torque via the axle housing to the stem section. Cross section of a stem-bone construct (B) illustrating retroversion torque (T) applied to the center of the stem. Interface micro-motion was measured using eight pairs of sampling points on either side of the interface. Locations where there was no bone, indicated here as open white squares, were not used in the measurements. The global response (C) was measured as applied torque (T) versus angular rotation (ϕ) with global stiffness and span as the main outcome measures.

Figure 2.

A stereology approach was used to determine contact fraction across the implant-bone interface. One hundred lines were projected across the implant-bone interface and the number of projection lines that indicated apposition at the tray-bone interface (red) were counted and divided by the total number of projection lines.

Figure 3.

Section images of representative uncemented femoral component postmortem retrievals at or near the less trochanter. Donor bone H (panel **A**) had a contact fraction of 68% along the implant-bone interface, donor bone F (panel **B**) had a contact fraction of 33%, and donor bone B (panel **C**) had no implant-bone contact. Donor bone B also had a femoral stem that was highly retroverted relative to the anatomy of the lesser trochanter. Inset images include a scale with 1 mm length.

Figure 4. Mean and standard deviation of micro-motion for eight uncemented postmortem retrieval sections subjected to torsional loads consistent with gait and stair climbing loading conditions. Donor bone B was not loaded with the stair climbing loading condition because of the excessive motion for this loose component.

Figure 5.

There was a power-law relationship between interface contact fraction and interface micro-motion for uncemented sections subjected to stair climbing loads ($r^2=0.75$).

Figure 6.

There was a power law relationship ($r^2 = 0.88$) between the median micro-motion measured at the implant-bone (for uncemented) and cement-bone (cemented) interface and the contact index for the gait loading condition. The two constructs indicated with an asterisk (*) are highlighted in Figure 7.

Figure 7. Digital image correlation mapping of total motion for an uncemented component (donor E, top) and a cemented component (bottom) subjected to stair climbing loads. The corresponding interface micro-motion and contact fraction are indicated in Figure 6 (*). Color key indicates motion in microns. Bone (B), cement (C), and titanium implant surface (I) are indicated on figures.

Donor and implant characteristics for the 8 uncemented (U) femoral stems and 11 cemented (C) femoral stems reported on previously [6]. Articulation type was total hip (TH), bipolar (B), or hemi-arthroplasty (H). The time in service was not available for 3 donor bones with uncemented hips and 2 donor bones with cemented hips. The range of measurements from three sections is given for contact fraction, texture fraction, perimeter length, and aspect ratio for the uncemented hips. Texture fraction was not relevant for the cemented hips.

Table 1

Fixation Type - Donor	Gender/Age	BMI	Cause of Death	Time in Service (years)	Radio- graphically Loose	Contact Fraction (%)	Texture Fraction (%)	Perimeter Length (mm)	Aspect ratio	Implant (type) - Manufacturer	Surface Finish
U-A	F/58	26.6	Renal failure	21	N	40 – 55	100 – 100	37 – 45	1. 2 – 1.6	AML (TH) - Depuy	Sintered beads
U-B	F/92	23.4	Renal failure	17	Y	0 – 0	0 – 0	46 – 59	1.6–2.2	3M (B)	Satin
U-C	F/78	26.2	Respiratory failure	NA	N	46 – 59	29 – 80	46 – 51	1.1 – 1.2	Secur-Fit Plus (B) - Stryker	CP Ti arc deposition
U-D	F/86	31.6	Septic shock	NA	N	25 – 28	35 – 64	49 – 59	1.0 – 1.3	Fiber Metal Midcoat (B) - Zimmer	Fiber Ti Metal
U-E	F/75	32.3	Lung cancer	NA	N	50 – 57	19 – 65	51 – 82	1. 1 – 1.6	Secur-Fit Plus (B) - Stryker	CP Ti arc deposition
U-F	F/87	23.9	Pulmonary infection	1	P	27 – 55	57 – 100	49 – 64	1. 3 – 2.3	Accolade TMZF Plus (H) - Stryker	HA coated plasma spray
U-G	F/87	23.9	Pulmonary infection	0.25	P	12 – 37	81 – 100	49 – 64	1. 9 – 2.5	Accolade TMZF Plus (B) - Stryker	HA coated plasma spray
U-H	M/61	32.9	Cardiac pulmonary infection	3	N	60 – 72	12 – 100	51 – 59	1. 4 – 1.8	Synergy Porous-Coated stem (TH) -Smith & Nephew	Sintered beads
C-A	F/76	26.6	Breast Cancer	2	N	20	-	43	1.1	Modular calcar replacement (B) – Zimmer	Satin
C-B	F/76	26.6	Breast Cancer	5	N	7	-	39	1.2	Precision long stem (B) – Howmedica	Grit blasted
C-C	F/87	27.1	Cardiac Arrest	0.9	N	1	-	36	1.2	Perfecta (H) – Wright Medical Technology	Grit blasted
C-D	M/88	20.7	Cardiac Arrest	0.2	N	4	-	46	1.2	Perfecta PDA (HE) – Wright Medical Technology	Grit blasted
C-E	F/80	26	Cardiac Arrest	20	N	14	-	38	1.2	Muller Curved (TH) – JRI Ltd.	Satin
C-F	F/77	15.4	Adeno-carcinoma	NA	P	16	-	43	1.0	Cemented F Series (B) - Implex	Satin
C-G	F/93	24.9	Renal insufficiency	NA	N	33	-	41	1.1	Omnifit w/proximal rough surface (B) - Osteonics	Grit blasted
C-H	F/92	23.8	Renal insufficiency	6	N	1	-	34	1.2	Endurance (TH) – Depuy Orthopedics	Satin
C-I	F/85	29.6	Bacterial endocarditis	8	P	3	-	38	1.0	Versys cemented (TH) – Zimmer	Satin
C-J	F/85	29.6	Bacterial endocarditis	8	Y	0	-	37	1.1	Versys cemented (TH) – Zimmer	Grit blasted
C-K	F/67	20.8	Alzheimer’s disease	14	N	7	-	50	1.1	Harris precoat (TH) - Zimmer	Satin

Interface contact and mechanical response parameters for uncemented and cemented implants that were radiographically stable. Measurements were made between metal implant and bone for the uncemented implants and between cement and bone for cemented implants. Two-sample t-tests with Hochberg-Bonferroni correction for multiple sampling was performed on log transformed data. Log transformation was performed to normalize data prior to statistical testing. Note that the cemented implant results were from a previously reported study [6].

Table 2

	Uncemented implants (n=7)			Cemented implants (n=10)			p-value
	Mean (SD)	Range		Mean (SD)	Range		
Interface contact fraction (%)	41.8 (14.4)	25 – 68		11.5 (10.2)	0.76 – 32.5		0.0033
Global torsional stiffness (Nm/deg)	3.07 (0.77)	1.86 – 4.02		1.49 (1.15)	0.07 – 3.72		0.0194
Torsional span (deg)	0.014 (0.014)	0.003 – 0.43		1.85 (3.28)	0.003 – 9.5		0.0048
Median shear micro-motion (µm)	0.56 (0.35)	0.21 – 1.21		26.8 (50.1)	1.9 – 166		0.0006
Median normal micro-motion (µm)	0.35 (0.24)	0.14 – 0.71		8.3 (12.6)	0.72 – 41		0.0016
Median total micro-motion (µm)	0.81 (0.79)	0.32 – 2.55		28.8 (51.1)	2.2 – 169		0.0006