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Damage in total knee replacements from mechanical overload

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ABSTRACT

The mechanical loads acting across the knee joint following total knee replacements (TKR) during activities of daily living have recently been measured using instrumented TKRs. Using a series of post-mortem retrieved TKR constructs we investigated whether these mechanical loads could result in damage to the implant bone interface or supporting bone in the tibia. Eighteen cemented *en bloc* tibial components (0 to 22 years in service) were loaded under axial compression in increments from 1 to 10 times body weight and digital image correlation was used to measure bone strain and interface micromotion during loading and unloading. Failure was considered to occur when micromotion exceeded 150 μm or compressive bone strain exceeded 7300 με. The results show that all retrieved specimens had sufficient bone strength to support most activities of daily living, but ~40% would be at risk under larger physiologic loads that might occur secondary to a higher impacts such as jogging or a stumble. The tray-bone micromotion (regression model R^2=0.48, p=0.025) was greater for donors with lower age at implantation (p=0.0092). Proximal bone strain (model R^2=0.46, p=0.03) was greater for donors with longer time in service (p=0.021). Distal bone strain (model R^2=0.58, p=0.005) was greater for donors with more time in service (p=0.0054) and lower peri-implant BMD (p=0.049). High mechanical overload of a single or repetitive nature may be an initiating factor in aseptic loosening of total joint arthroplasties and should be avoided in order to prolong the life of the implant.

1. Introduction

Total knee replacement (TKR) utilization continues to grow at a rapid pace in the US with nearly 1 million primary knee replacements predicted to be implanted in 2015 (Kurtz et al., 2014). Unfortunately, revision surgery to replace failed primary arthroplasties also continues to grow in number. Cumulative results from a large series of joint registries indicate that revision rates are 6% at 5 years and 12% after 10 years of in vivo service (Labek et al., 2011). Aseptic loosening continues to be the predominant mechanism of failure (Lombardi et al., 2014; Sharkey et al., 2014), but it is acknowledged that this is a rather broad descriptor and the underlying mechanisms leading to loosening are not fully understood.

Recent in vivo studies using instrumented total knee replacements have shown that the loading magnitude applied across the knee joint can vary considerably depending on activity (Bergmann et al., 2014; D'Lima et al., 2012, 2011). In addition, infrequent high demand loading situations such as stumbling (Bergmann et al., 2004) could potentially damage the fixation of TKRs and this might lead to eventual clinical loosening. However, it is not known whether the load magnitudes associated with the more aggressive activities of daily living or stumbling would damage the TKR construct. Mechanical overload could cause the supporting bone beneath the cemented tibial tray to mechanically yield resulting in subsidence of the implant. Excessive loads could also induce large relative motions between the implant and bone (micromotion) resulting in loss of mechanical fixation. The overall goal of this study is to assess the ability of functioning knee replacements to support high mechanical loads without damage to the implant-bone interface or supporting bone bed. Testing of high load scenarios is not possible in patients because of the obvious risk of causing damage to the TKR construct. However, *en bloc* post-mortem retrieved knee replacements, that have functioned for the life of the donor, could be used to directly assess the effects of mechanical overload for the scenarios of supporting bone damage and increased induced micromotion.

Clinically, the risk of aseptic loosening is known to be higher for younger TKR patients and revision risk increases for implants with

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longer time in service (Julin et al., 2010). Younger patients are likely to be more active and may generate larger loads on their knee replacement, while implants with long time in service may be associated with a reduction in the amount of bone supporting the knee replacement due to stress shielding (Huiskes et al., 1987; Levitz et al., 1995; Li and Nilsson, 2000) and loss of mechanical interlock between the implant and bone (Goodheart et al., 2014; Miller et al., 2014). It is possible that the overload damage scenarios described above may also depend on donor factors such as age at implantation, time in service, and amount of peri-implant bone. Using the tibial components from postmortem retrieved TKRs, we asked two primary research questions: 1) do high (but clinically relevant) mechanical loads result in damage to the implant-bone interface or supporting trabecular bone? and 2) is there greater micromotion and bone strain for donors with younger age at implantation, longer time in service, or lower peri-implant bone mineral density in cases in which there is mechanical overload?

2. Methods

2.1. Total knee replacements following in vivo service

Seventeen human knees with cemented total knee replacements (TKRs) were obtained from eleven donors (6 were bilateral pairs) from the SUNY Upstate Anatomical Gift Program. The components were retrieved *en bloc* and were frozen at −20 °C until time of testing. Constructs were radiographed in the anterior-posterior and medial-lateral planes, and radiolucencies along the implant-bone interface were scored as: well fixed (no radiolucencies), possibly loose (peripheral radiolucencies), and loose (extensive radiolunciesies). Peri-implant bone mineral density (BMD) was obtained using clinical dual-energy x-ray absorptiometry (DXA) (Lunar DPX-IQ, GE Healthcare, Waukesha, WI). An additional 'time 0' construct was created in the lab using a fresh cadaver knee using conditions that simulated the clinical operative environment to represent a case with no bone remodeling. Summary statistics for donor information are shown in Table 1. All tibial components (n = 17 postmortem retrieval + 1 'time 0') were metal backed and were cemented in place. Fourteen of the eighteen implants had stems or keels, while four did not. Details for each donor TKR are included in Supplemental Table 1. The polyethylene insert wear was documented using the scoring method described by Hood et al. (1983).

2.2. Mechanical loading procedure

The tibiae were cemented in a pot 70 mm below the tibial tray using PMMA and fixed to a mechanical test frame (MTS Systems, Eden Prairie, MN), such that the tray was orthogonal to the loading axis. Axial compressive loads were applied to the center of wear patches on the articulating polyethylene insert using distributed loading pads (20 by 24 mm) with a 60% medial/40% lateral load distribution (Fig. 1A). Axial compression was chosen as this represents the primary loading direction acting on the tibia as determined from instrumented knee replacement studies (Kutzner et al., 2010). The loading regime consisted of axial compressive loads applied in displacement control (5 mm/min, equivalent to ~0.75 BW/s) to load limits from 1 to 10 body weight (BW) in 1BW increments. Once the load limit was reached, the test was reversed until the tibia was unloaded (Fig. 1B). Testing was stopped if there was frank compressive failure of the construct. Loading was limited to a maximum of 10BW, because this was considered to be an upper limit of what might be anticipated during any ambulatory activities including stumbling, but excludes impact loading due to falls or automobile accidents.

2.3. Digital image correlation (DIC) strain and micromotion measurements

A pair of CCD cameras (Diagnostic Instruments, Sterling Heights, MI) with telecentric lenses (55 mm, Edmund Optics, Barrington, NJ) was used to capture images (0.053 mm/pixel) of the posterior and medial surface of proximal tibia during loading. Digital image correlation (DIC) software (MI-2D, MatchID, Merelbeke, Belgium) was used to quantify horizontal (u) and vertical (v) displacement components (Fig. 1A) for sampling zones on the tray and bone surface. A normalized sum of squares difference (NSSD) with subregions of 35 pixels and step size of 17 pixels was used for the DIC analysis. The axial (Δv) component, shear (Δu) component, and the vector sum ($|\Delta u+\Delta v|$) of micromotion across the tray-bone interface (gage length, $Y_2-Y_1=\sim 5$ mm) were calculated. Axial (eg. $\Delta v/(Y_3-Y_2)$) and shear strain (eg. $\Delta u/(Y_3-Y_2)+\Delta v/(X_3-X_1)$) measurements were made using a 'virtual extensometer' approach (gage length, $Y_3-Y_2=\sim 24$ mm). At the peak load for each loading step, the *total* tray-bone micromotion/bone strain was measured relative to the initial unloaded state (Fig. 1B). After unloading, the *permanent*

Table 1
Descriptive statistics for TKR donor information, total (with load applied) and permanent (after load removed) tray/bone micro-motion and bone strain measures at 3 and 6 times body weight (BW) ($n=18$). Sign convention with tensile (+) and compressive (−) axial strain, opening (+) and closing (−) axial micromotion was used. Shear micromotion and strain reported as positive values. Data represent micromotion with largest magnitude and minimum (most compressive) bone strain for each donor.

	Applied Load (BW)	*Mean*	*Median*	*Std Dev*	*Min*	*Max*
Age (years)		78.9	85	11.2	54	90
Age at Implantation (years)		69.0	72.5	10.8	42	82
Time in Service (years)		9.9	10	5.3	0	22
Mass (kg)		83.3	85	14.3	53	100
BMI (kg/m^2)		29.0	29	3.92	22.7	36.5
Peri-implant bone mineral density (gr/cm^2)		0.63	0.54	0.27	0.07	1.16
Total tray-bone micromotion [axial] (μm)	3	−53	−35	51	−175	−6
	6	−67	−48	57	−176	−12
Permanent tray-bone micromotion [axial] (μm)	3	12	9	12	2	46
	6	20	13	19	1	81
Total tray-bone micromotion [shear] (μm)	3	16	15	10	6	42
	6	25	17	27	6	142
Permanent tray-bone micromotion [shear] (μm)	3	12	8	12	1	38
	6	16	10	15	1	63
Total tray-bone micromotion [axial +shear] (μm)	3	57	39	50	9	176
	6	77	54	62	14	215
Permanent tray-bone micromotion [axial+shear] (μm)	3	17	13	16	2	59
	6	28	19	24	4	102
Total proximal bone strain [axial] (με)	3	−2030	−2000	860	−4150	−850
	6	−4400	−3840	2390	−10,900	−2000
Permanent proximal bone strain [axial] (με)	3	−610	−470	570	−2670	−140
	6	−1110	−830	930	−3360	−150
Total proximal bone strain [shear] (με)	3	419	290	350	39	1183
	6	507	531	375	11	1614
Permanent proximal bone strain [shear] (με)	3	199	135	174	31	590
	6	296	255	229	14	1016
Total distal bone strain [axial] (με)	3	−1550	−1230	940	−4440	−670
	6	−3050	−2550	1790	−8430	−1560
Permanent distal bone strain [axial] (με)	3	−200	−140	240	−730	140
	6	−470	−360	280	−1010	−110

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