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VALIDATION OF FE MICROMOTIONS AND STRAINS AROUND A PRESS-FIT CUP: INTRODUCING A NEW MICROMOTION MEASURING TECHNIQUE

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Abstract

Finite element analysis provides a useful tool with which to analyse the potential performance of implantations in a variety of surgical, patient and design scenarios. To enable the use of finite element analysis in the investigation of such implants, models must be experimentally validated. Validation of a pelvic model with an implanted press-fit cup in terms of micromotion and strain is presented here. A new method of micromotion has been introduced to better describe the overall movement of the cup within the pelvis. The method uses a digitizing arm to monitor the relative movement between markers on the cup and the surrounding acetabulum. Finite element analysis was used to replicate an experimental set up using a synthetic hemi-pelvis with a press-fitted all-metal cup was subject to the maximum loading observed during normal walking. The work presented here has confirmed the ability of finite element models to accurately describe the mechanical performance of the press-fitted acetabulum and surrounding bone under typical loading conditions in terms of micromotion and strain distribution, but has demonstrated limitations in its ability to predict numerical micromotion values. A promising digitizing technique for measuring acetabular micromotions has also been introduced.

Keywords

FINITE ELEMENT ANALYSIS; ORTHOPAEDIC SURGERY; PELVIS; CEMENTLESS ACETABULAR CUP; MODELLING; HIP REPLACEMENT

3. Introduction

Cementless, rather than cemented, implants are being implanted with increasing frequency in total hip replacements (THRs)¹. With full bone ingrowth they can provide a more stable connection to the underlying bone and accommodate larger head sizes than cemented implants. The long term success of a cementless implant is widely accepted to be heavily reliant on initial stability^{2,3}; thus initial post-implantation micromotions are considered an early indicator of long-term THR success⁴. This has motivated a number of studies on the primary stability of cementless implants e.g.^{5,6,7,8}. Finite element (FE) analysis offers the researcher a tool with which they can isolate the effects of a variety of parameters on chosen outcomes and carry out large-scale multi-factorial studies which are not feasible in *in vitro* cadaveric or *in vivo* work. There have been a range of studies developing and experimentally validating FE models able to predict the micromotions of implanted cementless femoral

stems e.g.^{9,10,11}. The UK National Joint Registry reports five year revision rates of 6.0% and 2.9% for all-metal and metal-backed cementless cups respectively, compared to 1.8% for cemented cups¹. With significantly low comparative performance, metal on metal hip replacements have become a controversial bearing type. FE analysis of cup stability can provide one aspect of investigation into the poor performance of these bearing types, alongside a range of other research methodologies.

A range of studies assessing acetabular implant micromotions have been carried out experimentally^{12,4,13} and using FE modelling¹⁴⁻¹⁶. In most reported experimental studies the acetabular cup micromotion was recorded with uni-directional linear variable differential transformers (LVDTs) at three sites; this is not enough to provide information on the overall movement of the cup. Kluess et al.¹⁵ attempted to validate their FE predicted micromotions using optical markers but could not attain acceptable measurement accuracy. Zivkovic et al.¹⁷ validated their FE micromotion predictions using six LVDTs, under a chair-rising loading, and with a significant proportion of the ilium fully constrained. The work presented here details an alternative digitization based method of micromotion measurement which does not require attachment of measuring devices to the given specimen, and from which micromotion in three orthogonal directions at a large number of sites can be measured. Micromotion is measured under a normal walking load case, without fully constraining the ilium.

FE validation of pelvic strains, rather than micromotions, is more commonly reported in the literature e.g.^{18,15,19}; though validation of strains on a pelvis implanted with a press-fitted cup could not be sourced. The mechanical situation and associated strains for a press-fitted cup are substantially different to those for a cemented or in-line fit cup. It is paramount to validate any FE model concerning press-fit cups before it is used to analyse implant behaviour.

The aim of this study is to test whether FE models can accurately predict cup-bone micromotions and the strain around an acetabulum implanted with a press-fit cup, and whether a new technique for micromotion measurement can accurately measure cup micromotions. These points will be explored through FE modelling and experimental measurements taken from a synthetic hemipelvis subject to joint loading representative of the maximum load occurring during normal walking²⁰. A synthetic hemipelvis was used rather than a cadaveric specimen to isolate the errors associated with press-fit implantation from those associated with estimation of cadaveric bone stiffnesses. Previous work has demonstrated errors of up to 30% between FE predicted and experimentally measured principal strains on cadaveric specimens¹⁹. A large amount of this error is likely to stem from the necessary approximations of material stiffness and directionality of the cadaveric bone. The homogeneity of the materials comprising the synthetic pelvis allow for a more certain description.

4. Materials and methods

4.1 Experimental set-up

A synthetic biomechanical hemi-pelvis (Large Left Fourth Generation Composite Pelvis, Item 3405, Sawbones, Pacific Research Laboratories Inc, Sweden) was CT-scanned at 0.75mm intervals with a resolution of 512x512 pixels (Sensation 16, Siemens Plc, Munich, Germany). The resulting CT-scan was semi-automatically segmented (MIMICS, v12.11, Materialise NV, Leuven, Belgium) and a tetrahedral mesh produced. The synthetic hemi-pelvis comprises a short-fibre-filled epoxy cortical shell and a polyurethane foam trabecular inner.

For spatial fixation in all degrees of freedom, a steel block was attached to the ilium, at the sacroiliac joint, through four steel pins which penetrated through the medial cortical shell and into the trabecular foam, but not through to the lateral side of the ilium (Fig. 1D). Bone cement (Kemdent, Associated Dental Products Ltd, Swindon, UK) was used to fill in the gaps between the flat steel block and the contoured cortical surface. Two alignment screws were inserted to aid positioning of the block.

A 62mm ADEPT[®] porous HA-coated cobalt-chrome hemispherical monoblock acetabular cup with a 4mm rim thickness (Finsbury Orthopaedics Ltd, Leatherhead, UK) was chosen for implantation after visual templating from the pelvic mesh. A series of reamers, up to 61mm, were used to ream the acetabulum of the hemi-pelvis; the trial cup confirmed adequate stability. A digitizing arm (FARO Gage, FARO GmbH, Munich, Germany) with a needle-point probe was used to spatially measure the location and size of the reamed acetabulum within the hemi-pelvis. The cup was orientated at 45° anatomical inclination and 15° anatomical anteversion²¹. A sphere was fitted to the digitized acetabular points using a least-squares sphere-fitting algorithm (MATLAB, vR2010a, The Mathworks Inc, MA, USA), the rms error between best-fit sphere and points was 0.14mm. The CT-mesh was updated to include the size and location of the reamed cavity by referencing the physical bone to the CT-mesh (MATLAB) and performing a Boolean operation on the mesh to subtract the fitted sphere (RHINO, v4.0, Robert McNeel & Associates, WA, USA). The cup was impacted into the acetabulum. The digitizing arm was used to locate the position of the cup within the reamed cavity (MATLAB), and superposition onto the reamed mesh ensured full seating had been achieved (RHINO).

Four 45°/90° three-element rosette strain gauges (GFRA-3-50, TML, Tokyo Sokki Kenkyujo Ltd, Tokyo, Japan) were attached to the cortical surface. The gauges were positioned around the acetabulum (Fig. 1), in areas of predicted high strain. An extra control gauge (gauge 5 in Fig. 1) was positioned in an area of predicted low strain. Strain data and load data were collected every second throughout the loading regime.

The test set-up is shown in Fig. 2B. The steel block was clamped to the bed of the test machine (Instron 5866 Universal Test Instrument, Instron Co, MA, USA), orientated to allow direct loading into the acetabular cup to produce a resultant force of 2kN in the same direction as the maximum force occurring during normal walking²⁰. Loading was applied through a 54mm modular femoral head which was vertically loaded through a compression piece. The compression piece interfaced with the modular head through a tapered block which was polished and PTFE coated on its upper side to allow free sliding in a horizontal plane; thus only vertical forces were transferred into the cup (Fig. 2A). Ten cycles of 2.5kN loading were applied to ensure settlement of the apparatus, before loading to 2kN. Measurements were taken for FE comparison at 250N and 2kN, after the load was maintained for twenty minutes to account for the viscoelastic creep of the synthetic pelvis²². The strain reported was therefore the difference in strain between the two loading points of 250N and 2kN. The test protocol was repeated three times, with complete dismantlement of the apparatus between each test.

4.2 Measurement of micromotion

Micromotion was measured using the digitizing arm (Fig. 2C) to spatially locate points on the pelvis and cup before and after loading; this enabled the calculation of relative implant-bone movement under loading at each of the points. Total cup rotation was also calculated by fitting a plane through each marker point and calculating the rotation of the plane normal of the cup relative to the bone. To allow repeatable spatial location of these points, the female part of a plastic snap fastener was applied to the chosen surface with Cyanoacrylate. A 3mm ceramic ball probe was used to locate the internal ring of the snap fastener. Twelve

snap fastener markers were adhered around the implanted cup rim. Twelve further markers were attached to the cortical surface, located as close to the rim markers as possible. The marker locations are shown in Fig. 3. Micromotion was reported as the relative movement between these twelve marker pairs. This method was chosen after investigation and verification of a number of different location methods. The mean error in repeatability of measurement *in situ* was 6µm, although errors as high as 30µm were observed (99% confidence) at marker 1. This is higher than the manufacturer quoted accuracy of 5µm for the robot because of the cramped situation on the test bed which necessitated the close proximity of the digitising arm base and the test piece (ideal measurement is taken perpendicular to the surface of the test sample and requires more space). All markers were digitized three times for each measurement, and the mean of these three measurements taken as the reading. Four markers were also attached to the steel fixture block, at the ends of the four pins inserted into the ilium, to monitor possible movement of the sacroiliac fixture.

4.3 Finite element modelling

An FE model was created and solved in ABAQUS (v6.8, Dassault Systemes Inc, Velizy-Villacoublay, France) to simulate the experimental set up. The reamed CT-mesh was orientated to match the position of the physical model, using spatial positioning data from the digitizing arm. The mesh was re-meshed with approximately 120,000 four-node linear tetrahedral elements. A three-node triangular shell mesh was extracted from the tetrahedral mesh surface to represent the cortical bone. The cup geometry was provided by Finsbury Orthopaedics and the 3D model was created in ABAQUS and meshed with approximately 45,000 ten-node tetrahedral elements. All mesh densities were verified with convergence studies. The porous HA-coated surface was not included in the model; the cup was modelled with a smooth surface and the roughened coating accounted for with a frictional coefficient. The modular femoral head was modelled as a sphere with approximately 61,000 ten-node tetrahedral elements. The properties of cobalt chrome ($E=210\text{GPa}$, $\nu=0.3$ ²³) were assigned to the cup and head. Three point bending tests were conducted on test samples of epoxy and polyurethane, obtained from Sawbones, to ascertain material Young's moduli, as well as on a test piece cut from an identical biomechanical hemi-pelvis from the same batch as that used for the experiment. The measured Young's moduli were 10GPa and 155MPa for epoxy resin and polyurethane; these were respectively assigned to the outer cortical shell mesh and the trabecular tetrahedral mesh. The measured polyurethane Young's modulus agreed with the values published by Sawbones, but the epoxy Young's modulus was less than that published²⁴. The temperature-dependency of epoxy Young's modulus has been reported previously for Sawbone's femurs²⁵; the epoxy is also short-fibre-filled, adding to potential variability. Both materials were assigned a Poisson's ratio of 0.3²⁶. An algorithm was developed to assign a varying cortical bone thickness on the shell mesh, as is commonly implemented in such models, e.g.^{18,19}. The algorithmic inputs were two meshes (produced from MIMICS) of the outer bone surface and the inner trabecular volume. The algorithm calculated thickness as the perpendicular distance between each outer mesh element and the inner trabecular volume. In areas of high curvature (e.g. acetabular rim) special consideration was required to derive the thickness; the approach used has been reported in work by Anderson et al.¹⁸. Free sliding was permitted at the cup-head interface, and penalty sliding was implemented at the cup-pelvis interface, with a friction coefficient of 0.5, representing that measured experimentally between porous coated metal and trabecular bone²⁷. As the reported friction coefficient was not derived from experiments involving synthetic bones, a sensitivity analysis was carried out which showed that varying the coefficient of friction by ± 0.1 altered output micromotions and strains by 5% and 2% respectively.

The FE simulation was implemented through three steps: cup implantation; loading to 250N and loading to 2kN. Cup implantation was simulated by introducing the cup into the acetabulum from a 50mm distance perpendicular to the cup rim. The cup was displacement driven into its pre-loading position^{28,17}, forcing the surrounding acetabulum to circumferentially expand to accommodate the cup, thus introducing the pre-strain required for the press-fit; a perfect fit was not enforced between the implant and acetabulum. Slight movements at the sacroiliac joint necessitated a displacement driven, rather than load driven, simulation; the model was therefore not fixed in space at any specific coordinate. The two loading stages were simulated with displacement criteria at the femoral head and at the four steel pins in the sacroiliac fixture (Fig. 1D). The required input displacements were measured *in situ* with the loading machine (in the case of the femoral head) and the digitizing arm (in the case of the sacroiliac pins); the mean values from the three experimental runs were used. The resulting load measured at the femoral head in the FE simulation was within 1% of that applied in the experiment. It is most common to control an FE model with load criteria, rather than displacement; two further FE simulations were therefore undertaken to test whether the displacement driven model responds in the same manner as the load-driven model. The comparative models were identical to the test set-up, but to enable comparison, both had rigid fixation at the sacroiliac joint.

5. Results

Cup-bone micromotions and maximum and minimum principal strains occurring between loads of 250N and 2kN at the implanted acetabulum were predicted by the FE model and compared to those measured *in situ*. A magnitudinal comparison between experimentally measured and FE predicted micromotions is provided in Table 1, with the match-up of all directional components shown in Fig. 4. The match-up of cup rotation is shown in Table 2.

The mean and peak percentage errors in predictive micromotion were 27% and 110% respectively. The highest percentage error occurred at marker 1 (Fig 3), all remaining errors were below 36%. The gradient of the regression line in Fig. 4 is steeper than 1, at 1.21 and the y intercept is $-6.4\mu\text{m}$, with an R^2 value of 0.90. This indicates that the FE model underpredicts micromotion, as can be observed from Fig. 4. Similarly the FE model underpredicts whole cup rotation, by 14%; this error is reduced to 4% when the 95% confidence value is reported. The intraclass correlation coefficient between methods was high at 0.92. The limits of agreement on the Bland-Altman plot (Fig. 5) are 81 and $-86\mu\text{m}$, indicating the level of magnitudinal error in the FE model micromotion predictions. The directions of cup-bone micromotion measured in the experiment and predicted in the FE model are shown in Fig. 6.

The FE predicted distribution of maximum principal microstrain and associated principal strain directions at the gauge sites are shown in Fig. 7. The control strain gauge (gauge five in Fig. 1) reported cup strains in the range 29-44 microstrain in an area of known low strain.

Comparisons of experimentally measured principal microstrains and those predicted are listed in Table 3 and graphically compared in Fig. 8. The mean percentage error in microstrain was 151%. Percentage error increased significantly at microstrains experimentally measured below 50 microstrain. The mean and peak percentage error in principal microstrain prediction above 50 microstrain was 27% and 67% respectively. The gradient and y intercept of the regression line are 0.94 and 28 microstrain with an R^2 value of 0.85. Unlike with the cup-bone micromotion, the FE model does not consistently under or overpredict microstrain. The intraclass correlation for estimating the microstrain between the two methods is 0.91. The Bland-Altman plot shown in Fig. 9 has limits of agreement of 199 and -88 microstrain.

The mean differences in principal microstrain and micromotion measurements between the load driven and displacement driven comparative models were 2.3 microstrain and 1.04 μm respectively; these represent 0.65% and 2.0% of the mean overall measurements.

6. Discussion

The primary aim of the work presented here was to verify the accuracy with which FE models can predict the cup-bone micromotion and the strain around an acetabulum implanted with a cementless cup. This was investigated by comparing micromotion and strain data measured experimentally with that predicted using FE analysis. The presented study is limited by a number of factors. The focus of this study is the acetabulum, not the pelvis, and as such the experimental and computational models include a single hemi-pelvis fixed at the sacroiliac joint, rather than a complete pelvis. Future modelling based on this validation experiment will be limited to the acetabular region, as the full bone was not physiologically represented in this experiment. The computational model was subject to further limitations by the simplification of the cup backside surface coating as a smooth surface with substantial friction, and the idealisation of the reamed cavity as spherical. Together with the assumption of free sliding at the bearing interface, these limitations may have impacted on the accuracy of model prediction. A further necessary limitation of this work is the assumption of material properties as linear elastic and isotropic; both of these simplifications may influence the match-up of results, though the repeatability of the experimental measurements demonstrate a lack of material yielding during the loading phase. Had cadaveric bone been used in this study, the consequences of this assumption are likely to be more dominant, though it should be noted that good strain and micromotion FE predictions have been achieved in published cadaveric-based studies which also assume linear elastic isotropic properties^{18,19}.

A secondary aim of this work was to test the suitability of a new micromotion measuring technique. The error bars plotted on Fig. 4 and Fig. 8 communicate the variability of measurements over the three experiment runs. This variability is attributable to both the measurement technique and the different environmental conditions. The low variation in measured data, along with a mean error in micromotion measurement repeatability of 6 μm verifies the accuracy of both measurement techniques. The use of a digitizing arm allowed accurate measurement of micromotion in three directions at twelve points, totalling 36 measurements. This technique is a viable alternative to LVDTs. Assuming rigid body motion, six LVDTs are required to describe cup-bone micromotion; arrangement of such a number is restrictive to the experimental set up.

Strong linear regression correlations ($R^2=0.90$) and a high intraclass correlation coefficient (0.92) were found between experimentally measured and predicted cup-bone micromotions. The plots in Fig. 4 show the correlation between all directional micromotion components confirming that the FE micromotion predictions captured the anterior shift of the cup within the pelvis as well as the scooping and turning mechanisms which were observed in the experiment (Fig. 6). The mean percentage micromotion error was 20% (95% confidence interval) and the error in prediction of whole cup rotation angle was 4% (95% confidence interval). The higher accuracy of cup rotation prediction suggests that the main source of error may be the axis of cup rotation. Fig. 8 shows that the FE model tends to under-predict micromotion. The cup-bone interface description has a dominant effect on the micromotions at this face. The uncertainties associated with the consistency of cup diameter considering the porous beaded surface, and the potential nonsphericity of the reamed cavity may have contributed to the under-estimation. When using FE to predict potential failure mechanisms of implanted cups it is paramount that the mechanism of movement is captured as well as the general magnitude to ensure full confidence can be placed in results. The only other

reported validation of FE predicted micromotion on a cementless acetabular cup compared their data in terms of the angle of cup rotation¹⁷ and managed to predict the rotation of the cup within 5% of the measured value under a chair rising load. FE predicted cup rotation for the current study was also within 5% (95% confidence interval) of the experimental value, though observed rotations were significantly lower than those measured in the previous study¹⁷. Reggiani et al.¹⁰ compared a single tangential micromotion measure of an implanted femoral head under a torque load, achieving an error of 12% between predicted and experimentally measured micromotion. A quarter of the predicted micromotions in this study were within this accuracy, but the remaining predictions exceeded it. There was, however, no directional component to measure in the femoral stem study, where torque loading was applied and measured directly perpendicular to the implanted stem. The acetabular cup in the present study was free to slide under the spherical head loading device and therefore was not constrained to move in any known direction. The results observed in this study demonstrate good correlations and accurate directional micromotion predictions and cup rotation, but less accurate magnitude estimation. This supports the use of FE to assess relative differences between acetabular cup modelling scenarios, but the authors would advise caution in using FE to predict explicit numerical micromotion values, such as the 150µm failure criteria posed by Pilliar et al.³ in the case of a press-fit implant.

Relatively strong correlations ($R^2=0.85$) and a high intraclass correlation coefficient (0.91) were observed between predicted microstrains and those measured experimentally. The strain directions were well predicted at all gauges. Very high percentage errors were found between all measured strains below 50 microstrain and those predicted. Such high errors have not been observed in other reported pelvic validation studies^{18,15,19}. The strain reported from the experiment is the difference in strain between two loading points: 250N and 2kN, not absolute strain. Before loading commences the cup is impacted into the acetabulum, which is therefore pre-strained. Cadaveric specimens have shown press-fitted cups causing an acetabular pre-strain of up to 700 microstrain²⁹. The FE analysis predicted impaction pre-strains in the region of 2000 microstrain. The absolute strains being measured are thus a lot higher than the reported values in Table 3. Percentage error is therefore an unsuitable metric to use in this circumstance. Predicted strain magnitude is highly dependent on the specific interference level and cup seating, which are in turn dictated by the diametrical difference between the reamed acetabular cavity and the porous-beaded cup surface. The support provided to the cup by the reamed acetabulum is likely to vary around the cup, leading to a varying distribution of pre-strain. This is demonstrated by the high variation in surrounding acetabular strains shown in Fig. 8. The predicted value in strain measurement is thus highly dependent on location and a slight difference in pelvic orientation and cup movement may lead to a large difference in strain prediction. The strength of the correlation is comparable to published validated pelvis models without press-fit acetabular cups^{18,15,19}, and principal component direction is well predicted. This again demonstrates the ability of FE analysis to accurately predict the strain distribution and direction around a press-fitted acetabular cup, but less well the magnitudinal values of strain.

The discrepancy between predicted and measured micromotions and microstrains may have arisen from a number of sources. The close agreement between the FE resolved joint loading force and that experimentally applied (within 1%) suggests that the model was well defined; together with the low error in cup rotation prediction this suggests that the source of error may have been the precise direction of cup translation and axis of rotation within the bone. The definition of the implant-bone interface was thus very important. The implant-bone friction coefficient was estimated, and both micromotions and microstrains have been shown to be sensitive to the choice of coefficient. Fixation at the sacroiliac joint provided a more physiological joint description than is provided by the common fixation of a large section of the ilium e.g.^{18,17}; this is more difficult to accurately describe with FE modelling, despite

use of the digitizing arm. The majority of discrepancies are likely to be attributable to the measured level of press-fit; this has been shown to have a significant effect on cup-bone micromotions¹⁷, but is difficult to accurately quantify. Effort was made to accurately describe the press-fit between the cup and acetabulum in this study; errors may have arisen from the assumption of the reamed surface as spherical³⁰, the assumption that the beaded cup surface was consistent and the final placement of the cup. The cup was assumed to be fully seated based on superposition of the cup onto the unreamed CT-mesh. Errors in the description of the reamed surface or the level of cup seating would impact significantly on cup stability and pelvic pre-strain. This study was also limited by the use of a synthetic, rather than cadaveric, specimen. Based on previous work^{18,15,19}, the authors do not have any reason to believe that the techniques and conclusions here would not be applicable to subject-specific FE models.

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Figure 1.
Positions of delta rosette strain gauges on synthetic pelvis.

Figure 2.

A. Close up of loading configuration – a tapered block is fitted into the modular head, the topside of which is free to slide against the compression piece. B. The whole experimental test set up, with digitizing arm clamped to the base. C. The digitizing arm. D. The sacroiliac fixture – four pins (highlighted) are embedded in the synthetic pelvis at the sacroiliac joint and held rigidly into a steel block.

Figure 3.
Positions of snap fastener markers around the rim of the implanted cup.

Figure 4.

Comparison of all experimentally measured (y axis) and FE predicted (x axis) directional micromotion components. Best fit line and R^2 value labelled on plot. Datapoints are marked with a point, and vertical error bars are added to communicate the range of measured experimental values.

Figure 5. Bland-Altman plot comparing FE micromotion prediction with experimentally measured micromotion, with limits of agreement represented by dashed lines.

Figure 6. Direction of micromotion measured experimentally (above) and FE predicted (below) at twelve marker points around the cup rim (arrows are not to scale). The crosses represent the location of the four sacroiliac pins, and the large arrow the direction of loading.

Figure 7. FE predicted distribution of maximum principal microstrain (top). Direction of maximum (red) and minimum (blue) principal microstrain measured experimentally (middle) and FE predicted (bottom).

Figure 8.

Comparison of all experimentally measured (y axis) and FE predicted (x axis) directional components of maximum and minimum principal microstrains. Best fit line and R^2 value labelled on plot. Datapoints are marked with a point, and vertical error bars are added to communicate the range of measured experimental values.

Figure 9. Bland-Altman plot comparing FE microstrain prediction with experimentally measured microstrain, with limits of agreement represented by dashed lines.

Table 1

Comparison of experimentally measured and FE predicted micromotion magnitudes at 12 marker points around the cup rim

Marker number	Experimentally measured micromotion magnitude (μ m)	FE predicted micromotion magnitude (μ m)	Error (μ m)	Error (%)
1	69.1 (range 66.3–71.3)	141	77	110
2	129 (range 128–130)	147	19	15
3	221 (range 205–242)	205	19	7.6
4	295 (range 291–299)	222	72	25
5	197 (range 195–203)	179	19	9.6
6	231 (range 228–234)	178	53	23
7	231 (range 227–238)	151	80	35
8	256 (range 247–264)	198	59	23
9	219 (range 216–224)	149	70	32
10	225 (range 220–230)	204	21	9.4
11	272 (range 262–279)	220	51	19
12	318 (range 316–320)	249	70	22
Mean error (95% confidence interval)			51 (48)	27 (20)

Marker point locations are shown in Fig. 3

Table 2

Comparison of experimentally measured and FE predicted cup rotation, radians

Experimentally measured cup rotation (radians)	FE predicted cup rotation (radians)	Error (radians)	Error (%)
0.0051	0.0044	0.0007	14
Error (95% confidence interval) *		0.0001	4

* The 95% confidence interval value was derived by using 95% of the marker points to calculate the cup normal

Table 3

Comparison of experimentally measured and FE predicted principal microstrain at four gauge sites

Gauge number	Experimentally measured principal microstrain	Maximum/minimum	FE predicted principal microstrain	Error (microstrain)	Error (%)
1	386 (range 369–393)	Maximum	389	3.2	0.82
2	590 (range 571–639)	Maximum	388	200	34
3	29.8 (range 17.3–42.1)	Maximum	112	82	280
4	215 (range 209–220)	Maximum	359	140	67
1	–317 (range –295 to –332)	Minimum	–223	94	30
2	–413 (range –410 to –418)	Minimum	–421	8	1.9
3	6.76 (range 3.98–11.9)	Minimum	15.3	8.5	130
4	37.4 (range 32.6–44.3)	Minimum	290	250	680
Mean error (measured strains below 50 microstrain removed)				99 (90)	151 (27)

Gauge locations are shown in Fig. 1