1. Introduction

Loosening in cemented total hip replacement (THR) has been identified as one of the most significant failure mechanisms affecting long-term stability of THRs [1,2], particularly in acetabular cups, where the late loosening rate is reported to be three times that of femoral components [3]. Given the significance of the problem, it is somewhat surprising that few studies [4,5] have been reported on the fatigue behaviour of cup fixation, as opposed to fixation of femoral components [6-12]. One of the apparent reasons may be that the 3D geometry of an acetabulum and the loading conditions of the cup do not permit ready 2D representations, experimentally or numerically, as opposed to those of a femur [13].

Evidence from retrieval studies and in vitro experiments seems to support the hypothesis that cement fixation fails by fatigue [4-14]. For femoral components, the accumulated damage failure scenario was thought to be responsible for fatigue failure in cemented THRs [14]. Mechanical damage in the form of microcracks accumulates under cyclic loading conditions, cracks coalesce and form macrocracks, leading to disintegration of the cement mantle and eventually to gross loosening of the implant. For acetabular components, however, the situations are more complex. In simple multilayer models [4] built to reproduce the cement stress distributions as those in the cement mantle of a plane strain pelvic bone model, radial cracks in the cement mantle were observed in the vicinity of the peak tangential stress under constant amplitude cyclic loading that simulating the peak hip contact force during normal walking. However, in 3D acetabular reconstructs under similar loading conditions, fatigue failure occurred at the bone–cement interface [5], while the cement mantle remained intact. Although the latter work featured 3D acetabular reconstructs, a step closer towards in vivo condition, simplified sinusoidal loading waveforms and fixed loading direction were used to simulate the peak hip contact force. These apparently different failure mechanisms warrant further studies, as the prevailing mode of failure in acetabular replacements must be established, if effective methods are to be found towards improving the longevity of the fixation.

In this work, an in vitro study has been carried out using a new hip simulator specially developed for the purpose of fatigue testing of cement fixation. The machine is capable of simulating the resultant hip contact force as a three-dimensional force vector during typical physiological loading conditions, including normal walking and stair climbing. A 3D finite element analysis has also been carried out to evaluate the accuracy of the simulator in reproducing the hip contact force as measured in vivo, as well as to obtain the stress distributions in the cement mantle and at the bone–cement interface. Preliminary experimental results from the hip simulator testing are presented and compared with the experimental results from conventional fatigue tests [4,5]. The results are discussed, utilising the FE results from 2D and 3D pelvic bone models, with regard to stress state, stress magnitude and associated failure modes.

4. Discussion

5. Conclusions

References

2. Experimental details

2.1. Development of the hip simulator

The resultant hip contact force, as measured in vivo using instrumented implants with telemetric data transmission [15,16], was reproduced in the hip simulator. The three-dimensional time-dependent force vector in a cup coordinate system was achieved by synchronising the magnitude of the force with two rotational displacements associated with cup and femoral head. The ranges of the motions allowable are ±35°, ±360° and ±60° for rotations on YZ, XZ and XY planes, respectively, although only two motions are required to simulate the resultant hip contact force, namely, rotation about the Y axis on the XZ plane and a swing operation about the Z axis on the XY plane. The ranges required for normal walking (XZ: 88°; XY: 35°), ascending stairs (XZ: 141°; XY:29°) and descending stairs (XZ: 193°; XY: 44°) are hence well within the capacity of the apparatus. The machine has four stations which are individually controlled to allow selected load profiles to operate independently and simultaneously.

The test cell was designed to hold a hemi-pelvic bone with an implanted acetabular cup. The cup was in articulation with a spherical femoral head through which the load was transmitted. A hemi-pelvic bone specimen was mounted upside down in order to facilitate lubrication during testing. The constraints of the specimen were achieved at iliac tuberosity via moulds of epoxy putty to apply uniform pressure. A partial constraint was also applied to the pubic joint to maintain the symmetry condition of the pelvis but allow movements on the sagittal plane. One of the test cells, including a specimen held in the loading device, is shown in Fig. 1. A manual smoothing procedure was applied to reduce the local irregularities of an average patient data [15], and to bridge the gaps at the beginning and the end of a cycle to enable continuous cyclic loading. Full details of the hip simulator design were reported elsewhere [17].

Fig. 1

The Portsmouth four-station hip simulator with a specimen loaded in one of the test cells.

Fig. 2 shows the comparison of the hip contact force as a command signal and the response of the machine for normal walking, ascending and descending stairs. The hip simulator seems to reproduce the physiological loading profiles well, with a relative error of 3.6% for normal walking, 1.5% for ascending and 6.2% for descending stairs, at peak hip contact force. The motions during physiological loading conditions are also evaluated and the results are comparable to the load responses hence omitted here.

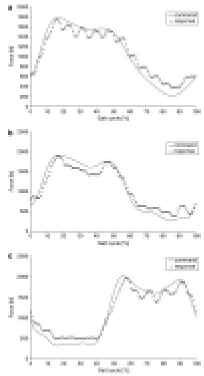


Fig. 2

Comparison of the hip contact forces as command signal (line) and response of the hip simulator (symbol) for (a) normal walking, (b) ascending stairs and (c) descending stairs.

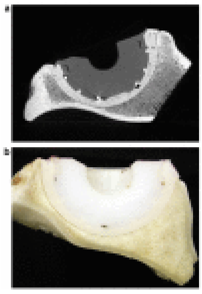
2.2. Experimental procedure

Composite sawbones, although adequate for constant amplitude fatigue testing [5], were found to be inadequate for hip simulator testing, as the bone failed prior to the failure of the cement fixation. Bovine bones were used instead in the experiments. The bones were retrieved from a local abattoir and processed to remove soft tissues and degreased. Sequential reaming of acetabulum was carried out to obtain the desired socket size with a congruent surface. Standard manual mixing technique was used to prepare the cement and to implant a Charnley cup with an average of 2–3 mm cement thickness. A purpose built device was used to mount the sample in the hip simulator, such that the cup coordinate system coincides with the coordinate system of the machine. The constraints were achieved at iliac tuberosity and the pubic joint. Lubrication was applied on the articulating surfaces prior to testing and topped up periodically during the testing to reduce wear, necessary to maintain accurate load transfer to the cement mantle. Tests were carried out under normal walking and descending stairs loading conditions at frequencies ~1 Hz, consistent with those occurred in vivo. Selected load levels were used, simulating a body weight of 75, 100 and 125 kg. Periodically, the samples were removed from the test rig to allow CT scanning [4] to be carried out for the purpose of monitoring the damage development in the cement fixation.

2.3. Preliminary experimental results

Preliminary results were obtained from three samples tested up to 17 million cycles. Damage was monitored using CT scanning at regular intervals, and the samples were sectioned and analysed using microscopy post testing. Debonding at the bone–cement interface was observed in all samples, irrespective of the loading conditions. Local debonding in the posterior–superior quadrant was revealed in one sample tested under normal walking for 15 million cycles; while multiple debondings were found at the bone–cement interface for another sample experienced 17 million cycles under normal walking loading condition. Extensive debonding at the bone–cement interface was identified by CT scanning (Fig. 3a) in a sample tested under the most severe condition, descending stairs, for merely 2.2 million cycles, where clear demarcation between the cement and the bone can be observed in the sectioned sample (Fig. 3b).

Fig. 3



Debonding at the bone–cement interface of a sample tested under descending stairs loading condition. (a) A CT scan image showing clear demarcation at the bone–cement interface; (b) almost complete separation of the bone–cement (more ...)

3. Finite element analysis

3.1. Methods

To evaluate the accuracy of the hip contact force reproduced by the hip simulator, and to study the stress distributions in the cement mantle and bone–cement interface in a sample tested in the hip simulator, three-dimensional finite element models were developed. Both sawbone full model and bovine truncated model were developed, although Fig. 4 shows only a truncated bovine bone model with constraints illustrated. A standard Charnley cup with outer/inner diameters 53/22 mm was inserted in a standard position with a cement layer of 3.5 mm. The femoral head was modelled as a rigid body. The cup and the cement layer were meshed with 8-noded linear brick and 6-noded linear wedge elements, with an element size of ~1 mm, and the total number of elements for cup and cement mantle were 2480 and 900, respectively. The bone structure was meshed with a total of 49878, 4-noded linear tetrahedron elements with the element sizes vary between 1 and 2.5 mm. Details of the material properties of the individual parts are given in Table 1.

Fig. 4

Finite element model of implanted bovine bone.

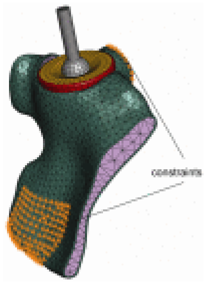


Table 1

A summary of the material properties used in the FE models

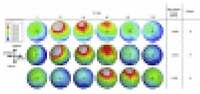
Material	Young's modulus (MPa)	Poisson's ratio	References
UHDPE	1000	0.4	Mak and Jin, 2002
PMMA	2000	0.33	Harper and Bonfield, 2000
Pelvis shell	10,000	0.3	Pacific Research Laboratories, Inc. US
Pelvis core	104	0.3	Pacific Research Laboratories, Inc. US
Bovine Trabecular	1000	0.3	Keaveny et al., 1993

Surface-to-surface discretisation was used for the contact analysis between the femoral head and the acetabular cup. To allow for contact pair motion, a finite sliding tracking approach was adopted. The cement surfaces were assumed to be fully bonded to the cup and the underlying bone structure using a tie contact formulation. The time-steps for the numerical simulations were chosen to be the same as those in the experiments. The initial increment for the step was 0.01 s. Simulations were assumed to be quasi-static, such that the inertial properties of the jigs and fixtures holding the specimen were not taken into account. Friction between the femoral head and the acetabular cup was ignored. The effect of constraint condition on the stress response was examined between that constrained in the hip simulator and in an anatomical condition, i.e. fully constrained at sacroiliac joint and symmetry conditions applied to pubic symphysis. Full composite bone geometry was also compared with truncated geometry of a bovine bone model. Convergence tests were carried out on a basic model that was constrained as in the hip simulator, and loaded statically at peak hip contact force, the selected mesh was then used for all the other models. Loading profiles of stair climbing as well as normal walking were simulated at six selected load cases (2%, 16%, 30%, 50%, 65% and 100% of a cycle).

3.2. Results

The stress distributions in the cement mantle were evaluated as they are of primary interest to the fixation of the acetabular replacements. The stress variation across the thickness of the cement mantle was found to be within 10%, hence only the stress distributions on the bone–cement interface are presented. In general, the von Mises stress distributions in all models at the selected load cases are found to be similar. The maximum Mises stress appears to occur consistently at ~16% of a gait cycle in the posterior–superior quadrant, with a normalised stress index, α $\left(= \frac{\sigma_{\max}^{\text{model}}}{\sigma_{\max}^{\text{A}}}\right.$, where $\sigma_{\max}^{\text{model}}$ and $\sigma_{\max}^{\text{A}}$ are the maximum von Mises stress in a given model and the basic model A, respectively) varies between 1 and 1.145 for models of different constraints (hip simulator vs. anatomical) and geometries (full hemi-pelvic bone vs. truncated). The influence of loading profile on the cement stress distribution is shown in Fig. 5. The stress distribution pattern in ascending stairs seems to be consistent with that in normal walking, although with a higher magnitude (α = 1.211) at a single-legged stance. The maximum Mises stress in descending stairs (model G) occurs at a later part of the loading cycle (~56%, as opposed to 18% in ascending stairs), with a higher still magnitude (α = 1.412). Clearly, descending stairs represents the worst case scenario amongst the three loading profiles examined here.

Fig. 5



Comparison of von Mises stresses in the cement mantle near the bone–cement interface under normal walking (E), ascending stairs (F) and descending stairs (G) loading conditions.

4. Discussion

4.1. Hip simulator as a preclinical simulation tool

Hip contact forces during normal walking, ascending and descending stairs have been realised and evaluated in a unique hip simulator for fatigue testing of cement fixation. This is the first time such an apparatus is purposely built for testing of bone cement as a fixation agent.

Conventionally, strength requirements for prostheses have been met by simplified mechanical testing, often simulating limiting locomotion patterns in loading [18]. The realisation of hip contact forces as measured in vivo is of significant advantages for endurance testing of implants and fixation, particularly as a tool for pre-clinical trials of novel implants and new fixation methods. Since the seminal publication of the database of hip contact forces by Bergmann [15], the utilisation of these data for endurance testing of the strength of fixation has yet to emerge. The present hip simulator has a greater operating envelope than most of the existing simulators, where stair climbing is simulated as well as normal walking. The successful trials of the new hip simulator open up realistic prospect in the application of combined loading cycles, such as those recommended in Bergmann et al. [19], for preclinical testing purposes.

The directional variation of the resultant hip contact force during physiological loading conditions may be important for fixation testing, as it defines the instantaneous primary load transfer region, which may in turn prompt micro damage in the cement or bone–cement interface. Indeed, the most severe loading condition, descending stairs, prompted almost complete failure at the bone–cement interface after only 2.2 million cycles, while similar failures, although not as extensive, were observed after over 15 million cycles under normal walking profile. Although the failure mechanism of bovine bone samples tested using the hip simulator is the same as that in saw-bone samples tested under conventional fatigue [5], i.e. debonding at the bone–cement interface, the number of cycles to failure is significantly greater in latter cases (20+ million cycles). The variation of both magnitude and direction of the hip contact force seems to have prompted earlier and more extensive failures, as observed in the bovine bone samples, even if the bond between the cement and the bovine bone might be stronger than the bond between the cement and sawbone, due to a greater penetration of cement into the open bone matrix. Utilisation of a hip simulator in the evaluation of fatigue endurance seems to be essential, if quantitative measures are to be obtained for new fixation/implants.

It is reassuring to note that the hip contact forces realised in the hip simulator are the same as those measured in vivo, and debonding was found to initiate in the posterior–superior quadrant, consistent with the high stress regions found from the FE analyses. Although the geometry and the strength of bovine pelvic bone differ from those of human, the morphology of the cancellous bone is surprisingly similar to that of human, as confirmed by two of the co-authors who are practicing surgeons. Bovine bones have been used for mechanical testing purposes for many years [20-22], one of the most recent work was in fact on the effect of bone porosity on the mechanical integrity of the bone–cement interface, utilising bovine bones [22]. The porosity of bovine trabecular bone was considered comparable with that of human bone and suitable for studies of bone–cement interface.

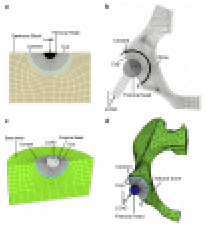
Admittedly, although lubricant was used in the articulation, the tests were carried out under dry conditions. Further work is to construct a serum chamber so that simulated body fluid may be used to bring the in vitro testing even closer to in vivo situations.

4.2. Stress state and failure mode

Although numerous studies of micro-cracking in bulk cement provide basic information on crack initiation and growth in cement specimens, the results from these studies can not be utilised directly in the study of cement fixation in cemented replacements, as the stress state of a cement mantle is inevitably multiaxial and variable, could not be adequately represented by, in most of cases, uniaxially loaded cement specimens. In fact, cement as a fixation agent has rarely been tested, particularly on the acetabular side, where the loading conditions differ fundamentally from those on the femoral side.

In our previous work [4], a simple multilayer model (Fig. 6a) was used to reproduce the variable multiaxial stress state as that in the cement mantle in a plane strain pelvic bone model (Fig. 6b). The predominant stress components were found to be tangential and radial stresses during a normal walking cycle. Cyclic tests of the experimental model confirmed that high combined tangential and radial stresses were sufficient in promoting radial crack growth under peak hip contact force during normal walking. The maximum principal stress of ~9 MPa at peak hip contact force during gait is comparable with the fatigue strength of CMW 1, as reported in [23]. The fatigue limit (at 106 cycles) of CMW was reported to be 14–17 MPa for fully compressive cycles, and 10 MPa for fully tensile cycles in [24]. This cracking pattern seems to be consistent with that observed in sections of cemented femoral replacements tested in vitro [25] and retrieved samples [6]. In contrast to femoral implants, however, cracks were observed to initiate from bone–cement interface, rather than prosthesis–cement interface.

Fig. 6



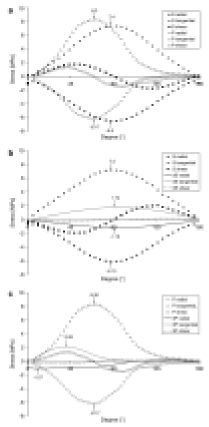
Finite element models developed for the evaluation of cement stresses: (a) Simple multilayer model (S) [4]; (b) plane strain pelvic bone model (P) [4]; (c) simple 3D multilayer model (3S); (d) 3D pelvic bone model built from CT scan images (3P) [26].

In a further development [5], 3D acetabular reconstructs were tested under constant amplitude fatigue loading conditions, simulating the magnitude and the orientation of peak hip contact force during normal walking. Composite hemi-pelvis samples were tested and extensive debonding at the bone–cement interface was observed in the dome region of the acetabulum. This failure scenario clearly differs from either that of 2D acetabular models [4] or that of femoral replacement models [6-12]. Amongst multiple parameters likely to have affected the fatigue behaviour, mechanistic aspect has been investigated using the finite element method. Fig. 6c shows a 3D simplified multilayer model while a non-homogeneous 3D pelvic bone model is shown in Fig. 6d. The last model was developed from CT scan images of a human pelvis, with variable bone densities scaled with a Hounsfield unit. The details of the model were reported elsewhere [26].

Fig. 7 shows the local stress distributions in the cement mantle near the bone–cement interface at peak hip contact force during normal walking for (a) simple multilayer and plane strain pelvic bone model [4]; (b) simple multilayer [4] and 3D multilayer models and (c) plane strain pelvic bone model [4] and 3D pelvic bone model [26]. It seems that although the patterns of stress distribution in the cement mantle and at the bone–cement interface are broadly similar for all models, the magnitudes of the stress components vary markedly for the same peak hip contact force. Whilst tangential and radial stresses generated in the simple multilayer model are comparable to that of the fatigue strength of the bulk cement [4], promoting radial cracking in the cement mantle, the stress levels in the 3D multilayer and 3D pelvic bone models are much lower (~25% for tangential stress; <20% for radial stress), such that the effective stresses in the cement mantle are far below the fatigue strength of bulk cement. Under these conditions, it is conceivable that cracking in the cement mantle may be completely prohibited, while other failure mechanisms, such as debonding at the bone–cement interface, may gain significance under favourable conditions.

Fig. 7

Local stress distributions near the bone–cement interface at peak hip contact force during normal walking. (a) Comparison of the results from the plane strain pelvic bone (P) and the simple multilayer (S) models; (b) comparison of the results (more ...)



Although 3D mechanics analysis of perfectly bonded bone–cement interface appears to suggest low shear stresses as well tangential and radial stresses (Fig. 7b and c), examination of virgin samples following standard implanting procedures revealed that defects at the bone–cement interface exist in majority of the samples prepared, possibly due to the shrinkage of cement post curing. It would seem to be plausible that debonding becomes the preferential failure mode at the bone–cement interface, as a result of these pre-existing defects. As only mechanical interlocking governs the bonding strength, the effectiveness of the cement penetration critically depends on the morphology of the bone. It seems reasonable to assume that any residual soft tissues or incomplete reaming may prevent complete cement penetration, hence facilitate premature failures of the bonding. Efforts should be directed towards improving the integrity of the initial fixation, so that the occurrence of initial defects can be at least minimised, if not completely eliminated. In standard acetabular replacements, stresses in the cement mantle are sufficiently low so that fatigue cracking may be largely suppressed.

Both cement–bone and cement–prosthesis interfaces were assumed fully bonded, this is necessary for the present analyses where the emphasis was on the validation of the function of the hip simulator. Further work is to develop models capable of handling progressive damage as crack growth at interface, as identified from the experiments, so that fatigue lives may be predicted from FE models and validated by the results from the hip simulator.

Other Sections▼

5. Conclusions

Fatigue failure in cemented acetabular reconstructs has been found to be at the bone–cement interface, based on a new hip simulator study. This failure mechanism is consistent with that of samples tested under constant amplitude fatigue [5], although fatigue lives in samples tested using hip simulator are much shorter than those tested under conventional constant amplitude fatigue. Finite element analysis seems to suggest that the stress magnitudes in 3D reconstructs are insufficient to generate cracking in cement mantle, while initial defects at the bone–cement interface may be responsible for eventual failure of the interface under physiological loading conditions.

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Footnotes

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