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Mechanical properties of cancellous bone from the acetabulum in relation to acetabular shell fixation and compared with the corresponding femoral head

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ABSTRACT

To gain initial stability for cementless fixation the acetabular components of a total hip replacement are press-fit into the acetabulum. Uneven stiffness of the acetabular bone will result in irregular deformation of the shell which may hinder insertion of the liner or lead to premature loosening. To investigate this, we removed bone cores from the ilium, ischium and pubis within each acetabulum and from selected sites in corresponding femoral heads from four cadavers for mechanical testing in unconfined compression. From a stress-relaxation test over 300 s, the residual stress, its percentage of the initial stress and the stress half-life were calculated. Maximum modulus, yield stress and energy to yield (resilience) were calculated from a load-displacement test. Acetabular bone had a modulus about 10–20%, yield stress about 25% and resilience about 40% of the values for the femoral head. The stress half-life was typically between 2–4 s and the residual stress was about 60% of peak stress in both acetabulum and femur. Pubic bone was mechanically the poorest. These results may explain uneven deformation of press-fit acetabular shells as they are inserted. The measured half-life of stress-relaxation indicates that waiting a few minutes between insertion of the shell and the liner may allow seating of a poorly congruent liner.

1. Introduction

The use of uncemented fixation for total hip arthroplasty (THA) varies from country to country but registries report it is gaining in popularity. In the US in 2012, 93% of THA constructs were cementless, increasing from 46% in 2001, and the hybrid construct, comprising a cemented stem and cementless cup, accounted for just 5% [1]. This is higher than in most countries. In Australia, cementless components are used in 63.2% and hybrid fixation in 32.4% of primary THA [2], whereas in Sweden cemented fixation is still more popular with only 20.9% of procedures reported as being uncemented and 3% hybrid [3]. In the latest report from the National Joint Registry of England and Wales, 39.0% of all primary hip replacements in 2015 had both components uncemented and 17.1% were classified as hybrid [4].

To gain initial stability, cementless acetabular components require a press-fit of an oversized shell into the acetabulum [5,6]. This approach can also be used for revision surgery [7] in cases of contained defects according to the American Academy of Orthopaedic Surgery classification [8]. There are concerns, however, that insertion forces may deform the acetabular shell making placement difficult, and this could affect liner insertion [9–11].

Despite the apparent importance of the underlying cancellous bone mechanical properties in providing initial stability we are aware of only two studies that have measured the mechanical properties of bone from this region of the pelvis [12,13]. The first was a comprehensive investigation of two whole pelves: a female from which 18 cubic samples were taken, and a male from which 39 cubes were obtained, although none was specifically taken from the acetabulum. The cubes hade sides about 6.5 mm long and were tested in all three directions across the faces in uniaxial unconstrained compression to 0.8% strain after pre-conditioning [12]. The second study investigated cement penetration into the reamed acetabular bone using cores taken from the articular surface of the femoral head and the acetabulum from patients with end-stage osteoarthritis (OA) undergoing THA. In addition to permeability, they

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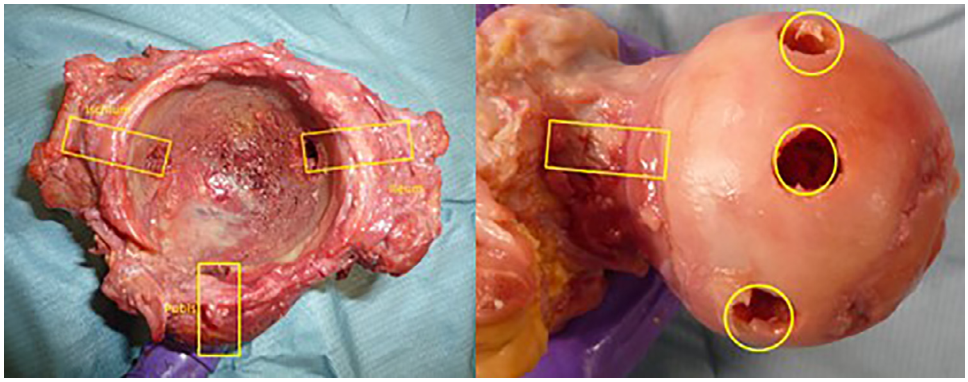

Fig. 1. Core locations from the acetabulum (ilium, ischium and pubis) and the proximal femur from the load-bearing area: superior, anterior, posterior and one drilled central to the femoral neck following resection of the femoral head.

measured the Young's modulus, apparent density and porosity [13]. Others have used CT scans to measure bone density and estimate the modulus [14,15], although once again these models were of the whole pelvis rather than just the acetabulum. In a previous study we investigated the effect of the stiffness of the bone on acetabular shell deformation and the ability of a surgeon to make a subjective estimate of the stiffness of the acetabular bone [16]. In that study, however, the design of the experiment precluded direct measurement of the properties of the acetabular bone.

To address this deficit, therefore, mechanical testing was performed on cores of bone from the ilium, ischium and pubis of reamed acetabula to answer the questions: (1) does the stiffness of the cancellous bone vary with location in the acetabulum? (2) Because bone is slightly viscoelastic, how quickly does a deformation relax and (3) to what extent? These data were compared with measurements from cores taken from selected sites over the corresponding femoral head.

2. Materials and methods

2.1. Bone samples

Four male, fresh frozen, whole pelves were obtained from Caucasian donors with mean body mass index of 26.3 kg/m^2 (range 20–31) and mean age 69 years (range 65–73). All specimens underwent CT scanning to exclude structural abnormalities prior to testing. Ethics committee approval for this study was obtained from the UK Human Tissue Authority, licensing number 12148, and all procedures were performed in accordance with the declaration of Helsinki.

Bone cores, approximately 9 mm diameter and of various lengths, were removed from selected sites on both femoral heads and acetabula from each pelvis (Fig. 1). Cores from the acetabula were drilled perpendicular to the articular surface into each of the three bones making up the innominate: ilium, ischium and pubis. Cores were removed from four sites in each femoral head; three from the load-bearing area: superior, anterior, posterior and one drilled along the axis of the femoral neck following resection of the femoral head. Samples were stored frozen wrapped in saline-soaked gauze. Before testing, each sample was thawed at room temperature and trimmed using a scalpel to remove any articular cartilage and ensure the ends were plane-parallel. The length and diameter of each core were measured using electronic Vernier calipers (Mitutoyo Digimatic, CD-6"CX). After mechanical testing, cores were cleaned of marrow by immersing in proteinase K (1 mg/ml in PBS, Fisher Scientific, UK)/ SDS (1% v/v) (SigmaAldrich/Merck, UK) solution. The apparent density of each core was determined by weighing and dividing the mass by the core volume. Material density was measured by weighing each core immersed in water and using Archimedes' principle [17].

Finally, bone cores were imaged using a Faxitron MX microfocal radiography unit (Faxitron, Tucson, AZ, USA) and a ScanX computed radiography scanner (Dürr NDT, Germany). Samples were imaged at 25 kV for 15 s exposures using a phosphor screen. Digital images were obtained by digital scanning of the phosphor screen using a ScanX laser scanner to release the stored image from the phosphor screen in the form of visible light photons. The photons were collected and amplified by the scanner and converted to a digital signal for processing and display. Images were acquired by Faxitron software and stored as DICOM images. Image J v1.50e was used to re-orientate images and convert to TIFF files.

2.2. Mechanical testing

Mechanical testing was done using an Instron materials testing machine (Instron Ltd., High Wycombe, model 5564) fitted with a 2 kN load cell. The calibration precision of the load cell was $<0.2\%$ from 20 N to 2 kN load. Two tests were performed in unconfined compression: a modified stress-relaxation test followed by a load-displacement test to yield. A modified stress-relaxation was performed by compressing at a displacement rate of 5 mm/min to a set load then holding the displacement for a total test time of 300 s. Femoral cores were loaded to 50 N. Acetabular cores proved to be much weaker and peak load was reduced first to 25 N for the first acetabulum, then to 10 N for the remaining acetabula. Loads were converted to stress by dividing by the cross-sectional area, engineering strains were calculated from the displacement divided by the original length. We used the loading part of the stress-relaxation test to calculate the modulus from the gradient of a straight line or a quadratic curve fitted to the stress-strain data. In the case of a non-linear relationship, the peak modulus was determined and the modulus at a load of 10 N also measured to enable comparison of femoral with acetabular data at a constant stress. The peak stress, the residual stress after 300 s as a percentage of the initial stress and the stress half-life, the time taken for half of the stress relaxation to occur, were calculated.

Stress-strain testing was done at a cross-head speed corresponding to a strain rate of 10% per minute (0.00167/s). Compression was monitored visually until the steepness of the load-displacement curve could be seen to be decreasing, indicating failure was starting to occur [17]. A fourth-degree polynomial was fitted to the stress-strain data in order to determine the maximum slope (maximum modulus) and the 3% yield point, i.e. the stress and strain at which the maximum modulus had declined by 3% [17]. The energy to yield, also called resilience, was calculated from the area under the stress-strain curve to the yield point.

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