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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.

Keywords

Cementless fixation; acetabulum; bone; mechanical properties, mechanical testing; viscoelastic

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 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 (Figure 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^{-1} 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^{-1}). 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. Analysis was done using Microsoft Excel software. Data are presented as mean (standard deviation). Statistical analysis was done using SigmaPlot 13.0 (Systat Software Inc.).

Repeated measures analysis of variance was used to explore differences between sites, with repeated measures being the 'within-subjects factor', and a value of $P<0.05$ was taken to be statistically significant.

3. Results

In total, 32 femoral and 23 acetabular cores were tested. The mean diameter of the cores was 8.89 mm (range 7.97 to 9.22 mm) and the mean length was 14.6 mm (range 6.7 to 22.2 mm). One pubic core broke while being extracted. It was immediately apparent that acetabular cores were much less robust than those from the femoral head. Testing protocols had to be adapted to test successfully the much weaker and softer cores, especially those from the pubis, two more of which broke during preparation or at an early stage of testing.

3.1 Stress-relaxation

Data from a typical stress-relaxation test from a femoral head core are shown in Figure 2. In this case a quadratic expression was fitted to the rising part of the loading curve and the peak modulus calculated at the end of this loading phase. The initial stresses were all close to 0.8 MPa for the femur and 0.16 MPa for the acetabulum. Stress relaxation was characterised by the final stress and the percentage residual stress, expressing the final stress as a percentage of the initial stress (Table 1). Stress relaxation generally proceeded very rapidly so that half of the stress-relaxation had occurred within a few seconds, although there was a wide spread of values and some notable exceptions to this generalization as shown by the range of values (Table 1). A compilation of all the acetabular relaxation data is shown in Figure 3 and that from the femoral head cores in Supplementary Figure 1. While many curves from each anatomical component are similar, several samples deviated markedly from the most common pattern for reasons that are not clear. Samples from the ilium and ischium of donor C relaxed almost to zero with half-lives of 9.3 s and 11.5 s respectively, whereas donor C is the only one to show reasonably consistent patterns from all sites over the femoral head. The right pubic sample of donor A was the first tested and was loaded to 25 N. This proved too high, the sample broke, and so data from the right acetabulum of donor A have been scaled by 10/25 for comparison with all the other acetabular samples which were loaded only to 10 N. The average modulus of all the acetabular cores at 10 N (13.2 (9.4) MPa) was about 25% of the average of the cores from the three loaded regions (posterior, anterior and superior) of the femoral head (56.2 (16.1) MPa). One core from the femoral neck of each of individuals A and B also failed early during testing.

3.2 Stress-strain

A typical stress-strain curve is shown in Figure 4 with the peak modulus and the yield point, the stress at which the modulus has decreased by 3%, marked. The peak moduli calculated may be found in Table 2 and show that the acetabular bone was considerably more deformable than that from the femoral head. Mean values differed significantly by site (P<0.001) and post hoc tests showed that these differences lay largely between femoral head and acetabular samples. There was large variation in the data and the cores from the anterior of the femoral head were, on average the stiffest. In contrast, the cores from the acetabulum had, on average, only 13% of the modulus of the average of cores from the loaded region of the femoral head. The ischial samples were the stiffest, followed by iliac and pubic.

Yield stress varied significantly with site (P<0.001) and reflected a similar order to the modulus, with the superior of the femoral head having the greatest median strength (Table 2). In the acetabulum, however, the ilium was the strongest and there was no difference between the strengths of the ischium and pubis although all the acetabular values were, again, considerably smaller, about 25% of those from the femoral head.

The energy to yield, resilience, showed a similar pattern to strength (Table 2). There was little difference between anterior, superior and posterior samples, with median values all around 15 kJ m^{-3}. Cores from the ilium were most resilient and those from pubis and ischium were similar. The overall difference between sites was not significant (P=0.22) due to the large variation in values although, on average, acetabular bone had only about 40% of the resilience of the femoral head bone.

3.3 Bone density

Measured values of densities are shown in Table 2. The apparent density of each core represents the amount of bone and values were significantly lower in the acetabulum than in the femoral head ($P < 0.001$). Apparent densities at the three sites in the acetabulum were very similar and approximately half of the average over the femoral head; values for the neck were slightly smaller than those for the femoral head. The material density of the bone matrix was slightly, but significantly ($P < 0.001$), greater in the acetabular bone than in the femoral head. The ratio of apparent to material densities was about 0.1 in the acetabulum but between 0.20 to 0.28 over the femoral head. X-ray images of cores from the acetabulum show the differences in quantity and texture of the trabecular bone compared with samples from the femoral head and the femoral neck (Fig.5).

4. Discussion

These data show that acetabular bone is considerably less stiff and less resilient than bone from the femoral head. Possibly because loads are spread over a larger area in the acetabulum whereas the femoral head and neck have an effect of concentrating the loads into a smaller area for transmission to the femoral diaphysis. Some of the pubic samples were not tested as they were so fragile that either they were already broken or broke during preparation. The smaller modulus of pubic bone means that deformation under a given load is larger than for ischial or ilial samples. Although the gradient of the curve for the pubic samples was smaller, the yield strength was not much different from ilial or ischial samples and hence the resilience, being the area beneath the curve, was not as dissimilar from those other sites as might initially be expected. Being incorporated into a larger structure will also enhance the properties over those measured in isolated, unsupported cores [18]. These data are in general agreement with the density measures of Dalstra et al. who reported the highest densities were to be found in the superior/anterior area of the acetabular wall, while the lowest densities were found in the ischial bone [12].

Bone from all sites showed viscoelastic behaviour, with stresses decaying by about 40% over a 5 minute period. Most of this decay appears to take place in the first approximately 10 s in most of the samples. The two outliers from donor C, right ilium and ischium, are unexplained. The subsequent testing to failure showed no unusual behaviour and there was no evidence of specimen damage before or during testing. Variations between samples could be evidence of natural variation as only four individuals were available and further testing, of samples from individuals with a wider range of ages and both sexes, is indicated, as discussed further below.

It needs to be noted that the test we performed is not a true stress-relaxation test, in which a predetermined constant strain is applied to each sample. In a pilot study it became apparent that there was large variability in the properties of bone from different regions of the acetabulum and we found that we could not identify a realistic and consistent value of strain to apply to all samples in order to standardize a stress relaxation test using constant strain. We could, however, load to a predetermined stress, although we still over-estimated this in the first samples in this study. A creep test might then have seemed the obvious test to perform but the screw-driven materials testing machine available to us is not well-suited to such a test. Hence, in order to obtain an element of comparability we allowed the stress to relax from a predetermined value.

The moduli and apparent densities we found for the acetabulum are similar to those reported by previous studies. Dalstra et al. reported mean values for moduli of between 30-60 MPa and apparent densities of 0.25 (0.10) g cm^{-3} for female bone, after removing some samples suspected of being subchondral bone rather than trabecular bone, and 0.195 (0.054) g cm^{-3} for bone from the one male pelvis tested [12]. Measurements, however, were made over the whole pelvis and identifying values from locations and directions that would correspond to those we measured is not possible. They measured density and fabric parameters over the whole pelvis with a view to improving finite element modelling of the pelvis, whereas we concentrated on properties, both instantaneous and time dependent, within the acetabulum as these will have the most direct effect on immediate cementless fixation of a press-fit cup. Another report found greater values for Young's modulus in the acetabulum, 116.4 (86.7) MPa, compared with the femoral head, 47.4 (29.5) MPa, with apparent densities of 0.35 g cm^{-3} in the acetabulum and 0.24 g cm^{-3} in the femoral head[13]. These samples were taken from patients with end-stage OA which may be the reason these acetabular moduli and densities are considerably higher than those we report here. Surprisingly, however, the moduli and apparent densities measured from the femoral heads are at the low end of the range compared with many previous studies [17,19,20]. For example, in earlier studies of OA bone we found that the average modulus of cancellous bone from the femoral head was 356 MPa and the apparent density was 0.71 g cm^{-3} reflecting the increased amount of bone commonly found [17], compared with non-diseased bone as reported here.

The apparent density of the trabecular bone in the acetabulum was approximately half that of the cores over the femoral head but, surprisingly, the material density was about 25% greater in the acetabulum. Strength and stiffness arise not just from the amount of bone but also its quality, and although there are clear differences in the densities these are not as great as those found in the moduli between femoral head and acetabulum. It is not clear from these measurements why the moduli are almost an order of magnitude different. One possible explanation is the bone structure, which was beyond the scope of this study. Differences in strength and energy to yield are smaller and may reflect more the differences in the amount of bone.

When pressing a shell into the acetabulum the greatest resistance will come from the stiffest bone. Consequently, it would be expected that insertion at the pubic location would progress most rapidly while that at ilial and ischial sites would meet most resistance. Unless the shell is uniformly supported by the insertion device this may cause some deformation of the shell making insertion of the liner problematic. The relaxation of the stress, however, may mean that, given sufficient time, the bone might ‘relax’ and the shell deformation reduce. The measurements made here suggest that most of that relaxation occurs rapidly, within about 10 s, but the variation shown in Figure 2 indicates that relaxation still proceeds in some samples after 5 minutes. If a liner does not fit the shell immediately, quality control means it is unlikely to be a mismatch in manufacturing and perhaps waiting a few minutes for the bone to absorb the deforming forces may allow the liner to be inserted.

There are a number of limitations to this study. The difficulty obtaining whole pelves and acetabula means that the number of individuals studied is small and studying both sides means that these samples are not then statistically independent. Consequently, repeated measures tests were used with the ‘within-subjects’ factor being the repeated measure as this test enables statistical inference to be made with fewer subjects [21]. All the donors were male and fairly close in age. A similar study would be required using tissue from female donors in order to provide an accurate representation of the similarities and differences between the sexes. The donors for this study were towards the younger end of those undergoing THR so including a wider range of older donors, and those with OA, would also be of interest. Although care was taken to cut the faces of the cores to be plane-parallel, a small angle between the loading platens and the core could explain some of the variation in strain. Bone marrow was left in situ, no attempt was made to clean the cores prior to testing, and this may have had a slight strengthening effect although, arguably, one that is present *in vivo*. The fragility of some of the cores, especially from the pubis, surprised us and made these samples difficult to test. Knowing this would lead us to redesign a future study to be more protective of cores during preparation. Finally, bone strength and stiffness depend not only on the amount of bone and its quality but also on the organisation of the trabeculae. In this preliminary study we did not measure bone trabecular structure but future studies would benefit from microCT measures of fabric.

In conclusion, this study provides unique data on the mechanical properties of cores from the acetabulum that not only will assist future modelling studies but also may inform surgical approaches to insertion of an acetabular shell for cementless fixation. Acetabular bone was considerably less stiff and much weaker than bone from the femoral head. The strongest and stiffest bone was found in the superior aspect, mainly the ilium, closely followed by the ischium with pubic bone having markedly inferior mechanical properties.

Acknowledgements

We thank Phil Jackson for help sourcing donated cadavers and transportation of samples and Dr F.R. Saunders for help with core preparation and advice.

Competing interests: Thomas Pandorf is an employee of Ceramtec GmbH. The other authors have no conflicting interests to declare with respect to the work published in this paper.

Funding

This research did not receive any specific grant from funding agencies in the public, commercial, or not-for-profit sectors.

Ethical approval: 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.

References

[1] Lehil MS, Bozic KJ. Trends in Total Hip Arthroplasty Implant Utilization in the United States. J Arthroplasty 2014;29(10):1915-8.

[2] Australian Orthopaedic Association National Joint Replacement Registry. Annual Report. . Adelaide: AOA, 2015.

[3] Garellick G, Kärrholm J, Lindahl H, Malchau H, Rogmark C, Rolfson O. Swedish Hip Arthroplasty Register Annual Report 2014. 2014.

[4] National Joint Registry 13th Annual Report. 2016.

[5] Kwong LM, O'Connor DO, Sedlacek RC, Krushell RJ, Maloney WJ, Harris WH. A quantitative in vitro assessment of fit and screw fixation on the stability of a cementless hemispherical acetabular component. J Arthroplasty 1994;9(2):163-70.

[6] Squire M, Griffin WL, Mason JB, Peindl RD, Odum S. Acetabular component deformation with press-fit fixation. J Arthroplasty 2006;21(6 Suppl 2):72-7.

[7] Elke R, Berli B, Wagner A, Morscher EW. Acetabular revision in total hip replacement with a press-fit cup. J Bone Joint Surg Br 2003;85(8):1114-9.

[8] D'Antonio JA, Capello WN, Borden LS, Bargar WL, Bierbaum BF, Boettcher WG, Steinberg ME, Stulberg SD, Wedge JH. Classification and management of acetabular abnormalities in total hip arthroplasty. Clin Orthop Relat Res 1989(243):126-37.

[9] Jin ZM, Meakins S, Morlock MM, Parsons P, Hardaker C, Flett M, Isaac G. Deformation of press-fitted metallic resurfacing cups. Part 1: Experimental simulation. Proc Inst Mech Eng H 2006;220(2):299-309.

[10] Meding JB, Small SR, Jones ME, Berend ME, Ritter MA. Acetabular cup design influences deformational response in total hip arthroplasty. Clin Orthop Relat Res 2013;471(2):403-9.

[11] Liu F, Chen Z, Gu Y, Wang Q, Cui W, Fan W. Deformation of the Durom acetabular component and its impact on tribology in a cadaveric model--a simulator study. PLoS One 2012;7(10):e45786.

[12] Dalstra M, Huiskes R, Odgaard A, van Erning L. Mechanical and textural properties of pelvic trabecular bone. J Biomech 1993;26(4-5):523-35.

[13] Thompson MS, Flivik G, Juliusson R, Odgaard A, Ryd L. A comparison of structural and mechanical properties in cancellous bone from the femoral head and acetabulum. Proc Inst Mech Eng H 2004;218(6):425-9.

[14] Clarke SG, Phillips ATM, Bull AMJ. Evaluating a suitable level of model complexity for finite element analysis of the intact acetabulum. Comput Methods Biomech Biomed Engng 2013;16(7):717-24.

[15] Anderson AE, Peters CL, Tuttle BD, Weiss JA. Subject-Specific Finite Element Model of the Pelvis: Development, Validation and Sensitivity Studies. J Biomech Eng 2005;127(3):364-73.

[16] Bone MC, Dold P, Flohr M, Preuss R, Joyce TJ, Aspden RM, Holland J, Deehan D. The influence of the strength of bone on the deformation of acetabular shells: a laboratory experiment in cadavers. Bone & Joint Journal 2015;97-B(4):473-7.

[17] Li B, Aspden RM. Composition and mechanical properties of cancellous bone from the femoral head of patients with osteoporosis or osteoarthritis. J Bone Miner Res 1997;12:641-51.

[18] Bryce R, Aspden RM, Wytch R. Stiffening effects of cortical bone on vertebral cancellous bone in situ Spine (Phila Pa 1976) 1995;20:999-1003.

[19] Goulet RW, Goldstein SA, Ciarelli MJ, Kuhn JL, Brown MB, Feldkamp LA. The relationship between the structural and orthogonal compressive properties of trabecular bone. J Biomech 1994;27(4):375-89.

[20] Brown SJ, Pollintine P, Powell DE, Davie MW, Sharp CA. Regional differences in mechanical and material properties of femoral head cancellous bone in health and osteoarthritis. Calcif Tissue Int 2002;71(3):227-34.
[21] Louwies T, Panis LI, Kicinski M, De Boever P, Nawrot TS. Retinal microvascular responses to short-term changes in particulate air pollution in healthy adults. Environ Health Perspect 2013;121(9):1011-6.

Table 1. Stress relaxation results. Samples were loaded to 50 N (femur) or 10 N (acetabulum). Data show the mean peak stress, the mean final stress 300 s after the start of the test, the final stress as a percentage residual of the peak stress and the stress half-life for four sites on the femoral head and three in the acetabulum. Also calculated during the loading phase was the modulus at 10 N load. Data are shown as mean (standard deviation), where normally distributed, or median [range] for skewed data. No data were available from 3/8 pubic samples and 2/8 neck samples due to failure of the fragile samples during preparation or testing.

	Femoral head					Acetabulum		
	Posterior	Anterior	Superior	Neck		Ilium	Ischium	Pubis
N	**8**	**8**	**8**	**6**		**8**	**8**	**5**
Peak stress /MPa	0.797 (0.023)	0.801 (0.025)	0.797 (0.012)	0.795 (0.043)		0.163 (0.008)	0.173 (0.020)	0.157 (0.006)
Final stress /MPa	0.485 (0.085)	0.480 (0.059)	0.420 (0.137)	0.484 (0.051)		0.088 (0.031)	0.091 (0.041)	0.089 (0.008)
Residual stress %	61 (11)	57 (14)	55 (14)	56 (14)		55 (19)	53 (23)	56 (5)
Half-life /s	3.0 [1.8:18.1]	2.6 [0.8:58.3]	3.8 [2.7:55.1]	2.6 [1.8:79.1]		2.1 [1.0:3.6]	3.2 [1.9:11.5]	2.2 [1.6:2.2]
Modulus at 10 N /MPa	45.6 [30.1:59.1]	52.5 [14.6:84.8]	67.1 [32.2:116.1]	40.6 [12.7:166.5]		12.1 [5.1:48.6]	13.8 [8.4:40.4]	5.5 [3.7:23.0]

Table 2. Stress-strain and density data. The peak stress, yield stress and the energy to yield are from four sites on the femoral head and three in the acetabulum. Apparent and material densities were measured from the same cores. Data are shown as median [range] or mean (SD)..

	Femoral head					Acetabulum		
	Posterior	Anterior	Superior	Neck		Ilium	Ischium	Pubis
N	**8**	**8**	**8**	**6**		**8**	**8**	**5**
Peak modulus /MPa	131.7 [14.8;291.5]	183.9 [24.9:433.4]	162.0 [56.9:379.0]	75.1 [29.1:142.0]		16.3 [9.42:34.8]	17.8 [0.83:89.0]	8.9 [1.2:20.3]
Yield stress /MPa	1.72 [0.55:4.1]	1.48 [0.69:9.04]	2.03 [0.60:4.92]	1.10 [0.68:2.25]		0.64 [0.07:0.80]	0.34 [0.10:0.93]	0.22 [0.20:0.77]
Energy to yield / kJ m^{-3}	15.4 [5.1:39.5]	14.8 [4.4:156.0]	15.6 [4.4:69.8]	13.3 [3.6:22.5]		9.1 [0.25:15.5]	3.7 [0.63:9.3]	3.5 [0.09:45.1]
Apparent density / g cm^{-3}	0.39 (0.15)	0.42 (0.16)	0.45 (0.13)	0.35 (0.10)		0.25 (0.05)	0.20 (0.04)	0.21 (0.10)
Material density / g cm^{-3}	1.93 (0.47)	1.80 (0.37)	1.64 (0.20)	2.03 (0.49)		2.32 (0.47)	2.37 (0.53)	2.28 (0.35)

Figure 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.

Figure 2. Example of a modified stress-relaxation test showing (a) the loading phase (blue, large dots) expressed as a stress-strain curve to calculate the modulus from the gradient of a fitted quadratic polynomial (black, small dots) and (b) the subsequent relaxation as stress versus time showing the decay from peak to final stress 300s after the start of the test. (Donor C, right femoral head, posterior).

Figure 3. Stress relaxation curves for acetabular samples. Samples were loaded to 10 N and the recording was ended 300 s after the start of loading. (Samples from the right acetabulum of donor A were loaded to 25 N and, accordingly, the data have been scaled by 10/25 for comparison).

Figure 4. Typical stress-strain curve illustrating the finding of the maximum modulus, yield stress and energy to yield as the area under the curve to the yield stress.

Figure 5. X-ray images (Faxitron) of bone cores from each of the main areas tested showing the differences in amount and texture of the trabecular bone at each site. Each core is 9 mm diameter.

Figure S1. Stress relaxation curves for femoral head samples. Samples were loaded to 50 N and the recording was ended 300 s after the start of loading. Two neck samples broke during testing.

Figure 1
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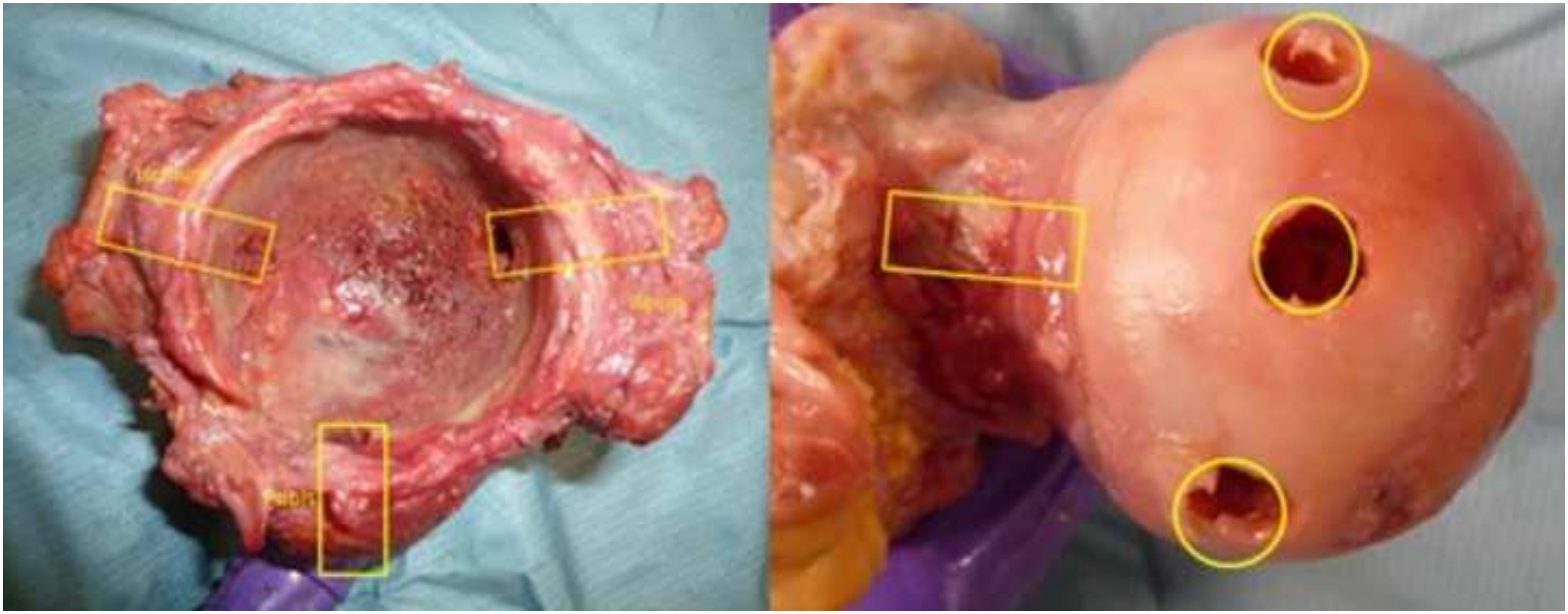


Figure 2a

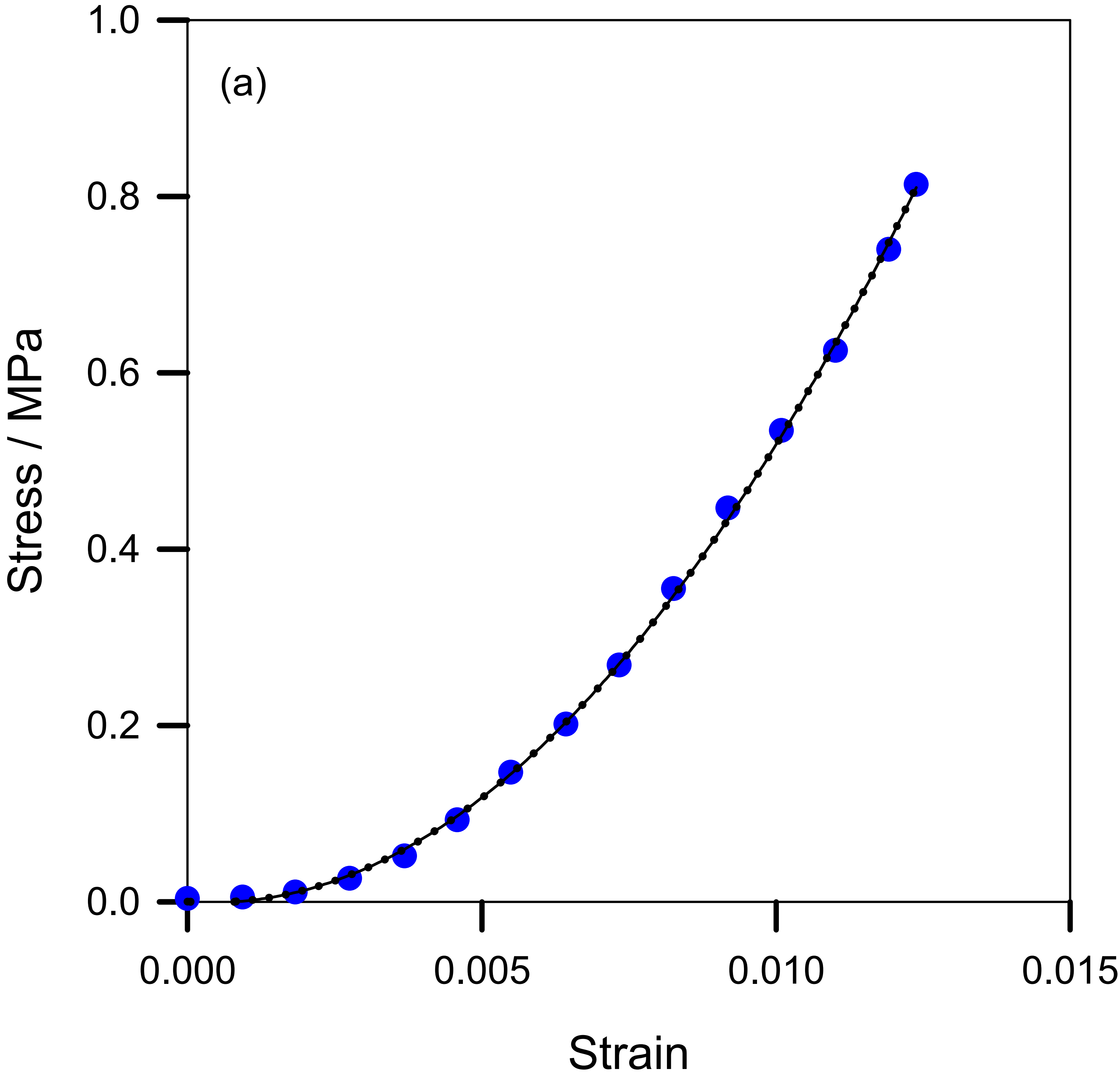


Figure 2b

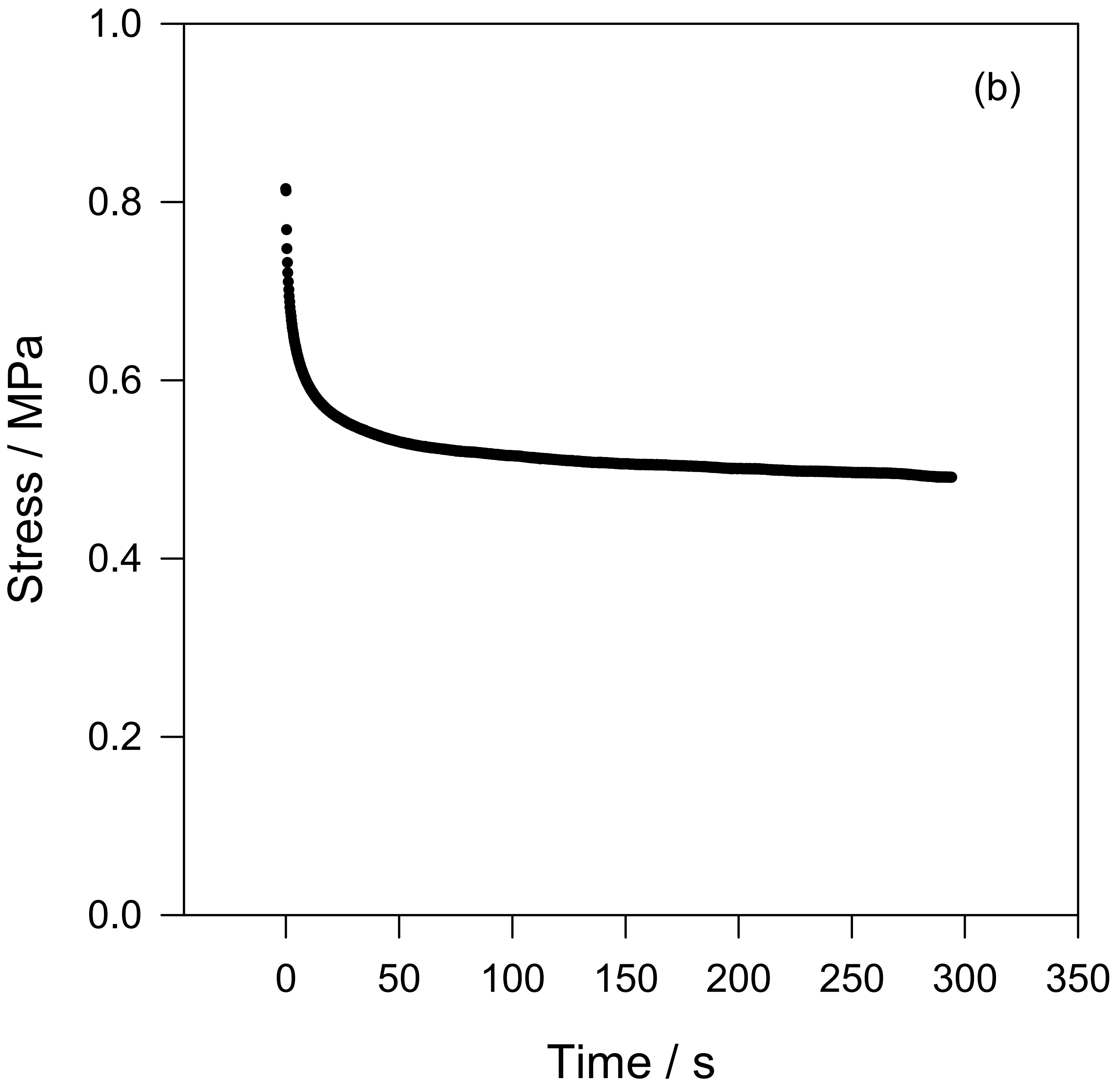
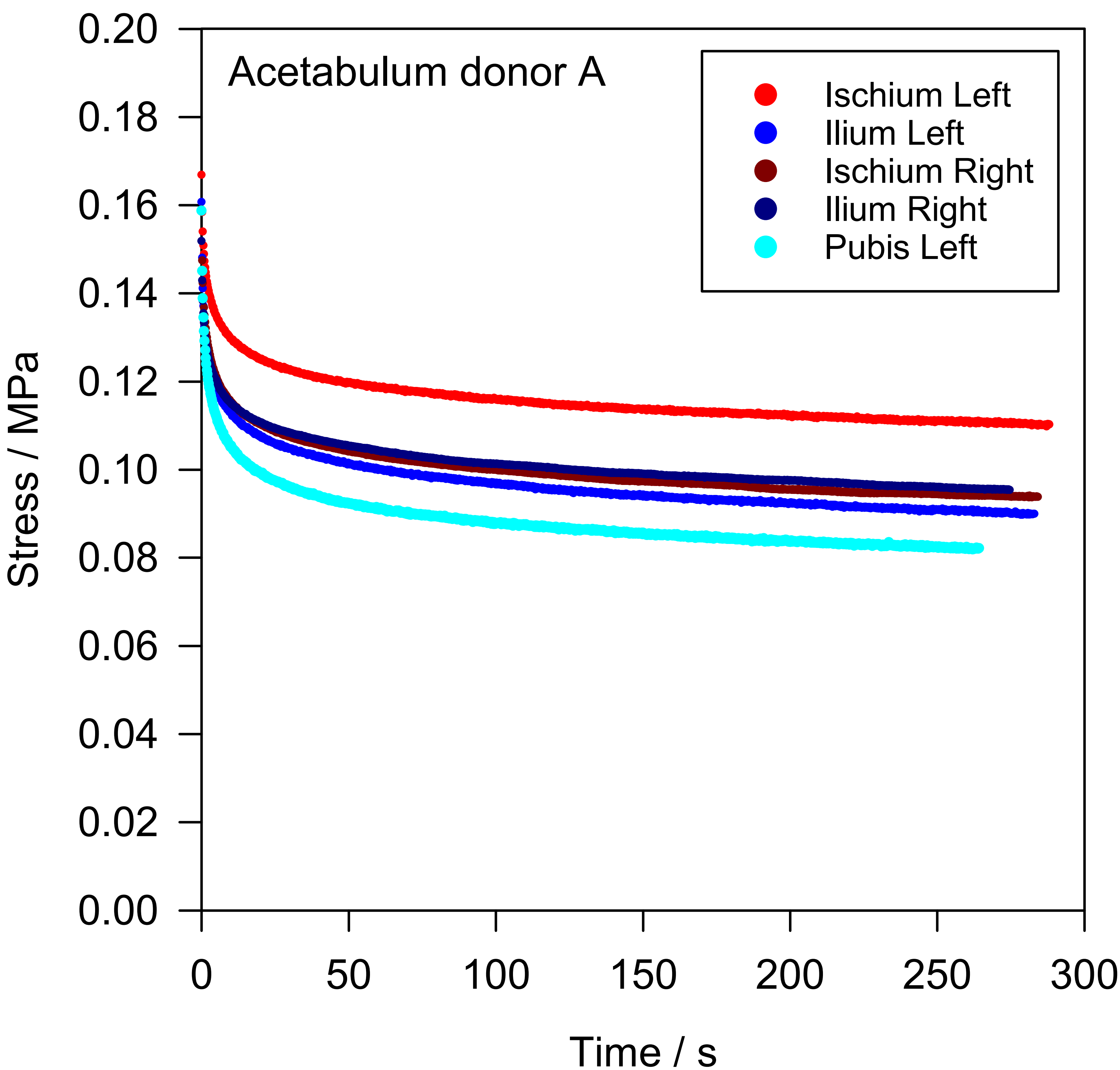


Figure 3a



Acetabulum donor A
Ischium Left
Ilium Left
Ischium Right
Ilium Right
Pubis Left
Stress / MPa
0.00
0.02
0.04
0.06
0.08
0.10
0.12
0.14
0.16
0.18
0.20
Time / s
0
50
100
150
200
250
300

Figure 3b

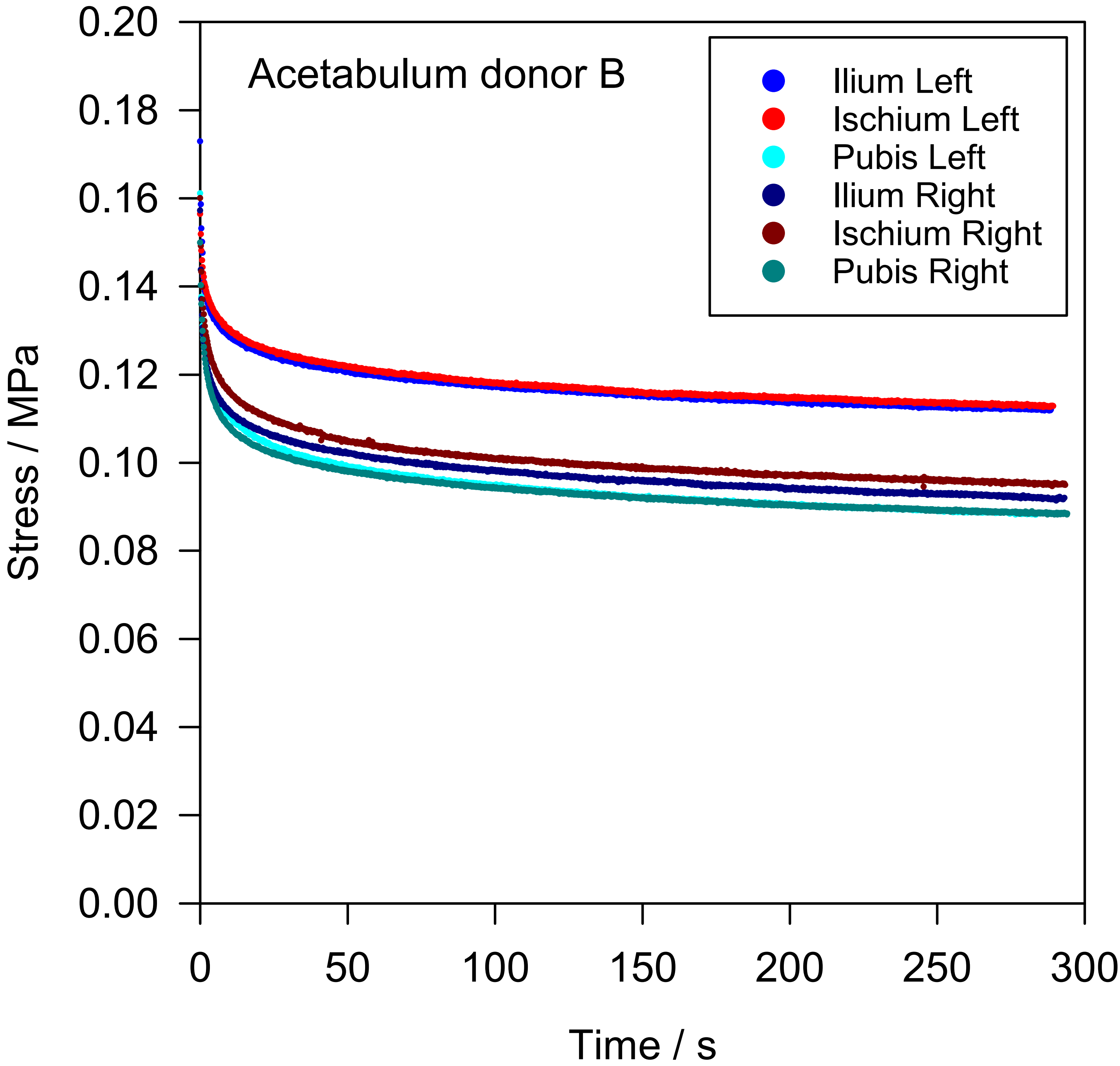

Acetabulum donor B
Ilium Left
Ischium Left
Pubis Left
Ilium Right
Ischium Right
Pubis Right
Stress / MPa
0.20
0.18
0.16
0.14
0.12
0.10
0.08
0.06
0.04
0.02
0.00
0
50
100
150
200
250
300
Time / s

Figure 3c

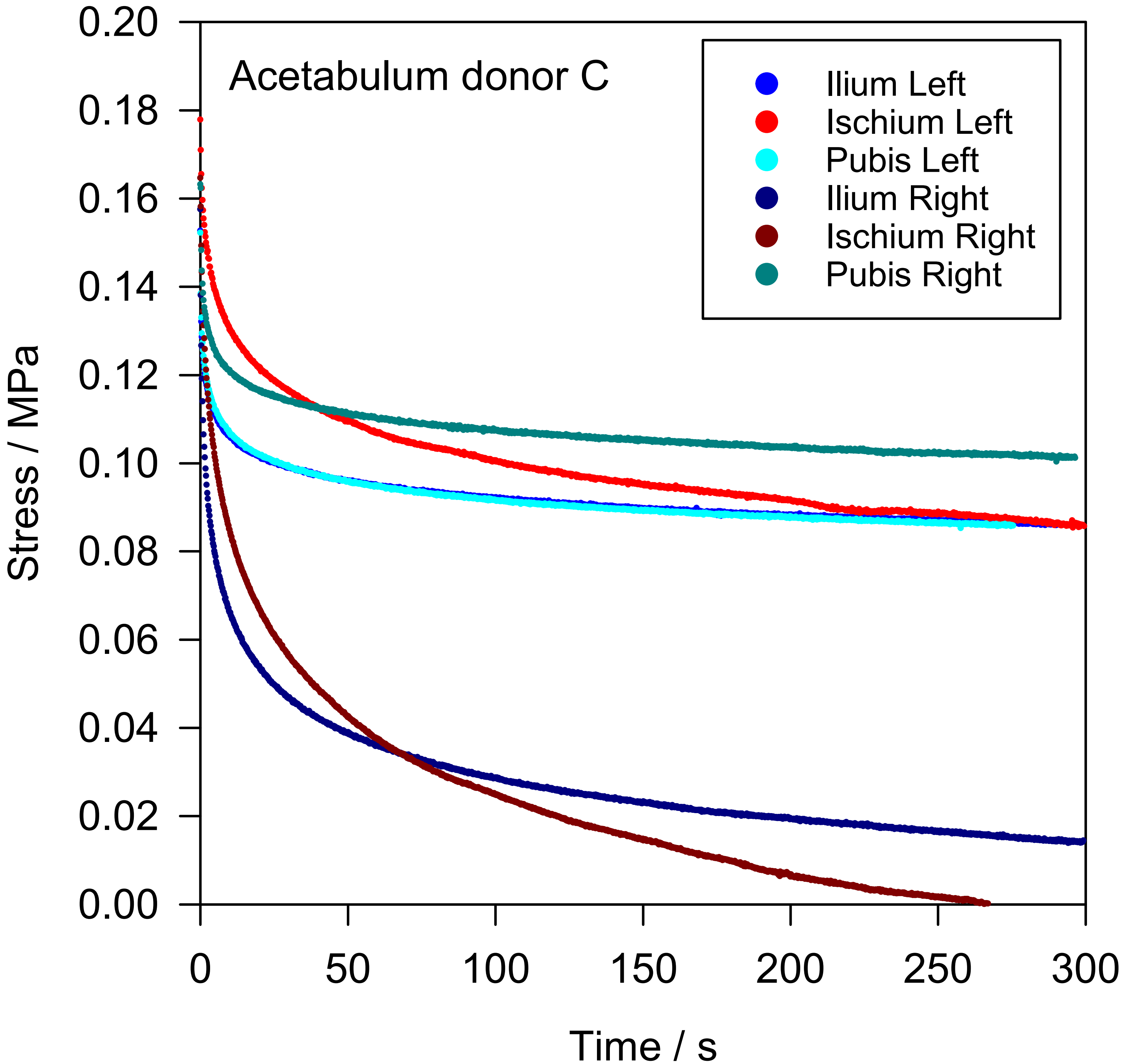
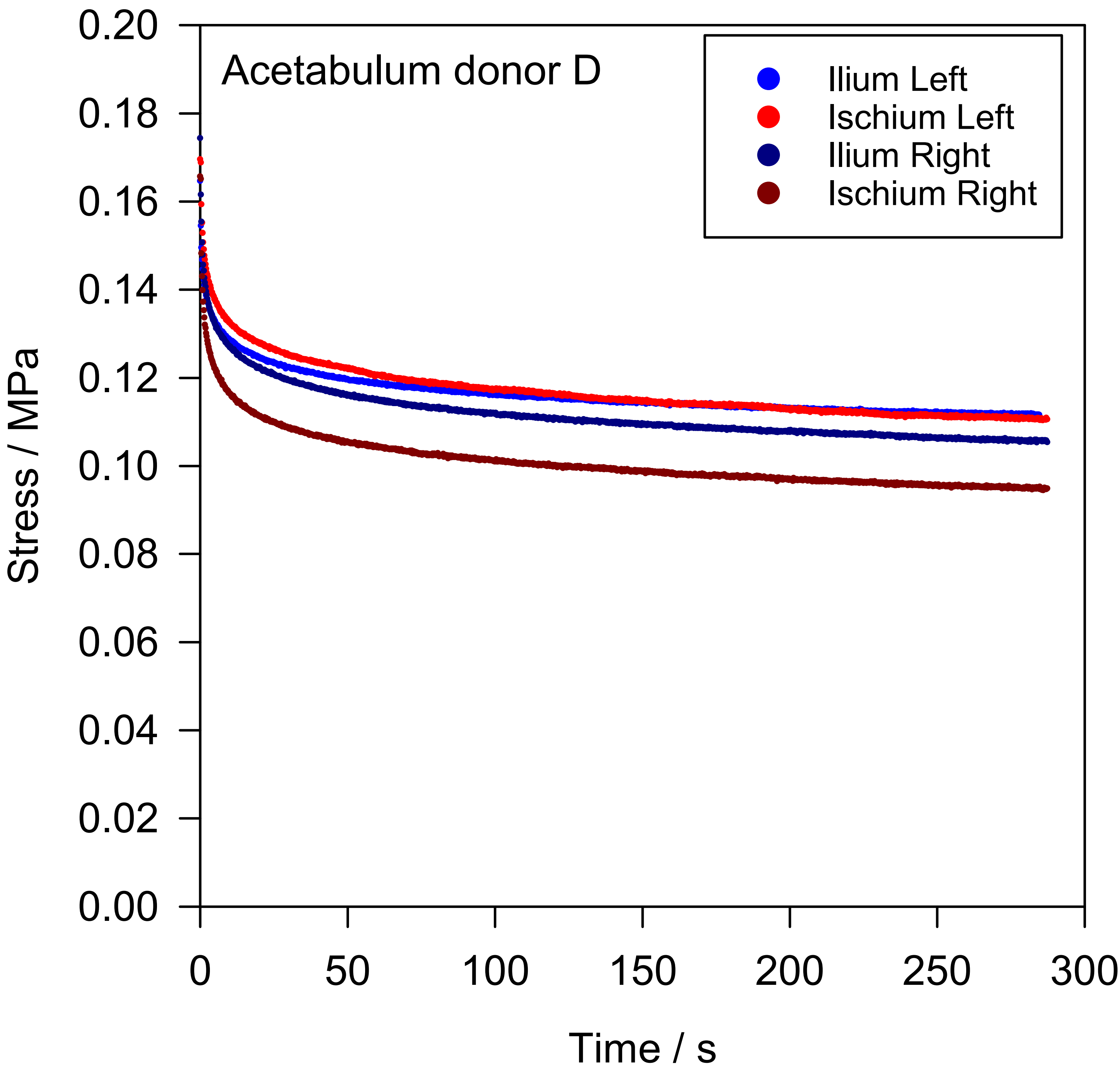
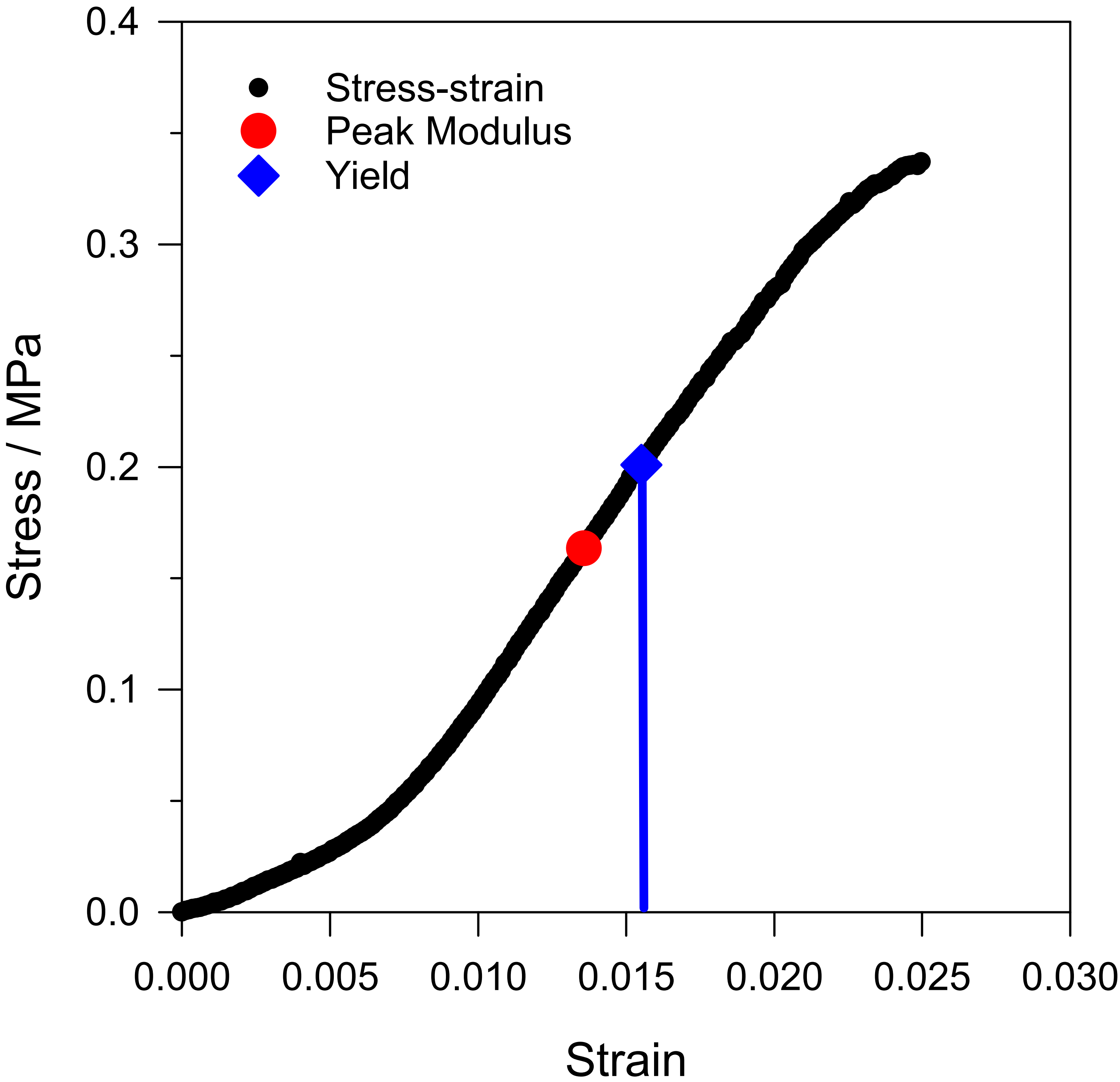

Acetabulum donor C
Ilium Left
Ischium Left
Pubis Left
Ilium Right
Ischium Right
Pubis Right
Stress / MPa
0.20
0.18
0.16
0.14
0.12
0.10
0.08
0.06
0.04
0.02
0.00
0
50
100
150
200
250
300
Time / s

Figure 3d



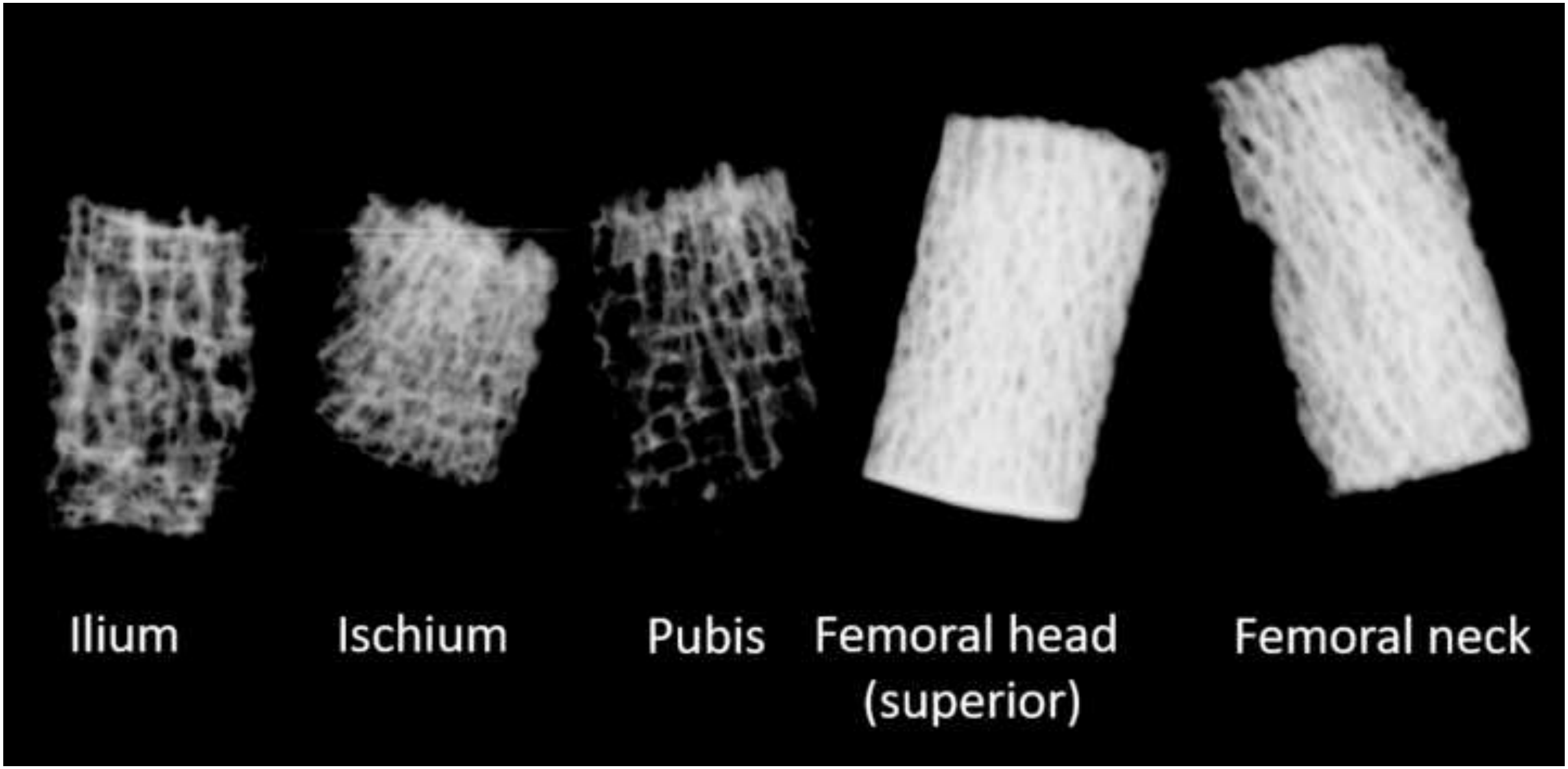
Acetabulum donor D
Ilium Left
Ischium Left
Ilium Right
Ischium Right
Stress / MPa
0.20
0.18
0.16
0.14
0.12
0.10
0.08
0.06
0.04
0.02
0.00
Time / s
0
50
100
150
200
250
300

Figure 4



0.4
0.3
0.2
0.1
0.0
Stress / MPa
0.000
0.005
0.010
0.015
0.020
0.025
0.030
Strain
Stress-strain
Peak Modulus
Yield

Figure 5
Click here to download high resolution image

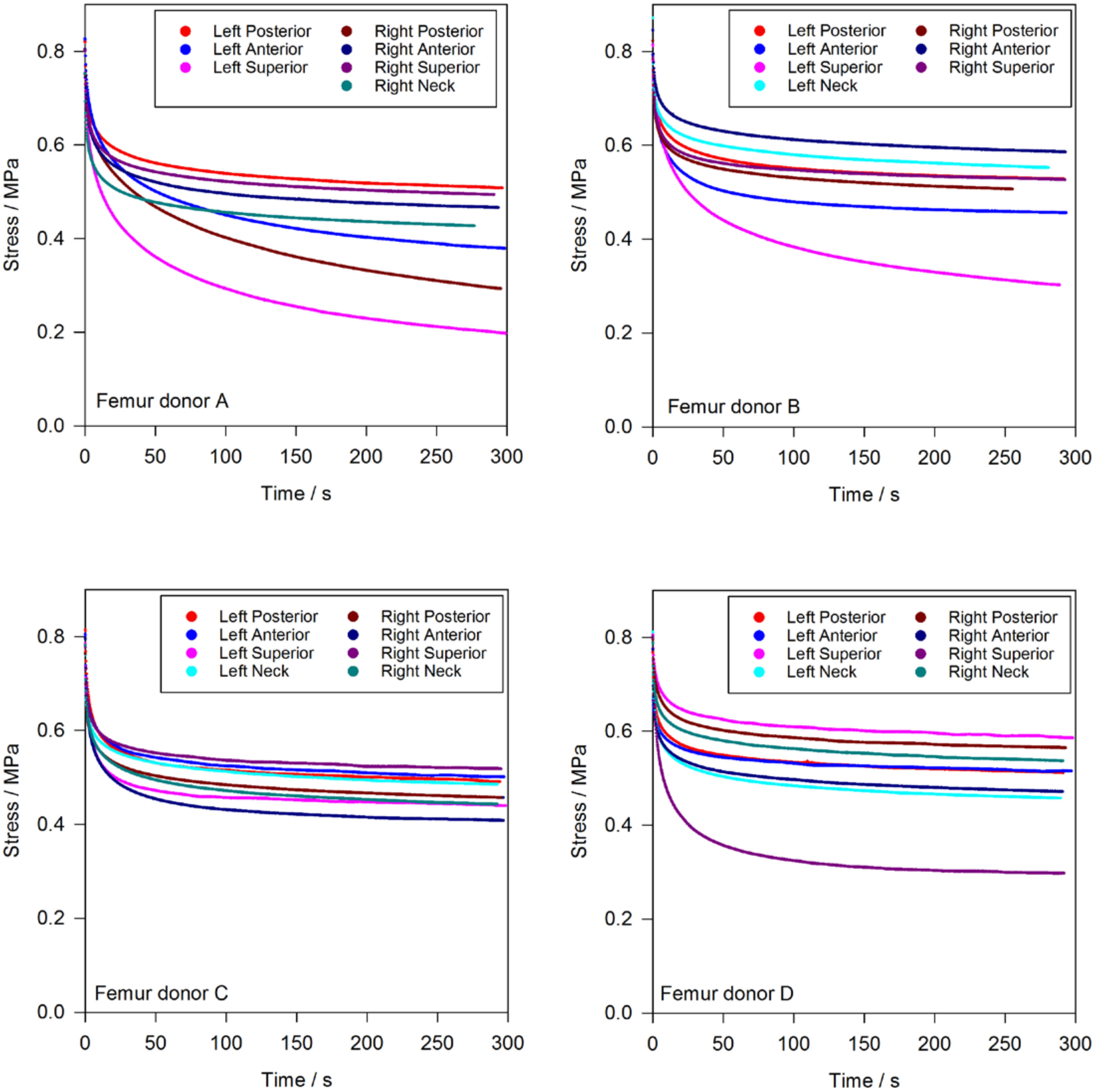


Mechanical properties of cancellous bone from the acetabulum in relation to acetabular shell fixation and compared with the corresponding femoral head

Rianne van Ladesteijn, Holly Leslie, William A Manning, James P Holland, David J Deehan, Thomas Pandorf, Richard M Aspden

Supplementary information

Figure S1. Stress relaxation curves for femoral head samples. Samples were loaded to 50 N and the recording was ended 300 s after the start of loading. Two neck samples broke during testing.


Left Posterior
Left Anterior
Left Superior
Right Posterior
Right Anterior
Right Superior
Right Neck
Femur donor A
Stress / MPa
Time / s
0.0
0.2
0.4
0.6
0.8
0
50
100
150
200
250
300
Left Posterior
Left Anterior
Left Superior
Left Neck
Right Posterior
Right Anterior
Right Superior
Femur donor B
Stress / MPa
Time / s
Left Posterior
Left Anterior
Left Superior
Left Neck
Right Posterior
Right Anterior
Right Superior
Right Neck
Femur donor C
Stress / MPa
Time / s
Left Posterior
Left Anterior
Left Superior
Left Neck
Right Posterior
Right Anterior
Right Superior
Right Neck
Femur donor D
Stress / MPa
Time / s