Current Orthopaedics (2008) **22**, 341—348

available at www.sciencedirect.com

ScienceDirect

journal homepage: www.elsevier.com/locate/cuor

Current ORTHOPAEDICS

BIOMECHANICS

Mechanical testing of intra-articular tissues. Relating experiments to physiological function

Christopher D. Smith [a,b], Spyros Masouros [a,b], Adam M. Hill [a], Andrew L. Wallace [c], Andrew A. Amis [b,c], Anthony M.J. Bull [a,*]

[a] *Department of Bioengineering, Imperial College, London, UK*
[b] *Department of Mechanical Engineering, Imperial College, London, UK*
[c] *Department of Musculoskeletal Surgery, Imperial College, London, UK*

KEYWORDS
Cartilage;
Labrum;
Meniscus;
Ligament;
Biomechanical properties;
Testing

Summary
There is a wealth of data published on the biomechanical properties of intra-articular tissues. However, much of this information is not intuitively applicable to clinical practice due to both methodological disparity between studies and the relevance of the methodology used to test the biomaterial. This inevitably results in comparison difficulties with other experimental data produced for the same tissue, or indeed different tissues.

Therefore, this review highlights the salient issues that need to be considered when trying to interpret biomechanical testing scenarios and how they influence clinical practice. As such, different testing protocols and their clinical relevance are scrutinised. The importance of re-creating the physiological loading conditions and the interpretation of the functional anatomy are highlighted.

Introduction

Clinical application of mechanical testing data of intra-articular biomaterials has been hampered by significant inconsistencies in the selection of testing environment, preconditioning conditions and loading protocols, and the presentation of measured variables. The physiological basis for these testing environments and protocols is often not apparent. Indeed, how mechanical terms relate to clinical scenarios also lacks clarity. Clinical questions of interest may include how tissues of interest function during normal physiological loading, how their biomechanical function translates into native tissue loading, and hence the desirable properties of reconstructed tissues and the loads that they should carry.

At present there are many different testing protocols; this makes it extremely difficult to interpret or compare and contrast results from different research groups. Protocols frequently bear little relation to the in-situ function of the whole tissue, which is the interest of the clinician. The ideal would be to conduct in-situ testing, but this is prone to significant errors due to experimental intricacy and practical or ethical limitations, and is thus rarely practiced. Therefore, intra-articular tissues are

* Corresponding author.
E-mail address: a.bull@imperial.ac.uk (A.M.J. Bull).

0268-0890/$ - see front matter © 2008 Elsevier Ltd. All rights reserved.
doi:10.1016/j.cuor.2008.07.010

primarily tested in-vitro, although it must be remembered that this is outside their normal biological and mechanical environment. In order for the resulting data to be clinically relevant, experimentation must take place in an environment recreating physiological conditions and physiological loading rates.

An example of the huge variation in results obtained can be seen for the tensile elastic modulus of human meniscal tissue, as presented in the literature. This ranges from 100 to 300 MPa for circumferential samples,[1–3] demonstrating how the testing regime and environment can increase the result three-fold.

Functional anatomy

To understand the relevance of specific mechanical measures that are presented for each material, the normal physiological loads placed on intra- and peri-articular soft tissues must first be understood. Biological materials are anisotropic, viscoelastic and inhomogeneous, so different results may be produced for the same tissue. This is dependent on the direction of force application (longitudinally, transversely or radially), the type of force applied (tensile, compressive or shear), the layer within the structure that is tested (articular surface or mid-substance) and the loading history or rate of loading. The most useful data are the biomechanical properties of the tissue in a situation that mimics its normal role within the joint. However, inconsistencies arise due to the difficulty in determining these parameters. Despite these inconsistencies, a number of different methods of approximating some of the biomechanical variables have been presented.

Joint reaction forces can be estimated using equilibrium analyses that use variables such as muscle physiological cross-sectional areas,[4–7] and the measurement of muscle moment arms from cadaveric studies.[8,9] However, this produces a static appreciation and does not account for differing patterns of muscle recruitment. By combining these techniques with EMG studies,[10,11] and predicting muscle forces using mathematical optimisation techniques[12–14] a more dynamic picture can be built. Even with this added information, load rates and the directions in which these forces act on the intra-articular tissues are difficult to predict.

Other methods have been used to measure tensions in soft tissues in-vivo. This usually entails placing a transducer that deflects the path of the load-bearing fibres. This was first described in animal models[15–17] using a pressure transducer placed in a split within the tendon, and limited studies have been performed in humans.[18–21] Direct muscle force measurements have also been obtained within the shoulder musculature.[22]

Mathematical models can assist in calculating muscle and joint forces, but the theoretical ideal is a direct method of measuring joint reaction forces. Pressure transducers placed within joints in-vivo are a possible solution. Obviously this data is sparse, but has been achieved within inter-vertebral discs.[23,24] Data has also been obtained in some prosthetic joint replacements, mainly within the hip[25–31] and more recently from the gleno-humeral joint[32,33] and knee.[34] Transducers within prostheses can measure the forces on joint surfaces during everyday activities. It is therefore possible to know what loads are encountered during specific activities in both magnitude and direction and at what rate these forces load articular tissues. In conjunction with data relating to contact areas and cartilage properties,[35,36] it is possible to estimate the stress on the articular cartilage and the loading rates involved. However, there are problems with data collected in this way; it will only give the gross force applied to the joint surface and is not able to supply information on the distribution of the force along each axis. Also, because the transducer is within a prosthetic implant or inside a tendon, it will undoubtedly change the characteristics of load transmission by the tissue within the joint.

By examining the anatomy of intra-articular tissue, data on the collagen bundle orientation, locations of ligamentous attachments, anchoring mechanisms, blood supply and nerve receptors can be obtained.[37–40] This not only gives information about likely forces passing through the tissue and its associated structures, but may explain other functions such as proprioception[41–43] and the potential for healing after injury. X-ray diffraction and polarised light microscopy can demonstrate the orientation and crimping of collagen fibrils within articular tissue, as well as split lines indicating the orientation of the highest tensile strength.[44–46] The anisotropy of intra-articular tissue has been demonstrated[47–49] and by interpreting these biomechanical results with a knowledge of the collagen microstructure, a picture of the structural behaviour of the tissue can be built.[3,50]

Imaging of the fine structure of tissues can give an indication of the function of the tissue by demonstrating the type, configuration and orientation of the collagen network within the tissue. For example, in addition to conventional histology, SEM studies can demonstrate distinct layers within the substance of a tissue. This has been shown for the glenoid labrum,[51,52] the meniscus,[53] the hip labrum[54] and articular cartilage.[55,56]

At a smaller scale Transmission Electron Microscopy (TEM) allows collagen fibril density and morphology to be examined, which relate to tissue properties.[51,57–60]

By interpreting the gross anatomy, histology and SEM findings, an appreciation of the functional anatomy of the tissue in question can be created prior to experimentation in-vivo.

Relating testing to physiological function

Testing regimes can be used to derive the compressive, tensile and shear properties of a material. It is important to understand the predominant function of the tissue being tested; for example the results of compressive testing of tendons offers little information to the clinician, because tendons usually function in tension. However, tendons that act by wrapping around bone such as the biceps tendon of the shoulder must also be able to resist transverse compression. Thus, it is important to state the specific scenario modeled. More intuitively, compressive testing is useful for compressive load bearing tissue such as articular cartilage, but it must be remembered that during localised compression the surrounding cartilage tissue will be subject to radial and circumferential tension and shearing.

Tensile testing must be conducted along the axis of force anticipated within the in-situ tissue; indeed, many biological

tissues can accommodate significant tensile load along one axis of loading, although failure at lower loads is seen if the optimal axis is not identified.

Shear testing can also be performed in compression or in tension. However, the multidirectional nature of loading and the complex soft tissue deformation induced can be difficult to interpret.

Structural and material properties

The most widely used method to look into the behaviour of tissues under load is the quantification of their structural and/or material properties.

When a tissue is subjected to a load it deforms. Uniaxial testing machines are commonly used to test materials under compressive or tensile loads. The output of a tensile or compressive test from such machines is a load-extension curve. This curve is dependent upon the material itself, but also upon the dimensions of the sample tested. For example, a whole tendon will break under a higher tensile load than the same tendon of half the thickness. Therefore, the load-extension curve provides information about the specific sample tested. This is information on the structural properties of the sample.

It is often useful to normalise for the dimensions of the specimen and acquire the material properties of the tissue. These are independent of the dimensions of the specimen tested and unique for the same tissue. To normalise for the thickness a measurement of stress is used and to normalise for the length a measurement of strain is used. Stress is the force divided by the cross-sectional area of the specimen and strain is the elongation of the specimen divided by its original length. So, a whole tendon will break at the same stress level as a sample of the same tendon which is half the thickness (Fig. 1).

The curve can be divided into several areas. At small loads the tissue behaves non-linearly; this region of non-linearity is referred to as the toe region. It has been shown that resting collagen fibres are crimped when relaxed.[61] In addition, due to the size of the attachment area of a structure such as a ligament, changing joint posture may cause differing patterns of fibre slackening or tightening across the cross-section. Therefore, within the toe region, the collagen fibres are both un-crimped and also recruited in transmitting the applied load.[61]

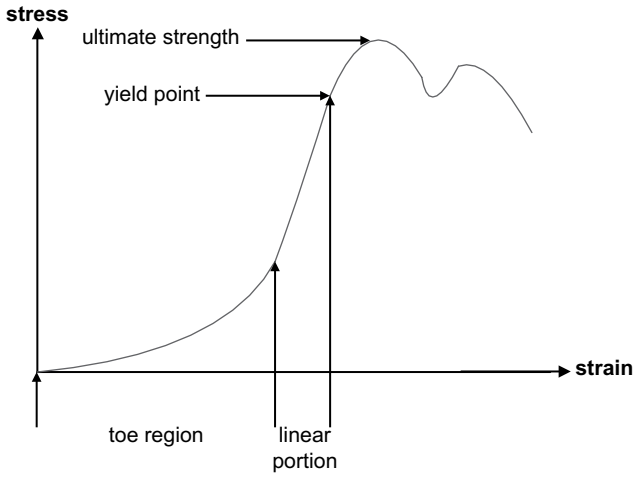

Figure 1 Demonstrates a typical stress-strain curve for a tensile test on an intra-articular tissue specimen.

Once the linear portion is entered, all the fibres are extended equally and there is a direct proportional elongation of all the fibres with respect to the load applied. Failure of some part of the structure starts to occur at the yield point, σ_y, and the linear relationship between stress and strain, and reversible behaviour, are lost. This may result from failure of the load-bearing collagen fibres, or of the substances linking the fibres to each another. However, the sample will continue to withstand further loading until the maximum or ultimate strength of the tissue, σ_u , is reached, after which catastrophic failure occurs.

Several different measures of a specimen's material properties are presented in the literature, commonly ultimate stress, the strain at ultimate stress and the Young's modulus. The Young's (or elastic or tangent) modulus is a measure of stiffness, defined as the slope of the linear portion of a stress-strain curve. However, previous loading conditions and the testing speed can significantly affect the elastic modulus of a viscoelastic material such as intra-articular tissue. This makes comparison between samples tested at different loading rates or a different loading history impossible.

Ultimate stress, or the strain at ultimate stress, reflect pathological loads where the tissue is failing and therefore do not reflect the function of the tissue during physiological activities. It has been suggested that normal daily activities actually take place within the toe-region,[62–64] but as this area is non-linear and ill-defined, standardised data cannot easily be drawn.

Time dependency and loading history

As all intra-articular tissues are viscoelastic, they demonstrate creep and stress relaxation. Creep is produced when a material is kept under a constant load and the dimensions of the material change with time. Undoubtedly all joints undergo creep; the most obvious example would be static standing for a period of time allowing a gradual deformation of the cartilage in any of the weight bearing joints of the lower limb (Fig. 2).

Stress relaxation is shown when a material is held at a constant deformation and gradually requires less force to maintain this configuration as time progresses. Cyclical stress relaxation can also be demonstrated, when the peak load reached decreases at the same position of elongation, as the number of cycles applied increases. Again this will occur during repetitive movements in day-to-day activities.

Both creep and stress relaxation are likely to aid in conditioning the tissues to prolonged force or elongation, reducing the stress in the tissues to avoid fatigue failure. These two mechanisms will occur simultaneously within the joint and will precondition the tissues by altering their performance.

When cyclical loading is performed on a viscoelastic material a hysteresis loop is produced. This represents the difference in the energy needed to load and unload the material and the area of the loop equals the energy dissipated as heat. If cyclical loading is continued, subsequent

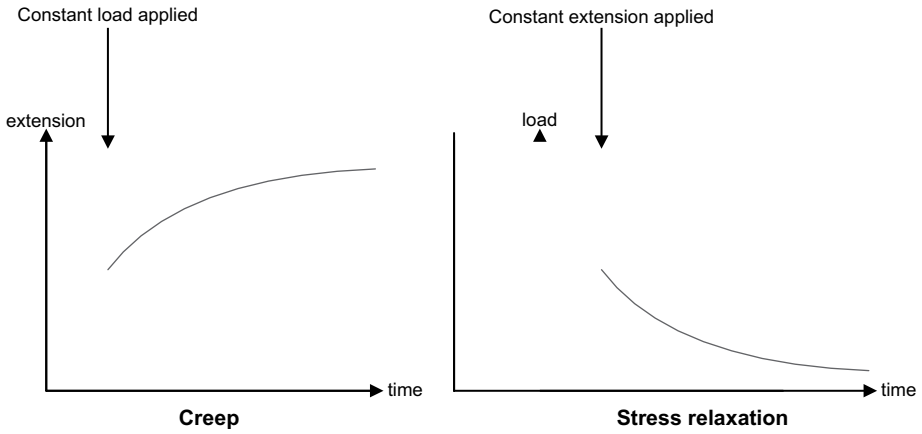

Figure 2 Diagrammatic representation of Creep and Stress relaxation for a viscoelastic material.

hysteresis loops become progressively closer in the amount of extension with load, until a quasi-static state is achieved and the material has been preconditioned.

Pre-cycling demonstrates that viscoelastic tissues have different responses to loading depending on their loading history and it has been shown that pre-cycling can increase the Young's modulus significantly in peri-articular and articular tissue.[65]

The key unknown in this is the amount of preconditioning that is present in the joints prior to loading. Undoubtedly all tissues will be preloaded to some degree by slight alterations in muscle tension even during relative inactivity. It is important to take this into consideration when testing specimens in-vitro, as the effects of deep freezing should be alleviated prior to testing to present the tissue in a more physiological state. The specimen should be brought to a quasi-static state prior to testing by pre-cycling to a low strain, typically below 5%, as strains above this level have been shown to cause permanent structural damage in ligaments (Fig. 3).[66]

It is important to control the testing environment and try to emulate normal physiological conditions. Certainly hydration status,[67,68] temperature,[69] bathing solution,[70] loading rate[68,71] and preloading regime[65,66,72] have all been shown to influence the results obtained. The ideal situation would be a bathing solution of synovial fluid at body temperature.

In-situ testing

Gross testing of intra-articular tissues can be performed 'in-situ' with the tissue in question still being attached to the joint. This can be useful when a specific pathology is in question. For example, pathology of the inferior gleno-humeral ligament has been related to clinical instability. In the literature, this ligament has been tested in the position of apprehension and information gathered regarding its tensile strength, plastic deformity and failure region in different age groups.[73–75] In this situation the ligament was still associated with the 'glenoid-soft-tissue-humerus complex'. This takes into account all the associated soft-tissues around the ligament that give it increased strength. This offers the most direct data to the clinician with regard to a very specific pathology, but provides little information about the biomechanical or material properties of the individual structure.

However, 'in-situ' testing can often test the junction between two tissues, rather than the tissue itself. An example of this is shear testing of the glenoid labrum[76,77] where failure has been noted at the interface between the labrum and the articular cartilage, rather than the mid-substance of the labrum itself.

Compression testing 'in-situ' is more limited, as it will usually involve compression of the tissue and underlying

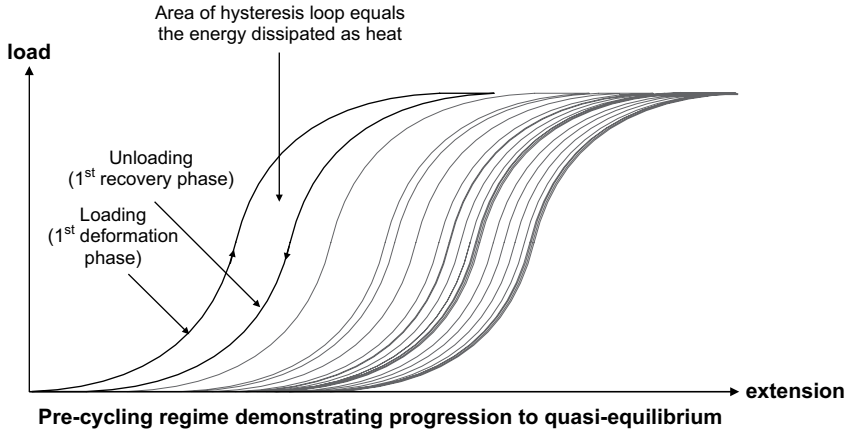

Figure 3 Diagrammatic representation of hysteresis curve for a viscoelastic material during cyclical loading.

bone as a composite. If the complete construct can be maintained then geometry artefacts can be accounted for. The approach most commonly taken is by indentation testing. This has been used 'in-situ' for intra-articular tissues. The result for an individual specimen will vary according to the size, shape and material used to cause the indentation, and at what speed this indentation takes place.

If the instantaneous indentation depth, indenter geometry and undeformed cartilage thickness at the test site are known, then the compressive modulus can be calculated.[78] It has been stated[78,79] that, despite the biphasic nature of intra-articular tissue, no movement of fluid occurs if it is loaded rapidly (20—30 ms); this means that the articular cartilage effectively responds as an elastic material, resulting in a substantially higher apparent modulus when compared to results obtained after 2 seconds of loading, which will include fluid flow and creep.

These types of tests can be conducted in-situ or with prepared specimens. Both approaches have drawbacks: because of non-uniform geometry, with in-situ testing the contact surface may not be flat or perpendicular to the axis of compression. This can lead to difficulties in accounting for the instantaneous area of contact between the indenter and testing surface. With prepared specimens and flat ended indenters there can also be issues relating to edge cutting or stress concentration effects rather than pure compression, so some studies have used spherical indenters.

If tissues are being tested in-situ, then the tests are often limited to loading the whole structure, such as the anterior cruciate ligament within the knee. This does not give information on the material properties, for which variables such as load must be normalised versus the cross-sectional area of the structure to give the stress. Because of this, it is often necessary to isolate smaller specimens from joints, in order to obtain their dimensions accurately.[80] This is essential if one is interested in tissue quality in order, for example, to measure the effect of therapeutic interventions. Also, because intra-articular tissues are not homogeneous, small specimens allow variations to be measured across, through, and along structures.

Specimen geometry

To allow for some control of the variables that can result from in-situ testing, tissues can be removed and made into uniform specimens. It is sometimes convenient to test the tissue as part of a larger specimen. For example, because it is difficult to grasp the fibres of a short, wide ligament such as the anterior cruciate ligament, it is usually tested as a link in a bone-ligament-bone complex. This will remove the problem related to failure at the gripping interface, but accurate assessment of specimen dimensions and deformations can be difficult and usually the outcome measure is defined only as stiffness properties (ultimate load, extension to failure), rather than material properties (Young's modulus).[81] To determine the amount of biomechanical variation within a tissue, the specimen can be separated into distinct fibre bundles for tests in-vitro;[82] else, for tests in-vivo, the location of the instrumentation within the anatomical structure can be defined precisely.[83]

By cutting and preparing specimens, the dimensions can be defined accurately and it can be ensured that only the specific layer of interest is tested. This will also allow multiple specimens to be obtained from a single structure and is useful in comparing results at different positions within the joint. Care must be taken in preparation of the sample for testing, as the natural boundaries of the specimen will have been removed during preparation, making the specimen more susceptible to altered environmental conditions. The functional structure is also important in preparing the specimen as the position and direction of collagen fibres should be considered when cutting the specimens. For tensile testing isotropic specimens should be dumbbell shaped to ensure failure in the mid-portion, but due to the anisotropic nature of the tissues this is often not possible; the collagen fibres may be cut obliquely during preparation and an already small sample will be relying on very few collagen fibres to be tested. Therefore the isthmus of the dumbbell will not consistently contain the same number of fibres and the specimen is likely to fail prematurely. Rectangular specimens can be used, which still allow the Young's modulus to be obtained, but will not reliably demonstrate ultimate stress and strain at failure (Fig. 4).[84]

Compression test specimens are typically tested between two flat plates, allowing their sides to bulge; this is classed as 'unconfined'. The two impermeable plates are larger than the sample used[85] and fluid is allowed to escape freely radially, which will rapidly dry out the specimen and in effect be testing only the solid matrix of the tissue.

Alternatively, the specimen may be fitted into a matching cavity in a well, and a punch/compressor bears down to compress the specimen—'confined' testing. A typical confined specimen will exactly fit the surrounding impermeable well and the compressor will be a semi-permeable flat indenter, which has a permeability of several orders of magnitude higher than the tissue being tested and is also an exact fit to the well.[86,87] In this situation, during compression fluid can only escape through the permeable indenter and the speed of release will be defined by the permeability of the material used. It can be presumed that unless the indenter has an identical permeability to the tissue being tested then the results will be altered by this change in boundary characteristics.

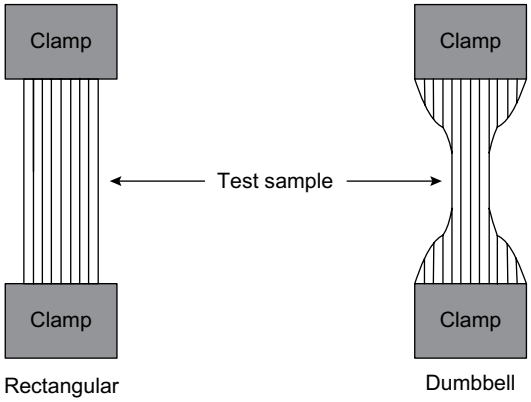

Figure 4 Examples of rectangular and dumbbell tensile testing samples.

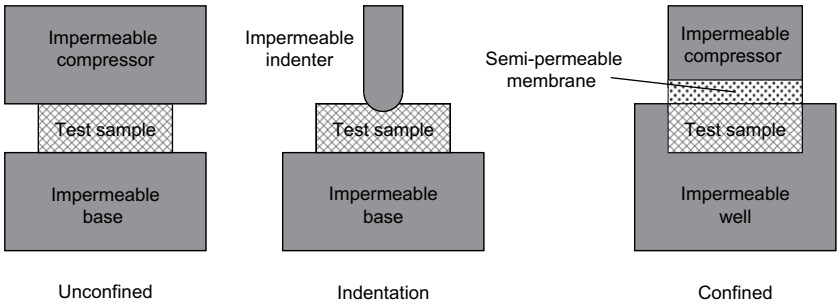

Figure 5 Examples of unconfined, indentation and confined compressive testing.

A third testing method is indentation testing, whereby a compressor/indenter is brought down on the specimen, which is placed on an impermeable plate. This type of testing can also be used in-vivo (Fig. 5).

Therefore, different moduli may be obtained from the different types of compression testing that includes different rates of deformation. How these are related to the in-situ, in-vivo situation is unknown.

It may be appropriate to divide different testing regimes into 'physiological' cyclical loading and acute loading. Physiological cyclical loading will keep the loading conditions well within the normal physiological limits found within everyday activities such as walking and other repetitive movements. Acute 'one-shot' loading endeavours to load the test samples rapidly, in order to recreate pathological conditions that will lead to failure of the tissue and would normally result in injury, giving data such as ultimate strength. These two types of testing are targeted at defining different properties of the tissue and will give very different data describing the properties of the same tissue due to the load rate and level imposed on the tissue.

Conclusions

There is no doubt that obtaining reliable and useful data for the mechanical properties of intra-articular tissues is difficult. To obtain clinically applicable data from in-vitro testing, the environmental conditions should be optimised to mimic the normal physiological conditions. Preconditioning to a quasi-static state should take place in the tissues in order to mimic the normal physiological conditions and to alleviate any detrimental effects of deep-freezing the tissues. Furthermore, the functional anatomy should be considered and the direction of loading should be appropriate to the known function and microstructure of the tissue.

During compressive and tensile testing the normal physiological rate of loading should be replicated, as the speed of testing is crucial and will affect the results considerably. With compressive testing, the confinement of the tissue will influence results and may result in only one phase of the tissue being tested.

Reference

1. Tissakht M, Ahmed AM. Tensile stress-strain characteristics of the human meniscal material. *J Biomech* 1995 Apr;**28**(4):411—22.
2. Lechner K, Hull ML, Howell SM. Is the circumferential tensile modulus within a human medial meniscus affected by the test sample location and cross-sectional area? *J Orthop Res* 2000 Nov;**18**(6):945—51.
3. Fithian DC, Kelly MA, Mow VC. Material properties and structure-function relationships in the menisci. *Clin Orthop Relat Res* 1990 Mar;(252):19—31.
4. Keating JF, Waterworth P, Shaw D, et al. The relative strengths of the rotator cuff muscles. A cadaver study. *J Bone Joint Surg Br* 1993 Jan;**75**(1):137—40.
5. Juul—Kristensen B, Bojsen—Moller F, Finsen L, et al. Muscle sizes and moment arms of rotator cuff muscles determined by magnetic resonance imaging. *Cells Tissues Organs* 2000;**167**(2—3):214—22.
6. Toshiaki A, Itoi E, Minagawa H, et al. Cross-sectional area of the tendon and the muscle of the biceps brachii in shoulders with rotator cuff tears: a study of 14 cadaveric shoulders. *Acta Orthop* 2005 Aug;**76**(4):509—12.
7. Amis AA, Dowson D, Wright V. Elbow joint force predictions for some strenuous isometric actions. *J Biomech* 1980;**13**(9):765—75.
8. Hughes RE, An KN. Force analysis of rotator cuff muscles. *Clin Orthop Relat Res* 1996 Sep;(330):75—83.
9. Amis AA, Dowson D, Wright V. Muscle strengths and musculoskeletal geometry of the upper limb. *Journal of Engineering in Medicine* 1979;**part H**(8):41—8.
10. Chang YW, Hughes RE, Su F, et al. Prediction of muscle force involved in shoulder internal rotation. *J Shoulder Elbow Surg* 2000;**9**(3):188—95.
11. Elias JJ, Bratton DR, Weinstein DM, et al. Comparing two estimations of the quadriceps force distribution for use during patellofemoral simulation. *J Biomech* 2006;**39**(5):865—72.
12. Heller MO, Bergmann G, Deuretzbacher G, et al. Musculoskeletal loading conditions at the hip during walking and stair climbing. *J Biomech* 2001 Jul;**34**(7):883—93.
13. Fisher IA. *A mathematical investigation of the influence of skeletal geometry on the mechanics of a prosthetic human hip joint*. London: Imperial College; 2000.
14. Duda GN, Schneider E, Chao EY. Internal forces and moments in the femur during walking. *J Biomech* 1997 Sep;**30**(9):933—41.
15. Xu WS, Butler DL, Stouffer DC, et al. Theoretical analysis of an implantable force transducer for tendon and ligament structures. *J Biomech Eng* 1992 May;**114**(2):170—7.
16. Glos DL, Butler DL, Grood ES, et al. In vitro evaluation of an implantable force transducer (IFT) in a patellar tendon model. *J Biomech Eng* 1993 Nov;**115**(4A):335—43.
17. Holden JP, Grood ES, Korvick DL, et al. In vivo forces in the anterior cruciate ligament: direct measurements during walking and trotting in a quadruped. *J Biomech* 1994 May;**27**(5):517—26.
18. Fukashiro S, Komi PV, Jarvinen M, et al. In vivo Achilles tendon loading during jumping in humans. *Eur J Appl Physiol Occup Physiol* 1995;**71**(5):453—8.

19. Ballard RE, Watenpaugh DE, Breit GA, et al. Leg intramuscular pressures during locomotion in humans. *J Appl Physiol* 1998 Jun;**84**(6):1976—81.
20. Dennerlein JT, Diao E, Mote Jr CD, et al. Tensions of the flexor digitorum superficialis are higher than a current model predicts. *J Biomech* 1998 Apr;**31**(4):295—301.
21. Beynnon B, Howe JG, Pope MH, et al. The measurement of anterior cruciate ligament strain in vivo. *Int Orthop* 1992; **16**(1):1—12.
22. Bull AMJ, Reilly P, Wallace AL, et al. A novel technique to measure active tendon forces: application to the subscapularis tendon. *Knee Surg Sports Traumatol Arthrosc* 2005 Mar;**13**(2): 145—50.
23. Lisi AJ, O'Neill CW, Lindsey DP, et al. Measurement of in vivo lumbar intervertebral disc pressure during spinal manipulation: a feasibility study. *J Appl Biomech* 2006 Aug;**22**(3):234—9.
24. Wilke H, Neef P, Hinz B, et al. Intradiscal pressure together with anthropometric data-a data set for the validation of models. *Clin Biomech (Bristol, Avon)* 2001;**16**(Suppl.1): S111—26.
25. Park S, Krebs DE, Mann RW. Hip muscle co-contraction: evidence from concurrent in vivo pressure measurement and force estimation. *Gait Posture* 1999 Dec;**10**(3):211—22.
26. McGibbon CA, Krebs DE, Mann RW. In vivo hip pressures during cane and load-carrying gait. *Arthritis Care Res* 1997 Oct;**10**(5): 300—7.
27. Bergmann G, Siraky J, Rohlmann A, et al. A comparison of hip joint forces in sheep, dog and man. *J Biomech* 1984;**17**(12): 907—21.
28. Bergmann G, Deuretzbacher G, Heller M, et al. Hip contact forces and gait patterns from routine activities. *J Biomech* 2001 Jul;**34**(7):859—71.
29. Davy DT, Kotzar GM, Brown RH, et al. Telemetric force measurements across the hip after total arthroplasty. *J Bone Joint Surg Am* 1988 Jan;**70**(1):45—50.
30. Hodge WA, Carlson KL, Fijan RS, et al. Contact pressures from an instrumented hip endoprosthesis. *J Bone Joint Surg Am* 1989 Oct;**71**(9):1378—86.
31. English TA, Kilvington M. In vivo records of hip loads using a femoral implant with telemetric output (a preliminary report). *J Biomed Eng* 1979 Apr;**1**(2):111—5.
32. Bergmann G, Graichen F, Bender A, et al. First in vivo measurements of contact forces in the shoulder joint. *J Biomech* 2006 Jan 8;**39**(Suppl.1):S134.
33. Bergmann G, Graichen F, Bender A, et al. In vivo glenohumeral contact forces-Measurements in the first patient 7 months postoperatively. *J Biomech* 2006 Dec 12;**39**(Suppl. 1):S134.
34. Zhao D, Banks SA, D'Lima DD, et al. In vivo medial and lateral tibial loads during dynamic and high flexion activities. *J Orthop Res* 2007 May;**25**(5):593—602.
35. Iglic A, Kralj—Iglic V, Daniel M, et al. Computer determination of contact stress distribution and size of weight bearing area in the human hip joint. *Comput Methods Biomech Biomed Engin* 2002 Apr;**5**(2):185—92.
36. Daniel M, Antolic V, Iglic A, et al. Determination of contact hip stress from nomograms based on mathematical model. *Med Eng Phys* 2001 Jun;**23**(5):347—57.
37. Cooper DE, Arnoczky SP, O'Brien SJ, et al. Anatomy, histology, and vascularity of the glenoid labrum. An anatomical study. *J Bone Joint Surg Am* 1992 Jan;**74**(1):46—52.
38. Putz R, Schrank C. Anatomy of the labro—capsular complex. *Orthopade* 1998 Oct;**27**(10):675—80.
39. Seldes RM, Tan V, Hunt J, et al. Anatomy, histologic features, and vascularity of the adult acetabular labrum. *Clin Orthop Relat Res* 2001 Jan;(382):232—40.
40. Edwards A, Bull AM, Amis AA. The attachments of the fiber bundles of the posterior cruciate ligament: an anatomic study. *Arthroscopy* 2007 Mar;**23**(3):284—90.
41. Kim YT, Azuma H. The nerve endings of the acetabular labrum. *Clin Orthop Relat Res* 1995 Nov;(320):176—81.
42. Vangsness Jr CT, Ennis M, Taylor JG, et al. Neural anatomy of the glenohumeral ligaments, labrum, and subacromial bursa. *Arthroscopy* 1995 Apr;**11**(2):180—4.
43. Mine T, Kimura M, Sakka A, et al. Innervation of nociceptors in the menisci of the knee joint: an immunohistochemical study. *Arch Orthop Trauma Surg* 2000;**120**(3—4):201—4.
44. Aspden RM, Yarker YE, Hukins DW. Collagen orientations in the meniscus of the knee joint. *J Anat* 1985 May;**140**(Pt 3): 371—80.
45. Muehleman C, Majumdar S, Issever AS, et al. X-ray detection of structural orientation in human articular cartilage. *Osteoarthritis Cartilage* 2004 Feb;**12**(2):97—105.
46. Kempson GE, Freeman MA, Swanson SA. Tensile properties of articular cartilage. *Nature* 1968 Dec 14;**220**(5172):1127—8.
47. Leslie BW, Gardner DL, McGeough JA, et al. Anisotropic response of the human knee joint meniscus to unconfined compression. *Proc Inst Mech Eng [H]* 2000;**214**(6):631—5.
48. Huang CY, Stankiewicz A, Ateshian GA, et al. Anisotropy, inhomogeneity, and tension—compression nonlinearity of human glenohumeral cartilage in finite deformation. *J Biomech* 2005 Apr;**38**(4):799—809.
49. Jurvelin JS, Buschmann MD, Hunziker EB. Mechanical anisotropy of the human knee articular cartilage in compression. *Proc Inst Mech Eng [H]* 2003;**217**(3):215—9.
50. Gabrion A, Aimedieu P, Laya Z, et al. Relationship between ultrastructure and biomechanical properties of the knee meniscus. *Surg Radiol Anat* 2005 Dec;**27**(6):507—10.
51. Hoerning EK, Brook KJ, Hill AM, et al. Microstructure of the glenoid labrum: A 3-D histomorphometric study. *J Bone Joint Surg Br* 2005 Apr 1;**87**—**B**(Suppl I):162c.
52. Nishida K, Hashizume H, Toda K, et al. Histologic and scanning electron microscopic study of the glenoid labrum. *J Shoulder Elbow Surg* 1996;**5**(2Pt 1):132—8.
53. Petersen W, Tillmann B. Collagenous fibril texture of the human knee joint menisci. *Anat Embryol (Berl)* 1998 Apr; **197**(4):317—24.
54. Petersen W, Petersen F, Tillmann B. Structure and vascularization of the acetabular labrum with regard to the pathogenesis and healing of labral lesions. *Arch Orthop Trauma Surg* 2003 Jul;**123**(6):283—8.
55. Chai BF, Tang XM. Ultrastructural study of hip joint osteoarthritis. Scanning electron microscopic study of articular cartilage of femoral head. *Chin Med J (Engl)* 1990 Apr;**103**(4): 290—4.
56. Hirotani H, Ito T. Scanning electron microscopy of the articular surfaces of the hip joint disorders. *Anat Anz* 1975;**138**(1—2): 29—38.
57. Merkel KH. The surface of human menisci and its aging alterations during age. A combined scanning and transmission electron microscopic examination (SEM, TEM). *Arch Orthop Trauma Surg* 1980;**97**(3):185—91.
58. Montella A, Manunta A, Espa E, et al. Human articular cartilage in osteoarthrosis. I. The matrix. Transmission electron microscopic study. *Ital J Anat Embryol* 1992 Jan;**97**(1):1—12.
59. Neurath M, Stofft E. New aspects of the functional anatomy of the menisci. *Unfallchirurg* 1992 Jan;**95**(1):17—20.
60. Kapitonova MY, Mansor O. Ultrastructural changes of the articular cartilage in some arthropathies with special reference to chondrocyte cell death. *Malays J Pathol* 2003 Jun; **25**(1):15—27.
61. Viidik A. Simultaneous mechanical and light microscopic studies of collagen fibers. *Z Anat Entwicklungsgesch* 1972; **136**(2):204—12.
62. Noyes FR, Butler DL, Grood ES, et al. Biomechanical analysis of human ligament grafts used in knee-ligament repairs and reconstructions. *J Bone Joint Surg Am* 1984 Mar;**66**(3):344—52.

63. Cerulli G, Benoit DL, Lamontagne M, et al. In vivo anterior cruciate ligament strain behaviour during a rapid deceleration movement: case report. *Knee Surg Sports Traumatol Arthrosc* 2003 Sep;**11**(5):307–11.
64. Beynnon BD, Fleming BC. Anterior cruciate ligament strain in-vivo: a review of previous work. *J Biomech* 1998 Jun;**31**(6): 519–25.
65. Schatzmann L, Brunner P, Staubli HU. Effect of cyclic preconditioning on the tensile properties of human quadriceps tendons and patellar ligaments. *Knee Surg Sports Traumatol Arthrosc* 1998;**6**(Suppl. 1):S56–61.
66. Provenzano PP, Heisey D, Hayashi K, et al. Subfailure damage in ligament: a structural and cellular evaluation. *J Appl Physiol* 2002 Jan;**92**(1):362–71.
67. Haut TL, Haut RC. The state of tissue hydration determines the strain-rate-sensitive stiffness of human patellar tendon. *J Biomech* 1997 Jan;**30**(1):79–81.
68. Race A, Broom ND, Robertson P. Effect of loading rate and hydration on the mechanical properties of the disc. *Spine* 2000 Mar 15;**25**(6):662–9.
69. Chae Y, Aguilar G, Lavernia EJ, et al. Characterization of temperature dependent mechanical behavior of cartilage. *Lasers Surg Med* 2003;**32**(4):271–8.
70. Chimich D, Shrive N, Frank C, et al. Water content alters viscoelastic behaviour of the normal adolescent rabbit medial collateral ligament. *J Biomech* 1992 Aug;**25**(8):831–7.
71. Pioletti DP, Rakotomanana LR, Leyvraz PF. Strain rate effect on the mechanical behavior of the anterior cruciate ligament-bone complex. *Med Eng Phys* 1999 Mar;**21**(2):95–100.
72. Wren TA, Lindsey DP, Beaupre GS, et al. Effects of creep and cyclic loading on the mechanical properties and failure of human Achilles tendons. *Ann Biomed Eng* 2003 Jun;**31**(6):710–7.
73. McMahon PJ, Dettling JR, Sandusky MD, et al. Deformation and strain characteristics along the length of the anterior band of the inferior glenohumeral ligament. *J Shoulder Elbow Surg* 2001 Sep;**10**(5):482–8.
74. McMahon PJ, Dettling J, Sandusky MD, et al. The anterior band of the inferior glenohumeral ligament. Assessment of its permanent deformation and the anatomy of its glenoid attachment. *J Bone Joint Surg Br* 1999 May;**81**(3):406–13.
75. Lee TQ, Dettling J, Sandusky MD, et al. Age related biomechanical properties of the glenoid-anterior band of the inferior glenohumeral ligament-humerus complex. *Clin Biomech (Bristol, Avon)* 1999 Aug;**14**(7):471–6.
76. Hara H, Ito N, Iwasaki K. Strength of the glenoid labrum and adjacent shoulder capsule. *J Shoulder Elbow Surg* 1996;**5**(4): 263–8.
77. Reeves B. Experiments on the tensile strength of the anterior capsular structures of the shoulder in man. *J Bone Joint Surg Br* 1968;**50**(B):858–68.
78. Shepherd DE, Seedhom BB. A technique for measuring the compressive modulus of articular cartilage under physiological loading rates with preliminary results. *Proc Inst Mech Eng [H]* 1997;**211**(2):155–65.
79. Shepherd DE, Seedhom BB. The 'instantaneous' compressive modulus of human articular cartilage in joints of the lower limb. *Rheumatology (Oxford)* 1999 Feb;**38**(2):124–32.
80. Robinson JR, Bull AM, Amis AA. Structural properties of the medial collateral ligament complex of the human knee. *J Biomech* 2005 May;**38**(5):1067–74.
81. Kennedy JC, Hawkins RJ, Willis RB, et al. Tension studies of human knee ligaments. Yield point, ultimate failure, and disruption of the cruciate and tibial collateral ligaments. *J Bone Joint Surg Am* 1976 Apr;**58**(3):350–5.
82. Butler DL, Guan Y, Kay MD, et al. Location-dependent variations in the material properties of the anterior cruciate ligament. *J Biomech* 1992 May;**25**(5):511–8.
83. Gupte CM, Smith A, Jamieson N, et al. Meniscofemoral ligaments-structural and material properties. *J Biomech* 2002 Dec; **35**(12):1623–9.
84. Ozkaya N, Nordin M. Chapter 7 stress and strain. In: Leger DL, editor. *Fundamentals of biomechanics: equilibrium, motion and deformation*. 2nd edn. Springer Science; 1999. p. 125–52.
85. Park S, Hung CT, Ateshian GA. Mechanical response of bovine articular cartilage under dynamic unconfined compression loading at physiological stress levels. *Osteoarthritis Cartilage* 2004 Jan;**12**(1):65–73.
86. Ferguson SJ, Bryant JT, Ito K. The material properties of the bovine acetabular labrum. *J Orthop Res* 2001 Sep;**19**(5): 887–96.
87. Armstrong CG, Mow VC. Variations in the intrinsic mechanical properties of human articular cartilage with age, degeneration, and water content. *J Bone Joint Surg Am* 1982 Jan;**64**(1): 88–94.