Biomechanics of the menisci of the knee

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Summary
The menisci of the knee are complex structures with various important functions within the knee. Loss of the menisci leads to a significantly increased risk of developing degenerative changes in the long term. To fully understand the role of the menisci it is necessary to have an appropriate understanding of their anatomy, microstructure, material/mechanical properties and biomechanical function. This review will give a brief outline of both the underlying principles involved as well as giving some detail explaining the behaviour in-vivo of these complex and important structures.

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Introduction
The menisci are two crescent-shaped cartilages that are found within each knee between the femoral condyles and the tibial plateaux. For many years the menisci were considered to be the functionless remains of a leg muscle. Indeed, in his paper in 1942 McMurray stated that "When the knee-joint is opened on the anterior aspect, and the suspected cartilage appears normal, its removal can be undertaken with confidence if the diagnosis of a posterior tear has been arrived at (clinically) prior to operation. A far too common error is shown in the incomplete removal of the injured meniscus."

Attitudes towards the menisci have changed dramatically, and since King’s pivotal paper in 1936, numerous studies have shown that the menisci do in fact play important roles in load bearing and shock absorption within the knee, and that they are also secondary stabilisers of the joint. Further roles in joint lubrication and...
nutrient distribution, and sensory function and proprioception have been proposed. Meniscal tears are probably the most common intra-articular injury of the knee, and it is now fully appreciated that meniscectomy leads to a large increase in the risk of subsequently developing degenerative changes within the joint.

The responses of the menisci to loads applied to the tibiofemoral joint result from their macro-geometry, their fine structure and the nature of their insertional ligaments. The menisci are fibrocartilaginous structures primarily composed of an interlacing network of collagen fibres (predominantly type I collagen) interposed with cells, with an extracellular matrix of proteoglycans and glycoproteins.

The collagen composition varies between the surface layer of the menisci and the deeper tissue. The collagen bundles of the surface layer are randomly orientated with a compositional similarity to articular hyaline cartilage. This relates to its function, which is to allow low-friction motion between the meniscal tissue and the femur and tibia as the femoral condyles slide against the meniscus and move it over the tibial plateau during knee motion.

Within the bulk of the meniscal tissue, beneath the surface layers, there are two structurally distinct regions where the orientation of collagen fibres is different: in the innermost third the collagen bundles predominantly lie in a radial pattern, whereas in the outer two-thirds the collagen bundles are orientated circumferentially (Fig. 1). This suggests that the inner third may function in compression and that the outer two-thirds function in tension. Further, less-frequent, radially-orientated collagen fibres can also be found within the bulk of the meniscal tissue and these may act as tie fibres, resisting longitudinal splitting of the circumferential collagen bundles.

To fully understand the functioning of the menisci of the knee, it is first essential to have a firm grasp of the basic underlying principles of biomechanics as they relate to meniscal tissue. These principles will be outlined and then expanded upon in the following sections.

**The basic biomechanics of tension**

When a force is applied to any material or tissue that is not completely free to move, there will be a resultant deformation. The behaviour of tissue as a stretching force is applied is referred to as its tensile properties.

As a stretching force is applied to a sample of tissue, such as meniscal tissue, the sample will show a resultant elongation. Equally, as a tissue is stretched, a force will develop within the tissue, opposing its elongation. Tissue extension can be plotted graphically against the measured resultant force, to give a load-elongation curve (Fig. 2).

The first region of the load-elongation curve is referred to as the 'toe region'. Little force is required to elongate the tissue initially, and the elongation at this stage occurs as the wavy pattern of relaxed collagen fibres within the meniscal tissue becomes straighter.

The second region of the curve is linear, and during this stage the collagen fibres themselves become stretched and are parallel, having lost their wavy appearance. During this phase, there is a linear relationship between elongation and load.

Towards the end of the linear phase of the curve, small dips representing force reductions can be seen. These are caused by the early sequential failure of some individual fibre bundles. Eventually, major failure of fibre bundles occurs and complete failure ensues. The point at which this begins to occur is referred to as the yield point for the tissue, and represents a change from elastic (reversible) to plastic (irreversible) deformation. The maximum load attained is referred to as the ultimate tensile load of the specimen.

The stiffness of the testing sample can be defined by

\[ k = \frac{\Delta F}{\Delta L} \]

and can be determined by calculating the gradient of the linear portion of the graph. The stiffness of the sample is a *structural property* of the specific tissue sample tested. This means that the stiffness value is dependent upon the material properties of the tissue itself and the dimensions of the sample. In order to make comparisons between different tissues or materials, measures which are independent of sample dimensions are used; these are termed *material properties*.

To adjust for specimen cross-sectional area, a measurement of stress is used. This is defined as

\[ \sigma = \frac{F}{A} \quad \text{(MPa=N/mm}^2\text{)} \]

![Figure 1](image1.png) Diagrammatic representation of the collagen fibre orientations within the meniscus. Taken from Bullough et al.

![Figure 2](image2.png) Typical Load versus Elongation curve for tensile testing of fibrous soft-tissue samples.
To adjust for specimen length, a measurement of strain is used. This is defined as

\[ \text{strain} = \frac{\text{elongation}}{\text{original length}} = \frac{\Delta L}{L} \]

The original length of the test sample is referred to as the gauge length. Strain is often multiplied by 100 to give percent strain, and has no units.

Using the definition of stress and strain given above, a new ‘stress-strain’ curve can be plotted. This has the same shape as the load-elongation curve. Calculating the gradient of the linear portion of this curve by dividing the change in stress by the change in strain, now gives a value referred to as the Young’s modulus of the material:

\[ \text{Young's modulus} = \frac{\text{stress}}{\text{strain}} = \frac{\sigma}{\varepsilon} \]

The modulus is a material property and can be used as a measure of comparison of the material stiffnesses of various tissues of different size and shape.

### The basic biomechanics of compression

In 1980, Mow et al. investigated the behaviour of articular cartilage and showed that it could be described as a biphasic material; composed of a solid matrix phase and an interstitial fluid phase. Favenesi et al., in 1983, examined the behaviour of bovine menisci and concluded that meniscal tissue also followed the expected behaviour of a biphasic material. They demonstrated that the water content of the meniscus was freely exchangeable with the surrounding fluid by mechanical means. The meniscal water content could be extruded either by compression or by direct application of a pressure differential. Favenesi et al. showed, however, that the meniscal tissue was only half as stiff, and one sixth as permeable as articular cartilage.

Spilker et al. used computer finite element analysis to model meniscal properties, and found that the biphasic representation, whereby the tissue was described as a mixture of solid and fluid components, was crucial, and that the fluid phase carried a significant part of the applied load.

Thus, when a load is applied to meniscal tissue the solid phase, consisting predominantly of circumferentially orientated collagen bundles, shows an elastic response. However, simultaneously, the fluid pressure carries the compressive load while the fluid component of the tissue is very slowly extruded at a rate dependent on the permeability of the tissue and the viscosity of the fluid. The tissue therefore exhibits viscoelastic behaviour.

If a constant compressive load is applied to a sample of meniscal tissue, there is an initial compression that is near-linear, and which predominantly represents the solid phase of the material, i.e. the elastic behaviour of the collagen bundles and the matrix material. This immediate phase is followed by a curved phase where there is a continued compression of the tissue, but at a diminishing rate. This second stage is where the fluid-phase predominates, and fluid is moving within and being expelled from the meniscal tissue. The continued but diminishing compression is referred to as creep, and can be seen when extension is plotted against time (Fig. 3).

![Figure 3](image1.png)

**Figure 3** Plot of Compression against time for a constant compressive load. An initial near-linear portion is seen (A), followed by a curved section of diminishing rate of compression (B), representing creep of the tissue as water is extruded.

Similarly, when meniscal tissue is compressed and then held, and the load required to maintain that compression is measured, it can be seen that the load diminishes with time. If load is plotted against time, then a characteristic curve is seen. As fluid is expelled from the meniscal tissue, the tissue relaxes and the required load to maintain any given compression decreases with time. The resulting curve is known as a stress relaxation curve (Fig. 4).

Creep and stress-relaxation are related characteristics of viscoelastic behaviour; this is a time dependent behaviour. The water content and permeability of meniscal tissue are responsible for the viscoelastic response of the tissue. It is important to note, however, that the permeability of meniscal tissue is very low when compared to articular cartilage, so that the menisci effectively maintain their volume under load, otherwise they would soon become functionless within the knee.

### The material properties of meniscal tissue

#### Tensile material properties

As far back as 1949, Mathur et al. took fresh meniscal tissue from amputated limbs, held the tissue in metal clamps,

![Figure 4](image2.png)

**Figure 4** Stress-relaxation curve, showing the behaviour of meniscal tissue when it is compressed to a set value of displacement.
suspended the tissue within a three foot high frame, and then hung weights on the samples, measuring the 'stretching' of the tissue and the load necessary to cause the tissue to 'fracture'.

However, in order to measure the material properties of meniscal tissue, samples of clearly-defined geometry are necessary. The standard shape of tissue sample used by most investigators is based on the basic materials testing dumbbell (Fig. 5). This has expanded sections at the ends for clamping the tissue, and a narrower gauge length in between. The cross-sectional area in this section is uniform, thus allowing calculation of stress. However, although this is appropriate for the testing of non-fibrous materials, dumbbell shaped test samples should be used with care for fibre-reinforced composites because they tend to split in the region where the width changes.

Whipple et al. studied the tensile properties of bovine menisci using dumbbell-shaped samples, 400 micrometers in thickness. They compared the properties of the tissue from samples taken either circumferentially or radially to the circumferentially-orientated collagen fibres (Fig. 6) and demonstrated that in the surface layer of the menisci, where the collagen fibres are randomly orientated, there were no differences between samples taken in different orientations. However, in the deeper, main bulk of the meniscal tissue, samples taken circumferentially had a much greater modulus (198.4 MPa) compared to radially-orientated samples (2.8 MPa). Similar findings have been observed by other authors.

This directional variation of material properties within the tissue is referred to as anisotropy, and the tissue is described as anisotropic.

Whipple et al. also observed differences in the material properties of circumferentially sampled meniscal tissue depending on the depth of the samples: tissue from the surface layer had a modulus of 48.3 MPa compared to a value of 198.4 MPa for the middle zone and 139.0 MPa for the deepest layer sampled.

Other investigators have demonstrated regional variations in the material properties of meniscal tissue. Fithian et al. found that the tensile modulus was significantly reduced in the posterior two-thirds of the medial meniscus, compared to the anterior third of the medial meniscus or the entire lateral meniscus. They inspected the meniscal tissue under polarised light to determine collagen fibre orientation, and found that whereas in the rest of meniscal tissue the fibres were highly orientated in the circumferential direction, in the posterior two-thirds of the medial meniscus the fibre bundles were orientated obliquely with respect to each other. Fithian et al. felt that the capsular attachments in the posterior two-thirds of the medial meniscus may interrupt the circumferential fibre organisation, and stated that this might explain in part the frequency of tears observed in the posterior horn of the medial meniscus.

In a more recent and highly comprehensive study, Lechner et al. observed that the posterior and middle regions of human medial menisci had a higher tensile modulus than the anterior region. They also noted that as the circumferential collagen fibres are continuous through the meniscus from the anterior to the posterior tibial attachments, and the anterior horn of the medial meniscus is not as wide in the radial direction as the central or posterior regions, the same number of collagen fibres must be packed into a smaller cross-sectional area of tissue anteriorly. This would increase the ratio of collagen fibres to matrix tissue anteriorly, and might explain the increased stiffness of the tissue anteriorly.

Lechner et al. also showed that the modulus of meniscal tissue varies inversely with test specimen thickness. They noted that standard materials testing assumes that the tissue being tested is actually homogenous. Meniscal tissue, however, is composed of collagen fibre bundles, ranging from 50 to 400 micrometers in diameter, embedded within weak matrix tissue (Fig. 7). Thus, thinner samples may consist predominantly of matrix tissue. Lechner et al. found that 28% of 0.5 mm thickness samples either failed prematurely, or could not even be tested. Only 17% of thicker, 3 mm samples were untestable. Thus with thinner samples, the samples that could be tested were likely to be those that contained a significant amount of collagen fibre tissue, as opposed to just mainly weak matrix. Hence the thinner samples that could be tested had an apparent higher mean modulus than the thicker samples.

The above issues emphasize how complicated it is to try to define the biomechanical properties of meniscal tissue. Thus, attempts to try to quantify absolute values for
material properties probably have little meaning unless taken within the context of the overall macrostructure of this highly complex organ.

However, for general comparison, if a value in the region of 150 MPa is taken as an approximation of the tensile modulus of meniscal tissue, Young’s modulus for the anterior cruciate ligament is approximately 200\,\text{e}^{300} \text{MPa}, and the modulus of high-density polyethylene is approximately 1000 \text{MPa}. The ultimate stress to failure of meniscal tissue is in the region of 20 MPa. The ultimate stress of ligament tissue is typically 50 MPa, and of mild steel is approximately 250 MPa.

Compressive material properties

Favenesi et al. examined the compressive material properties of bovine menisci using confined compression creep testing and direct permeability measurement.\textsuperscript{25} They found that the water content of the menisci was 73.9\%. The mean hydraulic permeability of the tissue was $6.4 \times 10^{-16} \text{m}^2/\text{N}\cdot\text{s}$, and the mean compressive modulus was 0.41 MPa. The water content, hydraulic permeability and compressive modulus were noted to vary according to the depth of the sample location (superficial vs deep) and also the location of the sample (anterior vs central vs deep).

Proctor et al. also investigated the compressive behaviour of bovine menisci to determine the compressive modulus and permeability of the tissue.\textsuperscript{31} Proctor et al. found a mean water content of 73.8\%, with a mean permeability of $0.81 \times 10^{-15} \text{m}^2/\text{N}\cdot\text{s}$, and a mean compressive modulus of 0.41 MPa. They found that values were uniform within the surface layer and did not differ according to location. However, for the deeper tissue, Proctor et al. did observe that the measured compressive material properties differed significantly according to sample location. The posterior deep specimens had a significantly higher modulus. No significant regional variations were observed in the permeability coefficients of the tissue, but the posterior specimens were found to have a significantly higher water content.

Functional biomechanics

Load transmission

The menisci transfer forces between the femoral and tibial joint surfaces by the development of hoop (circumferential) stresses within the meniscal tissue.\textsuperscript{11} These are tensile stresses transferred along the circumferential collagen fibres of the meniscus between insertions. As the near-circular femoral condyles bear down onto the meniscal tissue, the menisci have a tendency to be extruded peripherally.\textsuperscript{9} This extrusion is prevented by the firm attachments of the anterior and posterior horns to the tibia via the anterior and posteriorinsertional ligaments, because an increase of the radius of the circle of meniscus simultaneously increases the circumference. Thus, a circumferential tension develops within the menisci, opposing further displacement. This is referred to as ‘hoop stresses’.\textsuperscript{35} Thus the compression force across the knee, as it is transmitted from the femur through the meniscus to the tibia, causes tension along the circumferential collagen fibres within the meniscal tissue (Fig. 8).

A number of different experimental techniques have been used to determine intra-articular contact areas and pressure distribution. By loading cadaver knees in
a materials testing machine and using pressure transducers within the joint, Seedhom showed that on the lateral side the meniscus carries 70% of the load in the lateral compartment, and that the medial meniscus carries 50% of the load in the medial compartment.\textsuperscript{36} 

Walker et al. used methylmethacrylate casting and showed that under no load, contact across the knee occurred primarily on the menisci.\textsuperscript{8} With loads of 1472N, the menisci were shown to cover between 59 to 71% of the joint contact surface area. 

Using a casting method employing silicone rubber, Fukubayashi and Kurosawa found that under a load of 1000N, the menisci occupied 70% of the total contact area.\textsuperscript{5} Removal of the menisci more than halved the contact area and led to a doubling of the peak contact pressures. 

The importance of the menisci in load bearing was further shown by Baratz et al., who, using pressure-sensitive film in knees loaded by a materials testing machine, showed that after total meniscectomy, joint contact areas decreased by approximately 75%, and peak local contact stresses increased by approximately 235% (Fig. 9).\textsuperscript{37} 

**Shock absorption**

As has been described above, axially directed energy across the knee during loading of the joint is converted into hoop stresses within the circumferential collagen fibres.\textsuperscript{9} Furthermore, as well as energy being absorbed into the collagen fibres (solid phase of the meniscus), as the tissue is compressed energy is further absorbed by the expulsion of the joint fluid (fluid phase of the meniscus) out of the tissue. 

The shock absorbing capacity of the menisci has been measured by registration of bone vibrations resulting from gait, and it has been shown that without the menisci, shock absorption within the knee is reduced by approximately 20%.\textsuperscript{11} 

**Meniscal motion**

The menisci are dynamic structures, and to effectively maintain an optimum load-bearing function over a moving, incongruent joint surface, they need to be able to move as the femur and tibia move, to maintain maximum congruency. 

Thompson et al. were the first to describe meniscal movements through a full flexion-extension arc in the intact knee using MRI of cadaver knees.\textsuperscript{38} They showed that from full extension to full flexion, there was posterior excursion of the medial meniscus of 5.1 mm, and of the lateral meniscus of 11.2 mm, with the anterior horns moving more than the posterior horns. However, these observations were made in unloaded cadaver knees, and may therefore not be representative of the in-vivo weight bearing situation. Furthermore, Thompson et al. failed to comment on the medio-lateral movement of the meniscal tissue. 

More recent technical advances in the field of radiographic imaging have lead to the development of so-called ‘open’ magnetic resonance scanners. These scanners allow a subject to lie, stand, sit or squat within the imaging field, and thus permit imaging of the intact in-vivo knee under load in all positions. Using such a scanner, Vedi et al. described the details of meniscal motion in the normal knee in both the weight-bearing and non-weight-bearing situations.\textsuperscript{39} They found that the menisci moved less than had been reported by Thompson et al. However, in common with the findings of Thompson et al., Vedi et al. observed that the menisci do move posteriorly as the knee flexes. The anterior horns were noted to be more mobile than the posterior horns, and the lateral meniscus to be more mobile than the medial (Fig. 10). The posterior horn of the medial meniscus was found to be the least mobile. Vedi et al. also showed that there was significant movement of the bodies of the menisci peripherally with flexion. 

Vedi et al. compared the observed values in the unloaded knee with those found in the weight-bearing situation. They showed that statistically there was significantly greater movement in the anterior horn of the lateral meniscus when the knee was weight-bearing, but no significant differences were seen in the other meniscal movements observed. 

**Joint stabilisation**

The role of different anatomical structures in the stability of the knee has been studied extensively, both in cadaveric work and in-vivo. Tapper and Hoover reviewed 113 patients at between 10 to 30 years after meniscectomy (84% medial, 16% lateral) and found that 24% showed evidence of instability of the knee.\textsuperscript{40} In this study all patients that had evidence of collateral or cruciate ligamentous injuries during initial surgery were excluded. It was therefore postulated that the group of patients showing instability at follow-up either had unrecognised injuries, were re-injured later, had laxity due to degenerative changes, or that perhaps the loss of the meniscal mass might have caused a relative laxity of the ligaments. 

Levy et al. showed that anterior-posterior laxity and coupled rotation were unaffected by medial meniscectomy in the anterior cruciate-intact knee.\textsuperscript{15} However, after section of the anterior cruciate ligament subsequent further resection of the medial meniscus led to an additional increase in anterior laxity. In a further study, Levy et al. also showed that lateral meniscectomy after resection of the anterior cruciate ligament resulted in a small but
significantly further increase in anterior translation. The smaller increase in anterior laxity associated with removal of the lateral meniscus compared to that seen with the medial meniscus may well be related to the fact that the lateral meniscus is more mobile than the medial. The relatively immobile posterior horn of the medial meniscus, however, most probably acts as a ‘chock block’ resisting posterior translation of the medial femoral condyle.

Further work by Allen et al. confirmed that the actual forces within the medial meniscus in response to an anterior tibial load increased by up to 197% after section of the anterior cruciate ligament. It was also shown that after medial meniscectomy in the anterior cruciate deficient knee there was a significant further increase in induced anterior tibial translation of up to 5.8 mm.

It can therefore be considered that the medial meniscus does appear to be a significant secondary stabiliser to anterior drawer, of particular importance in the ACL-deficient knee, and this may well be a factor contributing to the poor prognosis that is seen in those patients with combined ACL tears plus medial meniscal deficiency.

**Associated ligaments**

To further complicate the issue of accurately describing the biomechanical properties of the menisci of the knee, there are various associated ligaments which, although variable in their presence, can have significant influence on the behaviour of the menisci.

The anterior intermeniscal ligament, also known as the transverse geniculate ligament, connects the anterior fibres of the anterior horns of the medial and lateral menisci (Fig. 11). In a cadaveric study of ninety-two knees performed by Kohn and Moreno, a transverse ligament was found to be present in 64% of specimens. However, the functional relevance of this ligament has not yet been determined.

Two ligaments have also been identified joining the posterior horn of the lateral meniscus to the lateral side of the medial condyle of the femur in the intercondylar notch. These are known as the meniscofemoral ligaments. One ligament runs anterior to the posterior cruciate ligament, and is known as the ligament of Humphrey. The other runs posterior to the posterior cruciate ligament, and is known as the ligament of Wrisberg (Fig. 12). The ligament of Humphrey has been found to be present in 74% of knees, the ligament of Wrisberg in 69% of knees, and both ligaments together in 50% of knees. The relevance and importance of these ligaments has been demonstrated by Gupta et al., who showed in a cadaveric study that the...
meniscocentral ligaments contributed 28% to the total force resisting posterior drawer at 90 degrees of flexion in the intact knee, and 70.1% in the PCL-deficient knee.\textsuperscript{44}

It should, therefore, perhaps be of some concern that such little, if any, consideration is currently given to the associated ligaments of the menisci during surgery of the knee and when undertaking meniscal repair or replacement.

Conclusion

This brief review has described how the morphology and structure of the menisci are adapted to their mechanical load-transmitting function. In particular, the circumferential collagen fibres allow meniscal radial extrusion under axial joint forces to be resisted by the build-up of hoop stresses. The anisotropic material properties are adapted to this role, but at the expense of the tissue being weak in the radial direction, which allows the meniscus to be torn relatively easily by splitting along the lines of the circumferential fibres. Meniscal surgery should aim to maintain continuity of the circumferential structure in order to ensure the load-bearing performance; without this, the articular joint surfaces are subjected to excessive contact stresses, resulting in premature onset of osteoarthritis.

The true structural complexity of the menisci, with their material inhomogeneity, their anisotropy and their various associated ligaments, emphasizes just how specialised they are in fulfilling their various roles within the knee. It also explains how complicated a task it is to try and model them accurately. However, a detailed and comprehensive understanding of the structural and biomechanical characteristics of the menisci is vital for future work such as the tissue engineering of meniscal replacements.

References

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