NC STATE UNIVERSITY
JTATM
JOURNAL OF TEXTILE AND APPAREL, TECHNOLOGY AND MANAGEMENT

Volume 3, Issue 2, Fall 2003

DESIGNING CARBON-POLYESTER BRAIDS FOR LIGAMENTS

Prasad Potluri*, William D Cooke, Alberto Lora Lamia and Edurdo Corral Ortega
Department of Textiles
University of Manchester Institute of Science & Technology,
PO Box: 88, Manchester M60 1QD, United Kingdom
*Prasad.Potluri@umist.ac.uk

ABSTRACT

Ligament prostheses are highly load-bearing structures that are subjected to both quasi-static and impact loading. They are expected to have a long service life without undergoing excessive creep. In addition, in vivo trials are very expensive and take a very long time. In view of this, numerical simulation techniques were developed in the present work to simulate the structural response of the ligaments. Non-linear load-deformation behaviour of a braided ligament was predicted based on the stress-strain relationships of constituent yarns of carbon and polyester. Tensile and transverse stresses were computed in order to establish failure criteria. A cyclic fatigue test scheme that maintains a desired load-amplitude is described here. This test scheme does not need expensive servo-hydraulics and hence can be used to test a large number of samples simultaneously to expedite a product development cycle.

Keywords: ligaments, anterior cruciate ligament, braid, carbon fiber. Polyester

INTRODUCTION

Ligaments are very important restraining members in the musculoskeletal system. According to the Oxford Dictionary, a “ligament” is a short band of tough flexible fibrous connective tissue linking bones together. Skeletal joints are kinematically constrained and stabilized by ligaments to minimise the transverse displacements while maintaining the rotational movements. Ligaments are subjected to shock-loads during sports and exercise programs sometimes resulting in rupture. In recent years, sports related injuries are on an increase as a result of populations participating in sporting activities well into their middle ages. As a result, prosthetic devices are used increasingly in joint rehabilitation.

Ligaments are essentially tensile structural members and hence offer very little resistance in compression. Hence, several ligaments are involved in stabilizing a joint. The natural ligament is made-up of aligned networks of collagen microfibrills giving high strength (2160N in the case of the ACL of a young adult). However, the key feature in the ‘design’ of the natural ligament is the presence of ‘crimp’, which is analogous to the crimp present in textile structures. The natural ligaments exhibit strain-hardening behaviour with a low initial modulus.

Anterior Cruciate Ligament in a Knee Joint

The work presented in this paper relates to the development of prosthetic devices to replace ruptured ligaments in a knee joint. The design methodology is equally valid for other skeletal joints. The knee joint is stabilized by four principal ligaments and the patellar tendon. Two cruciate ligaments, which run from back to front and cross each other like a ‘figure of eight’. These ligaments also absorb forward and backward shocks on the joint. Two collateral ligaments, which run up either side of the knee joint, provide lateral stability to the joint (Figure 1). The anterior cruciate ligament (ACL) is the most frequently injured, as it takes about 75% of the anterior shock load, and hence is the prime candidates for repair or prosthetic augmentation [1]. Although significant progress has been made towards understanding the anatomy, composition, biomechanics and healing of the ACL, today there is still no graft or prosthesis ideally suited for ACL reconstruction [2].

Figure 1: Knee Joint

DESIGN CONSIDERATIONS

The main design consideration should be to mimic the behaviour of a natural ligament. This is partly evident from the fact that almost all the ligament prostheses are made of fibers formed into rope-like structures. However, a number of early devices were designed mainly for ultimate strength rather than to match the load-extension behaviour of a natural ligament. For example, the Apex ligament was made with a parallel bundle of polyester fibers [3]. This ligament does not exhibit low initial modulus due to the absence of crimp, and so is prone to abrasion failure. Richards Polytex ligament, made with UHMW polyethylene, was totally unsuitable as it exhibits excessive creep [4]. The Stryker-Meadox ligament was manufactured from polyester fiber bundles with a knitted outer sleeve [5]. The Leeds-Keio ligament had an open polyester weave in the form of a flat tube. Although the initial results were good, longer term results were disappointing in comparison to the use of a patellar tendon graft [6]. A number of problems with the earlier ligaments would have been avoided if proper design steps were followed.

The authors identified the following design criteria:

1. The load-elongation curve should be similar to that of a natural ligament with low initial modulus. The modulus should increase with increasing load.
2. The load to failure should be comparable to that of a natural ligament. For example, load to failure of a typical young adult ACL is 2160N.
3. Creep or inelastic deformation should be kept to a minimum in order to avoid joint laxity.
4. The fatigue cycle life of a ligament is very important. For example, 6 million cycles is approximately equivalent to 4 years *in vivo*.
5. The ligament should have good resistance to abrasion against bones and joints [7].
6. The ligament should promote strong aligned tissue in-growth.

A number of commercial prostheses have failed to match the performance of the natural ACL resulting in post-operative failures and complications [8]. Most of these problems could have been avoided if accurate structural and life predictions had been made prior to clinical use. In order to improve the design process for ACL prostheses the authors have developed the modelling of the load-elongation behaviour until failure.

CARBON-POLYESTER BRAIDED LIGAMENT PROSTHESIS

Load-elongation behaviour of textile structures, especially woven and braided, is similar to that of a natural ligament. They exhibit low modulus during the initial stages; modulus increases gradually with de-crimping in the loading direction and reaches a maximum value. In addition, textile structures are less prone to abrasion in comparison to straight fiber bundles. Braiding is a preferred route for producing narrow rope-like structures. All the yarns in a braided structure share the axial load while only half the yarns (warp) share the axial load in a woven structure.

McLeod [8] reported the development of carbon-polyester ligament prosthesis, known as Active Bioprosthetic Composite (ABC). This prosthesis was constructed using 24 strands of carbon-polyester braided structure, referred in this paper as 'unit material'. The unit material is formed by over-braiding four strands of polyester yarn over a strand of carbon fibre tow (Figure 2). During the over braiding-process, the core fibres are disposed into a flat zigzag configuration. The degree of zigzag may be varied by adjusting the relative tensions of the core (carbon) and braid (polyester) yarns. A large number of structural and process parameters can be adjusted to achieve the desired load-deflection characteristic for the braided structure. Hence it is important that an accurate mechanical model be developed that is capable of dealing with the material and geometrical non-linearity.

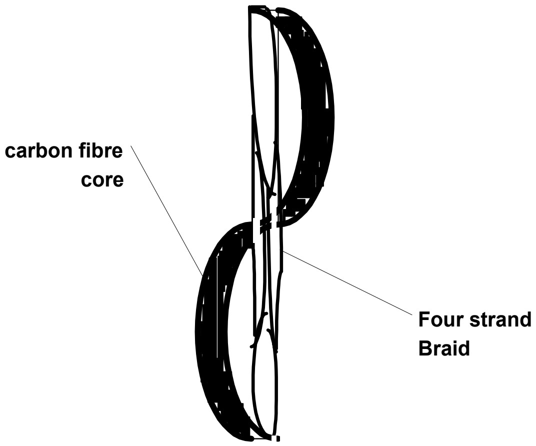

Figure 2: Carbon-Polyester braid

MODELING THE LOAD-ELONGATION BEHAVIOUR

The mechanical modelling aims to predict the load-elongation behaviour of the braided unit material and hence the entire ligament, based on the constituent material properties and the braid geometry.

Fiber assemblies are extremely difficult to model using Finite Element Methods (FEM) that are traditionally used for bulk materials. These materials undergo large deformations and behave in a non-linear manner. In addition, it is not convenient to model the material discontinuities that exist in fibrous assemblies, using commercial FEM packages. In the present work, finite-deformation based numerical modelling is used. A 'unit cell' (unit repeat of the braid, length between two points of intersection) is subjected to a finite deformation ($\delta_{u)}$, and the resulting forces and stresses are analysed during the entire stress-strain cycle.

Modelling the load-elongation of the braid

Figure 3 shows three distinct phases of deformation. During Phase 1, the polyester braid mainly generates the force, while the carbon core undergoes a slight change in curvature. The carbon core undergoes de-crimping during Phase 2 until it straightens. During Phase 3, the carbon core takes an increasing share of the load until the mechanical properties are dominated by the mechanical performance of the carbon component.

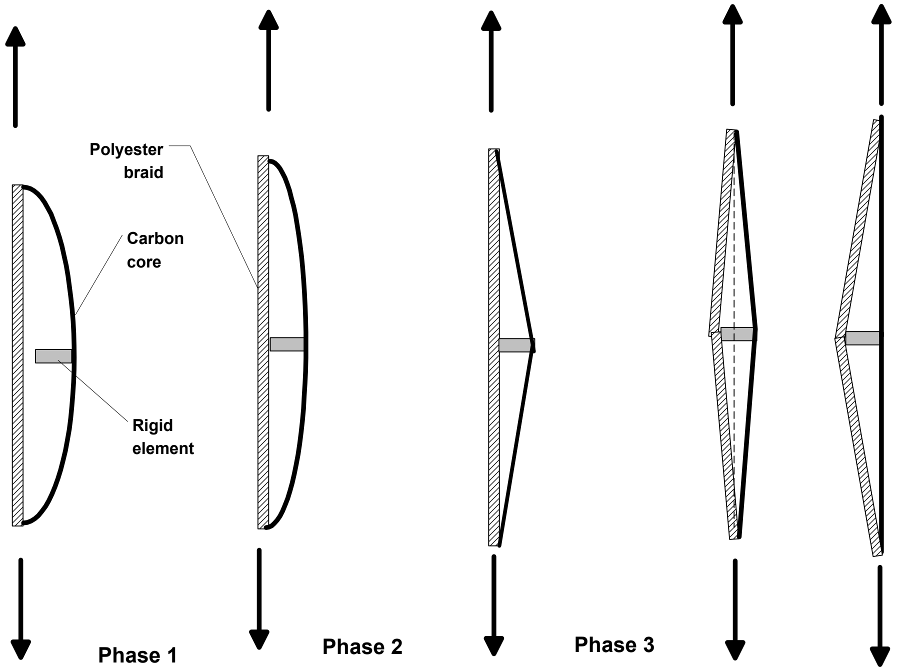

Figure 3: Kinematic model of the deformation process

Phase 1

The carbon filament tow is crimped and slack and hence does not take a significant load. The structure continues in this condition until the gap (initial value 0.086 mm) becomes zero (Figure 2). During this phase the polyester braid is subjected to stretching, and there are no contact forces between polyester and carbon structures.

$$\varepsilon_P = \varepsilon_b \left[\frac{1}{1+c}\right]^2 \left[\frac{1+\varepsilon_b}{1+\varepsilon_P}\right]^2 \qquad \textbf{(1)}$$

$$\varepsilon_c = 0 \qquad \textbf{(2)}$$

where,

ε_p = strain in the polyester fibres

ε_c = strain in carbon fibres

ε_b = strain in polyester braid(=strain of the unit material)

c = crimp ratio of the polyester yarns

During the stretching of the unit material, the helix angle decreases, and is dependant on the new strain values. The calculation proceeds until the gap is equal to zero. The force generated along the axis of the braid is:

$$F_b = 4 A_P f(\varepsilon_P).\cos\theta \qquad \textbf{(3)}$$

where,

F_b = force along the polyester braid

A_p = cross-sectional area of the polyester

$f(\varepsilon_p)$ = 5th order polynomial stress-strain relation of polyester fibres.(Figure 4)

ϑ = helix angle of the braid

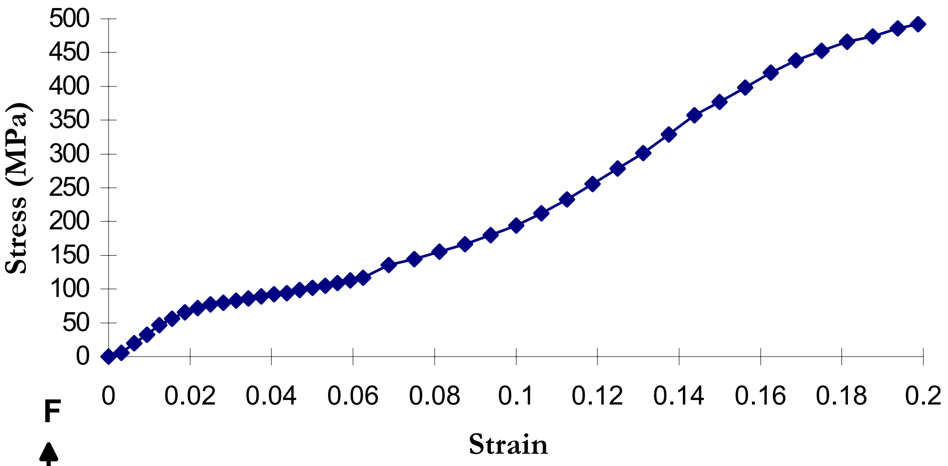

Figure 4: Stress-strain behaviour of a polyester braid

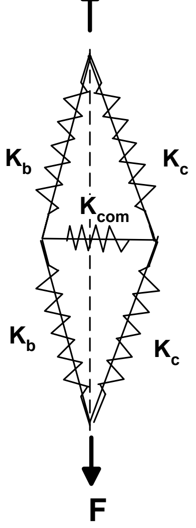

Figure 5: Saw-tooth spring model

Phase 2

During the second phase, the gap between the polyester and carbon strands is zero, but the carbon tow is still slack. While the polyester braid is stretching, the carbon tow is subjected to bending and hence loses its curvature. The carbon tow, within a unit cell, may be assumed as a *chord* with a constant curvature. The vertical force generated due to bending of the carbon tow is

$$F_{CV} = \frac{EI}{h}\left(\kappa - \kappa_o\right) \qquad \textbf{(4)}$$

where,

EI = flexural rigidity of the carbon tow.
h = centre distance between polyester braid and carbon core
κ = radius of curvature
κ_o = initial radius of curvature

Phase 3

During this phase, the polyester braid does not remain straight but deforms into a saw-tooth geometry as shown in Figure 5. The resulting position depends on the equilibrium of forces.

K_b and K_c are non-linear tension springs representing the polyester braid and the carbon core.

K_{com} is a compressional spring to represent the lateral compression between the polyester braid and the carbon core.

For a given strain value of the unit material (ε_u), one can compute the forces from the geometry and the non-linear material properties.

However, the geometry of the unit cell will change due to a new equilibrium of the forces, and hence it must be solved iteratively.

Force-strain relation

From Figure 6 it can be seen that the computed force-strain relation of the unit material closely follows that of the experimental curve which was generated in the laboratory using a universal mechanical tester. The force-strain curve shows a characteristic bi-linear relation with a smooth changeover between the two phases. The first phase up to about 0.04 strain is dominated by the stress-strain behaviour of the polyester braid, while the final part of the curve is dominated by that of the carbon core. It may be observed from the graph that the computed curve has a longer transition than the experimental curve. This is due to simplifications made to the model.

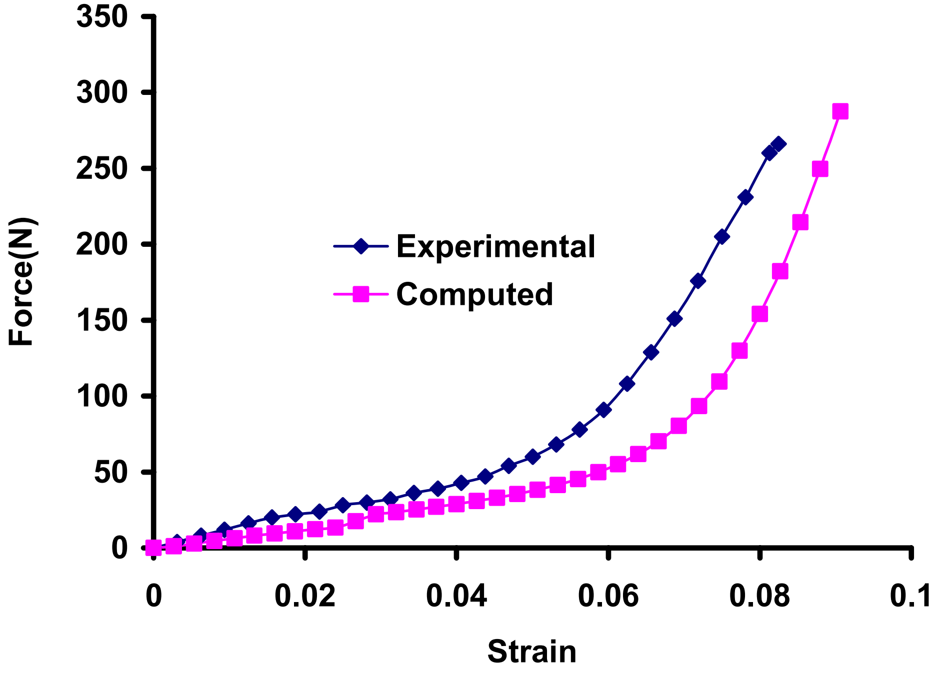

Figure 6: Theoretical and experimental stress-strain behaviour of a carbon-polyester unit material

Criteria for Failure

The tensile and transverse stresses generated in the polyester and carbon fibres (Figure 7 and Figure 8) can be computed from the numerical model. At failure, the carbon fibres are subjected to a tensile stress of 1800MPa, which is only 45% of their tensile strength. Similarly, polyester fibres are subjected to a tensile stress of 260MPa, which is about 50% of their tensile strength. Clearly, the mode of failure is due to applied forces other than tensile ones. Carbon fibres are weak in the transverse direction, and transverse strength data is seldom published. From the numerical model (Figure 7), it can be observed that carbon fibres have an average shear stress of 135MPa at the widest point on the unit material, where failure is normally observed. This compares to the short beam shear strength of a carbon fibre composite at about 124MPa, as supplied by the carbon fibre manufacturer.

During 5 years of clinical use of the Surgicraft ABC ligament in the period 1985-90, McLeod [9] reported an overall failure rate of approximately 12%. The majority of these failures occurred within 6 months of implantation and was shown by mode of failure analysis to be caused by incorrect anterior placement of the tibial exit hole [10]. Following the development of an accurate positioning jig [11], the number of early breaks was drastically reduced. Nevertheless unexplained breaks in well-established prostheses continued to occur despite the fact that the nominal (design) tensile strength of the ABC ligament was greater than that of a young healthy adult. The tensile strength of a ligament with 24 strands is 6.4 kN compared to the strength of a natural ligament at 2.2 kN. The numerical analysis reported here suggests that the transverse shear loads that develop within the braid structure during normal tensile loading could have contributed to a proportion of these breaks.

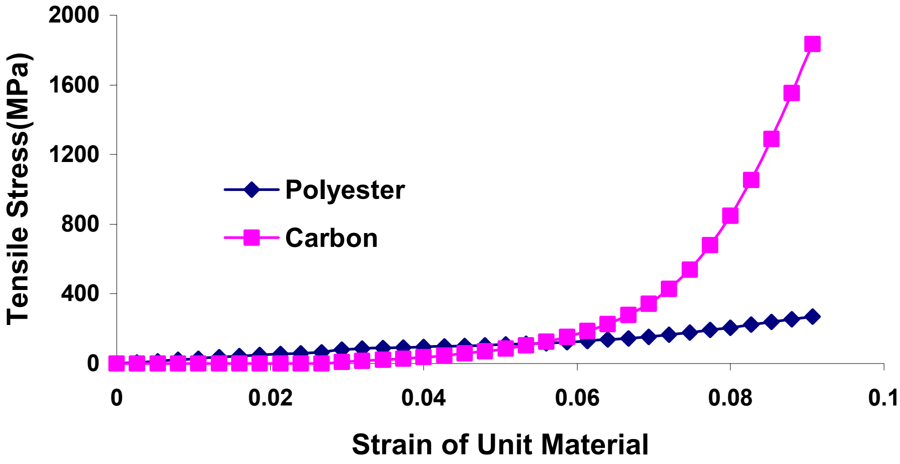

Figure 7: Tensile stress of polyester and carbon fibers

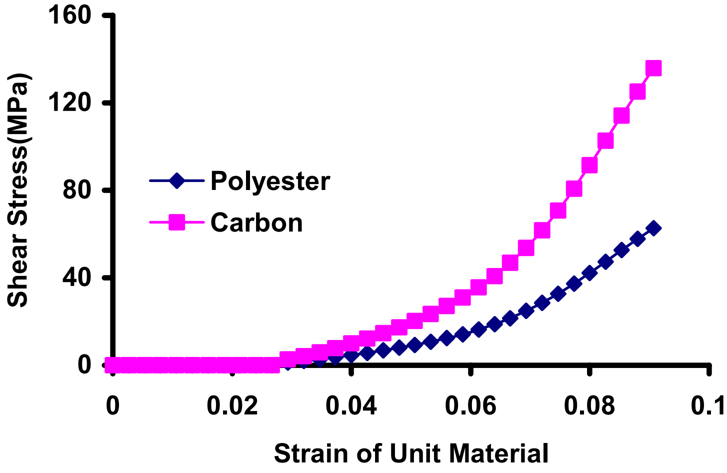

Figure 8: Transverse stresses in polyester and carbon fibers

DEVELOPMENT OF FATIGUE TESTING SYSTEM

In addition to structural modelling, it is important to conduct fatigue testing of a large number of samples to establish the service life of a prosthetic ligament. Servo-hydraulic test systems, commonly used at present, are very expensive, they apply a displacement amplitude rather than a load amplitude and hence need periodic adjustments to the test cycle as the sample experiences creep deformation. Furthermore, they need to be carried-out at low frequency, 1Hz in order to avoid heating. A test program involving a large number of samples can therefore take a very long time.

The authors have developed a novel, inexpensive electro-mechanical device for fatigue testing. This device, working on the principle of a moment arm applied by an oscillating mass, applies a load-amplitude and hence does not need any adjustments during the testing procedure. The level of creep can be continuously monitored. In addition, a static load can be applied for a specific duration if desired by simply switching off the machine. Perhaps, the greatest advantage of the new device is that a series of loading positions can be built at a limited cost. This can expedite the test program by running several specimens simultaneously under identical conditions. The new system has the following advantages that may be summarised as follows:

1. It operates with a constant load cycle, rather than a displacement cycle.
2. It does not need expensive load cells.
3. It is possible to have a bank of testing positions so that several specimens can be run at the same time.
4. It does not rely on conventional servo-hydraulic power packs which consume high energy.
5. If desired, a simple static load can be applied over a specified duration so as to measure creep as part of the test protocol. This can be achieved simply by switching off the machine at the set position which gives the desired static load.

Cyclic fatigue testing of a ligament prosthesis can be conducted between a load amplitude of 40 and 280 N at a frequency of 1Hz. The expected cyclic life is 6.3 million cycles, which represents about 4 years in vivo. The fatigue test machine developed in the present work is shown in Figure 7. The ligament prosthesis is mounted between a fixed point and a loading arm. A dead weight slides along the loading arm, actuated by a crank mechanism. The tensile force applied on the ligament is proportional to the distance of the dead weight from the pivot. Hence the load amplitude can be set by adjusting the displacement amplitude of the sliding dead weight. A load cell built into the fixed end is connected to a PC to monitor the load cycle and to count the number of cycles. Figure 10 shows a typical load cycle.

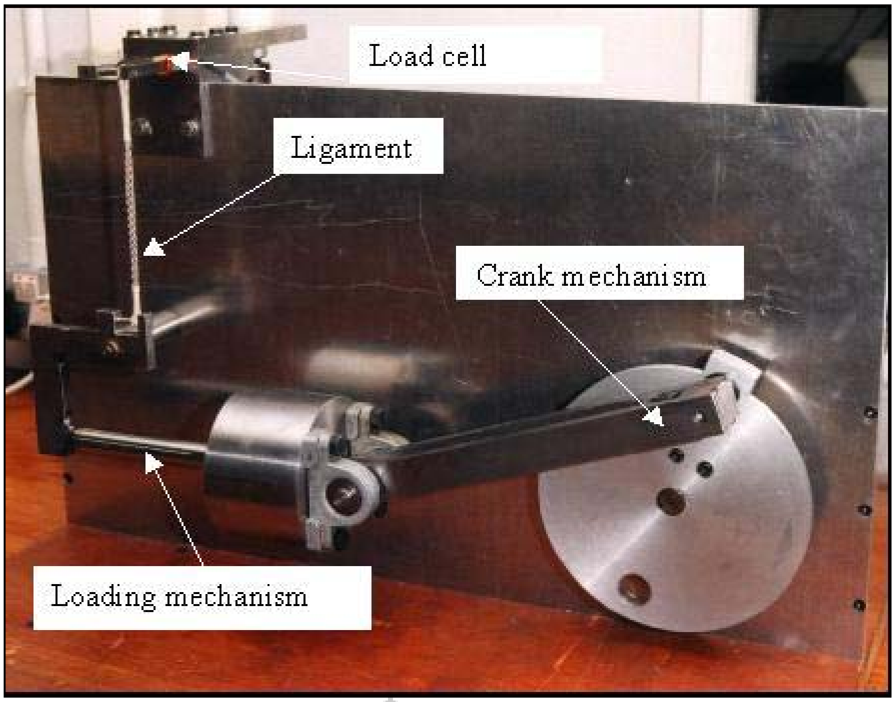

Figure 9: Prosthetic ligament fatigue test machine

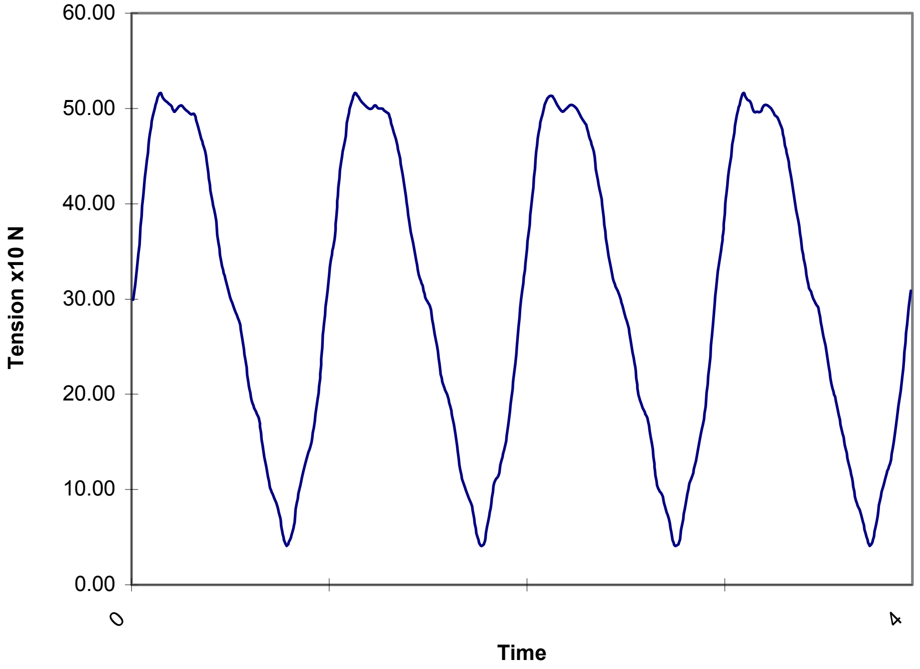

Figure10: Typical load cycle

DISCUSSION

This paper describes research that has identified three important issues in the design and development of an ACL prosthesis, namely, modelling the load-deformation behaviour, establishing the failure criteria, and fatigue testing to estimate the service life. A 'unit cell' of the braided structure has been modelled, using a combination of springs, rigid rod and gap elements. A computer model has been developed from first principles rather than using third-party software. This approach has assisted in the phenomenological understanding of the deformation process, and hence facilitated its use as a design tool. A number of additional refinements can easily be made to the model. For example, one could include a distributed contact force between the braid and the core rather than assume a single contact point. This would take into account the compresional strain at the contact point. The computer model predicts that the ligament may fail due to transverse stresses. A further investigation into the failure analysis is required. In the fatigue test rig, the prosthetic device is mounted between two points along a straight line. This should be modified to simulate the actual path and geometry of the prosthesis through the knee joint.

References:

1. Butler DL, Noyes FR and Grood ES, Ligamentous restraints to anterior-posterior drawer in the human knee: A biomechanical study*, J Bone Joint Surgery,* 62A: 259-270 (1980).

2. Dunn MG, Anterior Cruciate Ligament Prosthesis in Encyclopaedia of Sports Medicine and Science, Fahey, TD (Ed.)

Internet Society for Sport Science. http://www.sportsci.org, 1998.

3. Hunt and Heatley, Clinical results with the Apex ligament. *Orthopaedic Symposium*, London, May 1990.
4. Chen EH and Blacj J, Materials Design Analysis of the prosthetic anterior cruciate ligament, *J Biomedical Materials Research*, 14: 567-586 (1980).
5. Lopez-Vazquez E, Juan J and Vila E, Reconstruction of the ACL with a Dacron prosthesis, *J Bone Joint Surgery,* 73A: 1294-1300 (1991).
6. Dandy DJ and Grey AJR, Anterior cruciate ligament reconstruction with the Leeds-Keio prosthesis plus extra-articular tenodesis, *J Bone Joint Surgery*, 76B: 193-197 (1994).
7. Guidoin MF, Marois Y, Bejui J et al, Analysis of retrieved polymer based replacements for the ACL, *Biomaterials,* 21: 2461-2474 (2000)
8. McLeod ARM, The development of the ABC prosthetic ligament system following failure mode analysis, *PhD Thesis*, UMIST, Manchester, UK, 1995.
9. McLeod ARM, O'Brien, Cooke WD, Rees, Shafighian and Strover, Successes and Failures Following 5 Years of Clinical Experience with the Surgicraft ABC Prosthetic Anterior Ligament, *Interfaces between Engineering and Medicine, 2nd. International Meeting*, Bologne, Italy, Sept 1990, p.169-179.
10. McLeod ARM, O'Brien and Cooke WD, Structural Integrity of the ABC Ligament following Clinical Implantation. A Biomechanical Perspective through Mode of Failure Analysis, *European Society of Biomechanics, Proceedings of the 8th.Meeting*, June 1992, p.113.
11. McLeod ARM, Kong, Mowbray, Cooke WD, Strover and O'Brien, Mode of Failure Analysis Leading to New Instrumentation for the Enhancement of Cruciate Ligament Reconstructive Surgery, *International Society of Biomechanics, XIV Congress*, Paris, July 1993, pp.858-859.