STRENGTH OF CARBON AND POLYESTER FIBRE TENDON REPLACEMENTS

VARIATION AFTER OPERATION IN RABBITS

A. A. AMIS, J. R. CAMPBELL,[†] J. H. MILLER

From the Biomechanics Section, Imperial College of Science and Technology, London

The calcaneal tendons of rabbits were excised and either replaced with a carbon or polyester fibre implant, or left as controls. The strength of the neotendons and their mode of failure under tension were examined at intervals up to six months after operation.

Return to near normal strength took six months to develop, suggesting that patients having ligament or tendon reconstructions should not resume normal activity for several months. Carbon fibre-based neotendons showed progressive elongation which, unless avoided by a sufficient period of immobilisation, would affect the functional result.

Much interest was aroused by reports of the successful use of carbon fibres to repair defects in the tendons or ligaments of experimental animals (Jenkins *et al.* 1977; Jenkins 1978). These led to the clinical use of carbon fibres in both human (Jenkins and McKibbin 1980) and veterinary cases (Vaughan and Edwards 1978; Littlewood 1979).

In the original experiment of Jenkins *et al.* (1977), the calcaneal tendon of the sheep was excised and replaced by carbon fibres. The animals were then allowed to exercise freely, with no immobilisation of the operated limbs, and a return to normal locomotion was reported. An attempt to duplicate this work was unsuccessful (Amis *et al.* 1984). It was found that 56 days' immobilisation was needed to allow tissue ingrowth to augment the strength of the knotted anchorages of the implants and that, without splintage, progressive elongation of the repair led to deformity. Similar experiences with equine flexor tendons were reported by Valdez, Clark and Hanselka (1980), who recommended 90 days' immobilisation after implantation of carbon fibres. When carbon fibres were used by Jenkins and McKibbin (1980), in human patients, the joints were immobilised for 42 days.

In this initial experimental phase when contradictory results were likely, the only guide to the strength of these repairs and their variation after operation was provided by the work of Forster (1976) and Jenkins *et al.* (1977) on the calcaneal tendon of the rabbit. The testing

methods were rather primitive, but a graph of the variation of strength with time after operation showed a progressive increase from 25 N at the time of operation to 150 N after six weeks, with maintenance of this strength at 12 weeks. The contralateral tendons had strengths of between 100 N and 200 N, and it seemed reasonable to suggest that the carbon implants had allowed a return to normal strength in six weeks.

Later work by Aragona *et al.* (1981), also on the calcaneal tendon of the rabbit, suggested that a carbon fibre anastomosis across a 5 mm defect healed to normal strength by four weeks, although this was followed by a deterioration to below-normal strength, which had not fully recovered at 12 weeks. Aragona *et al.* found a mean normal strength of 340 N, at which level muscle fibres of the gastrocnemius ruptured. This result threw doubt on the results of Jenkins *et al.* (1977) who gave 150 N for normal tendons. Both sets of researchers had included the muscle belly in their tensile test specimens and found that this was the weakest point, but neither group mentioned the very rapid weakening of the muscle after death, or measures to avoid this effect. Viidik (1969) tested the calcaneal tendons of rabbits within one hour of death and found a mean strength of 377 N. The present authors therefore feel that the earlier experiments may not, in fact, have led to normal tendon strength either four or six weeks after carbon fibre repair.

The development of artificial finger flexor tendons has involved the study of tissue ingrowth into multi-filamentous implants to form tendon anchorages (Salisbury *et al.* 1974; King, Dunn and Bolstad 1975). These workers used polyester fibres, which had already gained acceptance for implants requiring tissue ingrowth, such as arterial grafts (Hoffman 1977). This work had involved tissue ingrowth in small anchorage areas, rather than along the entire length of the implant, so it was decided to compare the properties of polyester

A. A. Amis, PhD, C Eng, MI Mech E, Lecturer in Mechanical Engineering, Biomechanics Section, Imperial College of Science and Technology, London SW7 2BX, England.

[†] J. R. Campbell, PhD, BVMS, FRCVS, deceased.

J. H. Miller, FRCS Ed, FRCS Glas, Consultant Orthopaedic Surgeon Accident and Orthopaedic Division, Glasgow Royal Infirmary, Glasgow G4 0SF, Scotland.

Requests for reprints should be sent to Dr A. A. Amis.

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with carbon fibre when they were used in the manner reported by Jenkins *et al.* (1977). The relatively inert polyester fibres should function as a scaffold on which fibrous tissue could be laid down, in a manner similar to that demonstrated for carbon fibres, while also having several advantages. They are less brittle, and thus easier to handle at operation; they would not fragment and could therefore be removed in case of infection; and they could be designed to be elastically compatible with the natural structure, ensuring mechanical compatibility (carbon fibres are too stiff for this).

The strength of the normal calcaneal tendon of the rabbit was examined before polyester implants were designed, to provide data on elastic behaviour under tension. The normal rabbit calcaneal tendon had a mean strength of 505 N when premature failure of the muscle fibres was avoided. This suggested that the strengths of carbon fibre repairs reported by Forster (1976), Jenkins *et al.* (1977) and Aragona *et al.* (1981) had not reached normal values at either four or six weeks after implantation. The work of Jenkins *et al.* using carbon fibres was therefore repeated in parallel with a series of polyester fibre implants, with the aim of determining the variation in strength of both types of fibre after implantation, and also to provide guidance on the need for, and the duration of, immobilisation after operation.

METHODS

The carbon fibre implants were of Courtaulds Grafil type HM-S, a high modulus, high purity type; 20 000 filaments of 9 μ m diameter were used. The polyester implants were of ICI Terylene 50 f 20 type 100, undyed and with low delustrant content: 5000 filaments of 15 μ m diameter were used.

The calcaneal tendon complex of the rabbit consists of three closely associated but separate tendons (Nisbet 1960). It was excised under general anaesthesia from the left hind legs of mature California breed rabbits, from the superior surface of the os calcis to a point 5 mm distal to the musculotendinous junction, creating a defect about 15 mm long. The stumps of the gastrocnemius and plantaris tendons were elevated from the plantar aspect of the os calcis, which was then drilled upwards so that the hole emerged through the insertion of the soleus. A loop of the implant was passed distally through the drill hole, then opened and passed back over the superior aspect of the os calcis, to encircle it and the double ends of the implant, giving a secure knot-free anchorage (Fig. 1).

The two free ends of the implant were sutured to the musculotendinous junction by the Bunnell technique, with the foot plantarflexed, and the ends knotted together on the anterior surface of the muscle. The implant was cut above this knot and its ends secured by silk sutures. The wound was then sutured in layers with gut and monofilament nylon. The limbs were held in plaster for four weeks, with the ankle plantarflexed to

Fig. 1

Figure 1—Method of securing the fibre implant to the os calcis. The encircling loop is deep to the insertion of the gastrocnemius tendon and to the digital flexor tendon. Figure 2—Method of tensile testing. The bones of the hindfoot are embedded in methacrylate and the muscle fibres proximal to the tendinous junction are clamped by jaws frozen with liquid carbon dioxide (one jaw removed for clarity).

Fig. 2

relieve tension, after which the rabbits were allowed to move freely in their cages. A control group had the same operation, except that no implant was used. Groups of four animals were available for testing of each implant material at 6, 12 and 24 weeks, reduced to three animals in the 12 and 24-week carbon-neotendon groups by premature rupture. The four control animals were examined at 24 weeks.

Animals were killed at the timings given, the intact hind limbs were deep frozen for storage, then thawed for testing. After skinning the limbs the neotendons were measured and photographed. Specimens for tensile testing were prepared by embedding the bones of the foot in a pot of polymethylmethacrylate, then dissecting the tendon and its muscles from the limb. To ensure a secure hold on the specimen for tests, the muscles were clamped in freezing jaws (Riemersma and Schamhardt 1982), proximal to the implant knots, while the embedded foot was clamped in a vice (Fig. 2). The low conductivity of the tissue ensured that neither the implant nor its anchorages were affected by the freezing process. The variation of tensile load with time was plotted by a pen recorder while specimens were elongated at a rate of 50 mm per minute in an Instron 1122 materials testing machine. The mode of failure was noted. Tests were conducted at room temperature, care being taken to prevent dehydration of the surface of the specimens.

RESULTS

Gross appearance. The typical appearance of neotendons is shown in Figures 3 and 4, in which the carbon or polyester fibres have been encapsulated in an adherent layer of tissue. The skin also adhered to some of the implants close to the heel, where there was black pigmentation caused by penetrating particles of carbon fibre coming from the deep aspect.

Fig. 3

Fig. 4

Fig. 5

Fig. 6

Figures 3 and 4—The typical appearances of carbon fibre and polyester-based neotendons, 12 weeks after operation. Figure 5—Typical appearance 24 weeks after control excision, with a slack, elongated strand of fibrous tissue, dorsiflexed foot and atrophic muscle belly. Figure 6—Carbon fibre-based neotendon at 24 weeks, showing postoperative elongation and the similarity to Fig. 5.

Several results were excluded from the study: four because of technical errors, one misplaced drillhole and three early implants with inadequate proximal suturing, before the technique was improved; one polyester implant became infected and two carbon implants ruptured in use. These ruptures occurred at 9 and 10 weeks, near the middle of the implant, allowing excessive dorsiflexion of the ankle and resulting in a thin elongated central section of the neotendon.

The limbs of the control animals developed a fibrous structure linking the muscles to the os calcis; this structure was thinner and much longer than a normal tendon. A typical example is shown in Figure 5, with the foot dorsiflexed and the muscles atrophied and contracted behind the knee; the fibrous tissue bundle is of such length that it is still slack even when its origin and insertion are as far apart as possible, demonstrating that it cannot have acted as a tendon and transmitted tensile forces. This appearance was duplicated in a number of animals in which carbon implants had elongated progressively (Fig. 6). These carbon implants appeared to remain in continuity, but they must have fractured at many points, allowing the filaments to slide alongside each other. The elongation which occurred was shown distally by the thinning of the covering layers of fibrous tissue.

Length of neotendons. Before mechanical testing the lengths of the neotendons were measured from the superior surface of the os calcis to the edge of the test machine clamp, just clear of the proximal suture. Similar measurement of the contralateral normal limbs gave an average length of 42 mm. The differences gave results for the postoperative elongation of the reconstructed tendon

(Fig. 7). The spread of the results for polyester below the zero elongation axis does not imply that these neotendons had contracted, but that some replaced natural tendons were shorter than average. Since there are no data during the first six weeks, it is uncertain whether lengthening had occurred during the four weeks of immobilisation or after removal of the plaster. Despite this, it was apparent that the carbon-based structures had elongated 21 mm (50%) by six weeks and that this continued to increase. The polyester implants elongated

Fig. 7

Graph of the elongation of the neotendons with time after operation. The baseline represents the average length of normal tendons (42 mm).

6 mm (14%) by six weeks, with no significant increase (1 mm) after this. The average lengths of the tendon structures after 24 weeks were: carbon, 74 mm; polyester, 49 mm; and excision controls, 60 mm.

Strength of neotendons. The variation of the tensile failure strength of the neotendons with time is plotted in

Fig. 8

Graph of the variation of breaking strength of the neotendons with time after operation.

Figure 8, showing that strength continues to increase towards normal for 24 weeks. At this stage the control excision group ruptured at from 150 N to 343 N. Their mean strength (236 N) was less than 50% of that of normal tendon. The two types of implant gave similar strengths in relation to time, though the carbon implants had been disturbingly weak, only 25 N, at initial tests. By 24 weeks the implants had a strength advantage over the controls of about 275 N.

The carbon and the polyester implants had been shown to have tensile strengths of 800 N or 430 N respectively; thus neotendons based on carbon fibre never reached the strength of the original implant, whereas those based on polyester were, at 24 weeks, on average 50 N stronger than the implant fibres.

Elongation characteristics. The average load-extension characteristics of carbon-based and of polyester neotendons are shown in Figures 9 and 10 at various periods after operation, with curves for normal tendon, the implant material alone, and the excision controls, at 24 weeks, for comparison. Carbon implants were very much

stiffer than normal tendon whereas the polyester implant was similar to normal.

The neotendons based on both implant types, however, were more extensible than normal tendon. Carbon-based neotendons became more extensible from 6 to 12 weeks, with an increase in stiffness by 24 weeks. Polyester-based neotendons, however, increased progressively in stiffness towards normal, their results at 12 weeks being very similar to those of carbon-based neotendons at 24 weeks. The neotendons of the excision control animals without implants at 24 weeks extended more than twice as much as normal tendons for a given load. Under a force of 200 N, the normal tendon extended 3.9 mm, polyester neotendon 4.7 mm, carbon neotendon 6.2 mm and excision controls 9.5 mm.

Mechanisms of failure. Immediately after operation all carbon implants failed when they were cut by the bony edge of the hole through the os calcis. Fibrous tissue ingrowth in the next six weeks protected the carbon fibres from this, and failure was then at various sites between the os calcis and the proximal fixation knot, by transverse breaks through both implant and surrounding tissue. After six weeks the most common mode of failure of carbon neotendons was by rupture of the encapsulating tissue about 5 mm proximal to the os calcis, with the pull-out of a tail of carbon fibres from the ruptured central core of the implant, while the proximal knot remained embedded at the musculotendinous junction. Elongation of carbon-based neotendons, noted above, was associated with multiple internal failures of carbon filaments—both anchorages remained firm; these failures sometimes produced “clear” zones of neotendon tissue in which no carbon was present, though these zones were not necessarily at the plane of failure.

The polyester implants, being much more resistant to trauma, were not noticeably damaged during either their period of use *in vivo* nor by the subsequent mechanical testing to failure. It follows that failure occurred in the tissues adjacent to the anchorages, at either end of

Fig. 9

Fig. 10

Load-extension characteristics for carbon fibre-based (Fig. 9) and for polyester-based (Fig. 10) neotendons, 6, 12 and 24 weeks after implantation, compared with the behaviour of normal tendons, excision controls at 24 weeks and the original implant material.

the implant. Immediately after operation, the polyester either cut out of the musculotendinous junction or fractured the os calcis in the plane of the drill hole. At 6 and 12 weeks the former mode was most common with "unsleeving" of the implant, the encapsulating tissue rupturing near the os calcis and being left as an empty tube attached to the muscle proximally. In one specimen at 12 weeks, and in two of the four tested at 24 weeks, failure was by transverse fracture of the os calcis, possibly caused by the implant cutting into the bone. These tissue attachments only failed when applied loads reached the breaking strength of normal tendon.

The neotendons formed after control excisions failed at 24 weeks either by pulling from the os calcis (two cases) or by transverse fracture of the os calcis (two cases), the neotendons remaining intact. These failures occurred at less than half of the normal loads, suggesting that disuse atrophy of bone or insertion had taken place.

The design and use of freezing jaws avoided stress concentrations in the muscle fibres over which they were clamped, so normal specimens did not fail here, but, at a mean load of 505 N, by the tendon pulling away from the os calcis, sometimes with bone spicules. Higher figures would possibly have been attained by testing at a greater speed, because of the known strain rate sensitivity of bone (Wright and Hayes 1976).

DISCUSSION

The primary aim of the experiments was to examine the manner in which strength varied with time in calcaneal tendon reconstruction by multifilament implants. The only previous work on this, by Forster (1976) and Jenkins *et al.* (1977), had suggested that repair based on carbon fibre would reach normal strength in six weeks.

The calcaneal tendon is conveniently placed for experimental tendon surgery, but the choice could be questioned because it has an inherent capacity for repair. This could possibly overshadow any deficiency of an implant and produce over-optimistic results. For our study, however, we excised most of the tendon, creating a defect which did not heal adequately by 24 weeks in the absence of an implant. There is a capacity for self-regeneration of tendon-like tissue at this site, and this will be enhanced by the stimulus of functional tension (occurring when muscle is linked to bone by an implant). This means that our principal finding—that it took 24 weeks to regain normal strength after an implant (Fig. 8)—is likely to be an optimistic one. If these implants are used at sites where there is less pronounced tissue generation—for example, within a knee joint—it is probable that full strength is not attained for considerably more than 24 weeks. This has considerable implications for the postoperative management of patients after ligament repair or replacement.

The discrepancy between our results and those reported previously may be influenced by many factors: surgical technique, choice of experimental animal, and

type and length of immobilisation after operation. It has probably been caused by differences in technique of tensile testing, which caused premature failure even in normals, at forces significantly below those recorded in the present study.

It could be argued that carbon fibre neotendons will gain strength more rapidly if free movement of the limb is allowed immediately after operation. This is because fragmentation of the implant would provoke a more intense fibrotic reaction, before the relatively intact implant had been encapsulated. This regime was used by Jenkins *et al.* (1977, 1978) and Aragona *et al.* (1981) for animal experiments, but has not been carried through to clinical practice (Jenkins and McKibbin 1980).

The low strength of carbon reconstructions at the time of operation (20 N to 25 N) was caused by the brittle nature of the type HM-S carbon filaments, which were damaged by the edge of the hole in bone. This has been avoided in clinical work by the use of type A-S carbon, which is more resistant to trauma. Its drawback, a much higher level of impurities, has been overcome recently by the introduction of type XA-S fibres, which have both high purity and toughness. The handling characteristics of carbon fibres may also be improved by the use of protective coatings (Alexander *et al.* 1981). Premature failure of filamentous implants may be reduced by chamfering the sharp edges of bone holes, to reduce sharp angulation and fraying (Strover 1983). This was not practicable in the os calcis of the rabbit.

Although both carbon and polyester implants induced neotendons of adequate appearance (Figs 3 and 4), the graph of postoperative elongation of these neotendons (Fig. 7) suggests that carbon-based reconstructions are subject to progressive elongation in use. The experiences of Valdez *et al.* (1980) and Amis *et al.* (1984) suggest that this might have been avoided by prolonging immobilisation beyond four weeks. Since carbon fibres fracture at very low elongations, fracture must have occurred before six weeks in our tests, subsequent strength deriving from the soft tissues alone. The two failures at 9 and 10 weeks suggest that four weeks in plaster was barely adequate, and that patients should avoid strenuous activities after tendon implants for at least three months. The early 6 mm elongation found in polyester implants appeared to be due to tightening and re-arrangement of the Bunnell suture at the musculotendinous junction and not to implant elongation. This mode of elongation also applies to carbon implants, but there is an additional 14 mm of lengthening at six weeks; even a fraction of this would spoil the result of a ligament reconstruction.

The progressive tissue augmentation of the polyester implants, and their benign behaviour *in vivo* (Amis *et al.* 1984), produces a trend towards normal elastic behaviour (Fig. 10) and suggests that polyester-based neotendons may behave like normal tendons after about one year. In contrast, the carbon-based neotendons

behaved erratically, being less stiff at 12 weeks than at six (Fig. 9). Similar results were recorded by Aragona *et al.* (1981), whose histological work linked this phenomenon with the interval between the initial intense tissue reaction and the subsequent remodelling and deposition of tissue. The two implant failures in our series support the suggestion that carbon implants may be vulnerable between two and three months after operation.

Our main conclusions are that filamentous implants of differing but relatively inert materials can be the basis of repair of a tendon which will eventually attain normal strength, but that this may take more than six months. It seems that perhaps eight weeks' immobilisation is required to allow sufficient tissue augmentation, and that protection from excess loading is needed for an even longer period. This period could probably be reduced after ligament reconstructions by combining a bone anchorage system which will give immediate high strength with an implant material that is less vulnerable to damage.

It is with great sorrow that we record the death of Jimmy Campbell after a courageous fight against illness. That part of the work of this paper was done during a respite in that ordeal is a mark of his enthusiasm and determination.—AAA, JHM.

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