MEDICAL DEVICES: AN OVERVIEW

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INTRODUCTION

History of medical devices or healing aids in its broader terms; dates back to ancient Egyptians, Chinese and later the Greeks[1] . Egyptians used flax as sutures as far back as 2000 B.C. There are also numerous references to healing and healing aids in religious books such as the Bible and the Koran. Much later however, in 18th century magnets and their acclaimed influence on curing disease and aliments became highly popular[2]. Following the discovery of electricity later on and its perceived potential, a number of electrical and magnetic healing devices were produced for every conceivable disease and freely marketed with no scientific justifications[3]. The federal Food and Drug Act of 1906[4] and subsequent publication of the Flexner report in 1910[5] eventually put an end to sales and advertising of a large number of unregulated and unapproved medical devices. Today, medical devices have expanded to include a vast array of items generally referred to as implantable and non-implantables intended to be used on or in association with human beings. They by and large exclude products that achieve their primary intended purpose by chemical action or by being metabolised. Comprehensive lists of what is covered under medical devices is readily available[6] and will not be regurgitated in this short review.

To assist and protect the public against what is sold as a medical device, following the early history of these products; countries around the world have assigned regulatory bodies to inspect and monitor what is marketed as a medical device. In US, Food and Drug Administration (FDA) is responsible for ensuring that regulated products are honestly, accurately and informatively represented. They identify non-compliance and unlawful products and prevent their launch and/or remove uncertified products from the market.

In Europe, various European Directives introduced since 1990 have gradually replaced indigenous rules in an attempt to harmonise regulations and improve levels of protection for patients as well as assisting flow of trade. The first directive (90/385/EEC) relates to all active implantable medical devices and came into force in UK in January 1993. This has since been complemented by two other directives (93/42/EEC & 98/79/EC) extending the range to include items such as diagnostic X-ray machines and *in vitro* diagnostic (IVD) medical devices. These subsequent directives came into force in UK in January 1995 and June 2000 respectively. Medicines and Healthcare products Regulatory Agency (MHRA) is the UK responsible authority under the European Medical Devices Directive giving advice, disseminating information and ensuring full implementation. Classification scheme introduced by the Medical Devices Directive (MDD) allows manufacturers to decide whether their product is of low, moderate or high risk and directs them to act accordingly. Once EC mark of approval is granted, the product maybe freely marketed anywhere within the European Union.

Canada operates a similar system under Canadian Standard Association (CSA). To help manufacturers and health workers understand and apply standards, CSA offers product information as well as training. Japanese operate in accordance with Pharmaceutical Affairs

Law (PAL), under which all medical devices must pass safety tests and demonstrate medical efficacy before being approved, or "Shonin", to be sold in Japan[7].

Estimates of worldwide market for medical devices are not readily available given its broad diversity, however, the total market for all medical products and hospital supplies is put at $220 billions and Europe's share from this is estimated to be around $75 billion. Estimates for other countries such as Japan vary from $19 to $23 billion per annum[8]. National Health Service (NHS) in UK has a spending budget of about £6 billion or $10 billion for its medical devices per year. Given the ageing population and the increasing need for all round care, the global medical device and equipment market is expected to grow by no less than 4.6% over the next 5 years[9]. This is despite the recent and continuing economic turbulences experienced in world economy. The market continues to be led by US where demand is strongest.

The medical devices used today play a major part in every stage of human health from detecting symptoms and diagnosis to cure and general improvement in quality of life. They are becoming increasingly sophisticated whilst cutting down on hospital visits, unnecessary surgeries and prolonging material performances.

In orthopaedics for instance, replacement of movable joints such as elbows, ankles, hips and knees in patients suffering from osteoarthritis and rheumatoid arthritis has dramatically improved their quality of life and relieved them from agonising pains. Replacement of parts or whole of malfunctioning arteries in the heart as well as heart valves have now become routine hospital practices once disorder is identified and diagnosed. Slow or irregular heart beat is rectified by implanting pacemaker devices between the skin and pectorial muscle. The pacemaker generates precisely timed signals to mimic the naturally occurring heart rhythm and thus compensates for the fault. In Opthalmics, corrections of eyesight using synthetic or non-synthetic lenses have been around for a while but they are rapidly being replaced by laser powered corrective treatments. The whole lens of the eye can now be replaced by synthetic intraocular lenses when Cataract has caused damage beyond repairs. Artificial teeth fillings and permanent teeth implants made from organic and inorganic materials have drastically improved dental health and reduced trauma. Natural and synthetic sutures used for wound closures and internal repairs have undergone drastic technological improvements since their initial use by the early Egyptians. Drug delivery and localised drug release mechanisms have been built into many implantable and non-implantable devices with considerable successes.

Since medical devices can include anything from sutures to wheelchairs it is not easy to categorise the range and type of materials used. However, where direct contact and/or interaction with the human body are concerned, blanket properties such as; compatibility, lack of toxicity, non-allergic reaction as well as cost effectiveness are of major concerns. Synthetic polymers such as polyester, PTFE, UHMWPE, polyamide, polypropylene, polymethyl methacrylate, polyurethane and even PVC as well as natural polymers such as silk, collagen, alginates, chitin and keratin, to name but a few are used. Metals such as stainless steel, pure titanium and titanium alloys, cobalt based alloys and similar materials are regularly utilised in medical devices. Often materials made from more than one component or specially constructed composites are used in bone repairs and bone replacements. Carbon fibre for instance, is rigid whilst being strong, however, in conjunction with compatible resins can offer toughness, flexibility, lightness and high strength.

It is also interesting to note how ancient therapies and local remedies long dismissed as mystical, quackery and mischievous are making a comeback through scientific recognition. Leeches have been used by all civilizations as an old treatment for blood-letting, amputation, skin grafts and reattachment surgery but their use had sharply declined following the advent of antibiotics and antiseptics. However, Ricarimpex, a well established French leech breeding company has just won full approval from FDA for marketing medical leeches as a "medical device"[10]. Maggot therapy, another practice popularly used in the early 20th century to help heal wounds, faced a decline from mid 1940s when antibiotics and new surgical techniques took precedence. However, studies set up in 1989 at Veterans Affairs Medical Centre and University of California demonstrated that maggot therapy was more effective and efficient at debriding infected and gangrenous wounds than many of the other treatments commonly prescribed[11]. In January, 2004, FDA approved and allowed breeding and marketing of "Medical Maggots" as a medical device. Since February 2004, U.K. doctors have been permitted to prescribe maggots to National Health Service (NHS) patients who suffer from acute infected wounds[12]. Chitin and Chitosan, extracts from crab and lobster shells are another natural polysaccharides that had been known to ancient Chinese for their healing abilities. Although not strictly a medical device, Hemcon of Portland based in Oregon has produced bandages made from chitosan that are currently being used on injured troops based in Iraq. News from battlefield claims that they have been very effective in their almost instantaneous haemostic ability, preventing soldiers from heavy blood loss and therefore death[13]. They are also naturally antimicrobial.

Telecommunication systems and sensor probes integrated within garments and wearable products are other methods by which medical devices are making an impact on enhancing healthcare provisions around the clock. As well as monitoring bodily functions and regulating variables, these garments when fully developed will be capable of alerting and demanding attention if and when required thus minimising hospital resources and labour. Furthermore, they can play a major role in preventative ailments, health irregularities and unforeseen heart or brain disorders in apparently healthy individuals. This will be a major step towards preventative medicine and a whole new set of living standards for people around the world and a drastic reduction in governments' and private spending budgets' in this sector.

RESEARCH PAPERS

The proceeding papers in this chapter discuss the latest developments in what is generally classified as medical devices. Sutures in particular play a major role in surgery and wound closures; their material content, form of production and functional abilities are subjects of much research. The first five papers in this chapter are dedicated to this topic, the first one; authored by Jitendra R. Ajmeri reviews the world of sutures, providing statistics on worldwide production of sutures, their classifications and origins of their constituting materials. It further discusses their manufacturing methods, processes and the testing protocol they need to undergo prior to use.

Surgical operations are often synonymous with use of sutures and where sutures are involved there is normally a need for a knot or knots to secure a tie. However, type and quality of the knot plays a crucial part in achieving desirable results. E.Karaca et al in the second paper, go to some length investigating the performances of different knots on

monofilament and braided sutures. They conclude that both knot configuration and numbers play important and different roles on knot security in dry and/or wet conditions.
Jeffrey C. Leung et al recognise the problems associated with securing sutures after they have been sawn in and argue that conventional knots are cumbersome, prone to slippage/failure as well as being disruptive to the blood flow and tissue formations. A single barbed filament is proposed; where knots are eliminated. *In vitro* and *in vivo* experimental results using these sutures are presented and their potential advantages discussed.

In further reinforcement of knot's role in suture stability Pravin Patel et al stipulate that knot-pull strength is more important in sutures than straight tensile strengths and in the fourth paper describe the means by which knot strength can be improved when using resorbable monofilament copolymer of L-lactic acid and e-caprolatone [P(LLA/CL0]. Preliminary conclusions from this work suggest that molecular orientation and hence knot-pull strengths can be improved by acetone solvent treatment.

Fibrous structure and knot configuration could be a breeding ground for infection and micro-organisms during or after completion of operations, Martin King el al, in the subsequent paper discuss ways of addressing nosocomial infections specifically associated with braided sutures. It advocates the use of resorbable antibiotic coating on polyester braided sutures and reports on their *in vitro* reactions when exposed to Staphylococcus aureus. The study concludes that incorporation of an antibiotic within a resorbable coated suture can inhibit bacterial growth.

The final paper (R.Alimova et al) on this topic reemphasises the traditional use of silk as a suture material and highlights its advantages and potential benefits. It describes a new manufacturing methodology achieved by scientists from Uzbekistan in producing surgical threads made from silk to meet particularly acute local shortages for sutures.

Prostheses are manmade substitutes for body parts that have been damaged or removed and can include anything from artificial leg or breast to arterial/venous systems. Arterial/venous prostheses are often made from synthetic polymers and can be fabricated in to knitted or woven materials or injection moulded into shapes. Carmen Mihai et al discuss a new generation of vascular prostheses using circular weaving technology to fabricate made to size tubular configurations thus avoiding lengthwise seams and material breakdown. The new products are claimed to have reduced porosity, high resistance to tear and bursting as well as providing flexibility, softness and ease of handling.
Injuries to central nervous tissues damages nerve cells or neurons and prevent their axons to regenerate spontaneously. Nerve grafts or surgical transplantation from other parts of the injured individual or other people can help to repair and alleviate the damage. Bio-absorbable as well as biodegradable materials of natural or synthetic base can also be used to serve this purpose. Nilufer Yildiz introduces this topic and reviews the materials and methods used to implement these devices. The study explores the advantages and disadvantages that each of these materials can offer.

The medical device market in the UK is estimated to be worth billions of pounds and involves a large number of healthcare workers, professionals, trade associations and academia involved in research and developmental work. To coordinate and network such a vast community, Medical Devices Faraday, a government sponsored organisation has been instigated and setup. F.C. Smith in the final paper discusses the background to Medical Devices Faraday creation, reasons for its existence and more importantly its task in supporting, training and providing information to the community it represents.

REFERENCES

1 B Payne, *The body magnetic*, Santa Cruz, CA, 1990 (private publications).

2 F A Mesmer, *Between God and Devil*, by Wyckoff, James. Prentice-Hall, New York, 1975, First edition.

3 L A Geddes, 'A short history of the electrical simulation of excitable tissue including electrotherapeutic applications', *Physiologist,* 1984 **27**(1) S1-S47.

4 United States Statutes at Large (59th Cong., Sess. I, Chp. 3915, p. 768-772).

5 The Flexner Report and the Standardization of American Medical Education, Andrew H. Beck, *Brown* Medical School, Providence, RI, *JAMA.* 2004 **291** 2139-2140.

6 http://www.tremaine.fsnet.co.uk/

7 A Gross and S Lepage, 'Opportunities in Japan's Medical Device Market', Pacific Bridge, Inc. August 2001.

8 A Gross, 'Keeping up with changes in the Japanese Medical Device Market', *Medical Device and Diagnostic Industry Magazine*, 1997.

9 World Medical Market Report, Espicom Business Intelligence, 2004.

10 C Rados, 'Beyond Bloodletting: FDA Gives Leeches a Medical Makeover', *FDA Consumer magazine*, September-October 2004.

11 D G Armstrong et al, 'Maggot Debridement Therapy', *J of the American Podiatric Medical Association*, 2002 **92**(7) 398-401.

12 Press Release, Maggots on prescription, 'The World's Smallest Surgeons' Produced *by* the NHS *for* the NHS.

13 http://radio.weblogs.com/0105910/2004/08/28.html

KNOT PERFORMANCE OF MONOFILAMENT AND BRAIDED POLYAMIDE SUTURES UNDER DIFFERENT TEST CONDITIONS

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ABSTRACT

It is essential that quantitative information be available concerning the expected knot performance of a surgical suture during wound closure. In this study, the knot performance of monofilament and braided Polyamide sutures were tested by applying two different knots with two, three and four throws in dry and wet states. Instron Tensile Tester has been used to measure knot performance in this work. It was concluded that granny knot has better knot performance than square knot and once knot was secure, an additional throw did not increase the force to break as expected. It was also observed that in monofilament sutures and in wet state, more throws were needed to secure knot.

INTRODUCTION

Sutures remain the most common method of approximating the divided edges of tissue (10, 13). A suture is defined as a thread that either coapts adjacent cut surfaces of the wound or compresses blood vessels to stop bleeding (1, 10, 13). Surgical knots are recognized as important elements of surgical performance and surgical technique. During most operations many knots are made and knot tying consumes a substantial part of the duration of virtually all surgical procedures (8). When a force is applied to a suture loop, the quality of the knots is of the utmost importance because it is always the weakest link in a tied surgical suture (1, 12). When a knotted suture fails to perform its function, the consequences may be disastrous. Massive bleeding may occur when the suture loop surrounding a vessel becomes untied or breaks wound dehiscence may also follow (1, 10). Therefore, in recent years, many different kinds of sutures with different performance characteristics have been developed.

A tied suture has three components (Fig. 1). (1) Loop: The suture creates a loop of fixed perimeter secured in certain geometry by a knot. The loop compresses the adjacent surfaces by transfixing the tissue within the loop. (2) Knot: Ideally, the loop created by the suture in the tissue must be finished with a knot that advances to the wound surface and maintains tissue unification. After that is ensured, additional throws may be constructed to ensure knot security and prevent bleeding. So the knot is composed of a number of throws snuggled up against each other. (3) Ears: The ears are the cut ends of the suture. The ears act as insurance that the loop will not become untied because of slippage. Commonly, the ears are kept to 2-0 mm in length. The doctor's side of the knot is defined as the side of the knot with the ears, or the side to which tension is applied during tying. The patient's side is the side of the knot adjacent to the loop (1, 3, 10, 13).

Suture security is the ability of the knot and suture material to maintain tissue approximation during the healing process without slippage, untying and breaking. This is an important characteristic of sutures. Suture security is influenced by many factors,

including the coefficient of friction of the suture material, the size, construction, stiffness and type of knot, the number of overthrows, moisture absorption, the surgical technique, and the nature of the biological environment (7, 10).

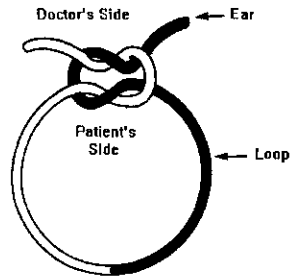

Fig 1. The components of a tied suture

Suture failure may be the result of the knot either untying or breaking. Untying results from slippage of the knot. The degree to which knots can slip is influenced by a variety of factors, including the coefficient of friction of the suture material, suture diameter, knot type, length of the cut ears of the knot, and moisture. Suture slippage can be counteracted by using more throws than are necessary for knot security, but these additional throws are time-consuming, prolonging unnecessarily the length of the operative procedure. In addition, considerably more suture is added to the wound, further damaging its resistance to infection. Therefore, we can conclude that these additional throws do not enhance the breaking strength of a secure knot. The surgeon must therefore tie a secure knot with each suture, using the fewest number of throws (6, 10, 11).

Knot breakage is the other cause of suture failure. The force necessary to break a knotted suture is lower than that required to break an untied suture. The tensile forces exerted on a tied suture are converted into shear forces by the configuration of the knot and rupture the suture. The magnitude of forces necessary to produce knot breakage varies with the size of the suture loop, the kind of suture material, and the diameter of the suture (10).

A knot stays tied due to the frictional force between the filaments comprising the knot so that both the strength and the security of the knot depend on the coefficient of friction of the suture material as well as on how the knot is tied. Braided suture materials exhibited higher values of the coefficient of friction and a slower decrease with increasing applied tension than monofilament sutures fabricated from the same material. The significance of these frictional effects can be seen from the observation that multifilament sutures, although the suture gauge, the method of tying, and the applied tension all were found to affect knot security (2, 5).

The handling characteristics of surgical sutures are one of the most important considerations in suture selection. Surgeons evaluate the handling characteristics of sutures by the number throws required for secure knot. They prefer a suture that permits two-throw knots to be easily advanced to the wound edges, providing a preview of the ultimate opposition of the wound edges. Therefore it is essential that quantitative information be available concerning the expected performance of a surgical suture during wound closure and the number of throws necessary for a secure knot must be optimised (4, 13).

The purpose of this study was to develop reproducible tests that provide practical insights into the knot performance of two different structure Polyamide sutures, commonly used in wound closure. With reproducible knot tests, we determined the influence of two different knot types, structure and size of suture on the knot security under dry and wet conditions.

EXPERIMENTAL

Material

In this study, synthetic, non-absorbable Polyamide (PA) sutures were tested. In order to see the effect of structure on knot security, monofilament and multifilament structures of Polyamide sutures were used. The sizes of these sutures were 0, 2/0 and 3/0 United States Pharmacopoeia (USP). The numbers in Tex of sutures were determined. Tex is thickness of textile yarns determined with weight (in gram) of length in 1000 meters of yarn. Table 1 shows the handling and mechanical properties of the tested sutures.

Table 1. Properties of the Tested Suture Materials

Suture	Surface Procedure	Size (USP)	Number (Tex)	Test Condition	Max. Tenacity (cN/Tex)	Max. Strain (%)	Modulus (cN/Tex)
		0	138	Dry	33,71 ± 1,53	62,15 ± 6,85	86,84 ± 3,14
				Wet	33,19 ± 1,94	74,29 ± 6,75	83,15 ± 2,96
Monofilament PA	None	2/0	118	Dry	33,78 ± 0,51	54,59 ± 456	86,92 ± 2,46
				Wet	30,82 ± 0,56	52,42 ± 1,97	89,49 ± 1,64
		3/0	48	Dry	39,29 ± 0,55	52,52 ± 2,13	127,19 ± 4,94
				Wet	34,98 ± 1,92	57,24 ± 9,02	112,64 ± 8,85
		00	130	Dry	41,54 ± 1,06	35,88 ± 3,14	145,89 ± 9,61
				Wet	37,39 ± 0,41	32,15 ± 1,14	161,37 ± 7,94
Braided PA	Wax	2/0	109	Dry	31,80 ± 1,14	30,01 ± 2,49	162,70 ± 5,36
				Wet	27,68 ± 0,78	30,46 ± 3,57	164,50 ± 8,88
		3/0	59	Dry	37,67 ± 1,11	39,10 ± 6,25	183,53 ± 9,92
				Wet	33,96 ± 0,88	36,54 ± 3,79	184,89 ± 4,54

Method

Prior to dry tests, all suture materials were conditioned under the standard atmospheric conditions (%65 humidity and 20 °C) for 24 hours. Prior to wet tests, all suture materials were conditioned in solution %0,9 NaCl for 24 hours. The instrument used to measure knot performance, shown in Fig. 2, can be mounted on the Instron Tensile Tester. Each sample placed on the tester after tying the knot, then pulled out until the knot breaks (force to break the knot) or unties (holding force of the knot).

A knot is composed of a combination of sequential throws. A single throw is formed by wrapping two strands around each other so that the angle of wrap equals 360°. The process of constructing two or more throws completes the knot. The configuration of a knot can be classified by the spatial relationship between the knot ears and the loop. In this study, square and granny knot types were used. When the right ear and loop of a two throw knot exit on the same side of the knot and parallel to each other, the type of knot is square. The left ear and loop come out from the square knot in a position that is directly opposite to that of the right ear and loop. The tension must be equal in both ears. The knot is considered a granny type if the right ear and loop cross or exit different sides of the knot. Similarly, tension is applied to the both ears (1, 10, 13).

A simple code has been devised to describe the configuration of the knot. The number of wraps involved in each throw is indicated by the appropriate arabic number.

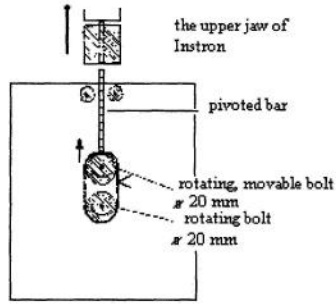

Fig 2. The instrument used to measure knot performance

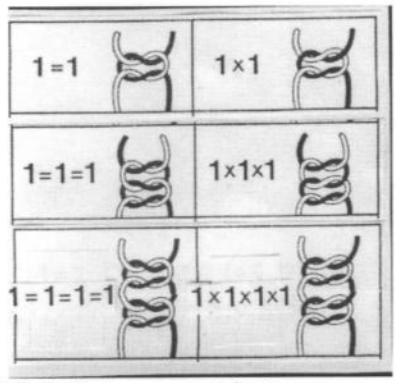

Fig 3. Knot types used for the sutures

The relationship between each throw to another, being either crossed or parallel, is signified by the symbols X or =, respectively. In accordance with this code, the square knot with two throws is designated 1 = 1, and the granny knot with two throws 1 X 1 (9). In this study, two, three and four throws for both of the knot types were formed as fig. 3.

The tests occurred in two stages. For the first stage, the knot type was formed on the suture that mounted around the bolts of the instrument. In order to produce the same knot tension in all samples, the knots were made mechanically with a dynamometer. To simulate real-life conditions, this knot tension was determined with a tension meter: To do this, ten surgeons were told to make a knot around synthetic tissue made from silicon using a dynamometer. Almost all the surgeons applied the tension of 10 N during knotting. Therefore, 10 N was designated as the knot tension. After the knot had been tied tightly enough, the ears of the knot around the two bolts were cut to leave 2 mm for the knot itself. At the second stage, the pivoted bar was clamped between the upper jaw of Instron and the knot was pulled out until it broke or untied. The gauge speed of the Instron was 200 mm/min.

The mechanical performance of the sutures were determined by the knot holding force, the force to break the knot and the number of throws required for a knot that will

reach break. All tests were repeated 10 times. The tenacity value (cN/Tex) was used for knot holding and knot breaking strengths of sutures.

RESULT AND DISCUSSION

Table 2 and Table 3 show knot performance results in dry and wet state respectively.

A-Dry Analysis

Monofilament PA

Four throws were necessary for the both granny and square knots to reach knot break regardless of the suture size.

When two and three throws were applied, knots slipped and the sutures did not reach knot break. Holding force of these two-throws knots was much lower than the holding force of three-throws knots and than the force necessary to break the square knot. This could also be concluded for all the suture sizes.

Table 2. Results of Dry Knot Performance

Suture	Size (USP)	Type of Knot	Number of Throws	Max. Tenacity (cN/Tex)	Result
Monofilament PA	0	Square	2 (1=1)	9,84 ± 0,49	Knot untying
			3 (1=1=1)	30,30 ± 3,29	Knot untying
			4 (1=1=1=1)	49,81 ± 0,95	***Knot breaking***
		Granny	2 (1X1)	10,28 ± 1,00	Knot untying
			3 (1X1X1)	22,36 ± 1,57	Knot untying
			4 (1X1X1X1)	52,60 ± 2,56	***Knot breaking***
	2/0	Square	2 (1=1)	9,74 ± 0,31	Knot untying
			3 (1=1=1)	28,36 ± 2,63	Knot untying
			4 (1=1=1=1)	48,62 ± 1,16	***Knot breaking***
		Granny	2 (1X1)	11,40 ± 0,89	Knot untying
			3 (1X1X1)	25,05 ± 1,67	Knot untying
			4 (1X1X1X1)	48,27 ± 1,96	***Knot breaking***
	3/0	Square	2 (1=1)	16,41 ± 0,58	Knot untying
			3 (1=1=1)	56,64 ± 6,46	Knot untying
			4 (1=1=1=1)	72,14 ± 2,55	***Knot breaking***
		Granny	2 (1X1)	24,62 ± 0,89	Knot untying
			3 (1X1X1)	50,71 ± 3,22	Knot untying
			4 (1X1X1X1)	75,91 ± 2,67	***Knot breaking***
Braided PA	0	Square	2 (1=1)	47,41 ± 3,50	Knot untying
			3 (1=1=1)	61,13 ± 1,19	***Knot breaking***
			4 (1=1=1=1)	64,18 ± 1,44	***Knot breaking***
		Granny	2 (1X1)	18,67 ± 0,40	Knot untying
			3 (1X1X1)	62,40 ± 4,01	***Knot breaking***
			4 (1X1X1X1)	66,45 ± 0,55	***Knot breaking***
	2/0	Square	2 (1=1)	27,61 ± 3,48	Knot untying
			3 (1=1=1)	57,11 ± 1,34	***Knot breaking***
			4 (1=1=1=1)	61,62 ± 1,08	***Knot breaking***
		Granny	2 (1X1)	13,94 ± 0,40	Knot untying
			3 (1X1X1)	55,17 ± 3,17	***Knot breaking***
			4 (1X1X1X1)	67,48 ± 1,32	***Knot breaking***
		Square	2 (1=1)	45,63 ± 5,82	Knot untying
			3 (1=1=1)	68,73 ± 2,38	***Knot breaking***

	3/0		4 (1=1=1=1)	69,57 ± 1,10	***Knot breaking***
		Granny	2 (1X1)	20,39 ± 1,53	Knot untying
			3 (1X1X1)	59,46 ± 4,58	***Knot breaking***
			4 (1X1X1X1)	75,02 ± 1,91	***Knot breaking***

*Results reported as mean ± SD

Once knot was secure and held, the force necessary to break the knot was almost the same for granny and square knots. It has essentially the same value in square and granny knots for 0 and 2/0 sizes. However, for 3/0 size the force to break the knot was much higher compare to that of 0 and 2/0 sizes.

Braided PA

Three throws were enough to reach knot break independent of the size for both knot types. When two throws were employed the knots slipped. The holding force of these two-throws was lower than the force to break the secure knot, but much higher than the same forces required in monofilament form. When three throws were applied the knot break was obtained for the granny and square knots. The force needed to break the four throws secure knot was not much higher than the force applied for three-throws knot. Here 3/0 size suture did not behave different than the rest of the sutures as explained in monofilament form.

B-wet analysis

It is very important to know the wet knot performance of sutures as they are in wet state after operation throughout healing. Therefore, wet analyses were also performed to complete the knot performance analysis.

Table 3. Results of Wet Knot Performance

Suture	Size (USP)	Type of Knot	Number of Throws	Max. Tenacity (cN/Tex)	Result
Monofilament PA	0	Square	2 (1=1)	11,36 ± 0,53	Knot untying
			3 (1=1=1)	26,40 ± 2,64	Knot untying
			4 (1=1=1=1)	47,10 ± 1,07	***Knot breaking***
		Granny	2 (1X1)	9,59 ± 0,85	Knot untying
			3 (1X1X1)	26,78 ± 2,09	Knot untying
			4 (1X1X1X1)	46,89 ± 2,15	***Knot breaking***
	2/0	Square	2 (1=1)	30,51 ± 2,03	Knot untying
			3 (1=1=1)	28,85 ± 2,35	Knot untying
			4 (1=1=1=1)	43,13 ± 2,09	***Knot breaking***
		Granny	2 (1X1)	9,06 ± 0,39	Knot untying
			3 (1X1X1)	28,77 ± 1,89	Knot untying
			4 (1X1X1X1)	46,69 ± 2,37	***Knot breaking***
	3/0	Square	2 (1=1)	9,14 ± 4,47	Knot untying
			3 (1=1=1)	55,33 ± 3,67	Knot untying
			4 (1=1=1=1)	65,16 ± 3,89	***Knot breaking***
		Granny	2 (1X1)	12,80 ± 1,39	Knot untying
			3 (1X1X1)	38,62 ± 1,89	Knot untying
			4 (1X1X1X1)	69,18 ± 4,08	***Knot breaking***
		Square	2 (1=1)	25,01 ± 3,64	Knot untying
			3 (1=1=1)	55,77 ± 1,38	***Knot breaking***

Braided PA	0		4 (1=1=1=1)	56,11 ± 2,33	***Knot breaking***
		Granny	2 (1X1)	12,06 ± 0,60	Knot untying
			3 (1X1X1)	43,93 ± 1,99	***Knot breaking***
			4 (1X1X1X1)	54,05 ± 1,62	***Knot breaking***
	2/0	Square	2 (1=1)	21,34 ± 1,89	Knot untying
			3 (1=1=1)	33,22 ± 2,25	***Knot breaking***
			4 (1=1=1=1)	48,25 ± 0,81	***Knot breaking***
		Granny	2 (1X1)	7,97 ± 1,46	Knot untying
			3 (1X1X1)	34,85 ± 3,56	Knot untying
			4 (1X1X1X1)	49,63 ± 1,26	***Knot breaking***
	3/0	Square	2 (1=1)	25,60 ± 1,10	Knot untying
			3 (1=1=1)	62,52 ± 2,33	***Knot breaking***
			4 (1=1=1=1)	63,64 ± 2,73	***Knot breaking***
		Granny	2 (1X1)	10,13 ± 0,73	Knot untying
			3 (1X1X1)	31,62 ± 1,63	Knot untying
			4 (1X1X1X1)	51,21 ± 1,06	***Knot breaking***

*Results reported as mean ± SD

Monofilament PA

Four throws were necessary for the both granny and square knot to get knot break as in dry state. The suture size seemed to have no effect on the number of throws necessary for the knot break. The force to break the knot was slightly less for wet sutures compared to dry state as water causes reduction in strength for most polymeric materials.

The holding force for two throws knot was much lower for granny knot than that of square knot and also was almost the same for three sizes in granny knot. However, the holding force for two-throws square knot was much higher for 2/0 and 3/0 sizes compared to 0 size.

Braided PA

Three-throws were enough to get knot break in square and granny knot for 0 and 2/0 sizes and in square knot for 3/0 size. However, four-throws were necessary for knot break. The size of the suture was important for braided PA sutures compared to monofilament sutures in terms of number of throws necessary for the secure knot. This was more obvious for wet braided sutures. It can be concluded that for the smallest size suture water can act as a lubricant between the filaments causing slippage resulting more throws for secure knot.

CONCLUSION

The type of knot configuration had influence on the security of only braid PA sutures in both dry and more obviously in wet state. The knot security was achieved with three-throws granny and square knot in 0 size, with a three-throws square and four-throws granny in 2/0 and 3/0 braid PA in wet state. And in contrast, four-throws square and granny knots in all sizes for monofilament PA failed by breakage in wet and dry state.

The knot security was obtained with a two-throws square knot and three-throws granny knot in braid PA for 0 size, and with three-throws square and granny knot in braid PA for 2/0 and 3/0 sizes.

In wet state four throws were needed for secure knot in braid PA sutures. This compared to dry states is higher. In dry state only three-throws were enough for knot break.

REFERENCES

1 E K Batra, D A Franz, M A Towler, G T Rodeheaver, J G Thacker, C A Zimmer and R F Edlich, 'Influence of surgeon's tying technique on knot security', *J of applied Biomaterials,* 1993 **4** 241-247.

2 J E Brouwers, H Oosting, D De Haas and P J Klopper, 'Dynamic loading of surgical knots', *Surgery Gynecology&Obstetrics*, 1991 **173**(12) 443-448.

3 C C Chu, J A Von Fraunhofer and H P Greisler, *Wound Closure Biomaterials and Devices,* Florida, CRC Press, 1997.

4 B C Faulkner, C G Tribble, J G Thacker, G T Rodeheaver and R F Edlich, 'Knot performance of polypropylene sutures', *J of Biomedical Materials Research,* 1996 **33** 187-192.

5 B S Gupta, K W Wolf and R W Posthlethwait, 'Effect of suture material and construction on frictional properties of sutures', *Surg Gynecol Obstet,* 1985 **161** 12-16.

6 E Karaca and A S Hockenberger, 'Investigating the knot performance of silk, polyamide, polyester and polypropylene sutures', *Textile Research J,* 2001 **71**(5) 435-440.

7 D P Mukherjee, 'Sutures', *Encyclopaedia of Polymer Science and Engineering,* 1987 **16** 473-487.

8 E J C Van Rijssel, J B Trimbos and M H Booster, 'Mechanical performance of square knots and sliding knots in surgery', *Am J Obstet Gynecol,* 1990 **162**(1) 93-97.

9 H Tera and C Aberg, 'Tensile strengths of twelve types of knot employed in surgery, using different suture materials', *Acta Chir Scand,* 1976 **142** 1-7.

10 J G Thacker, G Rodeheaver, J W Moore, J J Kauzlarich, L Kurtz, M T Edgerton and R F Edlich, 'Mechanical performance of surgical sutures', *The American J of Surgery*, 1975 **130**(9) 374-380.

11 N Tomita, S Tamai, T Morihara, K Ikeuchi and Y Ikada, 'Handling characteristics of braided suture materials for tight tying', *J of Applied Biomaterials*, 1993 **4** 61-65.

12 J B Trimbos, E J C Van Rijssel and P J Klopper, 'Performance of sliding knots in monofilament and multifilament suture material', *Obstetrics & Gynecology,* 1986 **68**(3) 425-430.

13 C A Zimmer, J G Thacker, D M Powell, K T Bellian, D G Becker, G T Rodeheaver and R F Edlich, 'Influence of knot configuration and tying technique on the mechanical performance of sutures', *J of Emergency Medicine,* 1991 **9** 107-113.

ENHANCING PHYSIO-CHEMICAL BEHAVIOUR OF P(LLA/CL) SUTURE BY SOLVENT TREATMENT

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ABSTRACT

Suture is polimeric biomaterial either in monofilament or multifilament form used in surgical application in medical and healthcare sector. The monofilament suture is very easy in handling and knot throw-down and also less risky in picking up infection. The paper described the experimental study on a recently developed resorbable monofilament copolymer of L-lactic acid and e-caprolactone [P(LLA/CL)] suture. For security of suturing, "knot-pull strength" is more important than tensile strength. The monofilament having higher denier makes it stiffer and difficult for easy knot throw-down. Using laboratory fabricated set-up, solvent treatment is given by immerging the fibre in the acetone: water mixture for different time periods. Birefringence measurement was made to study the change in the orientation of the fibre across its section. The tensile strength and knot-pull strength results of the treated fibres are compared to show enhancement in physio-chemical behaviour of the suture.

INTRODUCTION

Biomedical Textiles are fibrous material used in various applications in healthcare and related sectors. One of the important categories of these is implantable materials comprising **sutures**, vascular prostheses, artificial joints etc. The sutures are classified into two broad categories: absorbable and non-absorbable; of these absorbable sutures lose their entire tensile strength within 2-3 month. The absorbable surgical suture is a sterile, flexible strand prepared from collagen derived from healthy mammals, or from a synthetic polymer. The natural absorbable suture, catgut, is still being used, but it has several drawbacks, including low tensile strength and significant tissue reaction during the critical wound healing period which causes non-uniform surgical performance. The synthetic absorbable sutures are available as monofilament or as multifilament, twisted/braided[1]. The multifilament sutures are very good in handling, but have a risk of infection and significant undue friction with the tissue in contact. On other hand the monofilament suture are very easy in knot throw-down and less risky in infection. The synthetic absorbable monofilament sutures consist of strands prepared from a synthetic polymer, polymers or copolymers which, when introduced into a living organism, are absorbed by that organism and cause no undue tissue irritation. The present paper comprises study on a new absorbable monofilament surgical suture, developed from a copolymer of L-lactic acid and ε-caprolactone [P(LLA/CL)].

Poly-L-lactic acid (PLLA) fibres belong to the group of polymers that include poly glycolic acid or poly hydroxy butyric acid which because of their biodegradability to non-toxic products, may have wider practical application in medicine[2]. The PLLA is biocompatible and in the body, in course of time, it undergoes hydrolytic scission to the lactic acid, which is a natural intermediate in carbohydrate metabolism. This feature makes it suitable for use in producing the absorbable sutures, implants for orthopedic surgery or blood vessels, which finally can be replaced by the body's tissue. But it is

important that the product from PLLA should be sufficiently strong to maintain their stability during the healing process[3].

The melt-spun PLLA fibres give low tensile strength while the solution spun PLLA fibres give sufficient tensile strength depending on molecular weight and polymer concentration in the spinning solution. The rough surface of the structurised fibres spun from 10-20% solutions shows considerably higher knot strength as compared to fibres spun form dilute solution[4]. PLLA is a highly crystalline polymer and therefore its rate of degradation in a biological environment is rather slow. The crystalline domains remaining after partial degradation of the implant material give rise to a secondary inflammatory response of the surrounding tissue. For this reason, materials having a low crystallinity are preferred in surgical applications. The crystallinity of the L-Lactide homo-Polymer can be modified with other lactones and cyclic carbonates. For example copolymerisation of L-Lactic with tetrafunctional spiro-bis-dimethylene-carbonate (spiro-bis-DMC) formed very homogenous PLLA networks with higher tensile and impact strength and a lower crystallinity than the linear PLLA homopolymer. An elastic biodegradable nerve guide having inner diameter of 1.3 mm also have been synthesized from copolymer of L-lactide and 6-caprolactone(50:50,w/w)[5].

Monofilament sutures derived from p-dioxanone homopolymer (PDS II), and a copolymer of trimethylene carbonate and glycolide (Maxon) have been introduced in the market. But in general, monofilament does not handle as well as braids and therefore further improvements in handling and package memory are desired. A new ultra-pliable absorbable monofilament suture (Monocryl) has been derived from a segmented copolymer of ε-caprolactone and glycolide. The complex polymeric system having soft and hard segments in the same chain provided good handling characteristics and higher strength[1]. Recently another resorbable monofilament suture has been developed from a copolymer of L-lactide/ε-caprolactone at a molar ratio of 75:25 using octoate as a catalyst [P(LA/CL)][6]. The suture was implanted intramuscularly in the rat back using surgeon's knot to study its resorption, biodegradation, and biosafety of degradation by-products associated with chronic inflammatory reactions. The straight-pull tensile strength and knot-pull strength of the extracted sutures were measured. The histological observation was performed on the tissue surrounding the suture after explanting from rats together with the implanted sutures. The results show that the P(LA/CL) sutures lost most of their strength after 25 weeks, irrespective of the suture diameter. The tissue reaction, was found to be mild as compared to the silk sutures, which normally evolves severe tissue reaction. The suture is similar to PDS II with respect to biodegradation and tissue reaction, and hence seems to be promising as a resorbable suture.

In case of one of the P(LLA/CL) sutures, practical draw backs were observed after implanting in the human body. The one most important drawback of this surgical suture was that it was hazardous due to its nature of biodegradability. Another was its stiffness which makes it difficult for easy knot throw-down. The bending property of yarn depends upon on its diameter therefore fine suture would be preferred for easy knot throw-down. For higher knot strength, fine multifilament (of fine monofilament) was used but the multifilament was replaced by monofilament due to its risk of infection and friction with the body tissue. However, to get more strength of surgical knot using the single filament, coarser suture is being used which is difficult for easy knot throw down due to its high stiffness.

The present study is attempted to improve the practical working performance of one of the P(LLA/CL) monofilament sutures named 'Gunze' by diminishing its above mentioned drawbacks there by providing more comfort during surgery. The study is

focused on the strength and biosafety of degradation by-product associated with chronic inflammatory reactions.

EXPERIMENTAL METHODS

The surgical suture after the knot throw-down assumes the oval shape configuration and when body enzyme attacks on surface of suture to dissolve the suture from the outer layer, then at bending points(shown by arrows in Fig. 1), cracking effect occurs if the outer surface have higher orientation. The cracking effect may cause to break the copolymer into the oligomers rather than into the monomer. These oligomers may contain small crystalline domains which give rise to a inflammatory response of the surrounding tissue. For easy absorption of the suture by the human body, it is necessary to transform the copolymer into monomer so that smooth and gradual degradation of the suture takes place from outer surface to inner surface.

Fig 1 Configuration of suture after knotting

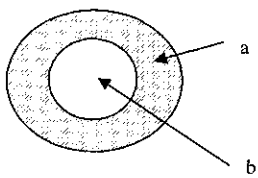

Fig 2 Cross section of suture

The suture also must have the sufficient tensile strength as well as knot-pull strength. Considering the fibre cross-section as circular, the sheath should be strong enough against the compression force along the transverse direction of fibre (Fig. 2a) while core should be strong enough against the tensile force along the direction of the fibre axis (Fig. 2b). This implies that the molecular orientation should be decreased from center to outer surface of the fibre. This can facilitate more smooth degradation along with good handling property i.e. it becomes soft due to less orientation in the outer surface. The suture thread should be easy to handle and knot throw-down but it must have the sufficient tensile strength as well as knot-pull strength.

Solvent treatment

In the attempt to achieve above mentioned mechanical properties of P(LLA/CL) monofilament suture, the experiment was designed to modify the molecular orientation distribution in the outer layer of fibre by solvent treatment. First, methyl alcohol (methylene) was used as solvent but the results were not noticeable. In the present experiment, acetone is used as solvent. The effect of solvent concentration and treatment time on the molecular orientation distribution across the fibre cross section has been studied to design the surgical suture. The solvent treatment of the P(LLA/CL) suture have been carried out in the solution of water and acetone mixture of various proportions. The suture was immersed in the solution for certain immersion time period with different immersion patterns. In case of repeated immersion treatment every time the sample was allowed to dry for a while. The details of water : acetone proportion, immersion time and immersion pattern for solvent treatment are given in the Table 1.

Table 1 Total immersion time and pattern of immersion for solvent treatment

Sr. No.	Water : Acetone	Immersion Time, Second(s)	Immersion Pattern
1	2 : 8	20 s	2 s × 10 time 5 s × 4 time 10 s × 2 time 20 s
		30 s	
2	3 : 7	20 s	2 s × 10 time 5 s × 4 time 10 s × 2 time 20 s
		30s	
		60s	
		180s	
3	4 : 6	30 s	
		60 s	
		180 s	
		300 s	
4	5 : 5	60 s	
		180 s	
		300 s	

Measurement of birefringence

The specific index of birefringence is a measure of orientation of fibre. The Polarized Microscope (Olympus BH2) is used for the measurement of birefringence of suture fibre. For all the 20 sample types, 5 specimens were considered to calculate the average birefringence value of the fibre. The values of total average birefringence and value of birefringence for each fringe position in the fibre have been measured as follows.

Average birefringence of outer and centre layers

Birefringence of fiber $\Delta n = n_1 - n_2$ where n_1 and n_2 are the refractive index of fibre in axial and transverse direction respectively. Retardation, caused by the difference of the speed of light between the two directions is a function of time difference i.e. $\Delta t = (t_1 - t_2)$.

Thus,

$$\Delta n = \frac{R}{D}$$

where R = retardation = c x Δt

c = speed of light in vacuum

Δt = time difference in axial and transverse direction

D = diameter of fibre

Birefringence value in each fringe position

The birefringence value for each position of fringe from outer surface to center of fibre also has been measured considering the thickness of each fringe. The birefringence in particular fringe is given by

$$\Delta n = \frac{R}{t}$$

where, t = thickness of the fringe

$t = 2r\ [1 - (x/r)^2]^{1/2}$

r = radius of fibre

x = distance of each fringe from center of fibre

The detail method of measuring the values of R, D, t and x is explained in a report[9]. The total average birefringence values for normal and treated fibres were compared, against 'the fringe distance from the centre'. Since the diameter of the fibre changes after the chemical treatment, birefringence (Δn) values also have been plotted against (x/r) value assuming cylindrical symmetry. The microscopic images of the normal and solvent treated sample with Water: Acetone(2:8 20s) have been shown in Fig. 3.

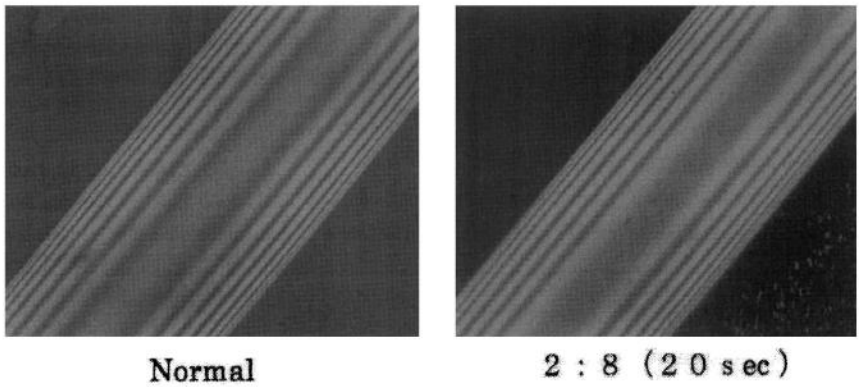

Fig 3 Microscopic images of normal and treated suture with water:acetone (2:8 20s)

Tensile strength tests

For surgical sutures, both tensile and knot strength are important. However for the security of suturing in the surgical use, knot-pull strength is generally more important than the tensile strength. Hence tensile tests for straight-pull strength as well as knot-pull strength were conducted. 20 specimens were tested for untreated (normal) suture fibre. After treating the suture fibre with acetone solvent with various treatment conditions, some selected specimen were used for the measurement of tensile and knot strength. 10 specimens in each case were tested for the fibres treated with various treatment conditions as given in Table 2.

Tensile tester 'Tensilon UTM-II-20' (Toyo Baldwin Co. Ltd., Japan) working on the constant rate of elongation principle have been used for the measurements of tensile and knot strength. For tensile strength test specimen of about 8cm length was taken and then mounted on cardboard window using suitable adhesive to get 50mm free length for gauge length. The specimen was then clamped between the upper and lower jaws of the machine. The crosshead speed was kept 20mm/s.

The specimen preparation for knot strength was some-what troublesome due to curliness of treated fibres. The fibre sample was knotted as surgeon's knot around a soft thin rubber tube resembling the human body's surgical organs (see Fig. 4). The average stress-strain curves for both straight-pull strength and knot-pull strength test results were obtained. The summary of tensile strength, knot strength and breaking elongations is given along with the graphical comparison in the results section.

RESULTS AND DISCUSSIONS

Birefringence (Δn)

After measuring birefringence values in each fringe position for all the samples, graphs were plotted putting 'Δn' against the fringe distance from the fibre center for normal and treated samples. The total birefringence values for all treated samples have been decreased as compared to that of normal (untreated) fibre. In case of sample treated for 20s with water:acetone 2:8 solution, Δn values for the surface of fibre were found lower for all type of immersion patterns, but with 3:7 solution, the reduction curve of Δn is found somewhat steep due to slight harsher effect on the surface probably due to more acetone % in the solution. In case of treatment time 30s, among the 3 samples (2:8, 3:7 and 4:6 water:acetone proportions), the birefringence values for 2:8 sample is lower than other two samples. It implies more deteriorations of fibre orientation due to high acetone % in the solution. Similar trend was seen for samples of 60s treatment time.

Table 2 Treatment parameters for strength measurements

Sr. No.	Water: Acetone	Immersion Pattern
1	2 : 8	20 s
2	2 : 8	2 s ×10 time
3	3 : 7	20 s
4	3 : 7	2 s ×10 time
5	3 : 7	60 s
6	4 : 6	60 s
7	5 : 5	60 s
8	5 : 5	180 s

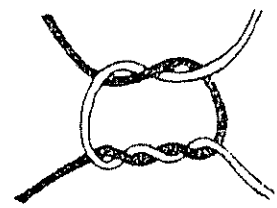

Fig 4 Surgeon's knot after throw-down

Birefringence (Δn) for treatment time of 180s and 300s reduces remarkably when concentrated solution is used for a long time. In 300s treatment time, it decreased tremendously. It was also found that in this case the deterioration was overall orientation of fibre. It is seen that there was no change in birefringence values in the center of the fibre, but it gradually decrease towards the outer surface for 20s treatment time and 2:8 water:acetone proportion, but for treating time 60s, Δn decrease drastically at the center.

As diameter of fiber changes due to chemical treatment, birefriengence values are also plotted on y-axis against x/r ratio on x-axis; for treating time 60s for the different solvent concentrations (Fig. 5a); and for the water:acetone proportion 3:7 for various treatment pattern(Fig. 5b). On x-axis, values -1.0 indicates the one outer surface, 0.0 is the centre and +1.0 indicates the other outer surface of the fibre. It can be seen from the graphs that 60s treatment time, Δn decreases at centre little for the ratio 5:5, moderate for 4:6 and in the case of 3:7 ratio, acetone attacks very severely due to higher concentration of the solvent. However for (2s×10) and 20s treatment time, Δn decreases gradually from the centre to fibre surfaces.

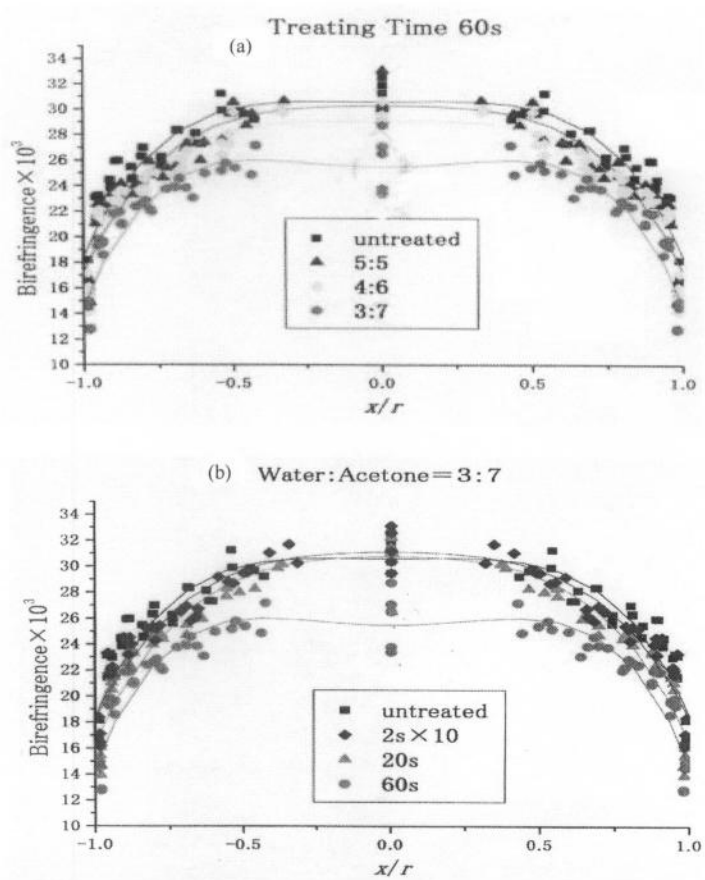

Fig 5 Birefringence across fibre section (a) Treating time 60s (b) Water : Aceton =3.7

Tensile properties

The tensile strength, knot strength, tensile elongation and knot elongation of normal and treated samples have been measured and the average results are summarised in Table 3. The tensile strength and knot strength for these samples are expressed graphically for easy comparison in Fig 6 and the tensile elongation and the knot elongation are shown in Fig 7.

It can be seen that for the cases of 2:8 (2s×10), 3:7(20s), 4:6(60s) and 5:5(60s), the tensile strength decreased and knot-strength increased. Fibre strength monotonically increases with increase in its molecular orientation. The reduction in tensile strength for above cases is mainly due to the reduction of orientation in outer surface of the fibre. In surgeon's knot due to four-fold configuration in the knotted portion and in addition rough surface created by solvent treatment, the additional friction force among the filaments is generated which enhance the breaking strength of surgical knot. The soft surface of outer layer of filament due to reduced orientation does not allow to slip, and also helps to change its shape to elliptical. This explains why knot strength increases in most treated samples. Both tensile and knot elongation also increases in most cases.

Table 3 Various tensile properties of suture

Sr. No.	Sample Type	Tensile Strength (MPa)	Knot Strength (MPa)	Tensile Elongation (%)	Knot Elongation (%)
1	Normal	325.56	273.72	38.41	39.09
2	2:8(20s)	271.41	268.23	53.08	57.39
3	2:8(2s ×10)	289.31	300.66	45.87	58.41
4	3:7(20s)	305.87	302.89	49.51	59.59
5	3:7(2s ×10)	324.25	283.68	45.44	51.31
6	3:7(60s)	235.36	223.73	47.86	71.70
7	4:6(60s)	303.03	292.27	45.32	61.84
8	5:5(60s)	297.21	305.94	40.34	58.20
9	5:5(180s)	247.68	-	49.10	-

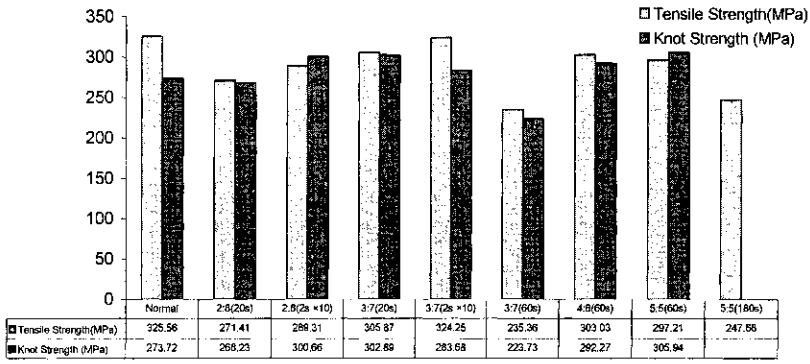

Fig 6 Comparison of tensile and knot strength of various samples

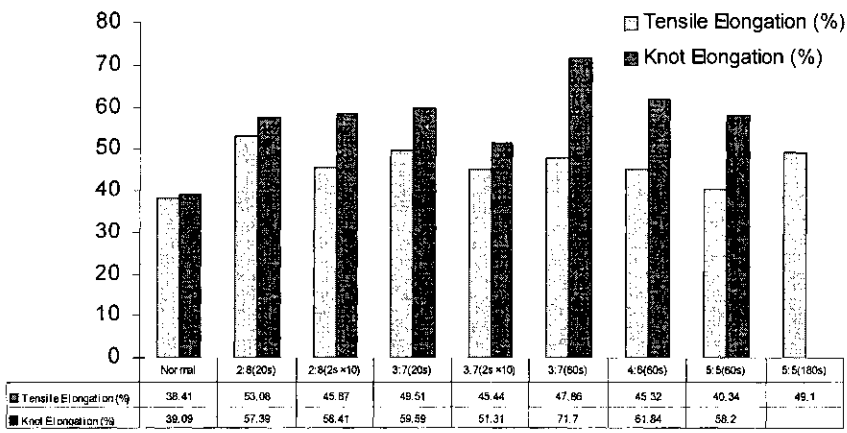

Fig 7 Comparison of tensile and knot elongation of various samples

CONCLUSION

Preliminary results show that there is a feasibility of enhancing the physio-chemical properties of P(LLA/CL) monofilament suture to get better surgical performance. Molecular orientation distribution can be controlled across the P(LLA/CL) fibre by dipping in acetone solvent and the knot-pull strength of P(LLA/CL) suture fibre can be improved by suitable treatment conditions. Thus soft and tough suture thread can be obtained in case of Poly L-Lactic Acid/Caprolactone suture by the solvent treatment.

REFERENCES

1 R S Bezwada, D D Jamiolkowski *et al.*,'Monocryl suture, a new ultra-pliable absorbable monofilament suture', *Biomaterials,* 1995 **16** 1141-48.

2 R L Kronenthal, Z Oser and E Martin, *Polymer Science and Technology* **8**, *Polymer in Medicine and Surgery*, Plenum Press, New York, 1975.

3 B Eling, S Gogolewski and A J Pennings, 'Biodegradable materials of poly (L-lactic acid): 1. Melt-spun and solution-spun fibres', *Polymer*, 1982 **23** (October) 1587-93.

4 G Perego, G D Cella et al, 'Preparation of a new nerve guide from a poly(L-lactide-co-6-caprolactone), *biomaterials,* 1994 **15** (3) 189-93.

5 S Gogolewiski, A J Pennings, 'Resorbable materials of poly(L-lactic). II. Fibres spun from solutions of poly(L-lactide) in good solvent', *J of Applied Polymer Science,* 1983 **28** 1045-61.

6 K Tomihata, M Suzuki, T Oka and Y Ikada, 'A new resorbable monofilament suture', *Polymer Degradation and Stability,* 1998 **59** 13-18.

7 Y Ohkoshi, H Shirai, Y Gotoh and M Nagura, *Sen'i Gakkaishi*,'Intrinsic birefringence of poly(L-lactic acid)' 21.

8 E Karaca, Bayraktar and A S Hockenberger, 'Investigating the knot performance of silk, polyamide, polyester and polypropylene sutures', *Textile Research J,* 2001 **5** 435-40.

9 A Jalal, *Master Degree Dissertation*, M. S. University of Baroda, India, 1999.

BARBED, BI-DIRECTIONAL SURGICAL SUTURES

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ABSTRACT

Surgical sutures are the most frequently used biomaterials for wound closure and tissue approximation. However, they rely on the surgeon's ability to tie secured knots, which is a challenging and time consuming process. Improper tying and handling can result in knot breakage or slippage, and potentially wound dehiscence. Further, the knot impedes wound healing, constricts blood flow, distorts tissue, and increases scar formation. To alleviate these problems, attempts have been made to design self-anchoring sutures. Recently, a novel knotless suture has been developed (Figure 1) in which bi-directional barbs are introduced into an absorbable monofilament suture using micro-machining techniques.

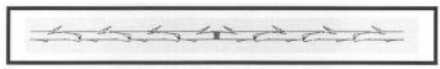


Fig. 1 - Magnified mid-section of barbed suture

This paper describes the analytical characterization of the barb geometry, and the biomechanical performance of the suture, including *in vitro* and *in vivo* wound closure testing. The former employs specialized microscopy and image analysis techniques. The latter entails tensile strength testing and apposition of tissues with a variety of stitch techniques, in comparison to commercially available sutures. These results will form the basis for further research into barb/tissue interactions and optimization of barb geometry for specific surgical applications.

INTRODUCTION

Surgical repair and conventional sutures

Each year in the U.S., some 50 million open surgical procedures are performed. This number is expected to grow to 63 million by the year 2005 (1). These procedures have one common challenge and that is the need to connect tissue. Typically, tissue connection involves closing a wound or incision across various levels of muscle, fascia, fat, or skin. Natural fibers such as cotton, silk and surgical gut were predominantly used until the development of synthetic polymers for suture use. These new polymers comprise a range of improved performance characteristics and absorption profiles.

To close a wound, sutures must be knotted. Suture failure occurs most frequently at the knot since local stresses weaken the fiber. Therefore, the US Pharmacopeia (USP) specifies minimum knot-pull tensile strength requirements for sutures. Tying knots requires time and extensive training. Because of the manner in which they are placed, conventional sutures are prone to various complications due to the presence of knots and excessive wound tension. Potential problems include:

a) Knot breakage and slippage – The more slippery the suture material, the more likely a given knot will slip (2, 3). This may lead to knot failure. Knots may also fail from improper tying or from damage caused by improper handling with surgical tools.

b) Suture extrusion or 'spitting' – A suture knot left below the skin, due to its bulk, may erupt through the wound causing infection, inflammation, and patient discomfort (4, 5). The rate of suture 'spitting' may be as high as 5%.

c) Infection – The interstices within a suture knot, as well as the many spaces and pores between the filaments of a braided suture, have been shown to offer a haven for bacteria (2).

d) Dehiscence – Wound failure at the closure site of tightly approximated wounds is primarily due to tissue pull-through. Up to 88% of suture loops in disrupted wounds may be found intact at the time of disruption (6).

e) Reduced Breaking strength and inflammation – Excessive tension has been shown to reduce wound strength by as much as 77% (7, 8, 9). Histologically, wounds closed under tension demonstrate a neutrophilic inflammatory cell infiltrate and increased tissue myeloperoxidase activity (9).

f) Ischemia and scarring – Overly taut sutures can produce pressure necrosis (10). With microangiographic examination, tightly tied sutures caused avascularity in the tissue within and around the suture loops (11). In addition to compromising wound strength, the resultant microinfarction leads to increased scarring.

Competing technologies in wound closure market

Other related devices in the wound closure market are broadly categorized as follows:

a) Staples and ligating clips – Although these can be placed with greater ease and speed, the surgeon has less feel of the pressure generated within the tissue. They are limited in the range of use and cannot replace conventional sutures in many instances.

b) Closure strips – They are used only topically. Though uses are limited due to inadequate holding strength, these non-invasive strips have high product acceptance.

c) Tissue adhesives and sealants – Dermabond™ is an example of a surface skin glue based on cyanoacrylate chemistry. Though it provides satisfactory cosmetic results, it is only indicated for uses comparable to that of a 5-0 suture (a relatively small size). Fibrin sealants are biocompatible for internal use but have many processing challenges and are of insufficient strength for most surgical needs (12).

d) Suture Tying Aids – Several devices, e.g. *Suture Assistant*™ and *Quick Stitch*,™ have been developed to facilitate suture knot tying. To eliminate the knot entirely, Axya has developed an ultrasonic system for welding suture ends together. These devices are limited by cost and extensive training requirements.

Clearly, all of the above commercial products have their own limitations and cannot totally overcome the complications and problems associated with suture knots. Thus, conventional sutures continue to dominate the tissue connection market with nearly 350 million uses each year in the U.S. alone (1).

Barbed sutures

Several self-anchoring suture concepts, particularly barbed, one-way sutures have been disclosed in U.S. Patents (13-15). Sutures with barbs in only one direction are limited by

their need to be anchored on one end. This compromises potential time savings and is unsuitable for many surgical applications. Because of these limitations, and probable manufacturing constraints, usage of these has rarely been reported (16).

Recently, a novel knotless suture has been developed (Figure 1) in which bi-directional barbs have been introduced into an absorbable monofilament suture using micro-machining techniques (17, 18). Some performance findings related to this suture has been reported previously (19, 20, 21), including strength and histopathology evaluations (20). The goal of this study was to analytically characterize the barb geometry, to evaluate the biomechanical properties of the suture, and to test its wound closure efficacy in vitro and in vivo.

MATERIALS AND METHODS

Barbed suture

While a range of suture materials, sizes, lengths, and barb geometry designs have been investigated, this paper only focuses on test results related to the suture described below.

A bi-directional barbed suture was fabricated from a monofilament fiber made of polydioxanone (PDO). The suture is 7" long and 0.45 mm in diameter (size '0' per USP). The middle 3" of the suture contains 78 barbs that are escarped into the fiber. The barbs are positioned in a spiral pattern around the circumference of the monofilament. They are divided into two equal groups that face each other in opposing directions from the suture midpoint. Extending beyond the barbed sections are unbarbed sections of the monofilament that are each 2" long. Depending on the stitch technique, a straight or curved needle can be attached to either end of these sections.

Characterization of barb geometry

The barb geometry of size '0' barbed polydioxanone (PDO) suture was characterized using an Optem Zoom 100 custom microscope with a CCD video camera with both ring and back lighting. Dimensions of the barbs – cut angle, θ, and the depth of cut, D_c (see Figure 2) – were measured from ten calibrated images and the averages were reported. This enabled the length of cut, L_c, to be calculated using the following formula:

$$L_c = \frac{D_c}{\sin(180 - \theta)}$$

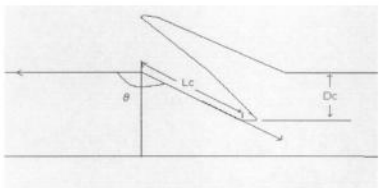


Fig. 2 – Geometry of individual barb

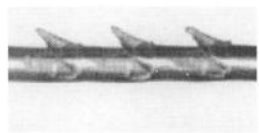


Fig. 3 – Twisted suture with barbs aligned

In addition, the spirality angle was observed and measured from the magnified images. The suture specimen was then mounted in a twisting device with one end clamped in a fixed position. Twist was imparted by rotating the other end of the specimen until the barbs

were aligned (Figure 3) and the spirality angle was zero. Then the longitudinal distance between cuts, P, was measured.

Suture tensile strength measurement

Suture tensile strength measurement was performed using a Test Resources Universal Tester, Model 200Q (Eden Prairie, MN, USA). Barbed sutures were held by two serrated jaws padded with cork gasket materials, whereas regular sutures were held by two capstan roller grips. The length of the suture specimen between the two grips was 5" (which in case of barbed sutures, contained all barbed sections). The specimen was pulled at a rate of 10 in/min until breakage occurred. The peak load was recorded as the straight-pull tensile strength. The average reading of at least 10 measurements was reported.

***In vitro* tendon repair**

Cadaveric porcine digital flexors, 0.7 cm in diameter, were transversely cut and repaired with one of the following methods: 1) traditional Kessler method (Figure 4) using 4-0 Ethibond polyester sutures (Ethicon); 2) 'switchback' method (Figure 5) with barbed '0' PDO sutures; or 3) 'finger-trap' method (Figure 6) with barbed '0' PDO sutures. Each method employed five tendon specimens.

Strength measurement of the repaired tendon specimens was performed. Each specimen was mounted between two serrated jaws on a Test Resources Universal Tester, Model 200Q. The distance between the upper and lower jaws was set to 2" and a pre-loaded of 1.5 N was applied. Each specimen was stretched to failure at a rate of 1 in/min. The force at which the tendon ends were separated by a mean of 2 mm was also recorded. The test data were captured by the WinCom software provided by Test Resources. They were further exported to Excel for analysis.

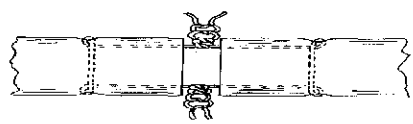


Fig. 4 – Kessler method (side view)

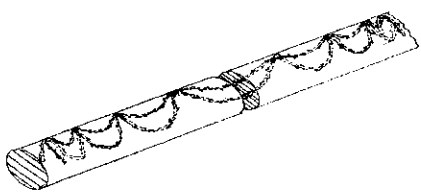


Fig. 5 – "Switchback" method

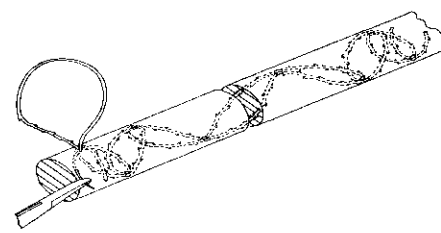
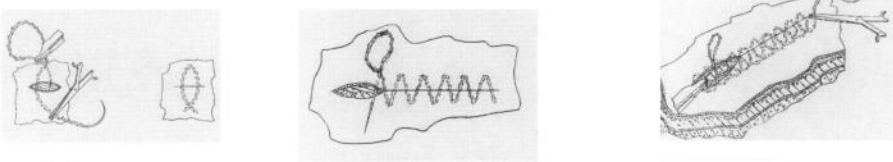


Fig. 6 – "Finger-trap" method

In vivo wound closure study

Three Mongrel dogs, each about 14 kg, were used. Each animal had seven incisions made at the thorax (twice), thigh (twice), flank, and abdominal midline and paramedian sites. The length of the incisions ranged from ½" to 4" and the depth of the incisions ranged from the dermis to muscular level. The incisions were closed with barbed sutures (PDO '0') and conventional sutures (2-0 silk, 2-0 nylon, and 3-0 PDS II) according to a premeditated randomized scheme. Three stitch methods—alpha, zigzag, and coil (Figure 7)—were used with barbed sutures. All animals were monitored daily for 14 days. At necropsy, the incisions were evaluated macroscopically.



alpha zigzag coil

Fig. 7 – Stitch methods with barbed sutures

RESULTS AND DISCUSSION

Barb geometry

Various barbed suture geometries were precisely measured using a specialized microscope and image analysis system. The ranges of the measured dimensions for the left and right directional barbs are shown in Table I.

Table I – Barb geometry measurements

Cut angle, θ	152 – 172°
Cut depth, Dc	0.08 - 0.23 mm
Cut length, Lc	0.34 - 0.59 mm
Distance between cuts, P	0.88 - 0.98 mm
Spirality angle	11 – 22°

For each suture geometry design, statistical analysis was performed on the data using a t-test with $\alpha < 0.05$. No significant differences were found between measured dimensions for the left and right directional barbs. Our results suggest that the geometry of the cuts in opposing directions is equivalent to each other. These data will now be used to develop and validate a mathematical model to predict the mechanical properties of barbed sutures, and to optimize specific design parameters for specific surgical applications. Some preliminary findings have been reported (21).

Suture tensile strength

Escarpment of barbs into a size '0' PDO monofilament, depending on the geometry design, reduced the straight-pull tensile strength approximately 45-60%, from 17.72 lb to 7.0-10.0 lb (Table II). This compares favorably with a conventional suture of equivalent size when the USP minimum knot-pull strength is taken into account. Even the weakest barbed suture of the range, with the smallest "effective" diameter, compares well with conventional 2-0 or 3-0 sutures. Since knot size and bulk frequently limit the maximum suture diameter chosen by the surgeon, barbed sutures should allow them to choose larger suture diameters when desired.

Table II – Tensile strength comparison of barbed vs. conventional sutures

Suture	Straight-pull (lb.)	USP Knot-pull (lb.)	Barbed, Straight-pull (lb.)
PDO '0'	17.72	8.60	7.0 – 10.0
PDO 2-0	11.86	5.91	NA
PDO 3-0	8.82	3.90	NA

The results of the *in vitro* and *in vivo* studies below were generated with a size '0' barbed PDO suture whose average tensile strength was 7.03 lb.

***In vitro* tendon repair**

For the repaired tendon specimens, the ultimate breaking strengths as well as the forces required to induce a 2 mm gap were determined. Typical load vs. time graphs for the three repair methods—Kessler, 'finger-trap,' and 'switchback'—are plotted in Figure 8.

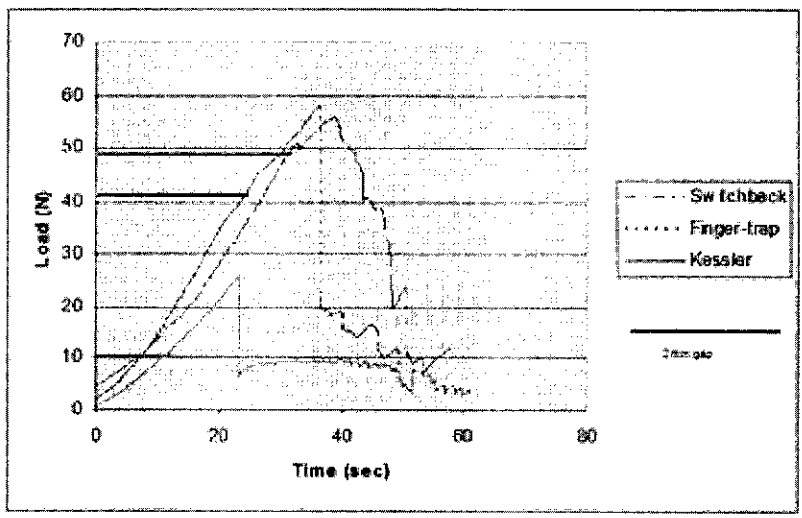


Fig. 8 – Comparison of three tendon repair methods

The commonly used Kessler method involves the placement of intricate stitch patterns to complete the tendon connection with one or two knots (22, 23). The suture knot may be placed on the outside surface of the repaired tendon where it could snag the surrounding tendon sheath and limit motion. If tied between the two ends of the tendon (as in this study), it presents a barrier between tendon sections that must appose in order to effectively heal. Furthermore the strength of the healed tendon is compromised by a gap between the ends. Kessler-type repairs resulted in a 2 mm gap (10 N) at less than half of the breaking strength (26 N).

The 'finger-trap' and 'switchback' stitch methods deployed with barbed sutures, on the other hand, not only had better holding power but also resisted gap formation. Particularly in the latter method, the force (49 N) required to induce a 2 mm gap was almost the ultimate breaking strength (56 N) of the repaired tendon! Smaller barbed sutures are expected to exhibit similar sustaining characteristics albeit perhaps at lower load values. These results suggest that the barbed sutures may significantly improve tendon repair by eliminating suture knots and minimizing disruption of the wound margin.

In vivo wound closure study

Wounds closed with size '0' barbed PDO sutures fared well compared to 2-0 and 3-0 control sutures. In various tissues, incision sizes and locations on the dogs, all incisions apposed with the barbed sutures stayed closed and appeared to heal normally throughout the observation period (14 days). No dehiscence occurred.

During the clinical observation period, three out of the six topical skin sites closed with nylon sutures experienced partial or complete suture loss, apparently due to the dogs' self-mutilation (Table III). These were treated with triple antibiotic ointment and healed secondarily without further incident. At necropsy, a small suture abscess was found in a ventral midline incision closed with barbed suture. This occurred in an animal which chewed out control skin sutures from the adjacent paramedian location, perhaps colonizing the barbed suture site with oral flora.

Table III – *In vivo* performance of barbed and conventional sutures

Suture Type	Tissue Layer	No. of Sites	Complications (No.)
Barbed Suture	Dermis	13	Suture abscess (1)
	Subcutaneum	6	0
	Muscle/fascia	5	0
Conventional	Dermis	8	Suture loss, dehiscence (3)
	Subcutaneum	6	0
	Muscle/fascia	1	0

CONCLUSIONS

A novel, self-anchoring suture, consisting of bi-directional barbs formed on a conventional monofilament suture, has been developed and its efficacy successfully demonstrated in two surgical repair models. The barb geometry of this new suture has also been characterized,

to facilitate research into optimization of design parameters for specific surgical applications.

The biomechanical performance of the barbed sutures was assessed in an *in vitro* tendon repair model and demonstrated superior wound retention when compared with a conventional suture. The effectiveness of the barbed suture in closing wounds was further manifested in an *in vivo* canine study wherein incisions in the dermis, subcutaneum, and muscle were successfully apposed and appeared to heal normally. The likelihood of improved therapeutic outcomes and cosmesis due to obviation of suture knots no doubt merits further investigation and longer term studies. Our findings suggest that a barbed, bi-directional suture may provide significantly improved strength and tissue repair in comparison to currently available sutures.

REFERENCES

1 Frost & Sullivan, Research Report: *U.S. Wound Closure Products Markets,* May, 1999, 5603-51. (www.frost.com)

2 R G Bennet, 'Selection of wound closure material', *J Amer Acad Dermatol,* 1988 **18**(4) 619-637.

3 R D Meyer, C J Antonini, 'A review of suture materials, Part I', *Comp Contin Edu Dent,* 1989 **10**(6) 360-367.

4 J J DuBois, 'A technique for subcutaneous knot inversion following running subcuticular closures'. *Mil Med,* 1992 157(5) 255.

5 D Miro, M V Julia, A Sitges-Serra, 'Wound breaking strength and healing after suturing noninjured tissues'. *J Amer Coll Surg,* 1995 **180**(6) 659-665.

6 H C Alexander, J F Prudden, 'The causes of abdominal wound disruption', *Surg, Gyn & Obstetr,* 1966 122(6) 1223-1229.

7 T Nilsson, 'Effect of increased and reduced tension on the mechanical properties of healing wound in the abdominal wall', *Scand J Plast Reconstr Surg,* 1982 **16** 101-105.

8 T Jonsson, H Högström, 'Effect of suture technique on early healing of intestinal anastomoses in rats'. *Eur J Surg,* 1992 **158** 267-270.

9 H Högström, U Haglund, B Zederfeldt, 'Tension leads to increased neutrophil accumulation and decreased laparotomy wound strength', *Surg,* 1990 107(2): 215-219.

10 L C Bartlett, 'Pressure necrosis is the primary cause of wound dehiscence', *Can J Surg* , 1985 **28**(1) 27-30.

11 M B Myers, G Cherry, 'Functional and angiographic vasculature in healing wounds', *Amer Surg*, 1970 **36**(12) 750-756.

12 Y Ikada, 'Tissue Adhesives in Wound Closure, Biomaterials, and Devices', C C Chu et al. (Ed.), CRC Press, Boca Raton, FL, 1997 317-346.
13 J H Alcamo, Surgical suture, US Patent Office, Pat No 3 123 077, 1964.

14 I Yoon, Suture devices particularly useful in endoscopic surgery and methods of suturing. U.S. Patent Office, Pat No 5 053 047, 1991.

15 H J Buncke, Surgical methods using one-way suture. US Patent 5 931 855, 1999.

16 A R McKenzie, 'An experimental multiple barbed suture for the long flexor tendons of the palm and fingers', *J Bone Joint Surg [Br]*, 1967 **49**(3) 440-7.

17 G L Ruff, Barbed bodily tissue connector, US Patent Office, Pat No 6 241 747 B1, 2001.

18 G L Ruff, Insertion device for a barbed tissue connector. US Patent 5 342 376, 1994.

19 J C Leung, G L Ruff, M A Megaro, 'Barbed, bi-directional medical sutures: biomechanical properties and wound closure efficacy study'. *2002 Society for Biomaterials 28th Annual Meeting Transactions*, #724.

20 J C Leung, S Pritt, 'Barbed, bi-directional surgical sutures: in vivo strength and histopathology evaluations', *2003 Society for Biomaterials 29th Annual Meeting Transactions*, #100.

21 P P Dattilo, M W King, J C Leung, 'Tissue holding performance of knotless absorbable sutures,' *2003 Society for Biomaterials 29th Annual Meeting Transactions*, #101.

22 J B Tang, Y T Gu, K Rice, F Chen, C Z Pan, 'Evaluation of four methods of flexor tendon repair for postoperative active mobilization'. *Plast Reconstr Surg*, 2001 **107**(3) 742-749.

23 J A Masson, 'Hand III: Flexor tendons', *Selected Read Plast Surg*, 1998 **8**(34) 1-40.

NATURAL SILK BASED SURGICAL THREADS PRODUCTION TECHNOLOGY EXPLOITATION.

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ABSTRACT

Production and processing of natural silk is one of the most complex technological processes in the textile industry. The natural silk goods produced with high sanitary-hygienic and exploitation properties are in great demand in the world and considered to be one of the most valuable among textile merchandises.

The Republic of Uzbekistan has got a powerful raw material supply and workout chain. Nevertheless, we have recently felt demand in such materials for surgical purposes. The natural silk, having high bactericidal and hygienically features and high fastness, is an ideal material for surgical threads, where biological complexion of the natural silk determines its compatibility with fabric of organism. For this purposes the thread should have the following properties:

- Smooth surface, which doesn't impair and without any drawbacks;
- High fastness either in wet and dry condition;
- Good perception towards wet-heating treatment when it is sterilized;
- Capability to retain its sterility for along period of storing;
- Low irregularity;
- Absence of oil and filthy spots.

Complying of the above listed features in stitch materials is possible when: an important mixture of raw materials is selected; the following defects such as knobs, mossiestion and oil spots are not allowed. The technology of production of the surgical threads from natural silk has recently been established and examined by the group of scientists of the Tashkent institute of textile and light industry for the production of diverse surgical threads. This method of extraction of the surgical thread has obtained a positive result from national patent agency. An experimental portion of the produced surgical threads was sterilized and has been tested in clinical conditions through several hospitals. The Ministry of healthcare of the Republic of Uzbekistan has given a certificate, which gives a license to produce and application of the surgical threads from the natural silk in medicine.

INTRODUCTION

How many operations are processed daily?

Every surgeon and doctor can tell, that the outcome of operation depends not only on skillfulness and experience, but also on the quality of stitch material, on surgical threads. Improper surgical threads can frustrate any attempts of the surgeon. That is why it has been of great importance for a long time.

It was 2000 years ago when intestinal stitch with appliance of natural threads was illustrated in one of the Chinese treatise. Description of appliance of flax by ancient Egyptians was given in Edwin Smith Papyrus, which is estimated to be of 4000 years old. Despite of the facts the exploitation of new materials was not considerable until XIX century.

Catgut threads, which are widely applied in surgery, had been developed by Helen and promoted in 1840 by Luigi Portal – professor of surgery in Pavia, and improved via chromating by Josef Lister in England in 1868. Catgut threads were one of the first self-resolving stitch materials.

Natural silk is the second widely applying stitch material. It was initially used by E.T. Kocher in 1887. The method of applying of surgical threads was perfected by W.S. Halsted later in 1913. Applying of silk as a surgical material had been mentioned in written sources of ancient Egypt. The more precise description of silk stitch had been illustrated in Indian medical source Ayurveda.

A number of researches had appeared in 40-60s dedicated to the problem of search of new surgical materials. The more great number of diverse materials, many of them might sound hilarious, had been offered, such as horse's hair, tendon threads of rats, of cats, of wale, of northern deer, kangaroo, using dogs' nerve, and human's umbilical cord. It was even a fishing wire, which was used as a surgical material. On the other hand, the defects (difficulties in extracting, their mutual reaction, mechanical features and infection possibility), of the above-mentioned threads, created obstacles for their wide implementation into surgical practice. New stitch materials began to appear since one after one and each of them claimed to be nearer to the ideal one.

A search for new materials led to establishment of a number of perspective trends. A great work has been carried out on the trends and it's still of great importance to work out the ideal stitch material which contains the following features:

- A stitch material must be of minimum harm or cause any infection;
- A stitch material must have a smooth and flat surface;
- A stitch material must not absorb the content on the wound and become a source of infection;
- The thread being considerably firm and elastic must not be volumetric and stick together with threads around

The 20th century has added more features. Nowadays, requirements to the ideal stitch material have considerably expanded and do imply the followings:

A. Optimal mechanical characteristics (which assert its ability to retain folded bend), such as, firmness, flexibility, friction factor and elasticity (for instance, the thread must stretch itself right after the operation to prevent its slitting, but, at the same time, provide certain compression)

B. Universality – opportunity to apply the thread in any operative interference.

C. Noninvasiveness – absence of sawing and tearing affect during the process of thread leading through the substance.

D. Absence of toxic, allergic, teratogenic, cancerogenic influence on organism.

E. Absence of capillarity that means its ability to absorb liquid and percolate it through fibers.

F. For self-resolving materials it's of great importance to resolve totally after completing its function without entailing of considerable changes in substances; terms of bio-hydration of stitch materials must be long-terming, than the time for formation of valuable cicatrix; products of destruction of threads must be included into metabolically processes in organism, not impacting them.

G. Sterility.

Observance of the above-given features of stitch materials might occur providing relevant selection of raw materials and preventing of the defects, such as fuzziness, oily flecks and so on.

First years of Independence of the Republic of Uzbekistan, after the frustration of the old system, when new economical links and development of democratic society

have recently begun, it has become obvious, that having a powerful raw material and manufacturing base those threads had not been produced and the Republic encountered sharp lack in stitch materials for surgical purposes.

A technology of producing of surgical threads from natural silk has been worked out by scientists of the Tashkent institute of textile and light industry for manufacturing purposes of the new range of surgical threads in Uzbekistan.

Natural silk, which contains high antibacterial and hygienic properties and, in addition, high firmness reckoned to be an ideal material for surgical threads, where biological origin of natural silk determines its compatibility with substances of living organism and implies a minimum reaction ability, which is shown below (See Fig.1).

Surgical threads produced from natural silk do provide drainage – an assigning factor of healing influence on the wound. They do not set off irritation, painful sensation, inflammatory processes on the body, they freely go through substances, pull together and keep edges of the wound in necessary position, fasten with each other without any obstacles and do not change their physical-mechanical features after sterilization.

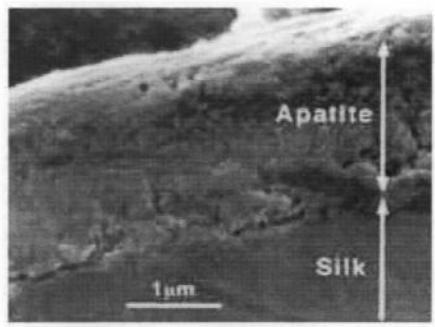

Fig. 1 Compatibility of Natural Silk with Substances of Living Organisms

The following technology for manufacturing of the range of surgical threads has been worked out:

- Raw material selection and preparation for rewinding;
- Rewinding of raw-silk from hanks into bobbins;
- Step by step mergence and twisting of threads;
- Boiling-off;
- Rewinding of surgical threads;
- Packaging into goods;
- Sterilization.

The range of produced surgical threads: 000; 00; 0; 1; 2; 3; 4; 6; 8.
New surgical thread is according to:

- Constitution – complex twisted (See Fig.2);
- Source – natural organic;
- Ability to resolve (bio-destruction) in substances of the organism – conditionally resolving.

While folding it is of great importance for surgeon to know and take into account surface features. It is commonly known, that twisted and weaved complex threads do keep the fold better than single thread and complex threads with covering. Threads, produced from silk, do play the role of "Golden standard" of reliability of the fold keeping, which have not been treated nor with wax, nor with silicon and give an opportunity to fasten a double-fold without warning that they can be unfasten.

Nowadays, an error in processing of folds can set off a development of hard and fatal deterioration. That is why acquaintance with both of new stitch materials and their features and rules of techniques in processing of stitch is necessary for every surgeon striving for optimal results of operation.

An experimental set of surgical threads has recently been sterilized and set for the clinical trial in number of Tashkent hospitals:

1. Thoracic surgery name by Vohidov;
2. Surgical clinic of the 1st –Tashkent State Medical Institute (TSMI);
3. The department of obstetrics and gynecology of the 1st -TSMI;
4. Children's stomatologic polyclinic #1, Tashkent.

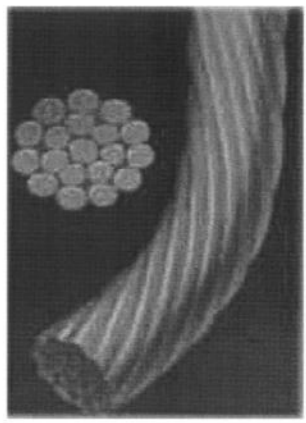

Fig 2 Surgical Thread with its Cross Section

- Volume of operations carried out – up to 1000 annually.
- Interferes carried out onto abdominal cavity and chest. Stitched substances: mucous, cardiac, muscular membrane, tendon, fascia, skin.
- Sterilization via simple scheme.
- Deterioration, caused by intolerance of stitch material is practically missing

Sterilization of surgical threads with radiation method has improved their physical-mechanical features due to more oriented position of intermolecular chain of the silk structure.

Surgical thread from natural silk entails a minimum inflammatory reaction with further step-by-step encapsulation of fiber fastening substances; extremely elastic material, high-quality structure and successful results during stitching. Applying in ophthalmology is recommended (glaucoma, scleroplasty). The product is delivered in disposable packaging with separated threads in certain length without needles.

The method of extracting of surgical threads from natural silk has received a positive result of examination conducted by Patent Agency of the Republic of Uzbekistan.

Nowadays, unsterilized threads from natural silk are solely produced by "Mosnitki" (Moscow, Russia) according to GOST 396-84 among the CIS countries. "Twisted, sterilized surgical threads" are included into nomenclature of allowed products for medical practice of the Republic of Uzbekistan and production as an item for medical purposes.

A registration certificate has been received from the Ministry of Healthcare of the Republic of Uzbekistan, which gives a right for producing and applying of surgical threads from natural silk in medical practice.

NEW GENERATION OF VASCULAR PROSTHESES ACCOMPLISHED THROUGH WEAVING TECHNOLOGIES

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ABSTRACT

The arterial system is a complex assembly of vessels that direct the blood from heart to other body organs. The identifying of biological requirements of the most complicated mechanism like the human body, connected to the possibilities of actual technique, has led to a successful accomplishment of an extremely varied range of medical products of textile materials.

At present, there are established and marketed systems of implanting arterial – venous prostheses by means of weaving, knitting and extruding technologies. Vascular prostheses accomplished by weaving technologies combine the advantages of knitted grafts with extruded grafts. In view of the athrombogenity of the artificial blood vessels walls there have been created the so called "prostheses –tight" which are obtained by using some biological substances such as: collagen, albumin, gelatin etc. At present, these products are used successfully, in surgical operations that refer to arterial-bifemural by-pass, arterial- femural by-pass; ilio-femural by-pass; femural-popitelial by-pass; axilo-femural by-pass; femural –axial by pass; femural-carotidian by-pass; subclavio-axial by-pass.

This paper presents the achievements of INCDTP correlated with the evolution at an international level, in the field of surgical implants, the level of biomedical and biofunctional performances of these implants, the terms of admissibility imposed by the international legislation in the field for the production and marketing of medical implants with textile structures.

GENERAL CONSIDERATIONS

The arterial system is a complex assembly of vessels that transport the blood from heart to the other body organs. At present there are a lot of synthetic or natural materials, that can be used as artery substitutes and that can be implanted into the human body. To make these surgical implants there is an increasing need for synthetic yarns characterized by: adequate ration resistance / weight, elasticity, flexibility, controllable porosity and permeability, surface texture compared to that of human tissue. The use of synthetic yarns caused a real revolution in medicine, introducing a different choice and approach of surgical techniques depending on the specific field of use.

The most usual operations are represented by: by-pass aorto – bifemural; by-pass aorto-bifemural; by-pass aorto-femural; by-pass ilio-femural; by-pass femuro-popitelial; by-pass axilo-femural; by-pass femuro-axilar; by-pass femuro-carotidian; by-pass subclavio-axilar.

Types of vascular implants

At present, Artero-venous prosthetics are successfully being produced and marketed using weaving, knitting and extruding technologies.

Grafts obtained by extrusion are: rigid and hard to be placed and handled by clinician; nonporous → impossibility to develop adjacent to implant

Grafts obtained by knitting present some advantages unlike those extruded, including: flexibility, softness, structure adapted to the tissue surface, porosity of velvet surfaces → adequate development of neotissue within the knitted graft

Disadvantages:

- due to great porosity , there is the risk to appear hemmorrhagies, a disadvantage that can be eliminated by performing a preliminary coagulation phase.
- high values of dilatation and elongation coefficients.

The film forming grafts present the following disadvantages: a higher rigidity that causes a difficult placement and handling; more expense; they should be kept in salty solution during storing period.

Vascular protheses accomplished through weaving technologies combine happily the advantages of the knitted grafts with those of the extruded grafts, including: reduced porosity; elimination of possibilities of ravelling when cutting in surgery room; dimensional stability; increased resistance when bursting; elastic and flexible; soft and pliability; reduced mass etc.

At present, worldwide production of vascular grafts obtained by weaving represent about 50% of total production of such products.

As a novelty woven vascular prosthesis using the gaze – technology has now been successfully produced. These fabrics can be obtained on Dobby looms or Jacquard looms. Fig. 1 presents a modality of combinig the yarns for the accomplishing of the gaze structure. This technology permitts the realisation of some products with special geometry (vascular cone and multi-branch prothesis) and that ensure unravelling of the textile materials in cutting operations.

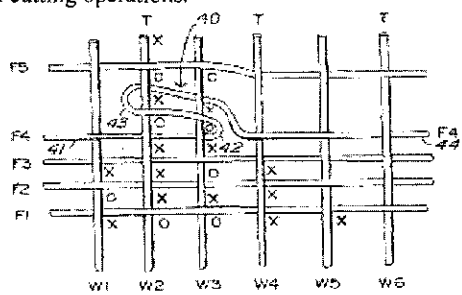

Fig 1. - Performing model of the gaze structure

Vascular prosthesis from composite yarns

The composite fibres can be obtained from compouds that are made of more than one polymers with different properties. The fibres thus obtained are called "conjugated fibres".

The compounds are made with the following purposes: attributing of a crimping (crepe) effect of an artificial blood vessel; the improving of the physical characteristics of the fibres, including : streching resistance and recovery of the streching; the forming of extra-fine fibres of the fibrillear type, the increasing of thermal adhesivity of the fibres with different melting points. In general, these composite materials are formed from polyester of superior stability in the human body and an elastic component with a

maximum elongation of 2%. The flexible component can be one of the following types: PET, PBT, copolymer polyester- polyeter or copolymer polyester – polyester.

Multilayered vascular prothesces

The multilayered vascular protheses are formed of 2 porous layers and one nonporous layer.

The graft can be tubular having the porous layers positioned inside and outside of the prothesis. This product is made of polyesteric fibres, continuously spun in electrostatic medium. The non-porous layer can contain a multitude of spun fibres in the electrostatic medium unified by blending with fibres in wet state (in a medium with solvent). The first porous layer can be formed by two blends of microfibres with different diameters, the first random oriented and the second circumferential against the longitudinal axis of the tubular graft. It is prefered that non-porous layer should be more close to the external surface than to the internal one.

Technological solutions for the improvement of the biomedical and biofunctional characteristics of the cardiovascular grafts.

"Sealed protheses" are obtained by use of certain biological substances: collagen, albumina, gellatin etc.

The impregnation of protheses of blood vessels with collagen fulfils 2 functions: it ensures the imposed porosity without necessarily using blood; it serves as support for antibiotic of which release is made step by step on a longer period of time.

The antibiotic effect is very important in the first week after the implantation, being still sensed after two weeks from the protheses implantation. In case of vascular protheses with small diameters under 6 mm, another method is used for the surface modification, and especially plasma treatments. The polymeric surfaces are usually modified by physical chemical or radiations methods, the plasma tratment technology, offering the following advantages: the modifications take place on the treated surface, the material properties remain unchanged, the modified surfaces can have any geometrical forms; the working temperature is that of the room; it offers flexibility, efficiency and protection for the environment. It has instead the disadvantage of relatively big costs and the reaction mechanism is not fully known. The accomplishing of the surface tratments with products in gel form with antibiotic content, has constituted the object of research of the company VASKUTEK – England. The purpose of accomplishing these treatments is to reduce the infections and to ensure the sealing of the vascular graft walls placed in the sanguin circuit. In the same way, the GELSOFT product was developed with the purpose of eliminating the precoaguling technological phases. Also, the gel form component has the capacity to form chemical bindings with different types of antibiotics. (i.g. rifenupin solution) with the effect of stopping the development of Staphylococcus aureus and S. epidermids.

THEORETICAL ASPECTS

The designing and accomplishment in the country of the vascular protheses by weaving technologies

The identification of the biological requirements of the most complicated mechanism in the human body, correlated with the possibilities of the current technique, has enabled the successful achievement in Romania for the first time, by a team of

specialists from The National Research – Development Institute for Textile and Leather, of a new generation of surgery implants destined for cardiovascular surgery.

The accomplishment of new vascular prostheses types has imposed high precision degree in the making of a tube with diameter included, so as to coincide with that of the blood vessel with which is to be coupled. In this respect, for the accomplishment of these products, there have been in view the following aspects: the choosing of the weaving model; the adequate choice of length density of the yarn that is to be mechanically processed; the exact establishing of the textile support density; the establishing of the yarn number in weft; the establishing of the internal diameter (D) of the graft that is to be made; the counting of the total number of weft yarns necessary for the obtaining by weaving of the tube with internal diameter (D). A special attention has been given to the selecting of the raw material destined for the manufacturing of these surgical implants, choosing the high purity filament polyesteric yarns, nonbleached optically and that are characterized by: good tolerance in human body, without determining the rejection phenomenon; inertia from the chemical point of view, maintaining for a while the functional properties; a very good dimensional stability; no allergic states or hyposensitivity; they are not cancerous; very high mechanical resistance. The designing of the tubular fabric consisted in: dimensioning of the contexture parameters (yarn fineness, binder type, thickness on systems etc); the establishing of the adjustment parameters of machines within the technological flow for its obtaining. To design the fabrics destined for the vascular prostheses, there have been taken into account the minimal requirements for the biofunctional characteristics, imposed by clinical usage field, from which there resulted the following: the product geometry imposes the use of the tubular structure and the maximal work-width; impermeability imposes main parameters of fabric designing (achievement of a structure for which there are used untwisted yarns, the densities of the two systems and the product mass, etc.)

EXPERIMENTAL ASPECTS

The tubular fabric is a double structure made of two overlayered simple structures, unified only at the two edges by transversal yarns (see Fig. 2).

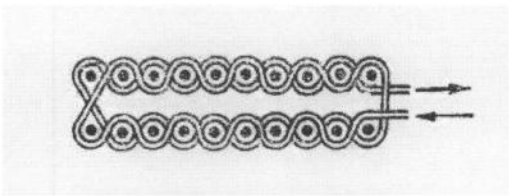

Fig. 2 - Cross section in tubular fabric

There have been introduced alternatively in the woven structure the weft elements, for the superior and inferior layer, both layers being separated.
Having in view the imposed conditions, there have been used for both of the fabric layers a plain weave, in which the scale of the two systems has to be 1:1.

The biological and physical-chemical characteristics imposed to the products in the field of clinical usage have been assured by a complex finishing process, consisting in a sequence of phases, represented by:
- scouring,

- oligomer extraction,
- neutralision,
- decontamination,
- embossing,
- thermosetting.

The obtained finished products (see Fig. 3) are characterised by biofunctional performances, represented by:
- reduced porosity;
- does not ravell when cutting in the surgery room;
- very good dimensional stability;
- Increased resistance when bursting
-adequate flexibility and elasticity
- softness and pliability
- easy to be handled and clinical placed, due to the mark yarn, inserted in the tubular structure
- reduced mass

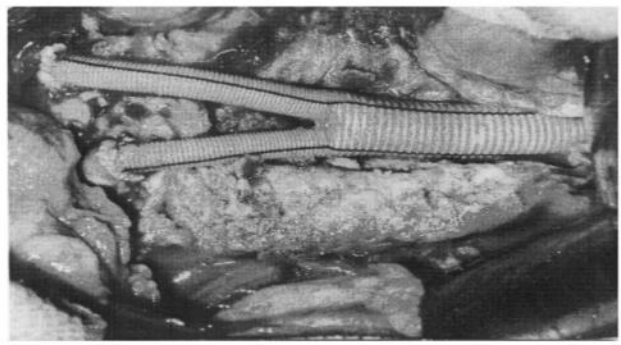

Fig. 3 - Vascular prostheses clinical placed

The level of the chemical – physical performances on aqueous extract is evaluated according to the methodology registered in the actual national normative for the category of the implantable medical devices, STAS 10914-89 is presented in comparison with the admissible limits imposed by this standard in table 1.

Table 1. Woven vascular prostheses – Physical – chemical characteristics on aqueous extract

ANALYSED PARAMETERS	NORMS	RESULTS
Aspect	Clear	adequate
Colour	colourless – faint-yellowish	adequate
Smell	lack	lack
pH ($pH_{test} - pH_{witness}$), max	±1	
Oxido-Reducing substances, ml $KMnO_4$ o,01N at 100 ml aqueous extract, max.	10	Within the limit
Content in NH_4^+, g at 100 ml aqueous extract, max.	0,0002	Within the limit
Content in CL^-, 100 ml aqueous extract, max	0,00025	Within the limit
Content in SO_4^{2-} g at 100 ml aqueous extract, max.	0,001	Within the limit

Content in in heavy metals, Pb^{2+} g at 100 ml aqueous extract, max.	0,0001	Within the limit
Content in Zn^{2+} g at 100 ml aqueous extract, max.	0,0003	Within the limit
Organic substances $E_{test} - E_{witness}$ in the range $\lambda = 220 - 360$ nm, max	0,3	Within the limit
Waste in evaporation, g at 100 ml aqueous extract, max.	0,005	0,0018

The mutagen material potential from which there are obtained the woven vascular prostheses destined for the cardiovascular surgery, was tested in collaboration with the body of accreditation of The Health and Family Ministry – The National Chemical-Pharmacy Research –Development Institute. The determination has been done by the test of micronucleus on the hematogen mouse spine. This is a test recommended by ICPEMC (International Commission for Protection against Environmental Mutagens and Carcinogens) and ensures a high degree of certainty. The test is part of the category of tests of short period and is based on the pointing out of some round citoplasmatic formations, similar in form and properties in coloring with the nucleous.

Micronucleous, known as Howell – Jolly corpusculis, can be formed of:

- acentric fragments of chromosomes that, in the lack of centrometer, can not be incorporated in the new nucleus during the cellular division;
- entire chromosomes that are left behind during mithosis due to the affectiong of the division spindle;
- complex chromosome configurations, that cause troubles during anaphasis.

Work method:

The mutagen potential of the test – product was determined by the i.p. administration of an extract made in 0,9 % NaCl solution according to the methodology recommended by ISO 10993. The extraction has been made at a concentration of 1 cm^2 /ml solution and the administration was made in dose of 25 ml/kg. As experience animals were used Swiss male mice of 3 month old, weighing 25-30 g. To determine a potential link between mithotic cycle and the forming of nucleus, the animals of each lot were sacrified at intervals, at 24 h and 48 h.

Besides the lots treated by the extract of the tested product, there were formed also witness lots:

- a witness negative lot, for the pointing out of the level of spontaneous apparition of cells with micronucleous, and which was given only saline solution,
- a witness positive lot, which received by embossing a product with known mutagen effect, cyclophosfamide.

From each lot, at times mentioned above the three animals were sacrified, from which the hematogen spine was taken from the femural bones. To avoid a subjective evaluation, the lamellas were examined at random, without knowing the lot they are part of. At the recording of micronucleus there have been into account the following criteria:

- there were considered only the micronucleus only the formations with regular form, having the same structure as main nucleus;
- micronucleus did not overpass in dimension ½ of the main nucleus;
- micronucleus presented the same colour intensity or they have been more decoloured than the main nucleus;
- unlike the nonnuclear particles, micronucleus must not present the phenomenon of light refraction;
- micronucleus counted only in the cells that have a good preserved cithoplasma;

- there were not taken into account the cells that presented more than two micronucleus, being considered artfacts. There is the possibility to confuse very often the micronucleus with granules released of mastocyts, that determines to a overestimation of number cells with micronucleus, in comparison with other coloring method (i.g. hemathoxiline- eozine), but the rating to the witness lots diminish the importance of this aspect.

Results:

- The frequency of erythrocytes with micronucleus (MN) of the adult cellular population (ENC) and young forms (EPC) is presented in table 2.
- the media obtained for micronuclear polycromathophile erythrocytes, both at 24 and 48 h do not surpass significantly the values of the negative witness lot. (p.0,05).
- the values of the positive witness lot, treated by ciclophosphamide, are significantly increased, that proves the availability of the used method.

CONCLUSIONS

In Romania for the first time we have made vascular grafts with variable diameters, by weaving technologies, in a variety of forms and configurations, without seam, avoiding the breaks of the textile material.

The obtained finished products are characterised by biofunctional performances represented by : reduced porosity, it does not ravell when cutting in the surgery room, a very good dimensional stability; an increased resistance to bursting, adequate flexibility and elasticity, softness and pliability, easiness in handling and clinical placing, due to the mark yarn, inserted in the tubular structure, reduced mass.

The physical – chemical characteristics on aqueous extract corresponds to the limits imposed by the national standards for medical devices implantable to humans.

The test regarding the mutagen potential points out that the obtained media for micronucleus polycromatophile erythrocytes, both at 24h and 48 h do not surpass significantly the values of the negative witness lot ($p > 0,05$) and corresponds to the limits accepted for the spontaneous mutagenity. (0,5 –1). The values of the positive witness lot , treated with ciclophosphamide, are significantly increased, that proves the viability of the used method.

Table 2 - Results of the micronucleus test on the mice hematogen spine

Lot	Dose /kg	no.animal	Sacrification Time, hrs	ENC	Nr.MN	%MN	Media %MN	EPC	Nr.MN	%MN	Mediaq %MN
Extract test-product	25 ml/kg	3	24	1880	0	-	0,02	1622	0	-	0,02
				2111	1	0,05		1619	1	0,06	
				1763	0	-		1498	0	-	
		3	48	1525	0	-	0,06	1364	0	-	0,02
				1975	1	0,05		1862	1	0,05	
				2106	3	0,14		1995	0	-	
white witness	0	3	24	1633	1	0,06	0,07	1570	1	0,06	0,04
				1984	2	0,1		1799	1	0,05	
				1542	1	0,06		1437	0	-	
		3	48	1833	2	0,11	0,08	1758	2	0,11	0,08
				1542	0	-		1567	1	0,06	
				1671	2	0,12		1587	1	0,06	
positive witness (ciclophasphamide)	20 mg/kg	3	24	1424	3	0,21	0,2	1514	23	1,52	1,78
				1619	4	0,24		1728	29	1,68	
				1541	3	0,19		1683	36	2,14	
		3	48	1520	6	0,39	0,43	1701	28	1,64	1,56
				1604	8	0,50		1624	27	1,66	
				1715	7	0,41		1807	25	1,38	

THE MEDICAL DEVICES FARADAY PARTNERSHIP

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ABSTRACT

The medical devices sector is critical both to the Nation's health and the UK economy. Life expectancy is increasing (0), giving an ageing population with attendant requirements for greater medical support for maintained independence. Government policy clearly aims to enhance the role of the National Health Service, requiring the effective introduction of new technological solutions.

The Medical Devices Faraday has been created with the aim of establishing a national framework, involving industry, academia, clinicians, RTOs and funding organisations, covering the Faraday Principles of research, training, people exchange and technology translation for the medical devices sector. The Partnership will complement and build on other UK initiatives and Faraday Partnerships concerned with many areas of research and technology including; biotechnology, medical electronics, textiles and plastics. Success will be achieved by focusing on delivery to the patient, through materials development, design and manufacture of medical devices and successful pre-clinical and clinical trials.

The following paper gives an introduction to the Medical Devices Faraday Partnership, outlining the structure of the Partnership, the aims and how they are to be achieved, along with deliverables already achieved. Particular attention is paid to describing what help and benefits this Faraday Partnership is intending to provide to organisations and individuals working in the textiles and healthcare fields.

INTRODUCTION

A Faraday Partnership is an alliance of organisations and institutions, which can include Research and Technology Organisations, Universities, Professional Institutes, Trade Associations and Firms, dedicated to the improvement of the competitiveness of UK Industry. Faraday Partnerships cover a wide range of disciplines and industry sectors

Each Faraday Partnership receives Core Funding consisting of:

- A grant from UK Government Departments (DTI, DEFRA, Scottish Executive etc.) to establish and operate the infrastructure of the Faraday Partnership.
- A ring fenced grant from UK Research Councils (EPSRC, BBSRC, PPARC etc.) for research projects that satisfy the twin criteria of excellent science and industrial relevance.

The Core funding establishes a platform from which the Partnership can bid successfully into other UK Government and EU funded schemes for research and technology transfer.

The four Faraday Principles, which provide the foundation for the operating models of the current 24 Faraday Partnerships, are:

- Promoting active flows of people, science, industrial technology and innovative business concepts to and from the science & engineering base and industry.

- Promoting the partnership ethic in industrially-relevant research organisations, business and the innovation knowledge base.
- Promoting core research that will underpin business opportunities.
- Promoting business-relevant post-graduate training, leading to life-long learning.

Faraday Partnerships strive to improve the interaction between the science and technology base and industry. Effective interaction requires the identification of industry needs and the subsequent synthesis of the knowledge and experience of those who can satisfy these needs. Faraday Partnerships use Technology Translators, people with broad experience of industry, academia and knowledge transfer, who can facilitate projects between Partnership members.

Faraday Partnerships are designated by a rigorous selection process following a national call for proposals. There are currently 24 Faraday Partnerships of which the Medical Devices Faraday Partnership is one of the newest, having formally started to operate in September 2002.

THE MEDICAL DEVICES INDUSTRY

The medical devices sector is critical both to the Nation's health and the UK economy. Life expectancy has increased and mortality and morbidity at all ages have declined (0), giving an ageing population with attendant requirements for greater medical support for maintained independence. Government policy clearly aims to enhance the role of the National Health Service, requiring the effective introduction of new technological solutions.

It is difficult to obtain consistent figures for the medical device market but it is substantial, with values of £2.4-2.7b for the UK (fifth largest in Europe)(0,0). The medical devices sector world-wide is heavily dependent on SMEs and the Healthcare industry sector in the UK employs over 3.3 million people (0).

Current projections show that the UK Medical Devices industry will begin to fall behind both the rest of Europe and, more particularly, the Far East and the US - and even become a net importer - unless a concerted effort is made to bring some cohesion to the UK's exploitation activities.

Increasingly, medical device development is exploiting knowledge and advances in biotechnology and bioengineering, with the UK a lead player in this aspect of development. To deliver biologically rational therapeutic interventions to the marketplace in a cost-effective and reliable way will demand tailored devices based upon a multidisciplinary approach. This will require input across the science, engineering and medicine spectrum at the macro to molecular scale. The major challenge, which has been taken on by the Medical Devices Faraday Partnership, is to focus UK efforts into a coherent approach.

MISSION OF THE MEDICAL DEVICES FARADAY PARTNERSHIP

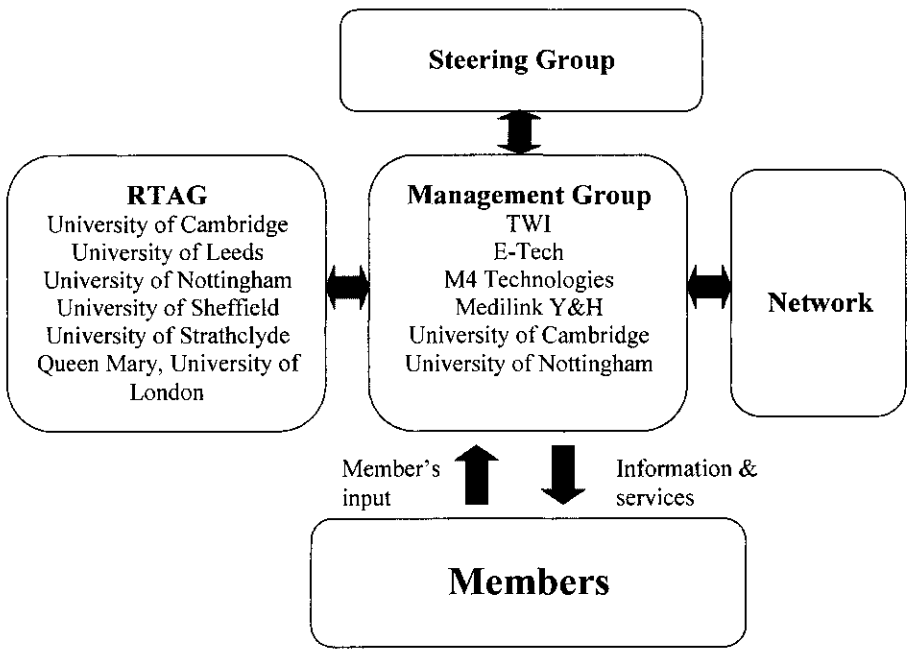
The Partnership mission is to 'provide the national infrastructure to facilitate the development and commercialisation of medical devices and biomaterials for new and improved patient treatment and increased competitiveness for UK healthcare industries'. This will be achieved by a programme of clinician and industry focused research, effective training programmes for post-graduates and students as well as major technology translation support direct to industry.

SCOPE OF THE MEDICAL DEVICES FARADAY PARTNERSHIP

The term 'medical devices', in the scope of this Partnership, encompasses all devices which are implantable or non-implantable, which act in some way on the human body and may be either permanent or disposable. This spans the range of medical disciplines from orthotics, orthopaedics and ophthalmics, to cardiovascular, urology and dentistry, and includes active as well as passive devices. It encompasses therapeutic and preventative care, as well as devices, which aid self-help for greater care in the community.

STRUCTURE OF THE MEDICAL DEVICES FARADAY PARTNERSHIP

The Faraday Partnership brings together a team of six leading academic institutions, including clinician representation, one experienced materials based RTO, two established biomaterials consultancy companies, and the national Medilink network. This powerful grouping has substantial delivery capability that, together with the industry, clinician and other research and professional organisation networks, is quickly establishing a national identity and strategic focus. The structure of the Partnership is shown in Fig. 1.



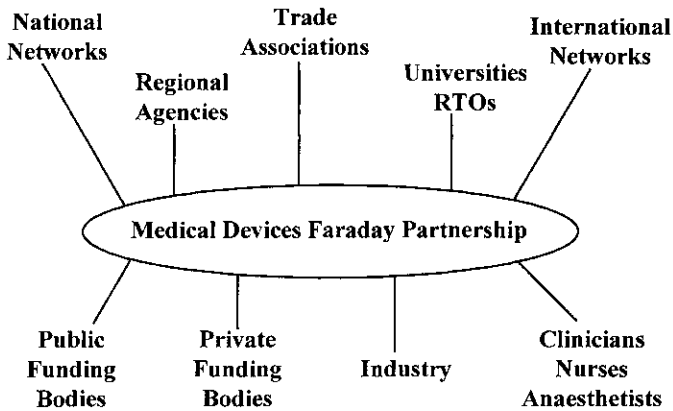
1. The Structure of the Medical Devices Faraday Partnership.

The Management Group of the Partnership is responsible for the delivery of the Partnership mission and objectives and is led by TWI, with E-Tech, M4 Technologies, Medilink Yorkshire and Humber, the University of Cambridge and the University of Nottingham.

The Research and Training Advisory Group (RTAG) consists of the six universities shown in Fig.1 and reports to and advises the Management Group on all matters related to research and training.

The Partnership reports to and is guided by a Steering Group that is chaired by Professor Alan Suggett, Smith & Nephew. This Group has a bias towards end-user representation through direct industry involvement including large organisations, SMEs, the Association of British Healthcare Industries (ABHI), together with clinical representatives from the Royal College of Surgeons and the NHS.

The term network, shown in Fig.1, is intended to show the inclusive nature of the Partnership. In order to get a true feel for the needs of the medical devices industry, the Faraday Partnership must listen to and interact with representatives from all sectors of the industry. Fig.2 gives an idea of the scope of the term network, but it is by no means a comprehensive diagram.



2. Types of organisations encompassed within the network.

MEMBERSHIP OF THE MEDICAL DEVICES FARADAY PARTNERSHIP

There is no set format for membership of Faraday Partnerships, therefore the Medical Devices Faraday Partnership has taken time to identify the membership format that would provide optimum benefit to the medical devices industry in the UK. The medical device industry is very segmented and there are a lot of regional networks that provide invaluable local support. By offering membership of the Medical Devices Faraday to members of the regional networks, the Partnership aims to help these regional networks grow and provide a link between them all to give one national voice for medical devices.

The Medical Devices Faraday has surveyed the UK to identify appropriate regional networks. These networks have been approached and the following networks have approved the idea in principle:

- Medilink Yorkshire and Humber,
- Medilink North West
- Medilink West Midlands
- Medilink East Midlands
- Medilink East

- Southern Medical Alliance
- MediWales
- Medical Devices in Scotland
- Medinet London

The first 'network of networks' meeting has been held and the process of setting up membership of the Medical Devices Faraday Partnership is under way. The list of networks included is expected to grow as other existing networks are identified and new networks are set up to cover areas previously unsupported. The intention is that people who would like to become members of the Partnership will do so via their local network. An announcement on membership of the Partnership will be made later in 2003.

RESEARCH

The Medical Devices Faraday Partnership has embarked on a research programme, funded by EPSRC and BBSRC. The Research Programme is designed to meet challenges that face the healthcare sector by developing solutions that can be exploited on a five-year horizon. This programme involved a call for outline proposals that was announced in November 2002:

Summary of requirement for outline proposals

Topics for research were identified with prime aims of:

- Enabling more effective clinical and nursing care.
- Providing models and data for an informed selection of materials, device design and manufacturing processes.
- Developing new devices for modern medical practice.

Within this framework, three priority platforms for research projects were:

1. Clinically robust devices.
    - New approaches to device design and materials development to minimise variability in clinical practice.
    - New pre-clinical testing techniques (e.g. durability) to assess performance and clinical efficacy, and the development of predictive models of clinical performance.
    - Development of cost effective devices which can replace or minimise clinician or nursing involvement, and thereby increase healthcare efficiency.
2. Improved materials and surface engineering concepts.
    - Enhanced biofunctionality in existing and new materials.
    - Validated models of device surface/tissue/cell responses at the nano-, micro- and macro-scales for rapid materials selection and device development.
    - Biologically validated materials properties.
3. Novel drug delivery systems.
    - Effective medical device/pharmacological solutions to drug delivery.
    - Customised drug release systems (tailored to patient).
    - Minimally invasive devices, therapies and surgical techniques.
    - Novel manufacturing solutions.

Other requirements for proposals included:

- Significant contributions (cash and/or in-kind) to the total cost of any project by industrial partners were required.
- Each project was expected to demonstrate multidisciplinary collaboration between academic researchers, ideally from more than one institution, and industry.
- The participation of SMEs was particularly welcomed.
- Proposals needed to demonstrate activities aimed at producing exploitable deliverables within a 3-5 year timescale, as well as scientific excellence.

Results of call

Twenty-six high quality outline proposals were received. Ten top quality proposals were selected to proceed to the next stage of full proposal. A number of other high quality proposals were also identified for submission to alternative funding sources, such as the Health Device Technology Programme.

The successful Core Faraday proposals will be announced in October 2003. The Medical Devices Faraday Partnership will appoint a Project Monitor to each successful consortium who will continue to play a supporting role, during and after completion of the research, to assist in the exploitation process.

Industrial CASE studentships

EPSRC's Industrial CASE are allocated to companies to enable them to take the lead in defining and setting up research projects with the academic partners of their choice. These CASE Studentships provide an excellent mechanism for engaging universities and industry in the Faraday Partnership.

Any UK university may be selected for one of these 'Industrial CASE Training Grants'. EPSRC funds will be allocated to universities who in turn pay the students. The Industrial Company will be expected to provide top-up funds, which should be a minimum of one third of the total EPSRC grant. More information about the specific requirements can be found on the EPSRC website:
http://www.epsrc.ac.uk/

The Medical Devices Faraday Partnership will be supporting up to six industrial CASE Studentships for awards, which are due to start in the academic year 2003-04. A call for proposals was announced in April 2003 with a deadline for submission on 4 July 2003. The results will be announced to all applicants by August 2003.

TRAINING

Training support for technical, sales and marketing staff has been identified as a key requirement if the UK is to maintain and extend its competitive position in the rapidly expanding field of medical devices. This is of particular concern in view of the fragmented nature of the UK medical device sector, in which over 90% of companies are SMEs, and it is particularly important that training provision is co-ordinated so as fully to address the industry's changing requirements.

The Medical Devices Faraday Partnership intends to develop a national skills-base to support the growth and competitiveness of the UK Medical Devices Industry by:

- Mapping of the sector's emerging skills requirements (Training Needs Analysis).
- Integration of training resources to meet emerging needs.
- Improvement of industrial and clinical relevance of training provision.

The Medical Devices Faraday Partnership has set up a Training Working Party (TWP) comprising of experts in the area of training related to medical devices. The TWP is performing a Training Needs Analysis and is using the results of this survey to formulate a proposal for new training activity. The aim of the training programme will be to develop a national skills base to support the growth and competitiveness of UK companies, using information from the ongoing Training Needs Analysis to provide guidance on the effective use of existing resources and on the development of new capabilities where appropriate.

Details of the programme of work will be made available when the programme has been formulated and agreed for funding by the Management Group later in 2003.

TECHNOLOGY TRANSLATION

This term captures a range of activities that facilitates the commercialisation of innovative research and also communicates industry and clinical needs to the research community. As part of its technology translation programme, the Medical Devices Faraday Partnership will provide:

- Direct support to industry through Technology Translators.
- Technology Roadmaps.
- Audits/Reviews e.g. Product and Process Reviews and Intellectual Property Reviews.
- Information on market trends.
- Events and networking opportunities.
- New business ventures.
- Specialist interest groups.
- Technology Watch to pick up on emerging opportunities and threats.

FRAMEWORK 6

The first call for proposals under the European Community Framework Programme was launched on 17th December 2002. The Medical Devices Faraday Partnership addressed the areas relevant to its field, and responded to specific proposals into this area. Under the Nanotechnologies and Nanosciences thematic area, a call for proposals addressing 'the interface between biological and non-biological entitities' was particularly relevant to biomaterials and implantable devices. The Medical Devices Faraday Partnership was involved in a consortium bidding into a Network of Excellence in this area.

For the thematic area of Information Society Technologies (IST), The Medical Devices Faraday took a more active role. Firstly, in participating in a submission for an Integrated Project related to implantable sensing devices able to measure certain specific properties required for respiratory, urology and cardiovascular needs. Secondly, a Network of Excellence was developed, in combination with the EPPIC Faraday Partnership, to provide a visionary network able to support medically related Integrated Projects, that was to communicate and respond to the ambient intelligence. This provided a far-reaching opportunity for Europe to lead in new healthcare developments with the body detecting its own needs and communicating with external support systems to administer therapeutic treatments.

New calls for further research will be launched in late 2003/early 2004, when biomaterials has been identified as a key theme for new activities. Interest in collaboration in future calls would be welcomed.

CONCLUSIONS

The Medical Devices Faraday Partnership is up and running and ready to help with the technology and commercialisation of medical devices. For more information about the Partnership or help and advice in the field of medical devices please visit our webpage:

http://www.medical-devices-faraday.com

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REFERENCES

1 'The Age Shift – Consultation Document and Task Force Report.' April 2002. http://www.foresight.gov.uk

2 Foresight: Health and Ageing Population Panels. http://www.foresight.gov.uk

3 M Siebert, 'Giving the industry a face', *Medical Device Technology*, Dec 2001, 36-37.

4 ABHI. 'A Competitive Analysis of the Healthcare Industry in the UK', CoMap II 2003.
http://www.abhi.org.uk/servlet/dycon/ztabhi/abhi/abhi/en/abhi/Communication_Publications_CoMapII_index

5 EPSRC Healthcare Sector Brief, April 2002, ISBN 1-899371-04-4.

THE POTENTIAL FOR LOCALISED DELIVERY OF ANTIBIOTICS FROM BRAIDED SURGICAL SUTURES

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ABSTRACT

In an attempt to address the problem of nosocomial infections associated with braided sutures, we have proposed the incorporation of a resorbable antibiotic coating. The suture's coating has been engineered to release its active ingredients locally during the first few days following surgery. Two broad spectrum antibiotics with superior Gram-positive activity, moxifloxacin and clindamycin were selected and each incorporated at two levels of loading into a resorbable epsilon-caprolactone glycolide copolymer which was then coated onto polyester braided sutures. In vitro exposure experiments to phosphate buffered saline (PBS) were undertaken at room temperature for a period of 5 days in order to determine the duration and efficacy of the released antibiotic against a standard strain of *Staphylococcus aureus*. Results from zone of inhibition tests demonstrated that both antibiotics provided effective prophylactic bacteriostatic properties for periods of up to 2 days. As might be expected, the higher drug loadings resulted in larger zones of inhibition and more prolonged inhibitory effects. In comparison to the clindamycin, the moxifloxacin showed greater antimicrobial activity against *Staphylococcus aureus*.

INTRODUCTION

It is estimated that suture related infections affect over 1 million patients each year in the US alone, with associated healthcare costs of $2.5 billion [1]. Since most nosocomial surgical wound infections occur at the time and site of incision, concern over the introduction of organisms, such as *Staphylococcus aureus* and *Staphylococcus epidermidis*, into the interstices of a braided suture preclude their use in high-risk procedures, such as cardiovascular surgery, inguinal hernia repair, bowel surgery and operations involving immunocompromised patients. In fact surgeons often prefer monofilament sutures despite their inferior handling and knot security properties [2].

Several studies have shown that activated macrophages release a large number of secretory products when they adhere to implanted biomaterials, which promote bacterial adhesion [3]. As a result, the biological response to biomaterials increases the likelihood of chronic infection in surrounding tissue. In an attempt to address this problem prophylactically, this study proposed to evaluate the feasibility of incorporating an antimicrobial agent into a braided suture by means of an absorbable coating. A recently developed epsilon-caprolactone-co-glycolide coating for multifilament surgical sutures (Poly-Med Inc., Pendleton, SC, USA) is a potential candidate. It differs from other commercial resorbable coatings because it has a significantly faster rate of resorption [4]. This coating, however, has yet to be incorporated with an antimicrobial agent.

In the past zone diffusion studies have been used to determine the antimicrobial behaviour of surgical implants, such as catheters [5,6]. In this study we proposed to show whether or not such a method is conducive to determining the pharmacokinetic profile of suture materials.

OBJECTIVES

This study was planned to determine whether covering a braided suture with a resorbable coating containing an antibiotic would provide a possible solution for the problem of high risk wound infections. In particular, the objective was to use a modified *in vitro* zone diffusion assay with a standard test micro-organism, *Staphylococcus aureus,* to evaluate the bacteriostatic properties of the sutures coated with two different antibiotics, each at two loadings, before and after being exposed to a phosphate buffer solution (PBS) for periods of up to 5 days.

MATERIALS AND METHODS

Materials

A size 3 polyester braided suture having a diameter of 0.66 ± 0.01 mm and a linear density of 220 tex (2.2 mg/cm) was braided on a 16 carriage Herzog tubular braider. In addition to the 16 ends of 11 tex flat multifilament yarn, two additional core yarns were included inside the braided structure. The polyester yarns were drawn and heat set from 15 tex partially oriented yarn (POY) supplied by Guilford Mills (Fuquay-Varina, NC, USA)

After scouring, five suture samples were coated with 5% by weight of a resorbable epsilon-caprolactone-co-glycolide polymer. The antibiotics selected for the study were moxifloxacin (Bayer Inc., Toronto, ON, Canada) and clindamycin (Pharmacia Inc., Kalamazoo, MI, USA) both with superior broad spectrum Gram-positive activity. Two of the samples had 0.1% and 1.0% of moxifloxacin by weight of suture added to the coating, while two other samples had 0.85% and 8.5% of clindamycin added. These two loadings corresponded to approximately 10x and 100x the minimum inhibitory concentration (MIC) value per cm length. One control sample of the coated suture had no active ingredient and another control had no coating.

Methods

After sterilizing in isopropanol, samples of the five coated sutures were exposed to a series of 0.1M phosphate buffered saline (PBS) solutions (pH 7.4) at 21±1°C for a period of 1 hour and then at daily intervals up to 5 days. At the end of each period the buffer solution was replaced and the antimicrobial activity of the suture and the buffer solution over the previous period were determined by a semi-quantitative modified NCCLS agar dilution procedure in which the 1 cm suture fragment and a 10µl aliquot of PBS were spotted onto Mueller Hinton agar plates previously plated with the inoculum suspension of the test organism, *Staphylococcus aureus* (ATCC 25923), cultivated in the same growth medium. *Staphylococcus aureus* is known to be the most widely pathogenic organism found in suture infections [1].

The plates were incubated at 35°C for 24 hours prior to measuring the bacteriostatic effect in terms of the diameter of the inhibition zones (Fig. 1 & 2). The mean and standard deviation of 9 replicates at each time point are shown in Fig. 3 & 4. These zone diffusion measurements were dependent on the rate of release of the drug through the agar and the generation time of the organism. An estimate of the concentration values for the zone diameters was derived from a calibration curve of zones for known concentrations of each drug (Fig. 5). These estimated concentrations were then compared with the minimum inhibitory concentrations (MIC) for moxifloxacin and clindamycin (0.030 and 0.488 μg/ml respectively).

RESULTS AND DISCUSSION

The results of this study demonstrated that the incorporation of an antibiotic within a resorbable coating did provide effective bacteriostatic properties for a limited period. The results of the sutures with moxifloxacin at a loading of 0.1% gave a zone only at 1 hour, whereas at 1.0%, zones were obtained at both 1 hour and 1 day (Fig. 1). In comparison, the sutures with 0.85% clindamycin gave zones at 1 hour and 1 day (Fig. 2), and with 8.5% loading the zones were still evident at 2 days. Overall, clindamycin produced significantly larger zones than moxifloxacin, and as expected, the higher loadings gave the larger zones. So effective antibacterial activity was observed for the 8.5% clindamycin suture and surrounding buffer for up to 2 days (Figs. 3 & 4). As expected, the coated suture sample that contained no drug gave no zone of inhibition for any exposure time.

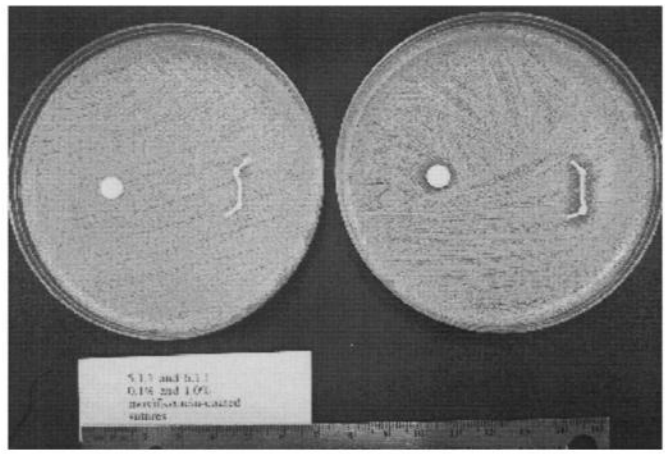

Fig. 1: Zones of inhibition for 0.1% moxifloxacin (left) and 1.0% moxifloxacin (right) coated sutures after 1 day in PBS buffer. For each plate the 1 cm suture fragment was on the right and the aliquot of exposed PBS buffer was spotted in the left.

Fig. 2: Zones of inhibition for 0.85% clindamycin (left) and 8.5% clindamycin (right) coated sutures after 1 day in PBS buffer. For each plate the 1 cm suture fragment was on the right and the aliquot of exposed PBS buffer was spotted in the left.

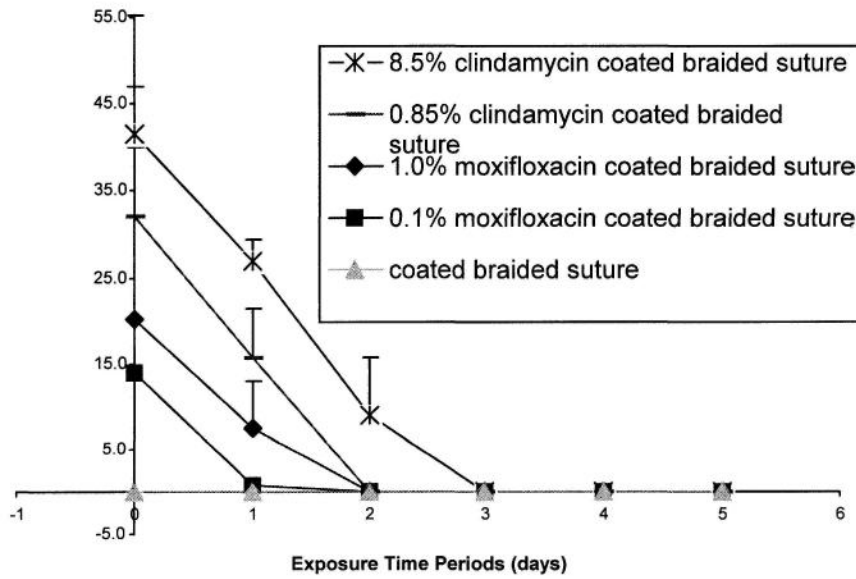

Fig. 3: Mean zones of inhibition of sutures against *S. aureus* after 0-5 days of exposure to PBS buffer.

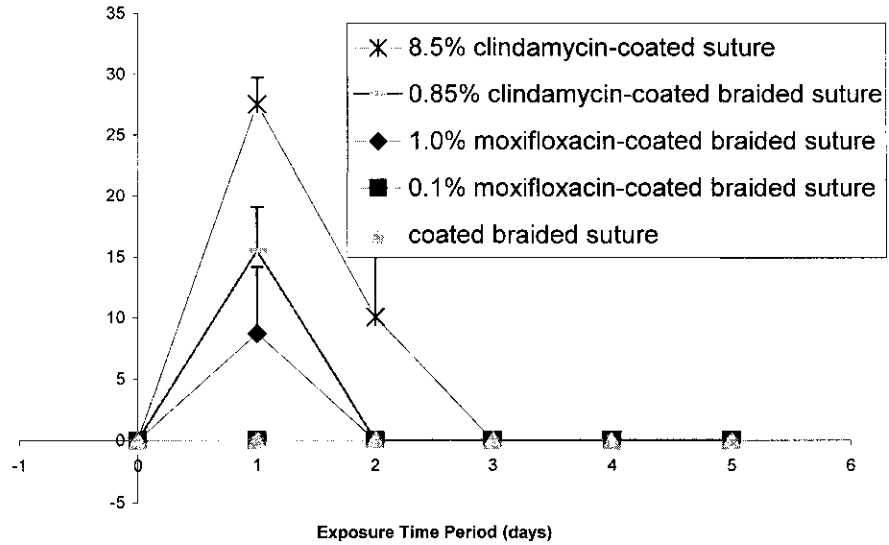

Fig. 4: Mean zones of inhibition of PBS buffer against *S. aureus* after 0-5 days of exposure to sutures.

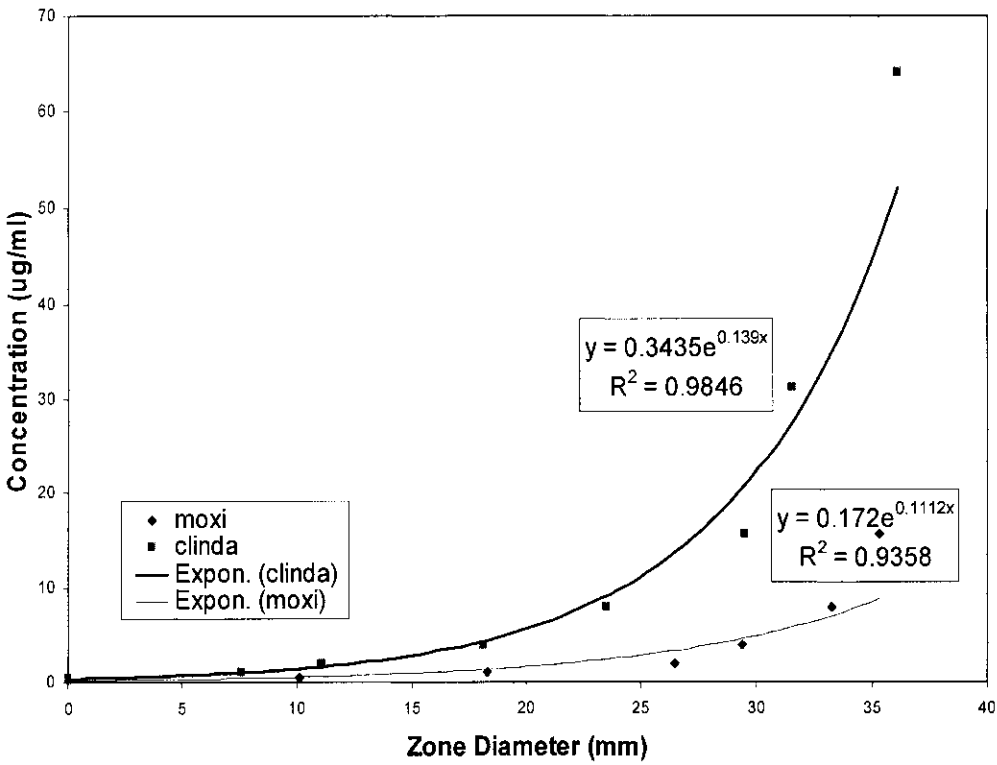

Fig. 5: Calibration curve of zone diameter against known concentration of antibiotic.

The release of the drugs from the coated sutures followed an exponential release curve for each type of antibiotic (Fig. 5). The MIC values measured for moxifloxacin and clindamycin against the standard test organism were in agreement with the values quoted in NCCLS guidelines. The measured MIC results per cm of suture showed that the moxifloxacin exhibited relatively more activity than the clindamycin under the conditions used for this semi-quantitative zone of inhibition assay [7].

CONCLUSIONS

This *in vitro* study has demonstrated that the incorporation of an antibiotic within a resorbable coated suture can inhibit bacterial growth both on the suture and in the surrounding fluid for a clinically relevant period of 2 days when prophylactic treatment is most effective. The zone of inhibition tests resulted in significant differences amongst the drugs and the loading dosages. Higher loading resulted in larger zones of inhibition and more prolonged inhibitory effects. While the clindamycin sutures achieved larger zone diameters, the standardized curves demonstrated that the amount of drug released was associated with a lower inhibitory value. The conclusion that moxifloxacin showed greater antimicrobial activity against *S. aureus* has since been confirmed by a scanning electron microscopy adhesion study.

Other broad spectrum antibiotics are also being considered as potential candidates for this application. Work is continuing to determine whether or not the coating process effects the activity of the antibiotic, and whether or not the addition of the coating influences the handling characteristics, mechanical properties, ease of knot tying and knot security of the suture.

ACKNOWLEDGEMENTS

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REFERENCES

1 D C Speller and H Humphreys, 'Hospital acquired infection', *Microbiology and Microbial Infections, 9th Ed.*, L Collier, A Balows & M Sussman (Eds.), Arnold & Oxford University Press, London, 1998.

2 R S Haluck, W E Richenbacher, J L Myers, C A Miller, A B Abt and J A Waldhausen, 'Results of aortic anastomoses made under tension using polydioxanone suture', *Ann Thorac Surg*, 1990 **50** 392-395.

3 C T Laurencin and H Elgendy, 'The biocompatibility and toxicity of degradable polymeric materials: implications for drug delivery', *Polymeric Site-Specific Pharmacotherapy,* AJ Domb (Ed.), John Wiley & Sons, New York, 1994, 27-46.

4 S W Shalaby, US Patent Office, Pat No 5 773 563, June 1998.

5 L M Vercaigne and G G Zhanel, 'Antibiotic-heparin lock: in vitro confirmation of antibacterial activity', *Can J Hosp Pharm*, 2000 **53** 193-198.

6 R O Darouiche, 'A comparison of two antimicrobial-impregnated central venous catheters', *New Eng J Med,* 1999 **340** 1-8.

7 G G Zhanel, A Walkty, L M Vercaigne, J A Karlowsky, J Embil, A S Gin and D J Hoban, 'The new fluoroquinolones: a critical review', *Can J Inf Dis,* 1999 **10** 207-238, 1999.

SURGICAL SUTURES : THE LARGEST TEXTILE IMPLANT MATERIAL

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ABSTRACT

There are numerous opportunities for textiles and related polymeric studies in the discipline of biomaterials science. Biomedical textile products begin with raw materials (polymers) which have been specifically constructed and treated for medical application. Common applications of biomaterials include orthopedic hip and knee replacements, intra-ocular lens, dental implants, heart valves, contact lenses, vascular grafts, blood bags, catheters, surgical gowns and drapes, sutures, drug delivery devices, adhesion prevention, wound dressings, membranes, tendon, and ligaments, etc.

The use of sutures which is one of the most common practices in the medical field and thus has direct effect on a great majority of the world's population. Thousands of years ago the Egyptians were among the first to use fibrous materials as biomedical devices using naturally derived materials for wound closure. With modern advances in synthetic filaments Sutures are the most common implant material today. The suture market currently exceeds $ 1.3 billion annually. Since the beginning of surgical history (5000 – 3000 BC), Sutures have been used as the means of repairing damaged tissues, cut vessels and surgical incisions.

By definition; a suture is a thread that either replicates and maintains tissues until the natural healing process has provided sufficient level of wound strength or compresses blood vessels in order to stop bleeding.

The United States Pharmacopoeia (USP), European Pharmacopoeia (EP) and British Pharmacopoeia (BP) are the official compendium for the suture industry. They sets standards and guidelines for suture manufacture. Suture sizes are given by a number representing diameter ranging in descending order from 10 to 1 and then 1-0 to 12-0, 10 being the largest and 12-0 being the smallest at a diameter smaller than a human hair.

Sutures can be classified into one of two groups, absorbable and non absorbable. For surgical sutures the predominant areas of concern are strength, capillarity, sliding and positioning of knots, knot security and handling characteristics. The recent focus of suture research has been on improving the structure of the braids (two recently proposed products are spiral and lattice – braided materials), reducing the difference in the elongation properties between the core and the sheath yarns, using finer denier filaments in the sheath yarns and improving knot security and performance by exposing a two-throw square knot to laser beam energy.

INTRODUCTION

Applications of biomaterials include orthopedic hip and knee replacements, intra-ocular lens, dental implants, heart valves, contact lenses, vascular grafts, blood bags, catheters,

surgical gowns and drapes, sutures, drug delivery devices, adhesion prevention, wound dressings, membranes, tendons, and ligaments[1,2].

Thousands of years ago the Egyptians were among the first to use fibrous materials as biomedical devices using naturally derived materials for wound closure. Since the beginning of surgical history (5000-3000 BC), sutures have been used as the means of repairing damaged tissues, cut vessels, and surgical incisions. As time passed, a variety of suture materials have been used: flax, hair, linen strips, pig bristles, grasses, mandibles of pincher ants, cotton, silk, the gut of an animal, nylons, polyesters, metals. The earliest use of gut can be traced back to the ancient Greek physician Galen. The first suture material specifically for surgical use was Joseph Lister's Catgut which he introduced in 1860. In all ancient Ayurvedic literature "vagbhata kaumudi" (AD 700) the use of dried sheep intestine is advised for ligatures in surgical operations. The eighteenth century brought the use of buckskin and silver wire, and the nineteenth brought the ability to chemically alter the properties of gut. By the twentieth century, cotton and treated natural materials had come to be the most widely used materials for suturing. Then came the era of synthetic absorbable sutures in 1931 with the production of Polyvinyl alcohol. A coated polyamide fibre - the Supramid was developed by BASF in 1939. In the fifities polyester was introduced into surgical practice. At the same time sterilisation of surgical suture material by irradiation was developed. In the second half of the sixties polyglycolic acid was processed and introduced in surgical practice in 1970 and 1971[3].

SUTURES

Sutures are probably the largest group of devices implanted in humans and can be used in skin, muscle, fat, organs, and vessels. Although they seem to be of small concern to the medical community, few devices have been made of so many different materials. The suture market currently exceeds $ 1.3 billion annually.

By definition : a suture is a thread that either approximates and maintains tissues until the natural healing process has provided a sufficient level of wound strength or compresses blood vessels in order to stop bleeding.

The United States Pharmacopoeia (USP), European Pharmacopoeia (EP) and British Pharmacopoeia (BP) are the official compendium for the suture industry, which sets standards and guidelines for suture manufacture. Suture sizes are given by a number representing diameter ranging in descending order from 10 to 1 and then 1-0 to 12-0, 10 being the largest and 12-0 being the smallest at a diameter smaller than a human hair[3]. The

Table 1 :Suture Sizes

USP	Diameter	Tensile Strength (surgeon's knot)
4-0	0.2mm	7.5N
3-0	0.3mm	12.3N
2-0	0.35mm	19.6N
0	0.4mm	22.3N
1	0.5mm	37.3N

more 0s characterizing a suture size, the smaller the resultant strand diameter (eg, 4-0 or 0000 is larger than 5-0 or 00000). The smaller the suture, the less tensile strength the strand possesses, see table 1. Sutures can be attached to different profiled needles of varying lengths and can have profiles: round (taper) body, taper cut, conventional cutting and reverse cutting.

CLASSIFICATION OF SUTURES

Absorbable (the assimilated type) sutures

Have ability to be "absorbed" or decomposed by the natural reaction of the body to foreign substances. They are used internally and are designed to lose strength gradually over time by chemical reactions such as hydrolysis. It is important to note that not all absorbable sutures have the same resistance level to absorption, but each can be formulated or treated in order to obtain a desired decomposition rate[3] and excreted in urine or faeces, or carbon dioxide in expired air[11,14,19].

Nonabsorbable (the non-assimilated type) sutures

Are not dissolved or decomposed by the body's natural action and generally used for closing cutaneous or oral incisions where the suture can be easily removed[3].

USP classification

Class I - Silk or synthetic fibers of monofilament, twisted or braided construction
Class II - Cotton or linen fibers or coated natural or synthetic fibers where the coating contributes to suture thickness without adding strength
Class III - Metal wire of monofilament or multifilament construction

MANUFACTURING OF SUTURES

Ideal suture characteristics

Sterile All-purpose; Minimal tissue injury or tissue reaction; Nonelectrolytic; Noncapillary; Nonallargenic; Noncarcinogenic; Ease of handling ; Holds securely when knotted (no fraying or cutting); High tensile strength; Favorable absorption profile; Resistance to infection. Unfortunately, no single material can provide all of the characteristics. Under different situations and given differences in tissue composition throughout the body, the requirements for adequate wound closure will require different suture characteristics. Suture materials can be monofilament or many filaments twisted together, spun together, or braided. They can also be dyed, undyed, coated, not coated.

Other suture characteristics

The following terms describe various characteristics related to suture material:
Capillarity - Extent to which absorbed fluid is transferred along the suture
Elasticity - Measure of the ability to regain original form and length after deformation
Knot-pull tensile strength - Breaking strength of knotted suture material
Knot strength - Amount of force necessary to cause a knot to slip (related to the coefficient of static friction and plasticity of a given material)
Memory - Inherent capability of suture to return to or maintain its original gross shape (related to elasticity, plasticity, and diameter)

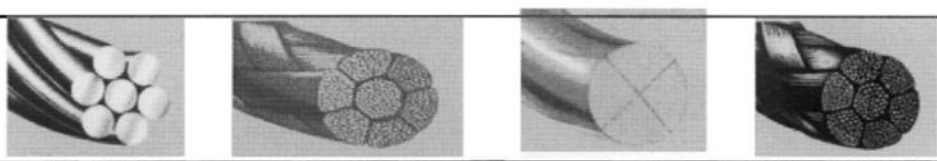
Plasticity - Measure of ability to deform without breaking and to maintain a new form after relief of the deforming force
Pliability - Ease of handling of suture material; ability to adjust knot tension and to secure knots (related to suture material, filament type, and diameter)
Sutures are manufactured with a wide variety of materials, "see Table 2"[3,4,6,9,16,20,21] and Figure 1.

Process

Mutli-filament yarn - twisted on a common ring twister- braiding a number of twisted yarn ends into a strand - stretching along with application of coating - testing of quality parameters - storing the spools for the needle attaching operation - cutting the braided yarn to desired lengths, generally from 150 mm to 900 mm - dipping one end of the suture about 25 mm in a polymer solution and placing in an oven for curing (makes the end stiff and easier to handle while attaching the needles) - sterilization and packaging. Generally, the

Table -2 : Classification of Sutures

Suture Materials			
Mono – filament		Multi-filament	
Absorbable	Non-absorbable	Absorbable	Non - absorbable
Surgical Gut - Plain and Chromic	Polyamide	Polyglycolic Acid	Surgical Silk
Collagen -Plain and Chronic	Polypropylene	Polyglactin 910	Surgical Linen
PDS	Polyester		Polyester Braided
	Stainless Steel		Polyester Braided Coated
			Polyamide Braided
			Stainless Steel
			Cotton



STEEL SUTURE | BRAIDED & COATED PGA SUTURE | CATGUT SUTURE | SILK SUTURE

Figure 1

completed suture is rolled into a circle of 25 mm in diameter and then placed in a small package and sealed and are ready for marketing[15].

General

Monofilament and multifilament sutures behave very differently in surgery. Monofilaments, which pass easily through living tissue and generate little frictional

resistance, contain no pores or interstices that might harbor infectious organisms. Multifilament sutures, on the other hand, are frequently coated to reduce frictional drag and damage to tissue, to fill the interstices between fibres, and to ease the repositioning of already-tied knots. Special processing steps have been developed for monofilaments to reduce the stiffness and memory without adversely affecting the strength. Braided or twisted multifilament sutures are inherently less stiff than monofilaments of comparable strength[11]. Un-coated braided sutures may provide increased resistance to passage through the tissues and may harbour bacteria within the braid. Natural materials generally result in greater tissue reactivity.

NATURAL ABSORBABLE SUTURES

Catgut chromic and plain

Composed primarily of collagen and are sold as plain, chromic, and mild chromic surgical sutures. Catgut Plain is manufactured from uniform, twisted strands of collagen packed in hydrating fluid containing isopropanol & water.

Catgut are Chromicized by treatment with Chromium salts in trivalent form to provide longer tensile strength retention than catgut plain and with oxidised pyrogallol (oxidised pyrogallic acid) to prolong its resistance to absorption, "see table 3". Range gauge sizes 8-0 to 3 (U.S.P) [11]; Sterilized by Ethylene Oxide Gas[10] excellent handling property and easy to knot, even in depth[3].

Table 3 : Absorbable Sutures - Surgical Gut

Type	Plain Catgut	Chromic Catgut
Tissue reaction	Within 24 hrs	3 days
Tensile strength loss	15 days	20-40 days
Absorption rate	6- to 120 days	60 to120 days

Table 4 a : Synthetic absorbable sutures[10, 11,12]

Type of suture	Braided & coated polyglycolic acid	Polyglactin	Polyglyconate
Introduced in	1967	1974	1985
Filament structure	Braided	Monofilament	Monofilament
Colour	clear or green	clear or violet	clear or green
Tissue reaction	-	low and slightly less than that of PGA	comparable to that of polydioxanone
Tensile strength loss	over 81% of the initial tensile strength is retained on 14th day	slightly greater than that of PGA	greater than that of polydioxanone
Rate of absorption	90-120 days	60-90 days	180-210 days
Method of absorption	Hydrolysis	Hydrolysis	Hydrolysis
Coating	Polycaprolactone, Calcium Stearate, N-Laurin & L-lysine.	Polyglactin 910 (Vicryl; Ethicon) is coated with polyglactin 370	-

SYNTHETIC ABSORBABLE SUTURES

Polyglecaprone

The newest one (Monocryl; Ethicon): introduced in 1993; very pliable despite its monofilament nature; handling and knot strength are excellent; highest tensile strength among all absorbable monofilament sutures; Complete hydrolysis occurs by 90-120 days, "see table 4a and 4b".

Table 4b : Synthetic absorbable sutures

Type of suture	Dexon	Vicryl	PDS
Basic polymers	Homopolymer of glycolide	Copolymer of glycolide and lactide (90% and 10%)	Homoploymer of polydioxanone (PDS)
Filament structure	Braided polyester	Braided polyester	Monofilament
Colour	Self-coloured or dyed green	Self -coloured or dyed violet	Translucent and violet - dyed
Tissue reaction	Minimal	Minimal	Minimal
Tensile strength loss	30 days	32days	56 days
Absorption rate	90 days	60-90 days	180 days
Method of absorption	Hydrolysis	Hydrolysis	Hydrolysis

NATURAL NON-ABSORBABLE SUTURES

Silk suture

Table 5 : Natural non-absorbable sutures

Type of Suture	Silk	Linen	Cotton
Origin	Cocoon of silk worm larvae	Flax	Hairs of seed of cotton plant
Basic material	Keratin protein of hair and silk	Cellulose	Cellulose
Suture formation	Braided round a core and wax coated	Fibre is twisted to form suture	Fibre is twisted to form suture
Tissue reaction	High	Similar to silk	Similar to silk
Tensile strength	Least of any suture material	Gains 10% strength when wet	Weaker than linen
Handling properties	Best	Well	Not so good as silk
Knotting	Easily and securely	Excellent	Easily

Made of filament of natural silk of 20-22 denier; a special degumming process removes extraneous material amounting to 30% of the original volume of raw silk.
Scientifically twisted in its finest form to make suture down to only 0.02 mm in diameter for opthalmic & microsurgery, high risk of infection due to high capillarity.

Cotton suture

Bleached with hydrogen peroxide and subsequently coated (finished or glaced) with starch and wax[11] , see table 5.

SYNTHETIC NON-ABSORBABLE SUTURES

Monofilament non-absorbable sutures

Polyurethane Polyether Suture: greater handling, knot security, lack of memory, smoothness & optimal tissue reaction.
Polybutylene terephthalate (PBT) Suture : popular due to strength and smooth surface[10].
Surgical Stainless Steel Wire: (iron-chromium-nickel-molybdenum alloy), monofilament, (known as fixation wire[11]), multifilament or braided, class III metallic suture. Used primarily in orthopedic, neurosurgical, and thoracic applications, "see table 6".

SUTURE DESIGN AND QUALITIES THAT INFLUENCE IT

Consideration to be taken in the manufacture and use of sutures are properties such as stress- strain relationship, tensile strength (and rate of retention), bending stiffness, intrinsic viscosity, wettability, surface morphology, degradation, thermal properties, contact angle of knots, and elasticity[17,18,19].

Suture application includes the knotting or tying off of the suture ends. Knotting causes a severe decrease of strength, break occurs most frequently at the knot. Bending stiffness is a critical which is affected by the shape of the fibre and the modulus, linear density and

Table 6 : Synthetic non-absorbable sutures

Type of Suture	Polyesters	Polyamides	Polypropylene
	Terylene	Nylon	Prolene
Suture formation	Chemically extruded from polymer and braided to form suture	Chemically extruded from polymer in monofilament form	Chemically extruded from purified and dyed polymer in monofilament form
Tissue reaction	Low	Low	Low
Tensile strength	High, retained indefinitely	Loss of 25% after two years	Extremely high
Handling properties	Good if PTFE coated	Stiff for handling	Good
Knotting	Easily and securely	Low security than terylene	Secure knotting
Passage through tissues	Tendency to cut through tissues	Easy due to low coefficient of friction	Slides easily due to low coefficient of friction
Availability	Eyeless needled sutures	Braided sutures also	Sutures and mesh
Other information	Teflon or PTFE coating for smooth surface	Stiffness is reduced by fluid addition in packets	Less thrombogenic, inert Non-biodegradable

specific gravity of the material. Stiff suture may not allow knotting and thus may be of little use in wound management. As some sutures will be blood contacting, the suture should not cause blood clotting and, as with any biomedical device, should behave favourably with surrounding tissues[3,5].

Suture failure usually occurs in one of the three ways : breaking of sutures, tearing of tissues, or untying of knots and may cause herniation of the incision or complete wound disruption. The risk of failure can be reduced by selecting the proper suture material and size, by using careful sewing technique, and by tying a knot of the proper configuration to keep it secure[11].

TESTING

Sutures are tested immediately after removal from their sterile packages without drying or conditioning.Diameters of sutures are measured using a gauge of the dead-weight type with a presser-foot 12.7 +/- 0.02mm in diameter. The diameter of each strand is measured at three points corresponding roughly to one-fourth, one-half, and three-fourths of the strand length.

For knot pull breaking strength test is based on the surgeon's knot and the shape of the clamps that fix the suture at jaws. The strength is defined as the maximum load recorded before failure, in kilogram force. Needle attachment tests are also done in a similar manner to the knot pull breaking strength tests.

Research test areas concerning the suture performance are breaking strength and elongation-to-break, Young's modulus, knot security, viscoelastic properties, tissue reaction and cellular response, cellular enzyme activity, suture metabolism, chronic toxicity, teratologics, mutagenicity and carcinogenicity, suture allergenicity and immunogenicity, and cell cultures.

CURRENT ADVANCES AND RESEARCH FOR SURGICAL SUTURES

For surgical sutures, the predominant areas of concern are strength, capillarity, sliding and positioning of knots, knot security, and handling characteristics. The recent focus of suture research has been on improving the structure of the braids (two recently proposed products are spiral and lattice-braided materials)[7], reducing the difference in the elongation properties between the core and the sheath yarns, using finer-denier filaments in the sheath yarns and improving knot security and performance by exposing a two-throw square knot to laser-beam energy[5].

REFERENCES

1 K K Leonas, *Biomedical Textiles*, June 7 1996.

2 Anon, Biomedical Textiles, Oct 17, 2000.

3 Satish Bhalerao, G S Lavekar and Y G Solanki, *Asian Textile J*, December 1998 81.

4 A Dayal and N Kumar, *JTA*, Nov-Dec 1999 187.

5 B S Gupta, *Medical Plastics and Biomaterials Magazine*, Jan 1998.

6 K P Chellamani and D Chattopadhayay, 'Yarns and Technical Textiles', *SITRA*, 1999.

7 K W Brennan, M Skinner and G Weaver, Braided surgical sutures, US Patent Office, Pat 4 959 069, 1990.

8 J G Ellis, *Proceedings of Textile Institute World Conference*, Manchester 2000.

9 K J L Burg, *International Fiber J*, August 2001 38.

10 T Hongu and G O Phillips, *New Fibres, Second Edition*, USA, John Wiley & Sons, Inc, 1997.

11 Encyclopedia of Chemical Technology, Vol. 23, Fourth Edition, 1997.

12 Encyclopedia of Chemical Technology, Vol. 20, Fourth Edition, 1996.

13 R Shishoo, *Nonwovens Industrial Textiles* 3/2001 24.

14 M Doser and H Planck, *Nonwovens Industrial Textiles* 3/2000 10.

15 S KBasu, *Mantra Bulletin*, Vol. 19/11, Nov, 2001 2.

16 U Sayed, M R Pratap and Y N Rane, *Asian Textile J*, June 2002 67.

17 H Mankodi, *The Indian Textile J*, November 2000 39.

18 M Jacob, The *Indian Textile J*, October 1998 88.

19 S Adanur, *Handbook of Technical Textiles*, Cambridge England, Woodhead Publishing Ltd, 2000.

20 A J Rigby and S C Anand, *Handbook of Technical Textiles*, Cambridge England, Woodhead Publishing Ltd., 2000.

21 S Adanur, *Wellington Sears Handbook of Industrial Textiles*, Lancaster, Technomic Publishing Co, Inc, 1995.

NERVE IMPLANTS

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ABSTRACT

Synthetic materials have been used in the medical field for surgical grafts due to lack of insufficient organic graft prosthesis. In the first years of their development, properties of an ideal graft were defined. "An ideal graft must be durable on implantation, must be produced at the required dimension, must be biocompatible, must be inert, must be non-toxic and non-allergic." It must also be elastic, comfortable, biostable and biomechanic. In this study nonresorbable and resorbable artificial nerve grafts are being searched as implantable textile materials based on various biomaterial in nerve tissue defects which occur in peripheral neurons in hands and arms and spinal cord neurons in the central nervous system. Evaluation of nerve implants are also being carried out according to their implantable biomaterial properties in medical textiles including electrophysiological studies and nerve regeneration of implants in nervous systems.

Key Words: Graft, nerve implant, biomaterials, nerve regeneration

HISTORY OF NERVE IMPLANTS:

Nerve grafting was first reported by Phillipeaux and Vulpien in 1870. However, reconstruction of a large defect in a major nerve trunk was a difficult problem because the use of a long nerve free graft had produced uncertain clinical results during the past 50 years. An attempt with autogenous nerve grafting was reported by Bunnell and Boyes in 1939 and the microsurgical nerve suture technique was developed by Smith in 1964. However successful transfer of a non-vascularized nerve graft requires an optimal recipient bed.

Since the first clinical application of vascularized nerve grafts by Taylor and Ham in 1976, many reports have suggested that this innovative technique has great possibilities to promote nerve regeneration. Mackinnon and Pho evaluated nerve regeneration after the nerve graft and compared the usefulness of both methods clinically and experimentally. In their reports, the most important factor in successful nerve grafting was to ensure a sufficient blood supply to the grafts. With sufficient blood supply to the graft, many schwann cells could survive and fibrosis was prevented in the grafted nerve (1).

WHAT IS THE FUNCTION OF NERVOUS SYSTEM?

Sensory cell generates an electrical signal in response to the stimulus via a process called sensor transduction. The electrical signals are forwarded to the spinal cord or brain where they are processed and sometimes returned to effector organs such as muscle to produce the appropriate response.

The basic signaling unit of the nervous system is the nerve cell or neuron which comes in many different shapes, sizes and chemical content (see Fig. 1).

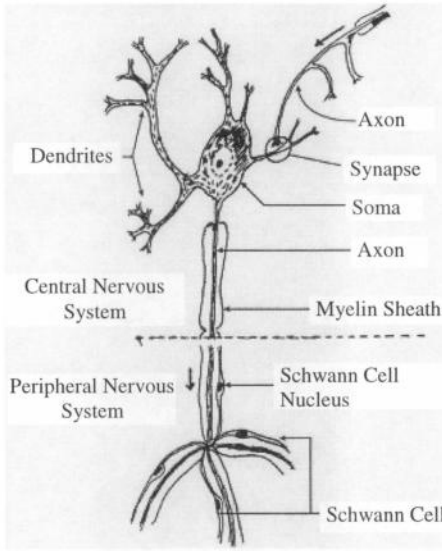

Fig 1 - Nervous system

An electrophysiological study of a nerve implant is determined by the nerve conduction velocity of an experimental animal.
The direction of nerve conduction is from dendrits to axons and conduction velocity is measured by stimulating or recording from two different sides along the course of a peripheral nerve.
Conduction Velocity = (Distance between two sides) / (Difference in conduction times between two sides)
And the other important functions of a nerve implant are histomorphological studies that the results are obtained according to exact quantative nerve fiber number and size. And the other function is walking track analysis. Shortly, the electrophysiological and histomorphological studies and walking track analysis of an experimental animal is evaluated together and compared with the optimum analysis while nerve implants are being studied (8, 12).

NERVE DEFECTS

Nerve defects may occur in peripheric nerves or central nerves. Making repair on peripheric nerve defects is more possible than making repair on central nerve defects due to the peripheric nerves are far away from central nervous system (7).

NERVE REGENERATION

What is a nerve graft?

A nerve graft is a piece of nerve whose extraneural support tissues will align and guide the outgrowth of axons from the proximal stump of a discontinuous nerve towards its target.

Grafting

Grafting is one method of surgical transplantation or implantation to replace a damaged part or compansate for a defect. And they are classified into three types which

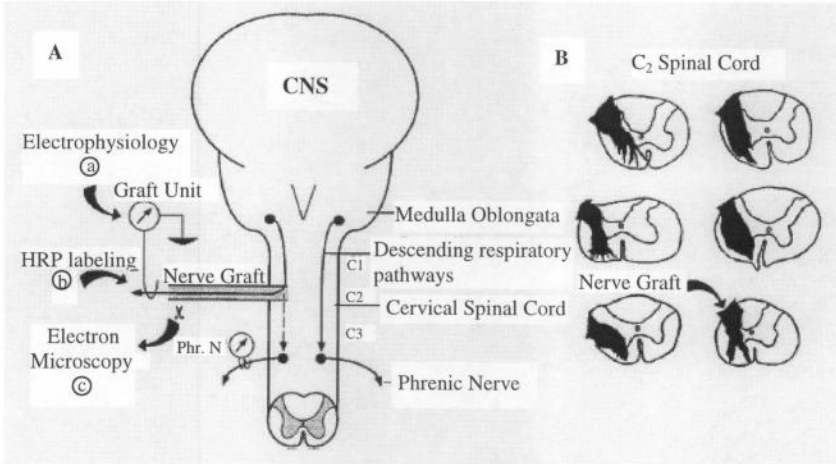

Fig 2 - Peripheric nerve graft usage in central nervous system

are autografts, allografts or homografts and xenografts or heterografts. Autografts are tissue from one part of the body and transplanted to another site in the same person and have very low rejection. Allografts or homografts are transplants made from one person to another in the same species and regimens are taken to minimize rejection. And xenografts or heterografts are donor and recipent are of different species and have higher incidence of rejection. An autologous peripheric nerve graft usage in central nervous system can be seen in Fig 2 (9).

Nerve grafting

Nerves are similar to electrical cable and contain many fibres (axons). Some of these fibres are sensory, providing feeling in a defined area and some are motor, activating muscles and sweat glands. When a nerve is repaired, the fibres sprout out of the nerve ending and grow across the gap towards the other end of the nerve at about 1mm/day. Recovery can fail for many reasons. If the nerve was not repaired, the gap is usually too large for the fibres to find the other end. If the nerve was repaired, good recovery can be prevented by factors such as infection or scarring. Nerve grafts are the traditional method of dealing with the problem. A length is taken from another nerve. These grafts probably give you the best return of nerve function (10).

WHICH MATERIALS ARE USED IN NERVE REGENERATION?

Biomaterials

Biomaterials are materials that are used in contact with tissue, blood, cells, protein and any other living substance. A variety of biomaterials can be classified into three types –

naturally derived materials (e.g, collagen and alginate), acellular tissue matrices (e.g, bladder submucosa) and synthetic polymers (e.g, polylactic acid, polyglicolic acid) (see Fig 3). Biomaterials also can be classified into two groups which are nonresorbable biomaterials and resorbable biomaterials. The polymers that are not considered to have biologically active surfaces are called nonresorbable polymers and include polyethylene, polypropylene, polyvinylchloride, polyester, polytetraflouroethylene, polyurethane, polyacrylnitrile, silicone rubber and polyacrylates.
Biodegradable and bioabsorbable materials are generally called bioabsorbable materials. They are composed in the body and their composition products are metabolized and excreted from the body. Biomedical applications of bioabsorbable polymers are absorbable surgical sutures, synthetic skin, adhesives and joints and nerve implants.
Bioabsorbable materials must satisfy other requirements such as degradability, biocompatibility, mechanical and chemical properties and also must be non-toxic.
Although performance requirements for a biomaterial factors - such as mechanical and physical properties and amenability to fabrication – differ from application to application, the universal requirement is 'biocompatibility'. Though lacking a rigorous and widely accepted meaning, biocompatibility may be fuctionally defined as the acceptance of the material by the surrounding tissues and fluids of the human body and by the body as a whole and the ability of the material to perform with an appropriate host response in a specific application. The degradation and absorption rate of a bioabsorbable material should be compatible with the healing rate of biotissues and organs. Bioabsorbable implant materials should maintain their mechanical properties and functions until the biotissues are completely cured. After complete healing of biotissues, the implanted materials should be degraded and absorbed quickly to minimize their side effects. Therefore, it is important to know the degradability properties of materials. The degradation products should not provoke inflammation or toxicity and must be removed from the body via metabolic pathways. The degradation rate and the concentration of degradation products in tissues surrounding the implant must be at a tolarable level (2, 3, 4, 11).

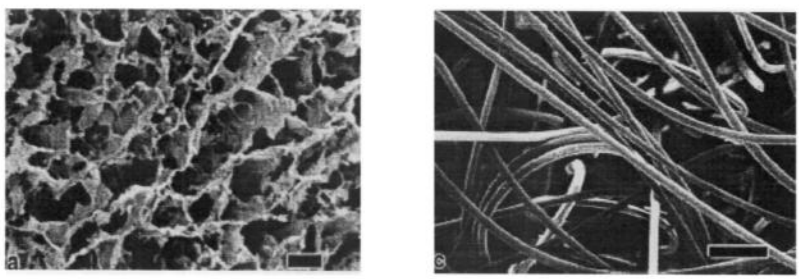

Fig 3 - a) Naturally derived biomaterial (collagen sponge)
b) Synthetic polymer (polyglycolic acid matrix)

Nonresorbable artificial nerve grafts:

These kind of grafts are made of synthetic polymers and are slowly absorbed within the body and take more than 6 months to degrade and include polyester, silicone rubber, polypropylene, PTFE, etc. Among these polymers silicones are long known to be biostable and biocompatible in most implants. Also frequently have the low hardness and low modulus that are useful in many device applications. In general silicone has favourable tissue compatibility and has easy formability. Especially because of its inert

and elastic properties, silicone tubing was one of the first and most frequently used synthetic materials for nerve grafts. The use of synthetic tubes as nerve guides to bridge a nerve gap has provided excellent in vivo experimental model to study the process of peripheral nerve regeneration (see Fig. 4). There are some advantages and disadvantages of nonresorbable artificial nerve grafts. Clinical intubulation of regenerating nerves often leads to long term complications including fibrosis and chronic nerve compression, requiring surgical removal of the conduit. These kind of grafts will remain in situ after the nerve has regenerated and may cause a chronic foreign body reaction. As a nonresorbable nerve graft, the silicone tube eventually becomes encapsulated as a part of a foreign-body response which can constrict the regenerating nerve and late nerve compresion with secondary complaints, fragments that break from these nerve guides may also cause injury to the regenerated nerve and impairment of nerve function. And a second operation might be necessary to remove these nerve guides, which may lead to injury to regenerated nerve. When we look at the advantages, silicone tube provides mechanical strength to allow surgical handling and suturing of the nerve ends into it. Despite diminishing clinical use, the silicone chamber has been a tremendous useful model for studying nerve regeneration in vitro. A nonresorbable nerve graft has allowed spatial and temporal examination of the regeneration process. For example, early experiments using the silicone chamber model indicated a maximum gap length between the proximal and distal ends of 10 mm (for rat neurons) across which regeneration could occur. And only by filling the graft with one or more neurotrophic substances could longer distances be bridged (5, 6, 7).

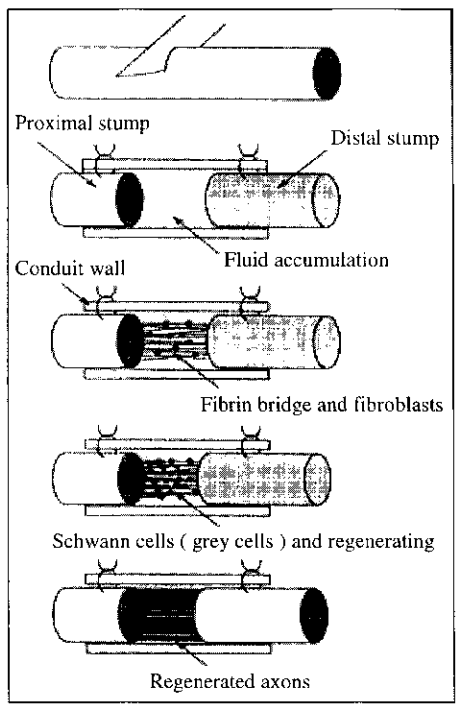

Fig 4 - Nonresorbable nerve graft (silicone-chamber model)

Resorbable artificial nerve grafts

A bioresorbable artificial nerve graft supports the reconstruction of a completely normal tissue without inflammation and removed from the body via metabolic pathways. A graft made of bioabsorbable materials is a promising alternative for promoting successful long term recovery, as has been seen both experimentally and clinically, because after serving as an appropriate scaffold for regeneration, the conduit eventually degrades.

The most important concern in designing a resorbable graft, apart from biocompatibility is choosing a material and processing conditions that will result in a graft that degrades slowly enough to maintain a stable support structure for the entire regeneration process but will not remain in the body much longer than needed. The time required for nerve regeneration will be a function of: nerve location, the species, the age of the patient and the gap length. For example, based on information gained from the silicone chamber model, a biodegradable graft for a 10 mm gap in the rat sciatic nerve should maintain its strength for 8 weeks or longer in order to ensure that axons have entered the distal stump and been myelinated. In addition, the graft should be flexible and its wall should have a thickness sufficient to hold the suture connecting to nerve epineurium and the graft. To avoid nerve compression, the inner diameter of the graft should be large enough to accomodate polymer swelling during degradation,which is seen with some polymers. Bioabsorbable grafts also have the significant advantage that, as they degrade, they can be made to release growth or trophic factors trapped in or absorbed to the polymer. A bioabsorbable nerve graft is seen in Fig. 5 (6).

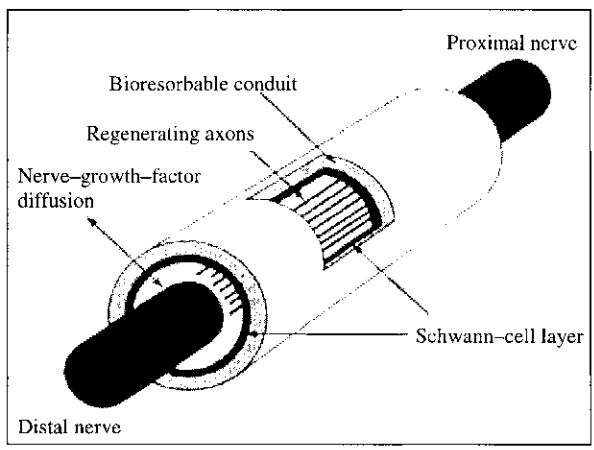

Fig 5 - Bioresorbable nerve graft

CONCLUSION

The nervous system reflects its complexity to the applications of nerve implants directly. So at first the nervous system and then the nerve injuries have to be understood as clearly as possible so that the nerve implants can be produced and implanted and examined correctly during implantation. We know that nerve injuries can cause serious problems to the human health that may not be ever repaired. As a conclusion a whole study of nerve implants are given as briefly as possible including the nervous system

function, nerve defects, nerve regeneration, nerve grafts and the biomaterials used in nerve defects as nonresorbable and resorbable artificial nerve grafts.

REFERENCES

1 M Iwai, S Tamai, H Yajima and K Kawanishi, 'Experimental study of vascularized nerve graft: evaluation of nerve regeneration using choline acetyltransferase activity', *Microsurgery,* 2001 **21**(2) 43-51.

2 B S Kim, C E Baez and A Atala, 'Biomaterials for tissue engineering', *World J Urol,* 2000 **18** 2-9.

3 S Adanur, *Wellington Sears of Handbook of Industrial Textiles,* Pennsylvania, Technomic, 1995.

4 The Swedish Institute for Fibre and Polymer Research, 'Opportunities and challenges for fibrous products in healthcare and medical applications' , 2nd int conf *Medical Textiles*, Cambridge, Woodhead, 2000.

5 A R Horrocks and S C Anand, *Handbook of Technical Textiles,* Cambridge, Woodhead, 2000.

6 http://www.biomed.metu.edu.tr/new_page_5htm

7 N Yıldız, 'Searching the usage of medical textiles in peripheric nerve tissue damages as reinforced grafts' , 1st int conf *Technical Textiles,* İzmir, DEU, 2002.

8 B L Seal, T C Otero and A Panitch, 'Polymeric biomaterials for tissue and organ regeneration' , *Elsevier Sci, 2001* **34**(4/5) 147-230.

9 P Decherchi, N Barreault and P Gauthier, 'Regeneration of respiratory pathways within spinal peripheral nerve grafts', *Experimental Neurology,* 1996 **137** 1-14.

10 http://www.pncl.co.uk/~belcher/ngraft.htm

11 Science and Technology Policy Institute Issue Paper, *Biomaterials Availability:Potential Effects on Medical Innovation and Healthcare,* 2000.

12 P Francel, T Francel, C Hertl and D Kline, 'Enhanced peripheral nerve regeneration across silicone conduits using ultrashort – segment nerve grafts', *American Association of Neurological Surgeons Annual Meeting.* 1996.