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Absorbable self-reinforced polylactide (SR-PLA) composite rods for fracture fixation: strength and strength retention in the bone and subcutaneous tissue of rabbits

A. MAJOLA, S. VAINIONPÄÄ, P. ROKKANEN
The Department of Orthopaedics and Traumatology, Helsinki University Central Hospital, SF-00260 Helsinki, Finland
H.-M. MIKKOLA, P. TÖRMÄLÄ
Biomaterials Laboratory, Institute of Plastics Technology, Tampere University of Technology, Tampere, Finland

The strength and strength retention of self-reinforced (SR) absorbable polylactic acid composite rods were evaluated after intramedullary and subcutaneous implantation in rabbits. Rods made of poly-l-lactic acid (SR-PLLA) and of poly-dl-lactic acid + poly-l-lactic acid composite (SR-PDLLA/PLLA) were used. The molecular mass (M_v) of PLLA was 260.000 and that of PDLLA 100.000. The bending and shear strengths were measured after a follow-up of 1–48 weeks. The initial bending strength of the SR-PLLA rods was 250–271 MPa and the shear strength was 94–98 MPa. After intramedullary and subcutaneous implantation of 12 weeks the bending strength of the SR-PLLA implants was 100 MPa. At 36 weeks the bending strength had decreased to the level of the strength of cancellous bone (10–20 MPa). There were no changes in the shear strength during 12 weeks' hydrolysis. The initial bending strength of SR-PDLLA/PLLA implants was 209 MPa and during the follow-up the implants lost their bending and shear strength faster than the SR-PLLA implants. The present investigation gave us the impetus to continue the studies with the fixation of experimental cortical bone osteotomies with SR-PLLA intramedullary rods.

1. Introduction

Since fixation devices are needed only for temporary support in the fixation of osteotomies and bone fractures, absorbable osteosynthesis materials offer a good alternative to metallic devices. Absorbable, polymeric materials retain their mechanical, tissue supporting properties for a given length of time, and, during that time, gradually lose their mechanical properties so that the stresses are stepwisely transferred from the implant to the healing bone tissue, decreasing the tendency to the stress protection atrophy. The implants undergo enzymatic and/or hydrolytic degradation into tissue-compatible metabolites which are used in the carbohydrate or protein metabolism, and the breakdown products are excreted in urine or in faeces or exhaled as CO_2. The biodegradation of the implants eliminates the need to remove the osteosynthesis materials, which is a substantial benefit both economically and medically. The most widely used absorbable osteosynthesis materials are aliphatic polyesters of α-hydroxyacid derivatives such as polylactic acid (PLA), polyglycolic acid (PGA), their copolymers (PGA/PLA), polydioxanone (PDS), and poly-β-hydroxy butyric acid (PHBA) [1–8].

Polylactic acid (PLA) is a synthetic, absorbable polymer which is synthesized by opening polymerization of the cyclic diester, lactide [9, 10]. In mammalian tissues, polylactic acid undergoes non-specific hydrolysis into lactic acid which becomes incorporated in the tricarboxylic acid cycle [5, 6] and is excreted through the lungs as CO_2 [1, 11].

In experimental studies, sutures, rods, screws, and plates made of non-reinforced poly-l-lactic acid (PLLA) have been found suitable for fixation of cancellous bone fractures and osteotomies of non-weight loading bones [1, 10, 12–17]. The mechanical properties of these non-reinforced materials are, however, not sufficient for fixation of weight loading bones [18–21]. In studies on the strength properties of the non-reinforced PLLA implants the initial bending strength has been reported to be 57–145 MPa and the shear strength 53–61 MPa [10, 19, 22–24].

The biocompatibility of polylactic acid has been shown to be good [1, 6, 7, 13–15, 18, 25–27].

Törmälä *et al.* have used the self-reinforcing (SR) techniques to develop polylactic acid composites which can attain much higher strengths than the earlier implants. The PLLA or PDLLA matrix was, for example, self-reinforced with highly oriented PLLA fibres [28, 29].

The aim of the present study was to evaluate the strength and strength retention of sintered SR-PLLA and SR-PDLLA/PLLA implants *in vivo* in the bone and subcutaneous tissue.

2. Materials and methods

Cylindral, composite rods (3.2–3.4 mm by 50 mm) made of self-reinforced poly-l-lactic acid (SR-PLLA) and poly-dl-lactic acid + poly-l-lactic acid composite (SR-PDLLA/PLLA) were used in the evaluation of the strength retention in the dorsal subcutaneous tissue and SR-PLLA intramedullary rods (4.5–4.8 mm by 60 mm) in the bone tissue of rabbits [30]. The SR-PLA rods were produced by sintering the PLLA fibres and PLLA or PDLLA matrix together in cylindral moulds at temperatures of 162–174 °C (PLLA rods) and 130–135 °C (PDLLA/PLLA rods). The compression time was 9 min for the PLLA rods and 5 min for the PDLLA/PLLA rods with the final pressure of 20 N mm^{-1}. The intramedullary implants were produced under N_2 atmosphere. The PLLA fibres were produced from poly-l-lactide by hot-drawing the melt-spun monofilaments to a draw ratio of 7. The complete SR-PLLA implant consisted of PLLA matrix reinforced with PLLA fibres and the SR-PDLLA/PLLA implants were composed of PDLLA matrix reinforced with PLLA fibres. The molecular mass (M_v) of PLLA was 260.000 (CCA Biochem B.V., Holland) and that of PDLLA 100.000 (Boering Ingelheim, FRG) before processing and sterilization. The implants were gamma-sterilized with a dosage of *ca.* 2.5 Mrad.

Twenty-one rabbits of both sexes with masses from 2800 to 4500 g (mean 3775 g) were used. Before anaesthesia the rabbits were given atropine 0.5 $mg\,kg^{-1}$ (Atropine[R], Orion, Espoo) subcutaneously (s.c.) as premedication. They were anaesthetized with s.c. medetomidine 0.3 $mg\,kg^{-1}$, (Domitor[R], Lääke-Farmos, Turku), ketalar 25 $mg\,kg^{-1}$ (Ketalar[R], Parke-Davis, Barcelona), and diazepam 0.5 $mg\,kg^{-1}$ (Diapam[R], Orion, Espoo).

Both hind legs of the rabbits were shaved and scrubbed with antiseptic fluid. A medial parapatellar incision was made, the patella was dislocated laterally, and the distal femur carefully exposed. A hole was drilled through the intercondylar region, and the SR-PLLA intramedullary rod was carefully tapped into the drill-channel (Fig. 1). The mouth of the drill-channel was sealed with Bone Wax[R] (Ethicon, FRG). The incision was closed in layers, and the rabbits were given 100.000 IU procainpenicillin s.c. (Procapen[R], Orion, Espoo) for infection prophylaxis. The cylindral rods were implanted into the dorsal subcutaneous tissue of the rabbits.

The follow-up times were 1, 3, 6, 12, 24, 36, and 48 weeks. Each follow-up group consisted of three rabbits. A total of 42 SR-PLLA intramedullary rods were implanted in bone tissue, six implants in each follow-up group, and 35 SR-PLLA and SR-PDLLA/PLLA cylindral implants were implanted subcutaneously, five implants in each follow-up group.

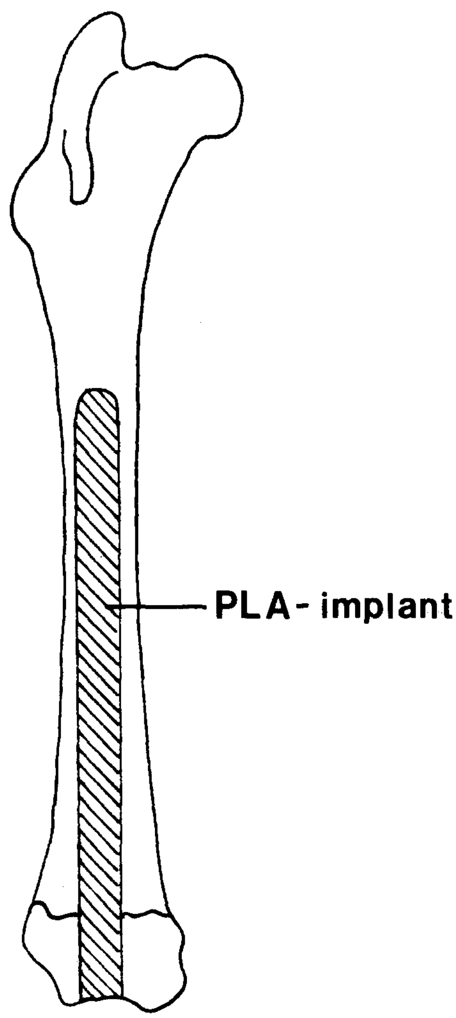

Figure 1 Anterior view of the femur of rabbit showing the operative technique. A channel was drilled from the distal part of the femur and the intramedullary space was reamed up to 4.8 mm. The SR-PLLA intramedullary rod was tapped carefully into the medullary cavity. The mouth of the drill-channel was sealed with Bone Wax[R].

The bending and shear strengths of the rods were measured after production and *in vivo* hydrolysis. After sacrifice the implants were removed from the subcutis, both femurs were split with a circular saw, and the intramedullary rods were carefully removed for the examination of the mechanical strength. The SR-PLA implants were immersed in saline until the strength measurements, which were performed within 12 h. The mechanical strengths were measured by using the JJT5003 tensile testing equipment (J.J. Lloyd Instruments, UK) and a tool constructed by modifying the standard BS 2782, Method 340 B (1978). The testing speed was 10 $mm\,min^{-1}$. The bending test was performed by using the three-point method; the distance between the supports was 42 mm. The shear strength was measured by a special tool described in Fig. 2. The strength values were measured at room temperature on the wet rods, since drying of the wet rods rapidly destroys their structure leading to a large decrease of their strength.

3. Results

The results of the bending and shear strength tests are

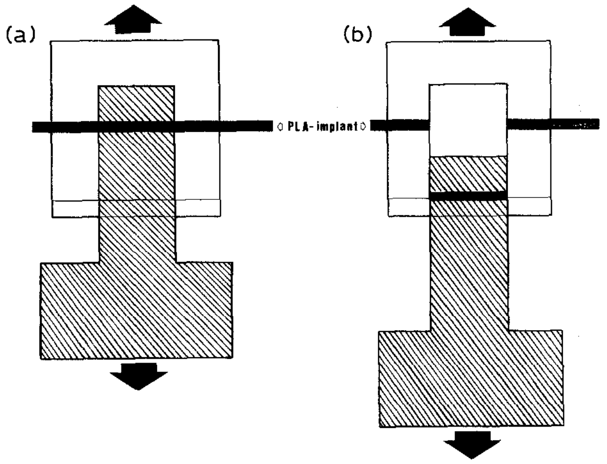

Figure 2 The shear strength measurement tool consists of two parts which were pulled apart with a universal Tensile Testing Machine. The implant was pushed through a joint holding the two parts together; during measurement the implant was cut into three pieces. (a) before test, (b) after test. The shear strength was measured from both ends of the implant.

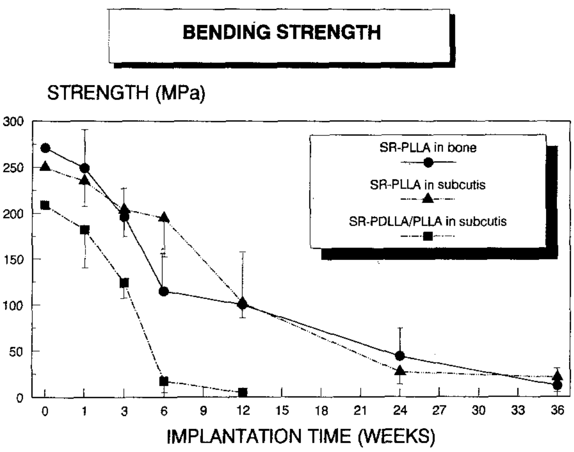

Figure 3 Loss of bending strength of SR-PLLA and SR-PDLLA/PLLA implants in bone tissue and in subcutaneous tissue of rabbits during a 36-week follow-up.

presented in Figs 3 and 4. The initial bending strength of the 4.5 mm SR-PLLA intramedullary rods was 271 MPa and the shear strength was 94 MPa. The strength values for the 3.2 mm SR-PLLA rods were 250 MPa and 98 MPa, respectively. The initial bending strength of the SR-PDLLA/PLLA implants was below the values of the SR-PLLA implants, the bending strength of the 3.2 mm SR-PDLLA/PLLA rods being 209 MPa and the shear strength 102 MPa.

3.1. Implantation in the subcutaneous tissue

The bending strength of the 3.2 mm SR-PLLA rods was 195 MPa after implantation of six weeks in the subcutis. At 12 weeks the bending strength was 101 MPa. During the follow-up the bending strength decreased slowly and was 21 MPa at 36 weeks (Fig. 3). At 48 weeks the implants were so elastic that the bending strength was not possible to evaluate. There were no changes in the shear strength during the 12-week follow-up. At 24 weeks the shear strength had

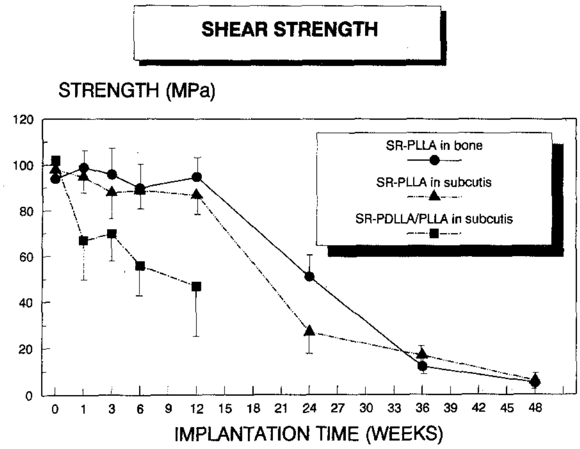

Figure 4 Loss of shear strength of SR-PLLA and SR-PDLLA/PLLA implants in bone tissue and in subcutaneous tissue of rabbits during a 48-week follow-up.

decreased to 27 MPa and at 48 weeks to 6.0 MPa (Fig. 4).

In the subcutis the 3.2 mm SR-PDLLA/PLLA rods lost their mechanical properties faster than the SR-PLLA implants (Figs 3 and 4). At three weeks the bending strength of the 3.2 mm SR-PDLLA/PLLA rods was 123 MPa and at six weeks it had decreased to 17 MPa. The shear strength was unchanged during three weeks and decreased to 47 MPa during 12 weeks' hydrolysis. At 24 weeks and later, the implants were so degradated that it was not possible evaluate the mechanical strengths.

The difference in the bending and shear strength retention of the SR-PLLA and SR-PDLLA/PLLA rods was statistically highly significant.

3.2. Implantation in bone tissue

The bending strength of the 4.5 mm SR-PLLA intramedullary rods after implantation of three weeks in bone tissue was 196 MPa and after six weeks 115 MPa. At 12 weeks the bending strength was still 100 MPa. After implantation of 12 weeks the bending strength slowly decreased and was 12 MPa at 36 weeks. At 48 weeks the implants were so flexible that it was possible to measure only the shear strength. The shear strength remained unchanged during the 12-week follow-up and decreased slowly later being 4.5 MPa at 48 weeks. (Figs 3 and 4).

The differences in the strength retention of the SR-PLLA implants between hydrolysis in the bone and subcutaneous tissue were not statistically significant, except at six weeks when the bending strength of the intramedullary SR-PLLA rods was lower than that of the subcutaneous SR-PLLA rods.

4. Discussion

Osteosynthesis devices made of self-reinforced polyglycolic acid (Biofix^R) have been in clinical use since 1984 [31–38]. SR-PGA-based materials lose their mechanical strength in four to eight weeks, which is sufficient for fixation of certain cancellous bone osteotomies and fractures. The initial bending and shear strengths of the SR-PGA implants are in the same order of magnitude as the corresponding values of the studied SR-PLLA implants. In the fixation of cortical bone fractures, the SR-PGA rods still lose their tissue-supporting properties too fast, though their initial strengths exceed the bending and shear strengths of cortical bone [8, 39].

In the present study the initial bending and shear strengths of the SR-PLLA implants were 250–271 MPa and 94–98 MPa. The corresponding values of the SR-PDLLA/PLLA implants were 209 MPa and 102 MPa. In previous studies the bending and shear strengths of the non-reinforced injection moulded PLLA implants with the same molecular mass have been reported to be 145 MPa and 53 MPa [23]. The tensile strength at break, yielded strength, and elastic modulus of melt-spun PLLA fibres were 410 MPa, 140 MPa, and 4.1 GPa [40].

The shear strength measurement arrangement, which was used in the present study, measures shear effects in the direction perpendicular to the long axis of the rod. This mode was selected, because this kind of load is important in practice where the slip of bone fragments in the fracture plane creates a similar shear. The SR-PLLA rods implanted in the subcutaneous and bone tissue sustained their mechanical strength longer than the SR-PDLLA/PLLA rods, which was coincident with previous studies. After implantation of 12 weeks in bone and subcutaneous tissue, the bending strength of the SR-PLLA implants was still 100 MPa, close to that of cortical bone [41, 42]. During the follow-up the bending strength slowly decreased and at 36 weeks it was at the level of that of cancellous bone (12–21 MPa). The SR-PDLLA/PLLA implants, instead, lost their mechanical properties significantly faster than the SR-PLLA implants; at six weeks the bending strength had decreased to the level of that of cancellous bone (Fig. 3). In the SR-PLLA groups there were no changes in the shear strength during the 12-week follow-up (Fig. 4). The rate of loss of the shear strength was slower than that of the bending strength. It was obviously caused by the faster degradation rate of the amorphous PLLA or PDLLA matrix than of the reinforcing PLLA fibres. This leads to delamination of the SR-structure and flattening of the rod during the bending experiment. During the shear strength experiment the rods retained their cross-sectional dimensions, and delamination could not contribute to the shear deformation of the rod, since, in the measurement arrangement used, the still strong PLLA fibres carried most of the shear load. This is a favourable phenomenon considering the clinical application of the SR-PLLA rods, because the shear-load carrying capacity of the rod performs an important role in the fixation of cancellous bone fractures and osteotomies [8, 40].

In the present study there were no statistically significant differences between the strength retentions of the SR-PLLA rods in the bone and subcutaneous tissue, except at six weeks when the bending strength values after implantation were slightly inferior in bone tissue than in subcutaneous tissue ($p < 0.05$). The difference was probably caused by the damage of the implants during removal from the intramedullary canal.

In the present study the strength retention of the SR-PLLA intramedullary rods was found evidently to be sufficient for fixation of experimental cortical bone osteotomies. On the basis of these promising results we have continued our research by experimental studies using self-reinforced polylactic acid screws and intramedullary rods in the fixation of osteotomies of weight bearing bones.

Acknowledgements

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