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ACL GRAFT FIXATION

Interference Screw Fixation of the Soft Tissue Grafts

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Academic Dissertation

ACADEMIC DISSERTATION
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To my family

CONTENTS

1. ABSTRACT

Rupture of the anterior cruciate ligament (ACL) is a very common consequence of a knee distortion injury. The treatment of choice is surgery in which the ruptured ACL is replaced with a tendon graft. The strength of fixation, rather than the graft itself, is currently suggested to be the weakest link in the early postoperative period after ACL reconstruction. This thesis focuses on interference screw fixation of the soft tissue grafts in reconstructing the ACL. Specifically, the project was planned to examine some common beliefs and recommendations concerning the fixation of a soft tissue ACL graft and factors influencing on it, concepts that currently lack scientific justification.

Compaction of the bone-tunnel walls (by compaction drilling or serial dilation) has been recommended as a means to enhance the fixation of the soft tissue grafts fixed with interference screws in the ACL reconstruction. In the first part of this study series, compaction drilling was compared to conventional extraction drilling by using thirty pairs of porcine tibiae randomly paired with thirty pairs of human cadaver hamstring tendons. The tendon grafts were fixed into tibial bone tunnels with interference screws. The specimens were tested using a cyclic-loading test and single-cycle load-to-failure test. The results revealed that compaction of the bone-tunnel walls by compaction drilling does not increase the initial fixation strength of a hamstring tendon graft in ACL reconstruction using a porcine model.

Porcine tissues have previously been used in many ACL graft fixation studies assuming that the results can be directly applied on human patients because porcine tissues have been believed to be highly similar with human tissues. However, in the first part of the study a significant difference was observed between the porcine tibiae and human controls for the volumetric trabecular bone density in favor for the porcine. In the second part of the study, to further study the suitability of porcine tissues as surrogates for human tissues in biomechanical evaluation of interference screw fixation of soft tissue grafts in ACL reconstruction, fixation strength of interference screw in both porcine and human cadaver tissues was tested. It was found that in comparison to young human cadaver tibia, the results obtained with porcine tibia significantly underestimate the graft slippage past the fixation and thus overestimate the fixation strength of the soft tissue graft in human ACL reconstruction.

To evaluate whether the conclusions of the first part of this study concerning the bone compaction were truly applicable to humans, the first part of the study was repeated using 22 pairs of human cadaver tibiae and hamstring tendon grafts. In addition, to evaluate further whether serial

dilation, the other means for compaction of the bone tunnel walls, could enhance the fixation strength of interference screw fixed soft tissue grafts, in the fourth part of the study series serial dilation was compared to extraction drilling using 21 pairs of human cadaver tibiae and anterior tibialis tendon grafts. The results showed that compaction of human bone-tunnel walls by compaction drilling or serial dilation does not increase the initial fixation strength of interference screw fixed soft tissue grafts.

Graft pretensioning, preconditioning and initial tensioning have been recommended for elimination of viscoelastic tendon creep and prevention of postoperative knee laxity. In the fifth part of the study, forty-two human anterior tibialis tendon grafts were pretensioned for 15 minutes using a Graftmaster board, drawn through the tibial bone tunnel, and then subjected to either no preconditioning (Group I), cyclic preconditioning (Group II, 25 cycles between 0-80 N in 100 seconds) or isometric preconditioning (Group III, 80 N for 100 seconds). Thereafter, an initial graft tension of 80 N was applied on the grafts and a bioabsorbable interference screw was inserted. The residual graft tension was recorded immediately and 10 minutes after the screw insertion. Additionally, ten pretensioned anterior tibialis and quadrupled hamstring tendon grafts were directly connected to the mechanical testing machine (without interference screw fixation into bone tunnel) and isometrically preconditioned. The residual graft tension was recorded 1 minute (corresponding to the time required to insert the interference screw in the Groups I-III) after the 80-N initial tensioning, and then after 10 minutes and 60 minutes. The results indicated that clinically applicable preconditioning protocols cannot fully eliminate the intrinsic creep from hamstring or anterior tibialis tendons. Regardless of the preconditioning status (whether or not preconditioned), the initially set tension decreases considerably (over 60%) postoperatively due to the remaining intrinsic tendon creep, thus questioning the reasonableness of preconditioning in anterior cruciate ligament reconstruction.

Bone density and insertion torque of interference screws are considered good predictors of postoperative fixation strength. Several studies have shown correlation between fixation strength, bone density, and insertion torque. However, the real predictive value of these parameters has not been previously analyzed using proper statistical analysis. In the sixth part of the study, the relationship between bone density, insertion torque of the interference screws, and the fixation strength of the ACL graft were evaluated based on the data recorded in the forth part of the study. Correlation and regression analysis were performed to evaluate how accurately fixation strength can be predicted with bone density and insertion torque. The results showed that despite relatively high group-level correlation between both of the predictor variables and the outcome variable, at an

individual level, bone density and insertion torque are poor predictors of the fixation strength when reconstructing the ACL with interference screw fixed soft tissue grafts.

2. LIST OF ORIGINAL ARTICLES

This thesis is based on the following original publications, referred to as I – VI in the text:

I.   Nurmi JT, Järvinen TLN, Kannus P, Sievänen H, Toukosalo J, and Järvinen M: Compaction versus extraction drilling for fixation of the hamstring graft in anterior cruciate ligament reconstruction. *Am J Sports Med 30*:167-173, 2002

II.   Nurmi JT, Sievänen H, Kannus P, Järvinen M, and Järvinen TLN: Porcine tibia is a poor substitute for human cadaver tibia for evaluating interference screw fixation. In press. *Am J Sports Med 32*: 000-000, 2004

III. Nurmi JT, Kannus P, Sievänen H, Järvinen M, and Järvinen TLN: Compaction drilling does not increase the initial fixation strength of the hamstring tendon graft in anterior cruciate ligament reconstruction in a cadaver model. *Am J Sports Med 31*: 353-358, 2003

IV.   Nurmi JT, Kannus P, Sievänen H, Järvelä T, Järvinen M, and Järvinen TLN: Interference screw fixation of soft tissue grafts in anterior cruciate ligament reconstruction. Part I: Effect of tunnel compaction by serial dilators versus extraction drilling on the initial fixation strength. In press. *Am J Sports Med 32*: 000-000, 2004

V.   Nurmi JT, Kannus P, Sievänen H, Järvelä T, Järvinen M, and Järvinen TLN: Interference screw fixation of soft tissue grafts in anterior cruciate ligament reconstruction. Part II: Effect of preconditioning on graft tension during and after screw insertion. In press. *Am J Sports Med 32*: 000-000, 2004

VI.   Järvinen TLN, Nurmi JT, and Sievänen H: Bone density and insertion torque as predictors of ACL graft fixation strength. In press. *Am J Sports Med 32*: 000-000, 2004

3. ABBREVIATIONS

ACL	*Anterior cruciate ligament*
ANOVA	*Analysis of variance*
AT	*Anterior tibialis tendon graft group*
BMD	*Bone mineral density*
BPTB	*Bone-patellar tendon-bone graft*
CT	*Computed tomography*
DXA	*Dual energy x-ray absorptiometry*
F	*Fixation strength*
HT	*Hamstring tendon graft group*
IT	*Insertion torque*
SD	*Standard deviation*
vBMD	*Volumetric bone mineral density, mg/cm^3*

4. INTRODUCTION

Anterior cruciate ligament is one of the most important structures in the maintenance of stability of the knee joint. Rupture of the ACL is one of the most common sports-related injuries, with an estimated incidence of 1 in 3000 per year within the general population in the United States, meaning more than 93 000 cases annually (Miyasaka et al. 1991). ACL deficiency results in significant knee instability and often secondary damage to other knee structures, such as the menisci and articular cartilage. Untreated, the ACL rupture can gradually lead to osteoarthrosis (Fithian et al. 2002, Kannus and Järvinen 1987, Smith et al. 1993). In the ACL reconstruction, the ruptured ACL is removed and replaced with an ACL substitute (Fig. 1). Autologous bone-patellar tendon-bone graft has long been the golden standard of the ACL substitutes. However, soft tissue grafts, mainly the hamstring and quadriceps tendon grafts, have recently emerged as a strong alternative, because morbidity associated with harvesting of these soft tissue autografts seems to be significantly lower than that with the BPTB graft (Brown et al. 1993, Fu and Ma 1999b, Kartus et al. 2001, Uh et al. 1997). Also, the use of soft tissue allografts, e.g., the Achilles, anterior tibialis and posterior tibialis tendons, has become more popular during the recent years especially in revision surgery (Donahue et al. 2002, Tom and Rodeo 2002, Vangsness et al. 2003). However, as opposed to the bone-patellar tendon-bone (BPTB) graft with bony blocks at both ends of the graft, the soft tissue grafts do not provide a favorable surface for the fixation device to tightly grip on. Rodeo et al. (1993) studied biomechanically and histologically tendon-healing in a bone tunnel using a canine model and found that before twelve weeks postoperatively, all specimens failed by pull-out of the tendon from the bone tunnel due to still incomplete tendon-to-bone healing. Accordingly, the strength of fixation, rather than the graft itself, is suggested to be the weakest link in the early postoperative period after ACL reconstruction, and progressive creep or slippage of the graft fixed in a bone tunnel are the most common concerns in the use of these no-bone grafts (Brand et al. 1999 and 2000a, Fu and Ma 1999b, Rodeo et al. 1993, Simonian et al. 1998). To alleviate this problem, researchers have used both animal and human cadaveric models to evaluate a great number of different novel fixation devices (such as interference screws, staples, cross-pins, washers, and endo-buttons), and to further develop their characteristics (e.g., material, position, length, and diameter) (Brand et al. 1999 and 2000a, Clark et al. 1998, Giurea et al. 1999, Kousa et al. 2003ab, Magen et al. 1999, Weiler et al. 1998).
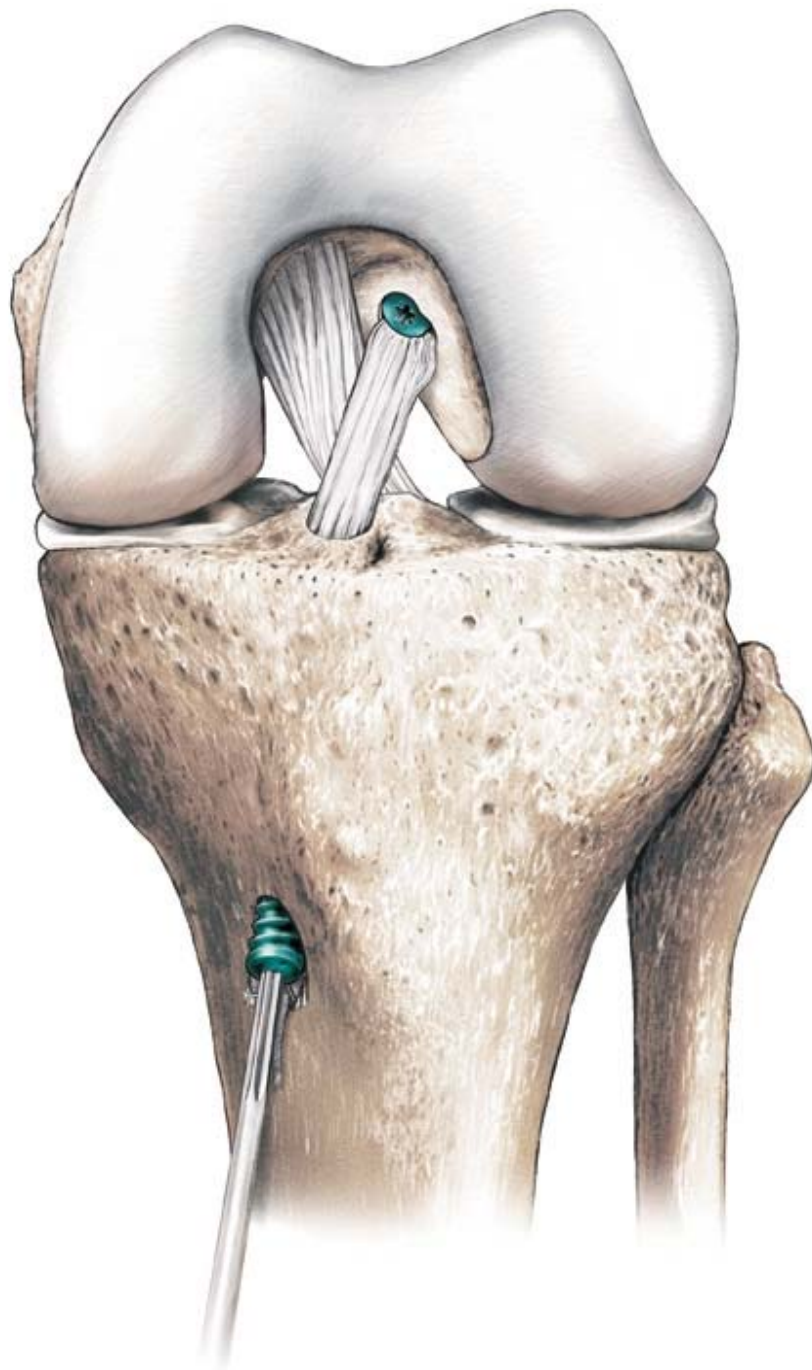
Figure 1. ACL reconstruction with interference screw fixed soft tissue grafts. Printed with the permission of Inion Oy, Finland.

5. REVIEW OF THE LITERATURE

In ACL reconstruction, the ruptured ACL is replaced with a graft. Optimally, the ACL substitute retains joint stability and kinematics postoperatively during graft healing and remodeling. However, it is not exactly known how strong the graft and its fixation should be to reach optimal clinical results. It is not known either how high forces are or should be applied on the graft and its fixations during healing and the postoperative rehabilitation. The currently used early accelerated rehabilitation protocols after an ACL reconstruction allow immediate full range of motion and full weight bearing, thus subjecting the graft to increased loading and placing special emphasis on secure initial graft fixation (Beynnon et al. 2002, Uh et al. 1997).

5.1. HEALING OF THE SOFT TISSUE GRAFT AFTER ACL RECONSTRUCTION

The deterioration of the soft tissue ACL graft begins immediately after implantation, although already in some weeks the intra-articular portion of the graft will be covered by synovium with an abundant blood supply that will eventually provide revascularization of the graft (synovial envelopment and neovascularization). Thereafter, repopulation of the graft occurs with an increasing number of fibroblasts and active nuclear morphology. Slowly, during several months of graft remodeling, fibroblast count continues to increase, and the active nuclear morphology and neovascularity remain increased with larger areas of degeneration as the percentage of mature collagen decreases. The following maturation stage is characterized by a slow decline in the number of nuclei and maturation of the collagen matrix. The random collagen fiber orientation progresses to a longitudinal orientation from the peripheral to the central areas of the graft and the soft tissue graft will achieve the appearance of the normal ACL (ligamentization or transformation into ligament-like tissue) over the initial year after surgery. (Arnoczky et al. 1982, 1991 and 1996, Bartlett et al. 2001, Goradia et al. 2000, Scranton et al. 1998, Yasuda et al. 1989)

Tendon-to-bone healing will occur in the femoral and tibial bone tunnels, depending on the fixation device used, either by indirect tendon-to-bone healing with a clear fibrous interzone between the tendon graft and bone, or by development of a direct ligament insertion where a zone of metaplastic fibrocartilage is present between the graft and lamellar bone instead of a fibrous interzone. Interference screw fixation provides direct compression of the graft against the wall of the bone tunnel to promote direct tendon-to-bone healing. With time, the bone tunnel remodels and replacement of fibrous tissue by new bone formation occurs. (Bartlett et al. 2001, Berg et al. 2001,

Eriksson et al. 2000, Grana et al. 1994, Lajtai et al. 2001, Nebelung et al. 2003, Pinczewski et al. 1997, Rodeo et al. 1993, Weiler et al. 2002a, Yamakado et al. 2002, Yamazaki et al. 2002)

5.2. BIOMECHANICS

5.2.1. Biomechanical properties of the normal ACL and soft tissue grafts

According to prevailing estimations and calculations, the ACL graft is loaded to approximately 150 to 500 N during normal daily activities (Fleming et al. 1992 and 1993, Holden et al. 1994, Magen et al. 1999, Markolf et al. 1996, Morrison 1968 and 1969, Noyes et al. 1984, Rupp et al. 1999a, Toutoungi et al. 2000). Woo et al. (1991) performed a biomechanical human cadaver study to quantify the effects of age on the structural properties of the femur-ACL-tibia complex. They reported the average (±SD) ultimate failure load of 2160 ± 157 N for the ACL in the specimens harvested from 22-35-year old donors. The structural and material properties of the BPTB graft have been shown to be comparable with those of the native ACL (Noyes et al. 1984, Woo et al. 1991), whereas the quadrupled hamstring (semitendinosus and gracilis) tendon graft and the single loop anterior and posterior tibialis tendon grafts have been shown to be stronger than the native ACL and the BPTB graft (Donahue et al. 2002, Hamner et al. 1999, Steiner et al. 1994). Therefore, the soft tissue grafts themselves have been assumed to be strong enough to be used as ACL substitutes.

Although the soft tissue grafts *per se* appear suitable as ACL substitutes in terms of their initial strength, animal studies have shown that the strength of the graft changes (first decreases and then increases) postoperatively due to graft necrosis, healing, and remodeling (Grana et al. 1994). In addition, tendons are known to be viscoelastic and elongate with time if kept under a constant load (graft creep), indicating that the length of the graft between femoral and tibial fixation sites is likely to increase with time after ACL reconstruction (Beynnon et al. 1994, Graf et al. 1994, Schatzmann et al. 1998, Smith et al. 1993, Woo et al. 1999, Yoshiya et al. 2002). Regarding the biomechanical properties of the fixation, slippage of the graft past the fixation at load levels below the failure load is always a major concern, in addition to the ultimate failure strength (Rittmeister et al. 2002).

5.2.2. Biomechanical evaluation of ACL reconstruction

Due to the differences in the previously used techniques to characterize the biomechanical behavior of the ACL reconstruction, comparisons between studies have been difficult (Beynnon and Amis 1998). Most of the previous ACL graft fixation studies have used some kind of pull-out (single-cycle failure) evaluations or several cycles of low-level loading (preconditioning) followed by single-cycle failure testing. To standardize the testing protocols, Beynnon and Amis (1998)

proposed standard guidelines for the evaluation of the knee and ligament biomechanics. In their review, they recommended conventional single-cycle load-to-failure evaluation, repetitive (cyclic) loading evaluation with low peak forces, as well as determination of cyclic-loading induced displacement in testing the properties of an ACL graft fixation. The single-cycle load-to-failure testing method provides the upper limit of the graft-fixation construct, which is useful information with regard to the behavior of the graft during unexpected loading events, such as a loss of balance or a fall of the patient. However, during early intense rehabilitation of the operated knee, the graft-fixation construct is subjected to thousands of cycles of repetitive submaximal loading, and thus, the time-zero failure loads may not appropriately reflect the possible changes that occur over time. The cyclic-loading testing, in turn, evaluates the cyclic behavior of the graft-fixation construct, and thus, better allows the determination of the changes possibly occurring immediately postoperatively (Beynnon and Amis 1998). Because it is impossible to determine the magnitude and nature of the (cyclic) loading test that would most appropriately mimic the loading during postoperative rehabilitation of the reconstructed ACL, especially as the rehabilitation itself also varies individually (Beynnon et al. 2002), the use of a loading regimen that is a bit overly strenuous (worst-case scenario) rather than overly cautious has been recommended (Giurea et al. 1999).

Many investigators have used whole knees (the entire femur-ACL graft-tibia complex) in the biomechanical evaluations of the ACL reconstruction. An alternative approach for pure fixation studies is to perform the biomechanical testing unilaterally; i.e., having one end of the graft fixed within a bone tunnel and the other secured to the moving actuator. An advantage of this approach over testing whole knees is that it is more controlled and allows one to focus on the monitoring the slippage of a single fixation site only (Beynnon and Amis 1998). The tibia is believed to be biomechanically more problematic of the two fixation sites in the ACL reconstruction since the bone quality in the tibial metaphysis is inferior to that of the lateral femoral condyle (Brand et al. 2000ab, Kohn and Rose 1994, Sievänen et al. 1998); the loading is applied to the ACL graft virtually parallel to the long axis of the tibial bone tunnel (as opposed to femur, in which the direction of the force does not come parallel with the bone tunnel until 100° of flexion) (Brand et al. 2000a, Malek et al. 1996); the fixation implant (here, the interference screw) has to be implanted in a less favorable outside-in direction; and finally, the biomechanically superior looped end of the soft tissue graft is generally used for the femoral fixation, leaving the end containing only the free tendon ends for the tibia (Brand et al. 1999, Pena et al. 1996). Consequently, fixation studies performed at the tibial site should provide a worst-case scenario, especially if the specimen is loaded with a distraction force acting parallel to the long axis of the graft and bone tunnel as recommended by Beynnon and Amis (1998).

Many animal models have been used in the previous biomechanical ACL reconstruction studies, porcine tissues being the most commonly used surrogate for human cadaver tissues, since they are readily available, free from human diseases, inexpensive, and have been used extensively (Black et al. 2000, Butler et al. 1994, Harding et al. 2002, Ishibashi et al. 1997, Kousa et al. 2001ab, 2003ab, Liu et al. 1995, McKeon et al. 1999, Miyata et al. 2000, Nagarkatti et al. 2001, Nakano et al. 2000, Paschal et al. 1992, 1994, Rupp et al. 1997, 1998, 1999b, Seil et al. 1998, Yamanaka et al. 1999). Both pure porcine tissue models and combination models of porcine and human cadaver tissues have been used. Some researchers consider porcine knees to good models for human specimens in terms of their size, shape, and bone quality (Aerssens et al. 1998, Black et al. 2000, McKeon et al. 1999, Mosekilde et al. 1987, Nagarkatti et al. 2001, Paschal et al. 1992, 1994, Seil et al. 1998, Xerogeanes et al. 1998), while others claim that biomechanical data obtained using animal tissues on interference screw fixation of soft tissue grafts in ACL reconstruction can not be directly applied on humans (Magen et al. 1999). Beynnon and Amis (1998) recommended that the use of animal tissues as surrogates for human tissues in biomechanical testing should be considered in light of the bone density, the soft-tissue material properties, and the size constraints of the joint. Regarding human cadaver specimens, the authors proposed that donors should be free from diseases and conditions that could affect the properties of the tissues (e.g., osteoporosis), and the age of the cadavers of the ACL graft fixation studies should be 65 years or less in males and 50 years or less in females (Beynnon and Amis 1998).

The biomechanical properties of the whole ACL reconstruction complex consist of a combination of the properties of the bone, the graft, and the fixation. Regarding the most important function of the ACL, to resist anterior displacement of the tibia on the femur (Uh et al. 1997), the clinical relevance of postoperative graft elongation due to graft creep and slippage of the graft past the fixation (cyclic-loading induced displacement) should be emphasized. However, since the fixation strength has traditionally been considered to be the weakest link of the ACL reconstruction and since sufficient or optimal fixation strength can not be defined as long as the true forces transmitted through the ACL graft are unknown, the ultimate goal has been to find as strong fixation methods as possible. Accordingly, most of the biomechanical ACL graft fixation studies have traditionally used maximum failure load as the number one outcome parameter (Fig. 2).
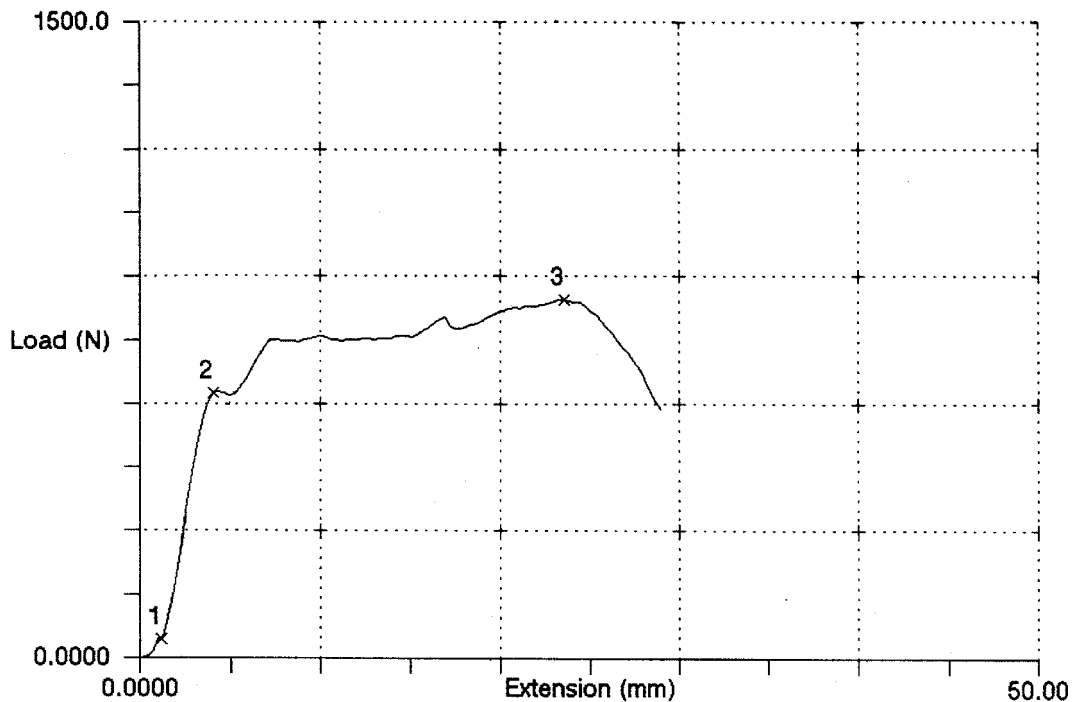
Figure 2. A load-displacement curve from a single-cycle load-to-failure test. 1. Preload level; 2. Yield point; and 3. Maximum failure point. After the yield point, the specimen undergoes significant stretching (presumably, both as the graft slips past the fixation and as the graft tissue itself elongates) but still shows a strong resistance to loading, as the load value keeps increasing until the maximum load is reached. Linear stiffness can be determined from the curve as the slope of the linear region (between points 1 and 2) corresponding to the steepest straight-line tangent to the loading curve.

5.3. INTERFERENCE SCREW FIXATION OF SOFT TISSUE GRAFTS

In an attempt to provide a more reliable fixation for the soft tissue ACL graft, numerous different ACL graft fixation devices have been developed (Brand et al. 1999, 2000a, Clark et al. 1998, Giurea et al. 1999, Kousa et al. 2003ab, Magen et al. 1999, Weiler et al. 1998), but the interference screw still remains the most commonly used device (DeLay et al. 2001, Harner et al. 2001). The strength of interference screw fixation relies almost entirely on the amount of friction the screw is able to provide between the graft and the surrounding walls of the bone tunnel. Because current accelerated rehabilitation protocols after an ACL reconstruction allow immediately full range of motion and full weight-bearing and thus subject the graft to increased loading, the need for a secure initial graft fixation is critical (Beynnon et al. 2002, Uh et al. 1997).

5.3.1. Biomechanical results

In addition to the current series of six experiments, twenty-nine studies evaluating biomechanically interference screw fixation of soft tissue grafts have been published since 1998. Pure human cadaver models were used in eleven of these studies (Aune et al. 1998, Brand et al. 2000b, Caborn et al. 1998, Rittmeister et al. 2001 and 2002, Scheffler et al. 2002, Selby et al. 2001, Simonian et al. 1998, Stadelmaier et al. 1999, Starch et al. 2003, Steenlage et al. 2002) and animal bones (porcine, bovine or ovine) in eighteen (Adam et al. 2001, Becker et al. 2001, Giurea et al. 1999, Harding et al. 2002, Harvey et al. 2003, Kilicoglu et al. 2003, Kousa et al. 2003ab, Magen et al. 1999, Miyata et al. 2000, Nagarkatti et al. 2001, Nakano et al. 2000, Seil et al. 1998, Weiler et al. 1998, 2000, 2001ab, 2002b). In four of the eleven human cadaver studies (Brand et al. 2000b, Caborn et al. 1998, Simonian et al. 1998, Steenlage et al. 2002), the average age of the donors exceeded the recommended upper limit (65 years) for the cadavers to be used in ACL graft fixation studies (Beynnon and Amis 1998), whereas the remaining seven human cadaver studies fulfilled the specimen requirements and are reviewed below (Rittmeister et al. 2001 under 5.3.3.).

Aune et al. (1998) compared biomechanically quadrupled semitendinosus-gracilis and BPTB grafts fixed into femoral bone tunnels with a metal interference screw. Five pairs of fresh frozen human cadaver femurs were used. The grafts were pulled out at a rate of 30 mm/s by an axially applied load. The mean failure load for the BPTB graft fixations was 505 N, 110% stronger than the mean failure load for the hamstring tendon graft fixations, which was 240 N (P=0.003). The authors concluded that the interference screw fixation principle used for hamstring tendon grafts is inferior to that for BPTB graft reconstructions.

The results of Aune et al. (1998) were confirmed in the study by Scheffler et al. (2002), where the fixation of BPTB grafts secured with metal interference-screws was compared to the hamstring tendon graft fixations with metal and bioabsorbable interference-screws and other fixation methods (endo-buttons, washers and sutures tied over a post). Also the impact of fixation level was investigated (anatomic, close-to-joint interference screw fixation versus non-anatomic, extra-cortical fixation with endo-buttons, washers and sutures tied over a post). Eight fresh frozen human cadaver femur-graft-tibia complexes were used in each study group. A cyclic anterior drawer with increasing loads of 20 N increments was applied at 30° of knee flexion. In the hamstring tendon group with metal interference screw fixation, slippage of the graft out of tibial tunnel occurred with the interference screw left in place and with no apparent tendon lacerations at loads around 200 N, and seven of eight specimens had failed at loads of 300 N, whereas only three of the eight metal interference screw fixed BPTB grafts and four of the hamstring tendon grafts with an attached bone block fixed with bioabsorbable interference screws had failed at loads of 300

N. However, based on their overall results, the authors concluded that anatomic fixation should be preferred for anchorage of hamstring tendons and fixation methods with linkage materials between the graft and the fixation device should be avoided. Direct soft-tissue fixation with interference screw was noticed to allow considerable graft slippage and the authors recommended the use of a bone block or application of a backup or hybrid fixation especially on the tibial fixation site.

Stadelmaier et al. (1999) studied the influence of screw length on the cyclic pull-out strength of hamstring tendon graft fixation with metal interference screws. Hamstring tendon grafts were fixed into the bone tunnels of eight pairs of fresh frozen human cadaver tibiae with 25 mm or 40 mm long metal interference screws. The specimens were loaded cyclically with the loading axis in line with the graft-tibial tunnel complex at a frequency of 0.5 Hz, with a 5-N increase in the amplitude of the applied load per cycle until failure of the graft fixation, as indicated by a drop in the force required for elongation of the graft in response to loading. The graft was then subjected to uniaxial loading to failure under a constant rate of displacement of 30 mm/sec to 25 mm. All grafts failed at the fixation site with the tendon being pulled past the screw. There were no measurable differences in the mean cyclic failure strength, pull-out strength (336 N in both groups), or stiffness between the two screw lengths.

The effect of screw length on interference screw fixation in a tibial bone tunnel was also investigated by Selby et al. (2001). Hamstring tendon grafts were fixed into bone tunnels of sixteen paired fresh frozen tibiae with 28-mm or 35-mm bioabsorbable interference screws. The grafts were then preloaded to 25 N of force and cycled from 0 to 50 N of force, after which they were subjected to 20 mm/min of traction force parallel to the axis of the grafts. The mean maximum load at failure of the 28-mm and 35-mm screws were 595 N and 825 N, respectively ($P<0.004$).

Rittmeister et al. (2002) studied the components of laxity in the interference fit fixation of quadrupled hamstring tendon grafts by using an infrared optical system. Twenty-three fresh-frozen human cadaver tibia-hamstring constructs were used in the study. The hamstring tendon grafts were pretensioned with 25 lbs. (111 N) for 20 minutes before fixation into tibial bone tunnels using bioabsorbable or metal interference screws. A preload of 5 N was then used before repetitive sub-failure loading in the direction parallel to the long axis of the bone tunnel. Each graft was loaded in tension at 25 N/sec to a peak value that was increased by 25 N per cycle from a starting load of 0 N. The construct was then unloaded in 1 second and left under zero load for 60 seconds before reloading, to allow time-dependent recovery of the graft. Loading was carried out until the displacement of the construct exceeded 15 mm. All constructs failed by slippage of the graft past the screw. On average, graft slippage accounted for 92% of total construct laxity (5 mm total construct elongation), whereas permanent stretching of the mid-substance of the graft accounted for

only 8%. Movement of the interference screw in the tibial bone tunnel was minimal, averaging 0.01 mm (range, 0 to 0.5) at construct failure. The authors concluded that inadequacy of isolated tibial interference screw fixation of soft tissue grafts may be overcome if early failure at the graft-screw interface at load levels as low as 150-200 N is prevented.

Starch et al. (2003) recently published their investigation on multi-stranded hamstring tendon graft fixation with a central four-quadrant (Intrafix) or a standard tibial interference screw for ACL reconstruction. Eight pairs of fresh-frozen human cadaver knees and hamstring tendon grafts were used in the study. ACL reconstructions were performed using either a metal interference screw or a plastic central sleeve and screw (Intrafix) on the tibial side, and a crosspin on the femoral site. During testing, the tibia was loaded at 25 N/sec in an anteroposterior direction with the knee flexed to 30°, simulating the Lachman test. After application of a 10-N pretensioning load, each specimen was cyclically loaded in anteroposterior direction. The amplitude of the loading cycles started at 25 N and increased by 25 N per cycle to a maximum of 400 N. After the peak load was reached in each cycle, the specimen was unloaded and allowed to rest for 30 seconds before reloading. All knees that remained intact during the incremental loading test phase and reached the maximum load of 400 N were pulled to failure at the rate of 25 N/sec. The load required to cause 1 and 2 mm of graft laxity, defined as the separation of the femur and the tibia at the points of graft fixation, was significantly greater with the sleeve and screw than with the interference screw (at 2 mm 216 N versus 167 N). The force at initial slippage for each of the graft strands was significantly higher with the use of the central sleeve and screw.

5.3.2. Clinical results

Shaieb et al. (2002) recently reported good clinical results using soft tissue grafts and metal interference screws. Seventy patients with patellar or hamstring tendon autografts were evaluated at least two years after surgery. No significant differences were noted between groups for Lysholm score, reduction in activity, KT-1000 arthrometer findings, quadriceps muscle size, return to sports, or ability to jump and do hard cuts and pivots. Significantly more patients in the patellar tendon group had patellofemoral pain at six months after surgery than did the hamstring tendon patients (48% versus 20%), and at the final follow-up the incidence of patellofemoral pain was 42% and 20%, respectively. Fourteen patients in the patellar tendon group and seven in the hamstring tendon group had loss of motion (approximately five degrees). Four patients (two in each group) had treatment failures and their results were not included in the clinical examination data. At the two-year follow up, 97% of patients with patellar tendon graft and 100% of patients with hamstring tendon graft rated the outcome of surgery as good or excellent. The authors concluded that the

interference screw fixed hamstring tendon grafts performed similarly to patellar tendon grafts, although fewer patients in the hamstring tendon group had patellofemoral pain and loss of motion.

Scranton et al. (2002) performed a prospective multicenter observation cohort study to evaluate metal interference screw fixation of quadruple hamstring tendon grafts in ACL reconstruction. One hundred-twenty patients were seen at two years follow-up. They were evaluated using Lysholm score, Lachman test, anterior drawer test, pivot-shift test, KT-1000, effusion assessment, and the Tegner Sports Activity Scale. The average Lysholm score increased by 42 points (range, 0 to 100 points) and the Lachman test, the effusion assessment, the anterior drawer test, the KT-1000, as well as the Tegner Sports Activity Scale scores all improved. Of the 120 ACL reconstructions, five failed. Of these, three failed from new late injury, one from technical error, and one from patient non-adherence to rehabilitation protocol. Some anterior knee pain was present in 30% of patients, and 22% had at some time experienced hamstring pain that did not interfere with athletic activity.

Colombet et al. (2002) reported a case series with 200 ACL reconstructions using four-strand semitendinosus and gracilis tendon grafts and metal interference screw fixation. All the patients included in the study were over 18 years of age, had a healthy contralateral knee, intact posterior cruciate ligament, and no peripheral surgical procedure or cartilage injury. Patients having undergone prior ligament reconstruction were excluded from the study. Clinical review allowed for documentation of International Knee Documentation Committee (IKDC), KT-1000 arthrometer laxity measurement, and isokinetic dynamometric analysis. The minimum follow-up was one year. Quadriceps and hamstring muscle strength loss was in average less than 17%, and, 50% of patients graded A, 44% B, and 6% C or D according to the overall IKDC evaluation (A the best grade). The anterior laxity was graded A for 157 patients with a median (side-to-side difference) of 1 mm. Of the 113 high-performance athletes, 98 (86%) had resumed a pre-injury level of sporting activity. The authors concluded that the ACL reconstruction using four-strand hamstring autograft is a safe, reliable and reproducible procedure.

Ejerhed et al. (2003) recently published their prospective randomized clinical trial with a two-year follow-up comparing BPTB and semitendinosus tendon grafts fixed with interference screws in ACL reconstruction. Seventy-one patients were seen at the two-year follow-up and the outcome assessed by physiotherapists not involved in the primary treatment. No differences were found in terms of the Lysholm score, Tegner activity level, KT-1000 arthrometer side-to-side laxity measurement, single-legged hop test, or International Knee Documentation Committee classification results. The knee-walking test was rated difficult or impossible to perform by 53% of the BPTB in comparison to only 23% of the semitendinosus patients, a difference that was

statistically significant. The authors concluded that the interference screw fixed semitendinosus tendon graft is an equivalent option to the BPTB graft for ACL reconstruction.

Pinczewski et al. (2002) recently published the results of their prospective clinical trial in 180 patients comparing outcomes of patellar tendon versus four-strand hamstring tendon autograft fixed with metal interference screw and followed for a minimum of five years after the reconstruction. According to the International Knee Documentation Committee assessment, more than 85% of each group had an overall score of A or B. The median Lysholm knee score was greater than 90 for both groups at two and five years, and the instrumented testing revealed no significant difference between the two groups beyond three years. Thirty-one percent of the patellar tendon group and 19% of the hamstring tendon group had a fixed flexion deformity at five years. Radiological assessment revealed early osteoarthritic changes in four percent of the subjects in the hamstring tendon group versus 18% in the patellar tendon group at five years. The authors concluded that arthroscopic reconstruction with either graft results in a similar surgical outcome, reliably restoring knee stability over a five-year period, and that patients with patellar tendon grafts are at a greater risk of developing early signs of osteoarthritis. However, some caution is warranted in the interpretation of the results of this study, as the authors did not use an intent-to-treat analysis, and included only subjects that had successful outcomes in their data analysis (removed failures from their data analysis).

5.3.3. Attempts to increase and predict the strength of interference screw fixation

In order to further increase the fixation strength of interference screws, modulation of the fixation conditions (in the vicinity of the implant), for example via compaction of the bone tunnel walls by specifically designed drill bits or sets of dilators of increasing diameter, has been advocated by some authors (Fu et al. 2000, Johnson 1998). Theoretically, bone tunnels with denser walls should provide optimal conditions for rigid fixation of soft tissue grafts by minimizing the chance of screw divergence, convergence, migration, and loosening when interference screw is used. Although bone compaction has been reported to increase the initial stiffness and ultimate fixation strength of the fixation of stainless steel porous-coated plugs in a canine model (Green et al. 1999), as well as the stiffness of the femur-post fixation of hamstring tendon grafts in ACL reconstruction in a human cadaver model (To et al. 1999), the only published study evaluating the effectiveness of bone-tunnel compaction on the fixation strength of interference screw fixed soft tissue graft in human cadaver model found no significant difference between the fixation strengths obtained using bone-tunnel compaction and conventional extraction drilling (Rittmeister et al. 2001). The investigators fixed hamstring tendon grafts with a metal interference screw into serially

dilated or extraction drilled tibial bone tunnels in fourteen pairs of fresh-frozen human cadaver knees. Each graft was loaded cyclically parallel to the long axis of the bone tunnel with a ramped amplitude function that increased the peak load applied to the graft by 25 N per cycle from a starting load of 0 N. Loading was performed until total displacement of the construct under load exceeded 15 mm compared to the initial value. The mean failure loads were 360 ± 120 N and 345 ± 88 N in the dilated and extraction drilled specimens, respectively (P=0.74).

Regarding the size of the bone tunnel, it is currently recommended clinically that for optimal fixation strength, the diameter of the bone tunnel should closely match the diameter of the graft (Steenlage et al. 2002). However, according to a canine study made by Yamazaki et al. (2002), graft-tunnel diameter disparity of up to 2 mm may not adversely affect intraosseous healing of the flexor tendon graft *in vivo* and surgeons need not to be overly concerned about minor graft-tunnel diameter disparities.

A novel and resourceful idea was recently proposed by Brand et al. (2000b), as they suggested that preoperative measurement of the femoral and tibial bone density by dual-energy x-ray absorptiometry (DXA), or alternatively, peroperative determination of the torque required to insert the interference screw into the femoral or tibial drill hole could actually be used to *predict* the strength of fixation of the ACL soft tissue graft. If these measurements actually accurately and precisely predicted the strength of fixation of the graft, the surgeons would then be able to identify those at risk of graft slippage and failure, and consequently, take the necessary precautions before (e.g., choice of graft and fixation implants), during (e.g., hybrid fixation in the tibia), and after the ACL reconstruction (e.g., less aggressive rehabilitation).

5.4. GRAFT PRETENSIONING, PRECONDITIONING, AND INITIAL TENSIONING

Progressive creep or slippage of the graft fixed in a bone tunnel is one of the most common concerns associated with the use of the soft tissue grafts as ACL reconstruction material (Brand et al. 1999, Fu and Ma 1999b, Simonian et al. 1998). Besides the slippage of the graft past the screw, the viscoelasticity of tendon grafts may also affect the stability of the knee joint postoperatively, as viscoelastic materials elongate with time if kept under a constant load (Beynnon et al. 1994, Graf et al. 1994, Schatzmann et al. 1998, Yoshiya et al. 2002, Smith et al. 1993, Woo et al. 1999). Because of this graft creep, the intra-articular portion of the tendon graft may elongate and the graft tension decrease postoperatively. This has been demonstrated in the primate patellar tendon where the stress in the tendon was reduced to 70% of the initial stress within 30 minutes (Butler 1989). Pretensioning and preconditioning of soft tissue grafts using graft preparation boards and tensiometers have been proposed as a countermeasure to this undesired effect, minimizing the graft

creep-induced graft elongation so that the initially set graft tension would be better maintained after the final graft fixation, and ultimately, the development of knee laxity after ACL reconstruction could be prevented (Amis and Jakob 1998, Schatzmann et al. 1998, Yasuda et al. 1997). According to Woo et al. (1999), preconditioning of the graft can reduce the amount of creep by approximately 50% when compared with non-preconditioning. On the other hand, total elimination of creep is probably not possible clinically, as Schatzmann et al. (1998) have reported that more than 160 cycles of extremely intensive loading (cycling between 75 and 800 N at 0.5 Hz frequency) of quadriceps tendons and patellar ligaments is required to reach a steady state where no further graft elongation occurs. Further, regarding the mode of preconditioning, although it has been proposed that cyclic preconditioning is particularly efficient in eliminating graft creep (Schatzmann et al. 1998), Graf et al. (1994) reported that no differences were observed in the relaxation behavior of specimens that were cyclicly or isometrically preconditioned when evaluating the effect of preconditioning on the viscoelastic response of primate patellar tendon grafts.

In addition to preventing joint laxity, appropriate graft tension is proposed to be needed as stimulus for proper orientation of the forming collagen fibers during the remodeling phase of the graft (Frank et al. 1983, Yamakado et al. 2002). Conversely, an inappropriate (too high or low) initial graft tension can lead to abnormal knee kinematics and have adverse effects on the postoperative remodeling process and the mechanical properties of the graft, which, in turn, can secondarily result in increased laxity (Amis and Jakob 1998, Arms et al. 1990, Katsuragi et al. 2000, Numazaki et al. 2002, Tohyama and Yasuda 1998, Yoshiya et al. 1987 and 2002). Any possible changes in the intra-articular graft length and tension after implantation are influenced by such a great number of different variables (e.g., graft material and its properties, fixation method, knee flexion angle at surgery, tunnel placement and initial fixation-to-fixation length, postoperative rehabilitation, biologic response of the graft consisting of initial ischemic necrosis and subsequent remodeling, revascularization and ligamentization) that it is difficult to precisely determine how much tension is (or should be) actually retained within the anterior cruciate ligament graft postoperatively (Amis and Jakob 1998, Beynnon et al. 2002, Burks and Leland 1988, Tohyama and Yasuda 1998, Yoshiya 2002).

Although an obvious controversy seems to exist concerning the ideal pretensioning, preconditioning and initial tensioning of the soft tissue grafts in ACL reconstruction, consensus prevails that there is a relatively wide safe window between a "too loose" and a "too tight" graft (Amis and Jakob 1998, Katsuragi et al. 2000, van Kampen et al. 1998, Yasuda et al. 1997). To date, no clinical study has managed to show any statistically significant differences in the postoperative clinical (functional) outcome between different pretensioning, preconditioning and initial tensioning

protocols (Fu et al. 1999a, van Kampen et al. 1998, Yoshiya et al. 2002), although Yasuda et al. (1997) observed in their study of autogenous doubled semitendinosus and gracilis tendons connected in series with polyester tape that patients with a 80-N initial tension showed significantly less laxity at two years postoperatively than patients with a 20-N or 40-N initial tension. Based on their results, Yasuda et al. recommended that 80-N initial tension should be used to obtain an excellent outcome in ACL reconstruction with the autogenous doubled semitendinosus and gracilis tendons, especially concerning the stability of the knee. However, initial tension higher than 80 N has been shown not to provide any further advantage (Numazaki et al. 2002), but rather to cause adverse effects on the graft (Katsuragi et al. 2000, Yasuda et al. 1997, Yoshiya et al. 1987). Concerning the current clinical practice, the average initial graft tension for hamstring tendon grafts used by the surgeons is approximately 70 N (Amis and Jakob 1998). Also, Cunningham et al. (2002) recently published a survey evaluating the graft tensioning and found that the average normal initial tension used by sports medicine-trained orthopedists is 60 ± 29 N. Finally, the clinical results associated with the most commonly used tensioning protocol (i.e., the unmeasured initial tension or maximal sustained one-handed pull) have been generally good (Cunningham et al. 2002, Howell et al. 2001, Nabors et al. 1995, Yoshiya et al. 2002).

Few published articles have dealt with the residual graft tension during and after final fixation with particular devices. Shino et al. (2002) reported gradually decreasing residual tension of isometrically preconditioned (49 N or 98 N for 5 minutes) double-looped bovine tendons fixed into a porcine bone tunnel with a Double Spike Plate. The graft tension was observed to increase immediately after fixation, but to decrease back to the initially set graft tension level already in five minutes.

6. AIMS OF THE STUDY

The aims of the study were to investigate:

1. the effect of bone tunnel compaction on the initial fixation strength of the interference screw fixed soft tissue grafts in ACL reconstruction (I, III, IV),

2. the suitability of porcine tissues as surrogates for human tissues in biomechanical ACL graft fixation studies (I, II),

3. the effect of soft tissue graft preconditioning on the maintenance of initial graft tension in ACL reconstruction with interference screws (V),

4. the suitability of bone mineral density and interference screw insertion torque as predictors of the ACL graft fixation strength (VI).
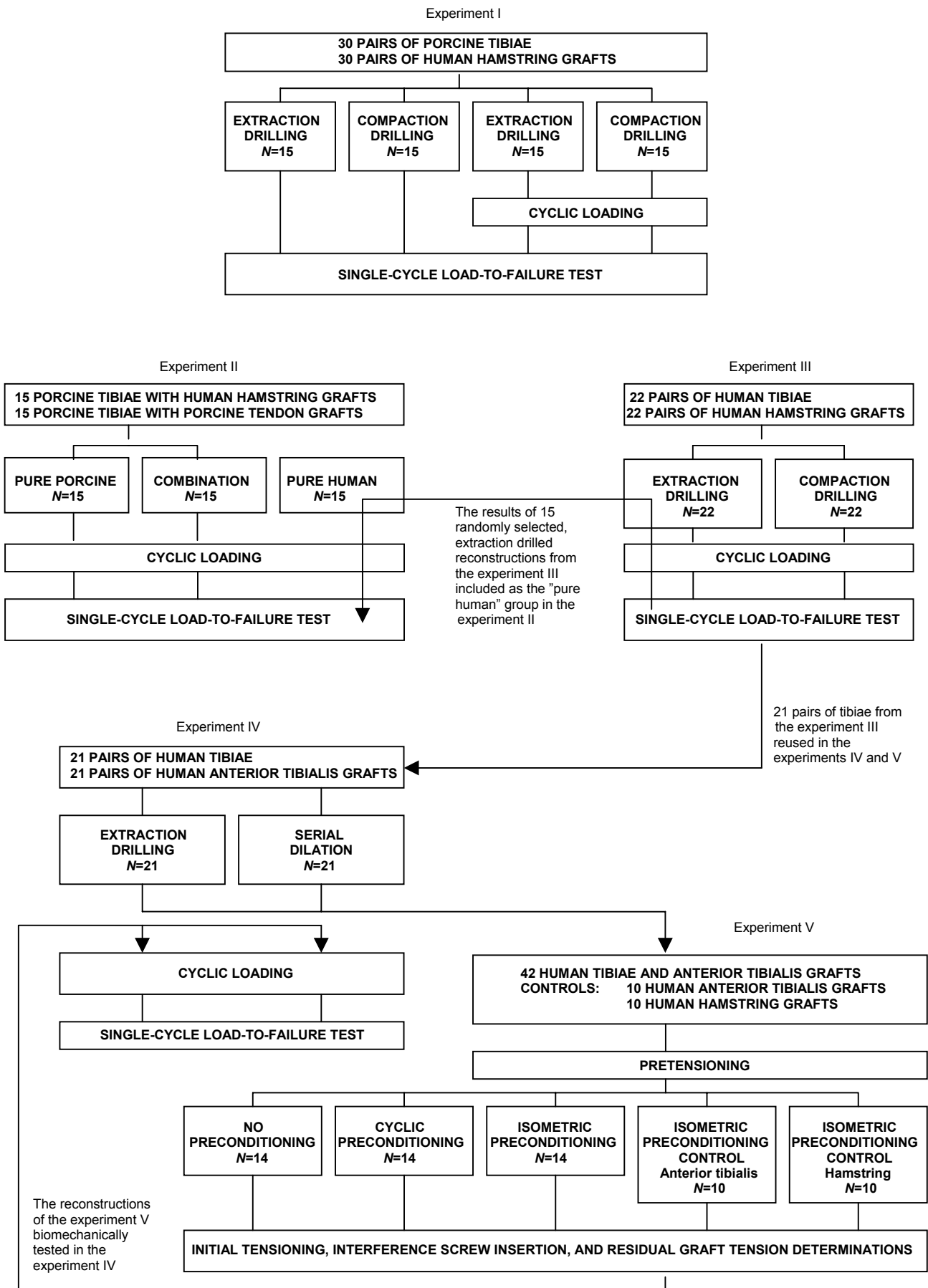
7. MATERIALS AND METHODS

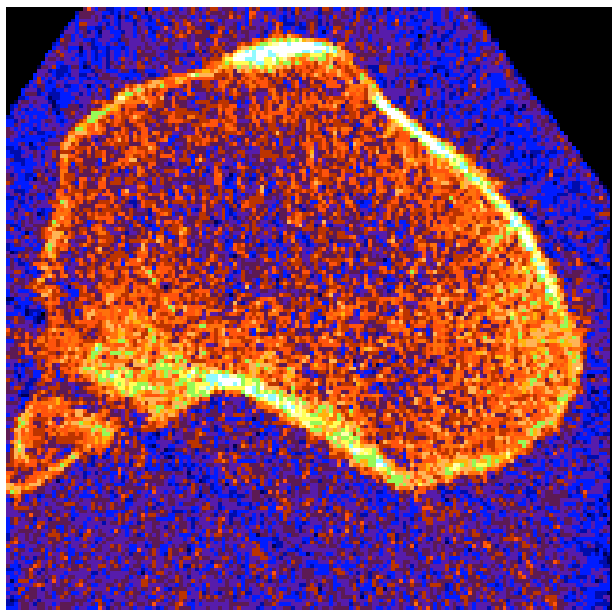## 7.1. SPECIMENS, STUDY GROUPS, AND CONTROL PERSONS

Human cadaver tibiae, hamstring tendon grafts (semitendinosus and gracilis) and anterior tibialis tendon grafts, and, porcine tibiae and tendon grafts were harvested and used in the experiments of this thesis (I-VI). The specimens and study groups (I-V) are summarized in Figure 3. The data of the sixth experiment was extracted directly from the results of the forth experiment (IV). The sample sizes (*N* of specimen per study group) in the experiments I-VI were 15, 15, 22, 21, 14 (10 in the control groups), and 21, respectively (Fig. 3). The harvesting procedure and use of human cadaver tissues were approved by the National Authority for Medico-Legal Affairs in Finland. Additionally, 21 women [mean age (± SD) 24 ± 6 years, range 19-38] and 22 men (24 ± 4 years, 19-33) were used as a control group for the bone density measurement (I). All participants gave their informed consent before enrollment to the study, and the protocol was approved by the Institutional Review Board and Ethics Committee of the UKK Institute.

7.2. PERIPHERAL QUANTITATIVE CT MEASUREMENTS

A peripheral quantitative CT scanner (XCT 3000, Stratec Medizintechnik GmbH, Pforzheim, Germany) was used to determine the trabecular bone density (in milligrams per cubic centimeter, mg/cm^3) at proximal tibia before the ACL reconstructions were carried out (Fig. 4). A cross-sectional portion of the proximal tibia approximately 2 cm distal from the articular surface, corresponding to the actual site of the tibial bone tunnel in an ACL reconstruction, was scanned (I-VI). In the first study, the trabecular density was determined in a 2 x 2 cm^2 region of interest corresponding to the site of the tibial bone tunnel in ACL reconstruction (I). In the remaining studies II-VI the trabecular density was determined from the entire cross-section of the proximal tibia to increase the repeatability of the measurement.

Experiment I

30 PAIRS OF PORCINE TIBIAE
30 PAIRS OF HUMAN HAMSTRING GRAFTS

EXTRACTION DRILLING *N*=15	**COMPACTION DRILLING** *N*=15	**EXTRACTION DRILLING** *N*=15	**COMPACTION DRILLING** *N*=15

CYCLIC LOADING

SINGLE-CYCLE LOAD-TO-FAILURE TEST

Experiment II

15 PORCINE TIBIAE WITH HUMAN HAMSTRING GRAFTS
15 PORCINE TIBIAE WITH PORCINE TENDON GRAFTS

PURE PORCINE *N*=15	**COMBINATION** *N*=15	**PURE HUMAN** *N*=15

CYCLIC LOADING

SINGLE-CYCLE LOAD-TO-FAILURE TEST

The results of 15 randomly selected, extraction drilled reconstructions from the experiment III included as the "pure human" group in the experiment II

Experiment III

22 PAIRS OF HUMAN TIBIAE
22 PAIRS OF HUMAN HAMSTRING GRAFTS

EXTRACTION DRILLING *N*=22	**COMPACTION DRILLING** *N*=22

CYCLIC LOADING

SINGLE-CYCLE LOAD-TO-FAILURE TEST

21 pairs of tibiae from the experiment III reused in the experiments IV and V

Experiment IV

21 PAIRS OF HUMAN TIBIAE
21 PAIRS OF HUMAN ANTERIOR TIBIALIS GRAFTS

EXTRACTION DRILLING *N*=21	**SERIAL DILATION** *N*=21

CYCLIC LOADING

SINGLE-CYCLE LOAD-TO-FAILURE TEST

Experiment V

42 HUMAN TIBIAE AND ANTERIOR TIBIALIS GRAFTS
CONTROLS: 10 HUMAN ANTERIOR TIBIALIS GRAFTS
10 HUMAN HAMSTRING GRAFTS

PRETENSIONING

NO PRECONDITIONING *N*=14	**CYCLIC PRECONDITIONING** *N*=14	**ISOMETRIC PRECONDITIONING** *N*=14	**ISOMETRIC PRECONDITIONING CONTROL** Anterior tibialis *N*=10	**ISOMETRIC PRECONDITIONING CONTROL** Hamstring *N*=10

The reconstructions of the experiment V biomechanically tested in the experiment IV

INITIAL TENSIONING, INTERFERENCE SCREW INSERTION, AND RESIDUAL GRAFT TENSION DETERMINATIONS

Figure 3. The specimens and study groups (Studies I-V).

Figure 4. Peripheral quantitative CT image of the proximal human tibia.

7.3. SPECIMEN PREPARATION

The harvested tendons and tibiae were cleared of adherent muscle fibers and surrounding soft tissues, wrapped in saline-soaked gauze and stored frozen in sealed plastic bags. These preservation procedures have been recommended for knee specimen intended for *in vitro* testing protocols of the cruciate ligaments and ligament reconstructions (Beynnon and Amis 1998) and have been shown not to affect the mechanical properties of bones (Pelker et al. 1984). On the day of testing, the tendons and tibiae were thawed to room temperature. All of the specimens were kept moist with physiologic saline solution during specimen preparation, fixation procedures, and biomechanical testing.

7.3.1. Hamstring tendon grafts (I-III, V)

A four-strand hamstring (semitendinosus and gracilis) tendon graft with a total graft length of 8 cm was constructed according to standard principles (Fig. 5). In short, the four strands were sutured at the free end for 40 mm with No. 2 suture material by using the running baseball stitch while maintaining constant tension on all four strands. The diameter of the graft was measured at the sutured end with a Graft Sizing Tube (Acufex Microsurgical Inc, Mansfield, Massachusetts, USA) and was identical for each set of paired specimens. In the study V, the graft was pretensioned using the Graftmaster II System board (Acufex) for 15 minutes, starting at 20 pounds (88 N) of tension.
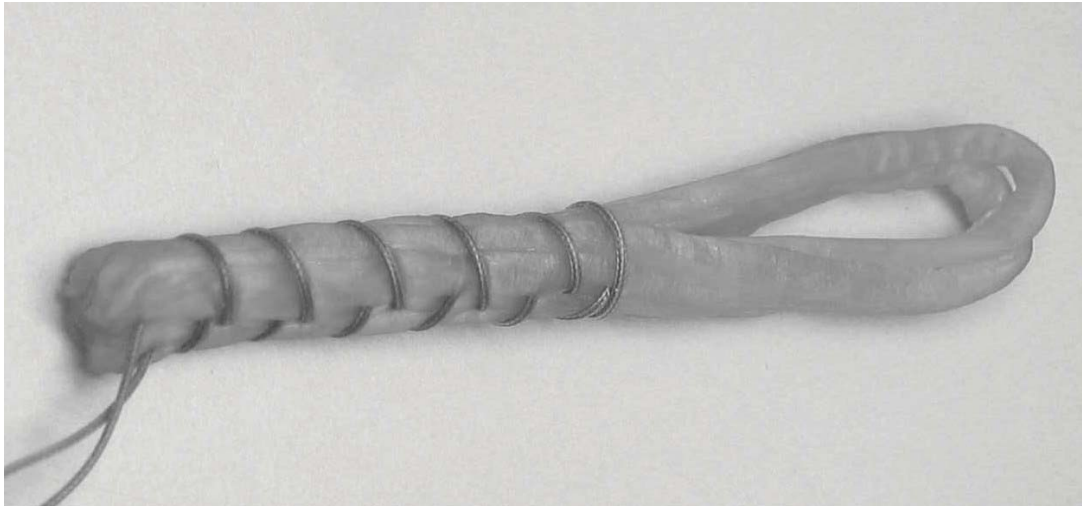
Figure 5. A four-strand hamstring tendon graft.

7.3.2. Anterior tibialis tendon grafts (IV-VI)

A looped (double-stranded) anterior tibialis tendon graft with a total graft length of 8 cm was constructed according to the technique described by Charlick and Caborn (2000). In short, the tendon was folded to form a graft with two strands and while maintaining constant tension on both strands, the graft was sutured at the free end for 40 mm with No. 2 suture material using the running baseball stitch. The diameter of the graft was measured at the sutured end with a Graft Sizing Tube (Acufex) to ensure that the diameters were identical for each set of paired specimens. The graft was pretensioned using the Graftmaster II System board (Acufex) for 15 minutes, starting at 20 pounds (88 N) of tension (Fig. 6).
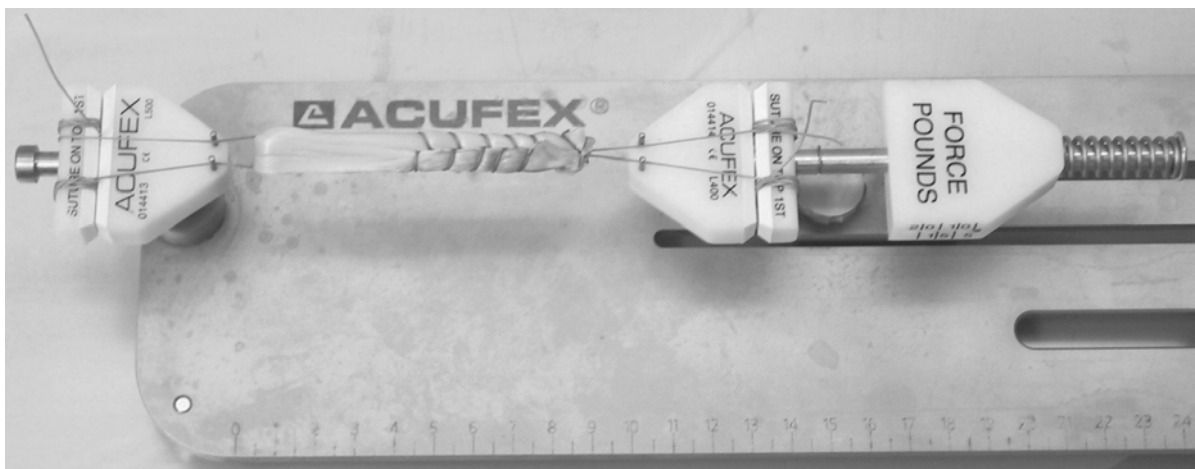
Figure 6. Pretensioning of a two-strand anterior tibialis tendon graft.

7.3.3. Porcine soft tissue grafts (II)

A porcine soft tissue graft model described by Ishibashi et al. (1997) and Harding et al. (2002) was used. Briefly, the porcine patellar tendon was cut approximately 8 cm distal from its patellar insertion and left attached to the patella (Fig. 7). The free end of each patellar tendon was cleared of soft tissue and sized to fit through a 8-mm Graft Sizing Tube (Acufex) corresponding to the average hamstring tendon graft diameter (Graham and Parker 2002, Howell and Taylor 1996). While maintaining constant tension on the graft, the graft was sutured at the free end for 40 mm with No. 2 suture material by using the running baseball stitch, and a 6-mm diameter hole was drilled through the center of the patella for rigid fixation of the patella into the mechanical testing machine.

Figure 7. Porcine patellar tendon used as a soft tissue graft.

7.3.4. Tibial bone tunnels

A bone tunnel equal to the diameter of the graft and approximately 40 mm in length was drilled into each tibia over a guide wire where the tunnel would be if an actual ACL reconstruction were being performed by using either extraction drilling with a conventional drill bit (Acufex) or compaction drilling with a Stepped Router (Acufex) (I-III) (Fig. 8). In the studies IV-VI, the previously (the study III) drilled bone tunnels ($\varnothing \leq 8.5$ mm, average 7.8 ± 0.6 mm), were enlarged to the desired final diameter of 10 mm either by successive placement of increased diameter Tunnel Dilators (Arthrex Inc., Naples, Florida, USA) with 0.5 mm increments (the previously extraction-

drilled specimens) or by drilling with a conventional cannulated 10-mm drill bit (Acufex) (the previously compaction-drilled specimens). The diameter of the bone tunnel was identical for each set of paired tibiae.
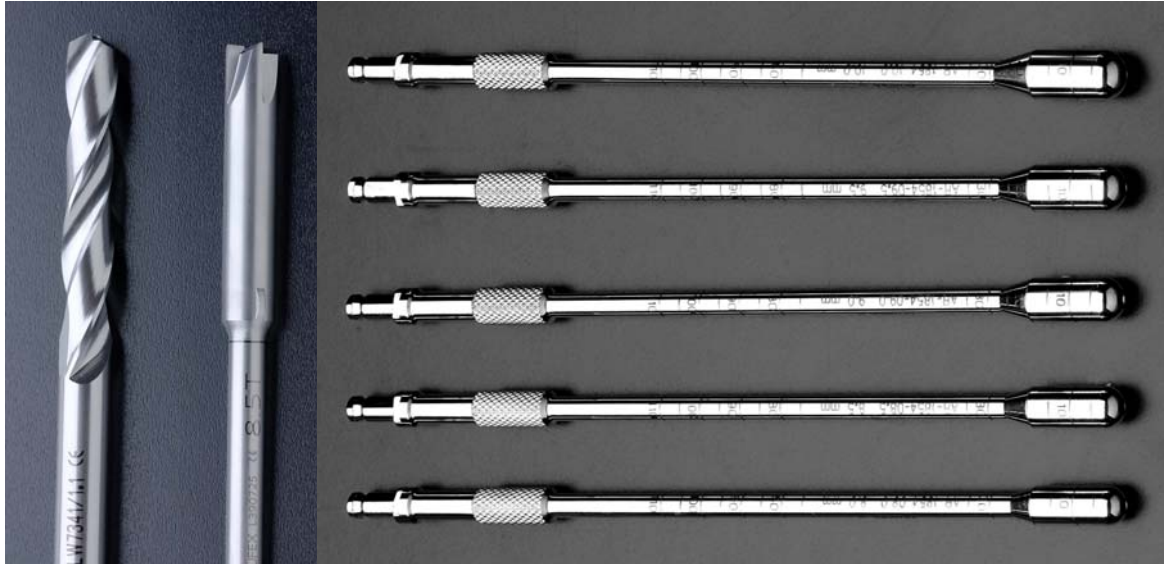

Figure 8. A conventional cannulated drill bit for extraction drilling (left), Stepped Router for compaction drilling, and Tunnel Dilators for serial bone-tunnel dilation (right).

7.3.5. Preconditioning, initial tensioning and residual graft tension (IV-VI)

The tibia was securely mounted to the mechanical testing machine (Lloyd LR 5K, J J Lloyd Instruments, Southampton, United Kingdom) by specially designed clamps in such a position that the bone tunnel was parallel with the direction of loading (Fig. 9). The graft was drawn through the tibial bone tunnel. The proximal end of the tendon graft was connected to the load cell of the mechanical testing machine by placing a steel bar through the unsutured, looped portion of the graft. The sutured, distal end of the graft was temporarily connected to the mechanical testing machine by tying the sutured free end of the graft to the mount holding the tibia. Then, mimicking the situation where a surgeon is tensioning the graft manually (pulling the sutures by hand or using a tensiometer), the graft was preconditioned with the mechanical testing machine according to one of the following three protocols: No preconditioning, cyclic preconditioning (25 cycles between 0-80 N in 100 seconds) or isometric preconditioning (80 N for 100 seconds). The same preconditioning protocol was used in both grafts of each specimen pair. After preconditioning, an

initial graft tension of 80 N was applied to all grafts and a bioabsorbable interference screw was inserted. Immediately after the completion of the screw insertion, the sutures were cut to make sure that the tension relied solely on the interference screw fixation without any possible support from the sutures. The residual graft tension was recorded immediately and 10 minutes after the screw insertion. Additionally, to eliminate the imminent bias introduced by potential graft slippage past the interference screw, ten pretensioned anterior tibialis and quadrupled hamstring tendon grafts were connected directly (no interference screw fixation or tibial bone) to the mechanical testing machine in study V. Briefly, a steel bar was placed through the unsutured, looped portion of the graft and the other end of the graft (sutured portion) was mounted to the mechanical testing machine with a clamp specially designed to securely fix a soft tissue graft (Fig. 9). Thirty to 35 mm of the graft, corresponding to the normal ACL length, remained between the clamp and the steel bar. The grafts were held parallel with the direction of loading and isometrically preconditioned (80 N for 100 seconds). The residual graft tension was recorded 1 minute (corresponding to the time required to insert the interference screw in the other groups) after 80-N initial tensioning, and 10 and 60 minutes thereafter.
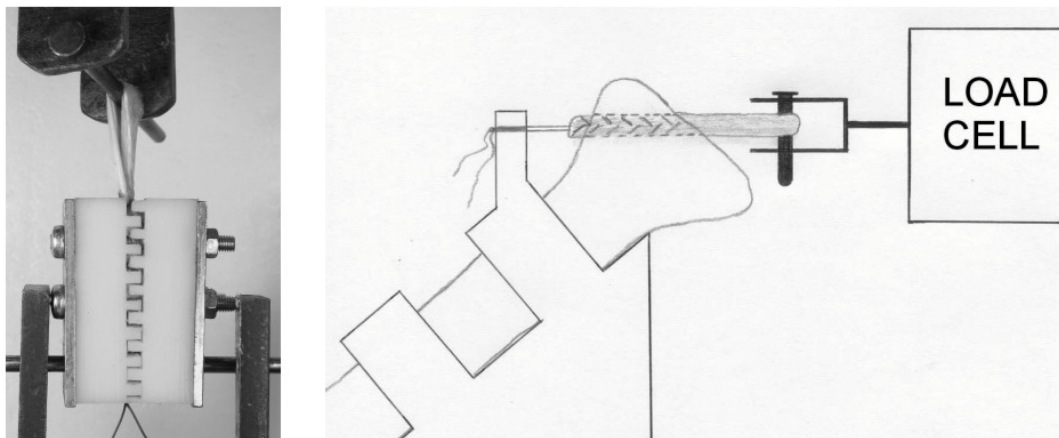
Figure 9. Preconditioning without (left) and with (right) fixation to tibial bone tunnel.

### 7.3.6. Interference screw fixation and insertion torque

Each soft tissue graft was placed through the tibial bone tunnel with 30 to 35 mm of the unsutured, looped portion of the graft protruding from the proximal opening of the bone tunnel. This length corresponded to the normal length of the ACL (Beynnon and Amis 1998). An interference screw (Table 1 and Fig. 10) was inserted in an outside-in fashion over a guide wire

between the graft and the anterior aspect of the tibial bone tunnel. The screw was advanced until its tip reached the proximal bone-tunnel opening. As with the diameter of the soft tissue graft and tibial bone tunnel, the diameter of the interference screw was also identical for each set of paired tibiae. Maximum screw insertion torque was determined using a digital electronic torque meter (Torqueleader TSD 350, MHH Engineering Co. Ltd., United Kingdom) mounted on the screwdriver (IV-VI).

TABLE 1
Interference screws used in studies I-VI

Study	Interference screw	Material	Length (mm)	Diameter (mm)
I	Softsilk (Acufex)	Titanium	25 mm	8 mm
II	BioRCI (Smith & Nephew)	Bioabsorbable	35 mm	8 mm
III	BioRCI (Smith & Nephew)	Bioabsorbable	35 mm	8 mm or 9 mm*
IV-VI	Delta Tapered Bio-Interference Screw (Arthrex)	Bioabsorbable	35 mm	11 mm

* Determined to be 0.5-1.0 mm greater than the diameter of the graft and bone tunnel.
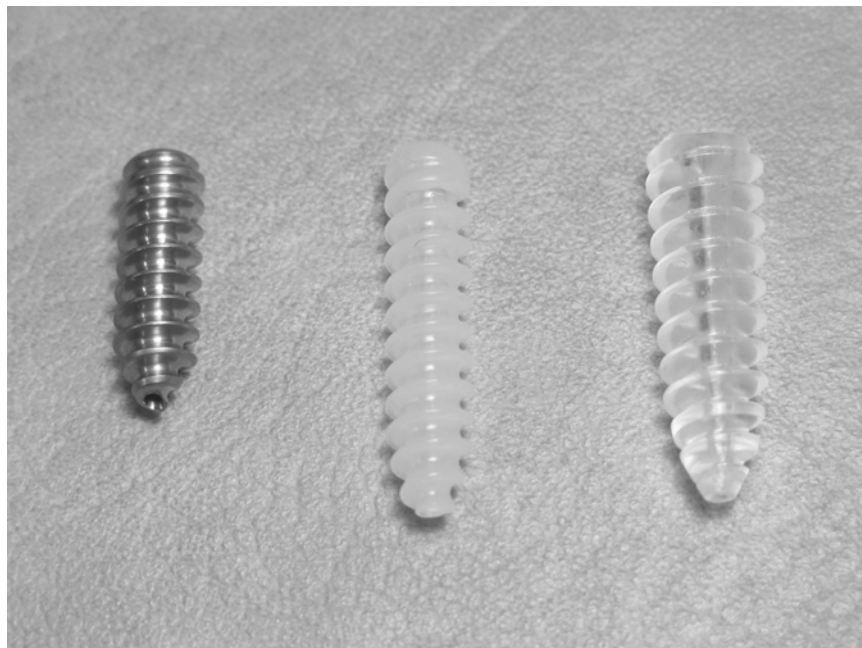
Figure 10. The interference screws used in the study: Softsilk (left), BioRCI, and Delta Tapered Bio-Interference Screw (right).

7.4. BIOMECHANICAL TESTING AND DATA ANALYSIS

The biomechanical tests were performed by using a Lloyd LR 5K mechanical testing machine (J. J. Lloyd Instruments, Southampton, United Kingdom) (Fig. 11). The tibiae were securely mounted to the testing machine by specially designed clamps, and the tendon grafts by placing a steel bar through the unsutured, looped portion of the human cadaver soft tissue grafts or the patellar drill hole of the porcine soft tissue grafts. The biomechanical testing protocol consisted of the cyclic-loading test followed by the single-cycle load-to-failure test.

Figure 11. The mechanical testing machine.

7.4.1. Cyclic-loading test

In the cyclic-loading test, the specimens were first subjected to a 50-N preload for 1 minute. Thereafter, the specimens underwent 1500 loading cycles between 50 and 200 N at a frequency of 0.5 Hz. The loading was parallel to the long axis of the bone tunnel. The response to loading was automatically obtained in the form of a load-displacement curve (Fig. 12). The rate of data acquisition was four times per second. The fixation was evaluated by determining the initial stiffness and the loading-induced increase in the displacement from the preload level to that after 1, 10, 50, 100, 250, 500, 1000, and 1500 cycles of loading, respectively. After 1500 loading cycles, the specimens that survived the cyclic loading were tested with a single-cycle load-to-failure test.
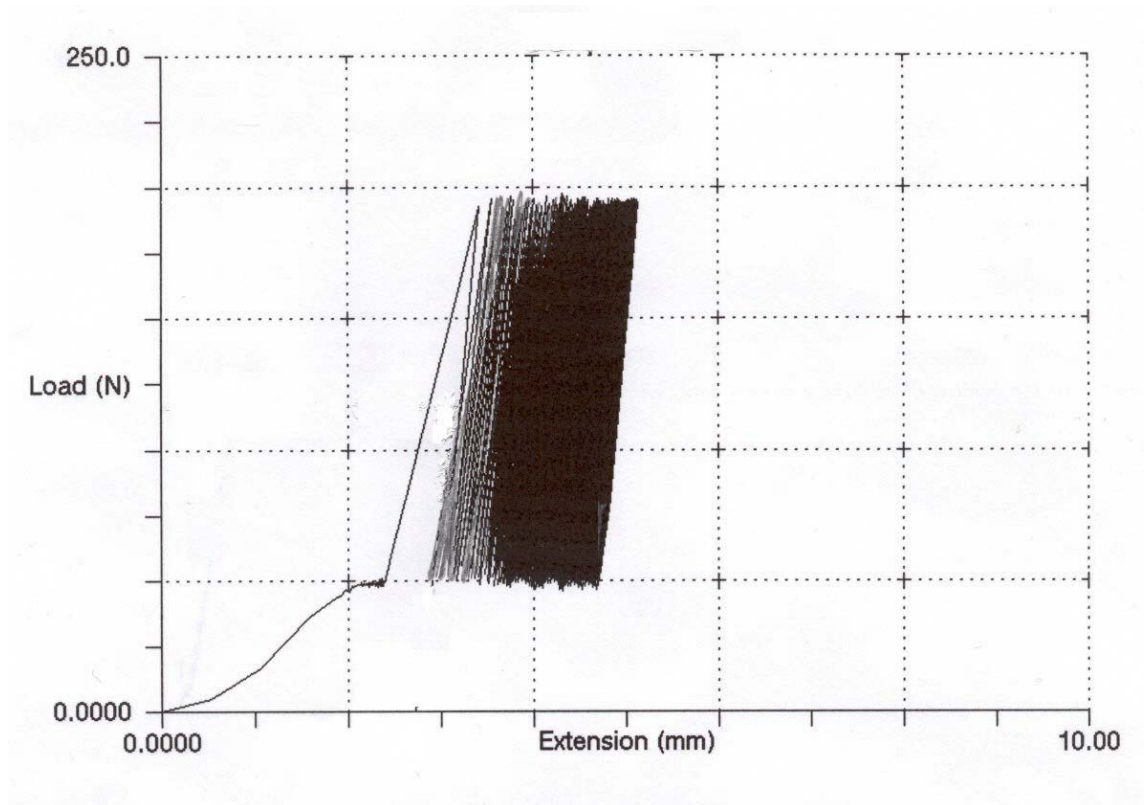
Figure 12. A load-displacement curve from a cyclic-loading test.

7.4.2. Single-cycle load-to-failure test

In the single-cycle load-to-failure test, a vertical tensile loading parallel with the long axis of the bone tunnel was performed at a rate of 1.0 m/min until failure of fixation. The specimen's response to the loading was automatically obtained in the form of a load-displacement curve (Fig.

2). The stiffness (determined as the slope of the linear region of the load-displacement curve corresponding to the steepest straight-line tangent to the loading curve), yield load (described as the load at the point where the slope of the load-displacement curve first clearly decreased), and displacement at yield load were determined. The mode of failure was determined visually.

7.5. STATISTICAL ANALYSIS

The results of the variables were reported as mean and standard deviation (SD) (I-VI). According to a statistical power analysis conducted for the studies III and IV, 21 specimen pairs were needed to obtain a 90% statistical power to detect a difference in the strength of the fixation between the two drilling groups of about 1.0 standardized difference at a significance level of $P<0.05$. In addition to the 21 specimens required according to the power analysis, an extra pair was included in the study III.

A paired sample t-test was used to compare the differences between the extraction drilling and bone-tunnel compaction groups (I, III, IV). The difference between the AT and HT groups was determined by using an unpaired sample t-test (V). One-way analysis of variance with Tukey's test as the post hoc test was used to test the differences between the preconditioning study groups (V). One-way analysis of variance with Bonferroni's test as the post hoc test was used to test the differences between the study groups in the study II. A P-value less than 0.05 was considered statistically significant (I, II, V). Using a Bonferroni correction, a P-value less than 0.0038 (0.05/13) was considered statistically significant in the studies III and IV, as the number of statistical comparisons between the groups was 13.

To determine the associations (correlation) between *volumetric* bone mineral density (vBMD), insertion torque (IT) and the fixation strength of ACL graft (consisting of both cyclic-loading induced displacement after 1500 loading cycles in the cycling-loading test, and yield load in the single-cycle loading test), prediction equations for vBMD and IT based on linear regression analysis ($y = ax + b$) were derived from the learning data. These equations were then validated in terms of accuracy of prediction using the validation data by estimating the fixation strength of each tibia of the validation group using the above described prediction equations. The *accuracy of prediction* was then evaluated by comparing the estimated load values with actual values (obtained in the mechanical testing of the validation group) using the procedure suggested by Bland and Altman (1986), in which the differences between the actual, measured values and estimated values are plotted against their mean, and the 95% limit of agreement are determined as the mean ± twice the standard deviation of the differences. (VI)

8. RESULTS

The main results are summarized in the following tables (2-4) and figures (13-20).

8.1. TRABECULAR BONE DENSITY (STUDIES I-VI)

The trabecular bone density of the proximal porcine tibia was found to be significantly higher than that of the proximal human tibia (Table 2).

TABLE 2
The mean trabecular bone density determined from the entire cross-section of the proximal tibia[a]

Experiment	Group	N	Trabecular bone density (mg/cm^3)
I[b]	Porcine	18	210 ± 45[c]
	Human ♀	21	129 ± 30
	Human ♂	22	134 ± 34
II	Porcine I	15	323 ± 41
	Porcine II	15	337 ± 44
	Human	15	177 ± 32[c]
III	Compaction	22	179 ± 39
	Extraction	22	173 ± 32
IV	Serial dilation	21	176 ± 29
	Extraction	21	182 ± 37
V	No preconditioning	14	180 ± 30
	Cyclic preconditioning	14	182 ± 43
	Isometric preconditioning	14	176 ± 27
VI	Learning data	15	186 ± 28
	Validation data	15	187 ± 39

[a] Mean ± SD
[b] The mean trabecular bone density determined in a 2 x 2 cm^2 region of interest corresponding to the site of the tibial bone tunnel in ACL reconstruction
[c] Significantly different from other groups of the experiment ($P<0.001$)

8.2. THE EFFECT OF BONE TUNNEL COMPACTION (STUDIES I, III, IV)

No significant differences between bone-tunnel compaction (by compaction drilling or serial dilation) and extraction drilling were found with regard to any of the parameters determined during the cyclic-loading and single-cycle load-to-failure tests (Figures 13-16, Tables 3 and 4).
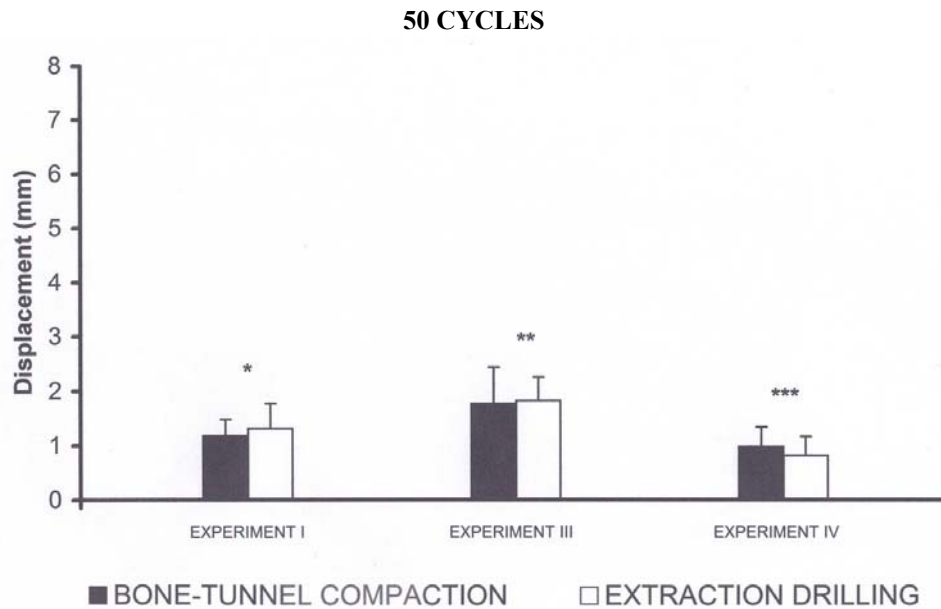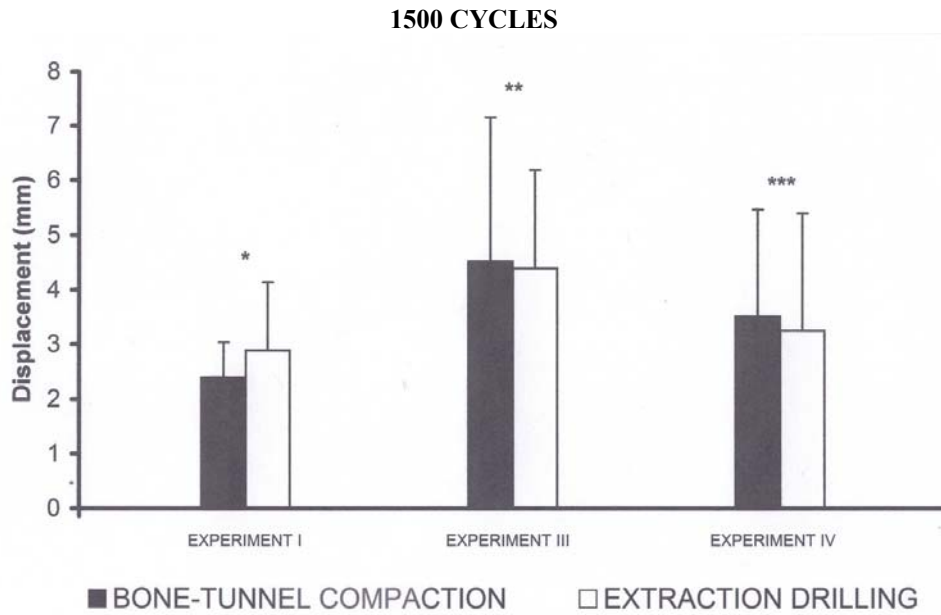
50 CYCLES

Figure 13. Cyclic-loading induced displacement (mean ± SD) after 50 loading cycles.
$* P=0.61 \quad ** P=0.84 \quad *** P=0.30$

500 CYCLES
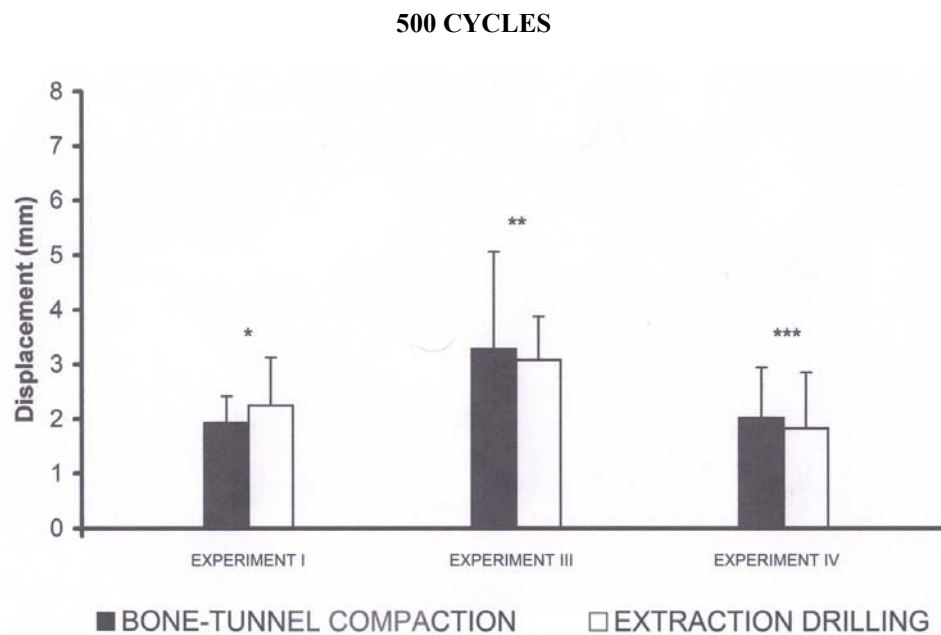
Figure 14. Cyclic-loading induced displacement (mean ± SD) after 500 loading cycles.
$* P=0.42 \quad ** P=0.34 \quad *** P=0.66$

1500 CYCLES

Figure 15. Cyclic-loading induced displacement (mean ± SD) after 1500 loading cycles.
* P=0.36   ** P=0.46   *** P=0.82

TABLE 3
Failures during cyclic loading (I, III, IV)[a]

Experiment	Group	N	Failure mode
I	Compaction	2	1 x tendon rupture 1 x graft slippage
	Extraction	2	2 x tendon rupture
III	Compaction	3	3 x graft slippage
	Extraction	-	
IV	Serial dilation	3	3 x graft slippage
	Extraction	6	6 x graft slippage

[a] All failed specimens and their pairs were excluded from the statistical analysis.
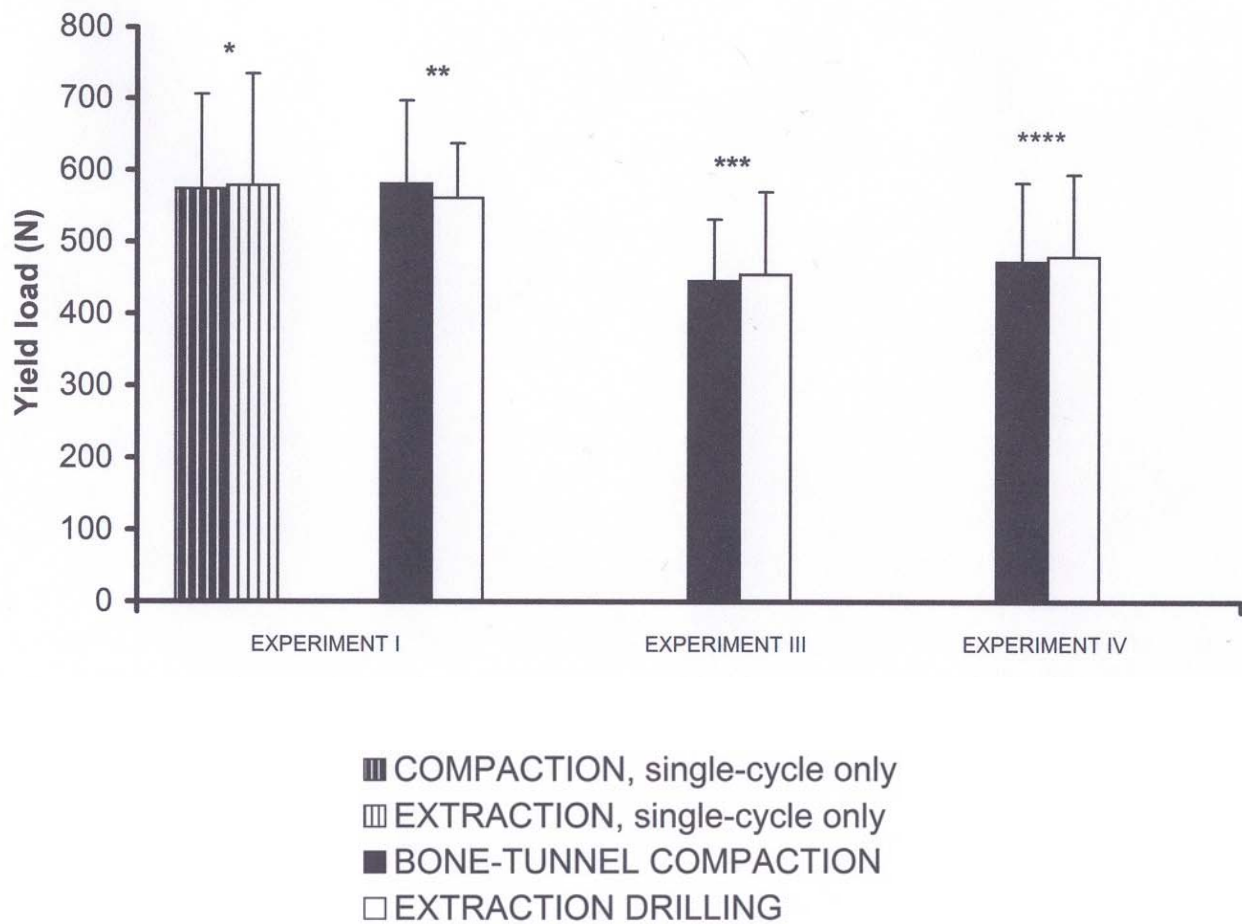
Figure 16. Yield load values (mean ± SD) obtained from the single-cycle load-to-failure tests after cyclic loading. In the first experiment, single-cycle load-to-failure tests were performed both with and without prior cyclic loading. * P=0.88   ** P=0.52   *** P=0.33   **** P=0.97

TABLE 4
Interference screw insertion torque[a]

Group	N	Insertion torque (Nm)
Serial dilation	21	1.7 ± 0.5
Extraction drilling	21	1.6 ± 0.6
P		0.39

[a] Mean ± SD

8.3. PORCINE TISSUES AS SURROGATES FOR HUMAN TISSUES (STUDIES I, II)

In the cyclic-loading test, significant displacement difference was observed between the two porcine groups (Group I and II) and the pure human group (Group III) (Fig. 17). In the subsequent single-cycle load-to-failure test, all groups were significantly different from one another with regard to yield load (Fig. 18).
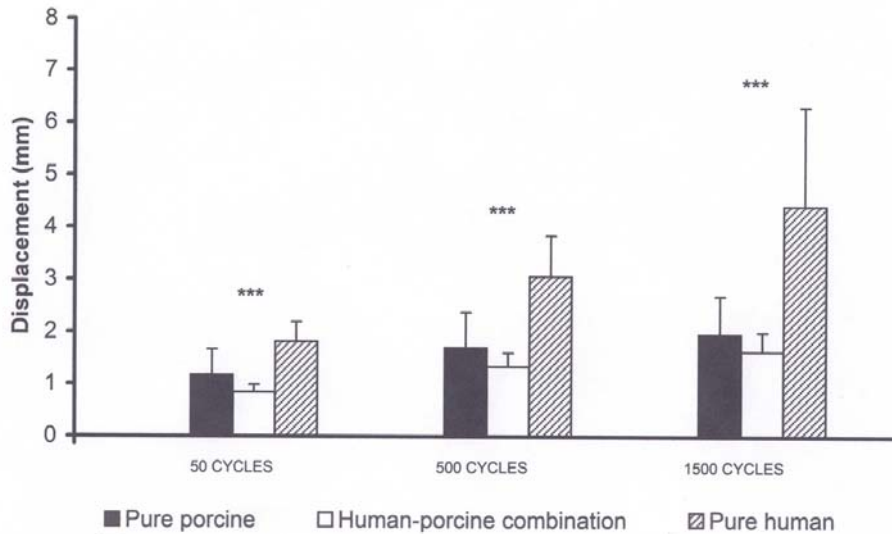

Figure 17. Displacement (mean ± SD) during the cyclic-loading test. *** denote to a statistically significant ($P < 0.001$) difference between Group III and both Groups I and II. After 1500 cycles, the cyclic-loading induced displacement was 2.0 ± 0.7 mm, 1.6 ± 0.4 mm and 4.4 ± 1.9 mm in Groups I, II and III, respectively. No failures of fixation occurred in any of the groups during the cyclic-loading test.

Figure 18. Yield load values (mean ± SD) obtained from the single-cycle load-to-failure tests after cyclic loading. The mode of failure was almost entirely graft slippage past the screw, although some graft laceration (partial rupture) and "graft stretching" was also observed, mainly at the screw-graft interface. In the human-porcine combination group one specimen failed by graft rupture.
* Significantly different from the human-porcine combination ($P<0.001$) and pure human group ($P<0.01$).
** Significantly different from the other groups of the experiment ($P<0.001$).
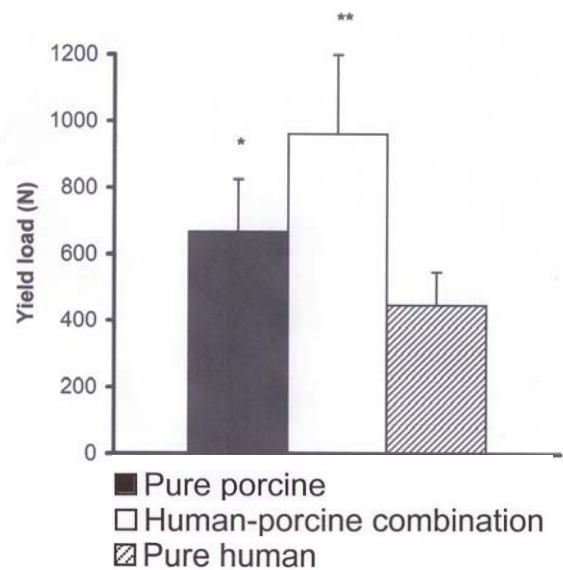
## 8.4. THE EFFECT OF PRECONDITIONING (STUDY V)

Immediately after interference screw insertion, the residual graft tension was nearly unchanged in the Group I, while it was significantly higher in the Groups II and III. Ten minutes after the screw insertion, the corresponding values had decreased, and the mean graft tension in the Group III was significantly higher than that observed in the Group I. For the anterior tibialis and hamstring tendon grafts connected directly to the mechanical testing machine (without interference screw fixation into bone tunnel), the residual graft tension also decreased over time (Fig. 19).
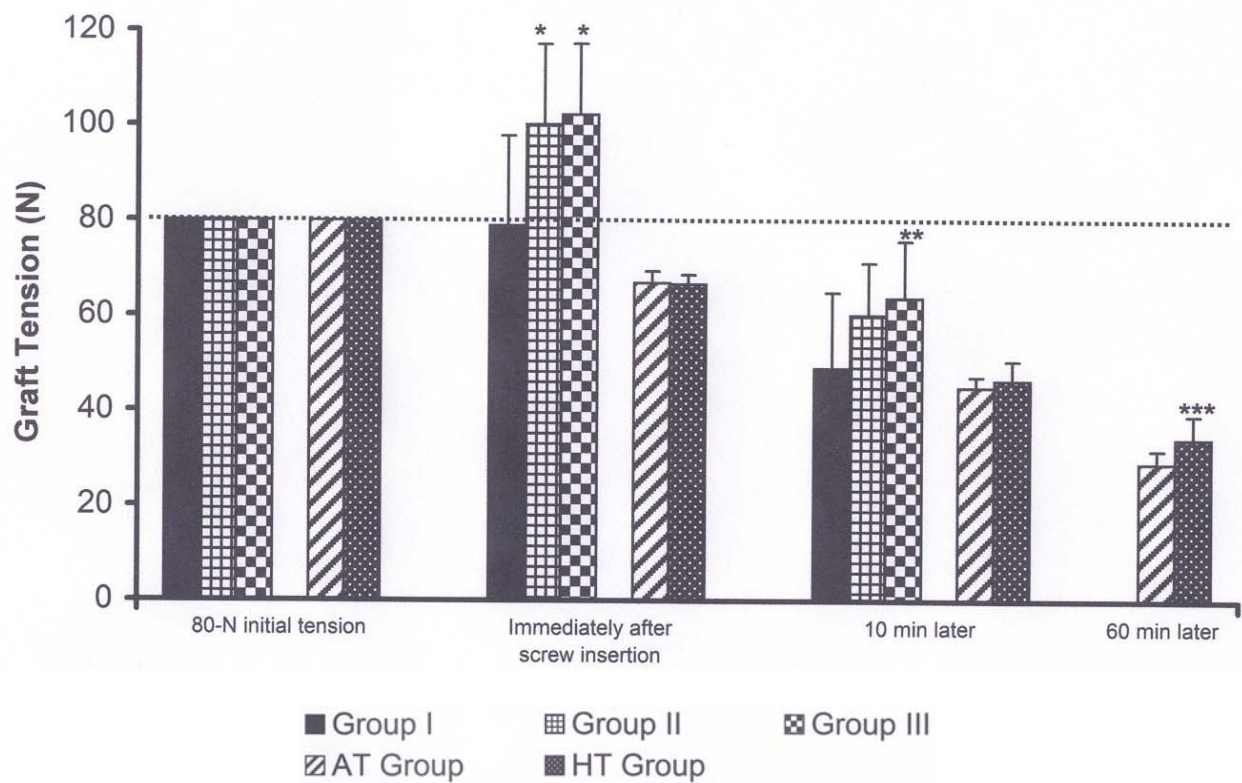
Figure 19. The average graft tensions (mean ± SD) in Groups I-III, AT, and HT after initial tensioning (80 N), immediately after interference screw insertion (1 minute after the initial tensioning in AT and HT), 10 minutes after interference screw insertion, and 60 minutes after initial tensioning (AT and HT only). *Significantly different from Group I ($P<0.01$). **Significantly different from Group I ($P<0.05$). ***Significantly different from Group AT ($P<0.02$).

8.5. BMD AND INSERTION TORQUE AS PREDICTORS OF FIXATION STRENGTH (STUDY VI)

Despite the relatively strong group-level correlations (R^2 from 0.33 to 0.54), the ability of the vBMD or IT to accurately predict the actual strength of fixation of an individual construct remained relatively poor. The errors ranged from –150% to 92%, –22% to 50%, –56% to 121%, and –23% to 50% when bone mineral density and insertion torque were used to predict cyclic-loading induced displacement and yield load, respectively (Fig. 20).
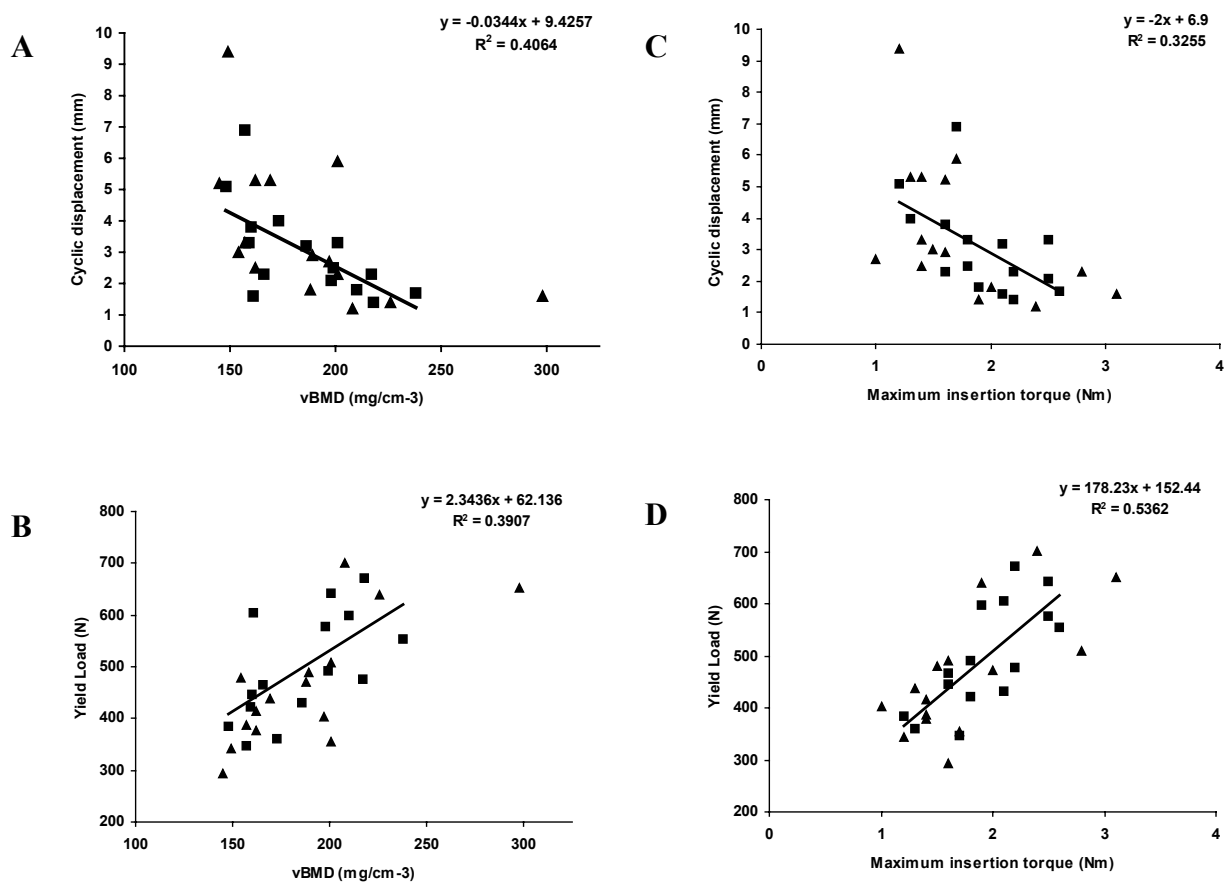
Figure 20. Relationship between the strength of the ACL graft fixation (i.e., the cyclic-loading induced displacement after 1500 loading cycles, and the yield load at single-cycle loading) and the volumetric BMD (Figs. 1A and 1B) and the insertion torque (1C and 1D). The squares denote the learning data from which the corresponding prediction equations were derived. The triangles denote the validation data. Note the considerable spread around the prediction lines.

9. DISCUSSION

Considering the high number of published ACL graft fixation studies and the fact that the optimal, golden standard fixation for soft tissue grafts has not been determined, literature provides surprisingly few studies reporting negative results. Also, as recently pointed out by Reider (2003), only few articles declare a conflict of interest in print despite the fact that federal funding for ACL research is minimal compared to that of industrial sponsors. Although industry-sponsored research is here to stay, critical independent evaluations are needed and also negative findings should be published to avoid the research becoming biased. Unfortunately, it is generally easier to publish (and get future funding for) studies reporting positive findings, a phenomenon called publication bias (Copas and Shi 2000, Montori et al. 2003).

Fixation strength of ACL grafts is believed to be the weakest link in ACL reconstruction. This belief has lead to a commercial surge for new improved fixation devices and techniques. This, in turn, has resulted ACL graft fixation to become one of the most popular research topics in sports traumatology. In addition to being industry sponsored, it appears that many of the ACL graft fixation studies could be considered commercially driven, instead of being independent and purely scientific. Accordingly, regrettably many of the previously published ACL graft fixation studies present weak and biased study designs and report results clearly aiming to serve the needs of instrument manufacturers. Common limitations include a small sample size, inadequate specimen age, poor or unknown bone quality, and the use of poorly validated or simplified models (e.g., porcine bones) assumed to adequately simulate the situation in a human knee joint. As a consequence, many confusing controversies can be found in the existing ACL graft fixation literature and several commercial products lacking real scientific justification have been used in the treatment of ACL patients. Good examples are the instruments sold for the bone tunnel compaction and graft preconditioning, and, for measuring the interference screw insertion torque. According to the results of the current study series, in which the above mentioned limitations and confounding effects were controlled by using adequate number (power analysis) of young, paired human cadaver specimens with confirmed bone quality (peripheral quantitative CT measurements), all of the above noted instruments were shown to fail under scientific scrutiny. Additionally, the porcine tibiae were shown to be poor substitutes for human tibiae despite being commonly used in the ACL graft fixation studies.

Bone tunnel compaction by either specific compaction drill bits or tapping with serial dilators of increasing diameter has been speculated to create dense-walled bone tunnels, to enhance

the "interference fit" between the bone tunnel walls and the fixation implant (interference screw), and ultimately, to provide a more rigid fixation of the ACL soft tissue graft. However, in this study, neither compaction drilling or serial dilation increased the initial fixation strength of soft tissue grafts fixed with interference screws in comparison to conventional extraction drilling. This lack of effect of tunnel compaction is in agreement with the results of Rittmeister et al. (2001), the only previously published study evaluating the effectiveness of bone-tunnel compaction on the fixation strength of interference screw fixed soft tissue graft in ACL reconstruction. Using human cadaver knees, hamstring tendon grafts, titanium interference screws and a biomechanical testing protocol very similar to that of ours, Rittmeister et al. showed that dilation of the tibial tunnel did not significantly improve the strength of fixation in comparison to extraction drilling. In contrast, Cain et al. (unpublished data, 1999) have previously observed in their preliminary report that tibial tunnel dilation had a clear positive effect on pullout strength of quadrupled hamstring tendon grafts secured in human cadaver knees with bioabsorbable interference screws. Rittmeister et al. suggested that the apparent controversy between their findings and those of Cain et al. could be attributable to the use of different biomechanical testing protocols, definition of the construct (fixation) failure, or difference in the interference screw (design, material) *per se*. However, our current study quite persuasively shows that bone-tunnel compaction does not provide any increase in the initial fixation strength.

We, among the others, have previously believed that porcine specimens well mimic human knee specimens in terms of their size, shape, and bone quality, and accordingly, strongly advocated their use as surrogates for human tissues in the evaluation of ACL graft fixation devices. However, as illustrated by our peripheral quantitative CT data, a clear difference exists between the species in the volumetric trabecular bone density of the human and porcine bone at the site corresponding to the tibial drill hole in the ACL reconstruction and in the cross-sectional structure of the entire proximal tibia of human and porcine bone. More importantly, the results of the current study indicate that porcine tibiae do not provide a valid surrogate for human cadaver tibiae in the biomechanical evaluation of interference screw fixation of soft tissue grafts in anterior cruciate ligament reconstruction. According to our results, the difference between these two tissue sources in relevance to ACL graft fixation studies can be summarized as follows: The trabecular bone mineral density of proximal porcine tibia is actually considerably higher than that of human cadaver tibiae (about 90%) and if porcine tibiae are used, the yield load values can be expected to be approximately two times higher than those of human cadaver specimens and the displacement values only one third of the values when using human cadaver tissues. These results are in perfect agreement with the recent findings by Magen et al. (1999) who obtained approximately two times

higher failure load values using animal tissues compared to their results with human cadaver specimens.

Although an obvious controversy seems to exist concerning the ideal pretensioning, preconditioning, and initial tensioning of the soft tissue grafts in ACL reconstruction, consensus prevails that there is a relatively wide safe window between too loose and too tight grafts (Amis and Jakob 1998, Katsuragi et al. 2000, van Kampen et al. 1998, Yasuda et al. 1997). Only few studies have previously studied the residual graft tension after an ACL graft fixation. According to our results, the graft tension always decreases significantly shortly after the ACL reconstruction procedure irrespective of whether preconditioned or not. It is most likely that the clinically applicable pretensioning, preconditioning or initial tensioning protocols cannot fully eliminate the intrinsic tendon creep or provide any clear advantage over simple tensioning of the graft (e.g., pulling by hand) in ACL reconstruction, and thus, the clinical reasonability of graft preconditioning *per se* is seriously questioned. If the initial graft tension and knee laxity conditions do not differ at the baseline despite the use of different initial tensions (van Kampen et al. 1998), there is no reason to believe that any clinically relevant differences could be seen over time either. The clinical results associated with the most commonly used tensioning protocol, the unmeasured initial tension or maximal sustained one-handed pull, have been generally good (Cunningham et al. 2002, Howell et al. 2001, Nabors et al. 1995, Yoshiya et al. 2002). It therefore appears that if only tensioned intraoperatively within the relatively wide safe window, the soft tissue grafts will adapt to the appropriate postoperative tension for ideal healing through the intrinsic creep of the tendons.

Bone mineral density and interference screw insertion torque have been proposed to be good predictors of interference screw fixation strength due to the correlation between the parameters (Brand et al. 2000b). However, the actual accuracy of prediction has not been investigated before. According to the results of the current study, although correlation between the parameters does exist (R^2 from 0.33 to 0.54), the ability of the bone mineral density or insertion torque to accurately predict the actual strength of fixation of an individual construct remained relatively poor. The strongest association was found between the insertion torque and yield load. However, 46% of the variation in the model was still not explained, and thus, attributable to something unknown, such as random error, measurement error, or other unknown sources of variation (Greenfield et al. 1998). For example, if the estimated fixation strength is 400 N (according to the insertion torque value measured during screw insertion), the true value can reside somewhere between 224 N and 576 N at 95% probability. From a clinical perspective, a much better accuracy of prediction at individual level would be needed.

Considering the cyclic-loading induced displacement noted during the cyclic-loading tests of this study, the concept of backing up the interference screw fixation (combined or hybrid fixation using another fixation method or implant) seems to be warranted. However, as mentioned before, we do not know if the real forces affecting the ACL graft are similar to those used in the cyclic-loading tests of this study. It has to be kept in mind that several authors have reported the clinical results of interference screw fixation of soft tissue grafts in ACL reconstruction to be at least as good as those obtained with interference screw fixed BPTB grafts (Ejerhed et al. 2003, Shaieb et al. 2002). Furthermore, Weiler et al. (2002b) recently reported in their comprehensive *in vivo* ovine study that not the strength of interference screw fixation of an ovine Achilles tendon split graft but the soft tissue graft itself was the weak link of the ACL reconstruction during the early healing stage. Accordingly, it has to be seriously questioned whether graft fixation strength really is the weakest link of reconstruction, as believed today, or even more importantly, if it indeed is the weakest link, is it actually a real problem or are the continuous attempts to increase fixation strength being done mainly for commercial purposes? According to the recent clinical studies, even professional athletes are usually able to continue their career after ACL reconstruction with interference screw fixed soft tissue grafts (Colombet et al. 2002). Although some reruptures do take place, they usually occur in situations where even the original ACL might have been torn, and thus, these cases do not really warrant the questioning of the success of the reconstruction.

Traditionally ACL graft fixation has been investigated *in vitro* by using worst-case-scenario models where the *in vivo* support from the other knee joint stabilizing structures (e.g., other ligaments, menisci and muscles) has been eliminated. The biomechanical testing has mainly consisted of load-to-failure tests. The obtained maximum failure load (determined in Newtons) has been considered to be the most important outcome parameter. Although failure load values provide results that can be easily illustrated and understood (10 N corresponding approximately to 1-kg weight), their true clinical relevance can be questioned as long as we do not know the magnitude and mode of the actual forces transmitted through the ACL graft during the postoperative rehabilitation and tendon-to-bone healing. The same problem exists even with the estimation of adequate load levels for cyclic-loading tests. On the other hand, considering the failure mode of interference screw fixation of the soft tissue graft, the slippage of the graft past the screw, the cyclic-loading induced displacement should provide a more relevant outcome parameter for biomechanical evaluation of ACL graft fixation than the single-cycle failure load. However, based on the current knowledge, it is impossible to say exactly what is the relationship between an increase in displacement of the graft and the resulting increase in anterior knee laxity, or what limit of increased anterior laxity would be considered harmful for a healing graft. In this context, one

should recall that only a proportion of the increase measured along the axis of the bone tunnel and graft (a worst-case-scenario *in vitro* model) projects into an increased anterior displacement of the tibia relative to the femur *in vivo*. Furthermore, the biomechanical ACL graft fixation studies have traditionally been performed in room temperature. However, as for example bioabsorbable devices could behave differently in body temperature, testing in 37°C water bath might provide an improvement to the current *in vitro* protocols.

In the future ACL graft fixation experiments, attempts should be directed to moving from *in vitro* to *in vivo* research (Beynnon et al. 1994 and 2001). Although well-designed *in vitro* models can be used successfully to compare the initial effects of different techniques or devices in well-controlled circumstances with minor bias, the most important outcome parameter, the clinical result after healing, can only be determined in *in vivo* studies. The *in vitro* models always include numerous "simplifications" to the *in vivo* situation, and thus, regardless of the biomechanical testing protocol used, caution is warranted in extending the results directly to clinical practice.

10. CONCLUSIONS

1.   In comparison to conventional extraction drilling, compaction of the bone-tunnel walls by compaction drilling or serial dilation does not increase the initial strength of an interference screw fixation of soft tissue grafts in anterior cruciate ligament reconstruction. Accordingly, there does not seem to be biomechanical basis to advocate the use of stepped routers and serial dilators in the anterior cruciate ligament reconstruction. (I, III, IV)

2.   In comparison to young human cadaver tibia, the results obtained using porcine tibia significantly underestimate graft slippage past the fixation and overestimate the fixation strength (failure load) of the soft tissue graft in the ACL reconstruction. Porcine tibia does not provide a reasonable alternative to young human cadaver tibia for evaluating interference screw (or other intratunnel) fixation in the ACL reconstruction, and therefore, human cadaver tibiae should be used in the future studies. (I,II)

3.   Clinically applicable preconditioning protocols do not fully eliminate the intrinsic tendon creep. Whether the soft tissue graft is preconditioned or not, the initially set tension decreases considerably postoperatively due to the remaining intrinsic tendon creep thus seriously questioning the reasonableness of preconditioning *per se* in anterior cruciate ligament reconstruction. (V)

4.   Bone mineral density and interference screw insertion torque cannot be used as clinically applicable predictors for the ACL graft fixation strength. (VI)

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Lempäälä, November 2003

Janne Nurmi

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