Experimental and numerical analysis of screw fixation in anterior cruciate ligament reconstruction

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Abstract—**This paper reports the results of an experimental and finite element analysis of tibial screw fixation in hamstring ACL reconstruction. The mechanical properties of the bone and tendon graft are obtained from experiments using porcine/calf bones and tendon. The results of the numerical study are compared with those from mechanical testing.**
Analysis shows that the model may be used to establish the optimum placement of the tunnel in anterior cruciate ligament reconstruction by predicting mechanical parameters such as stress, strain and displacement at regions in the tunnel wall.

Index Terms— **ACL reconstruction, screw fixation, finite element modelling, experimental data**

I. INTRODUCTION

A computer model has been developed to advance the state of the art of finite element analysis of knee joint in terms of geometry, material properties and loading. This model has direct application for tibial implant design but can also be used to better understand postoperative internal bone stresses in anterior cruciate ligament (ACL) reconstruction surgery.

The correct position of the bone tunnel and the choice of fixation type are the most important issues in ACL reconstruction [1]. Surgically created tunnels cause changes in stress pattern of the surrounding bone during loading. These stress changes can lead to post-operative tunnel enlargement possibly due to bone stress deprivation around the tunnels [2], resulting in graft failure and subsequent revision ACL surgery [3].

Furthermore, screw fixation can cause excessive compression from the threads, leading to accumulation of micro-damage [4] and eventually bone fracture at the fixation site.

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The purpose of this paper is to report preliminary results of a finite element model of bone tunnels representative of those used during ACL reconstruction using interference screw fixation.

The study examines the mechanical aspects of an interface screw fixation both experimentally and numerically, with its aim being to minimize deleterious effects in ACL reconstruction. The tibial cortical/cancellous bony tunnel and the stress pattern resulting from the screw fixation in the tunnel are investigated.

II. MATERIALS , TESTS AND MODELLING

A. Experimental Setup

Porcine and calf bones were used as a bone model of the knee joint in all experiments performed. Bovine flexor tendon, split into 2 strands with a total length of 100mm was doubled and used to represent a 4 strand hamstring graft. The tendons and tibiae were cleared of adherent muscle fibers and surrounding soft tissues. The ends of the graft were sutured in a standard whipstitch fashion to allow a constant tension to be maintained on all four strands during fixation. A 10mm tunnel was drilled in the tibia using a standard tibial guide (Conmed, Linvatec). The tunnel was placed in the centre of the ACL footprint at an angle of 45 degrees. With the aid of a passing suture, the looped end of the graft was pulled through the tibial tunnel, leaving the sutured four-tailed end of the graft protruding from the tunnel. A 9mm diameter x 25mm titanium screw (Arthrex) was inserted between 4 strands of the graft in an attempt to achieve concentric fixation. The screw was advanced until it was flush with the proximal bone-tunnel opening. Finally the looped end of the graft and the tibia were then secured on the testing machine with the used of a specially designed jig (Fig. 1).

The mechanical tests were performed using a screw-driven universal testing machine. The specimen's response to the loading was obtained in the form of a load-displacement curve as shown in Fig. 2, which shows a plot of pull-out force versus displacement using porcine tibia bone, calf tendon graft and a metallic screw (Arthrex). The results from the model (which is described in the next section) is also plotted for comparison.

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Fig. 1. Mechanical testing of porcine/calf tibial bone and tendon graft (tunnel is 45 degree from central axes of the bone). The pulling force and axis of the tunnel are in the same direction.

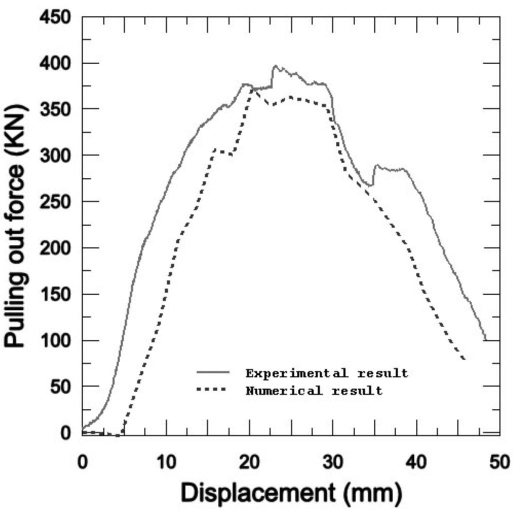

Fig. 2. Pulling force versus displacement of a porcine bone and calf tendon fixed at one end.

B. Finite Element Modelling

A finite element model was created for a section of tibial bone containing a tunnel and an advancing screw (Fig. 3). The materials for cortical, cancellous and subchondral bones were assumed linear elastic, which is adequate for most studies of bone stress and strain [5]. The modelled bone was assigned a stiffness value from the experimental data. The shear modulus was calculated from empirical relationships reported by Ashman et al. [6]. An isotropic Poisson's ratio was used [7]. Due to scarcity of experimental data, subchondral bone was assumed as isotropic [8,9] and homogeneous.

A commercial code, ABAQUS explicit was used to simulate the dynamic turning and advancing of the screw. The bone was modelled using 3-dimensional solid elements C3D4 element, a 4-node linear tetrahedron provided by the code. Reduced integration and hourglass control was applied. Mesh adequacy was validated using a convergence analyses. The screw was modelled using R3D4 elements which are 4-node 3-dimentional bilinear rigid quadrilaterals [10]. The length of the bony tunnel was set to 30 mm.

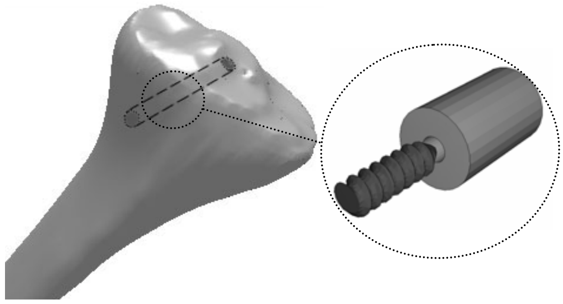

Fig. 3. Schematic of a tibia bone with a cylindrical tunnel. The model assumes a cylindrical section of bone around the tibia tunnel.

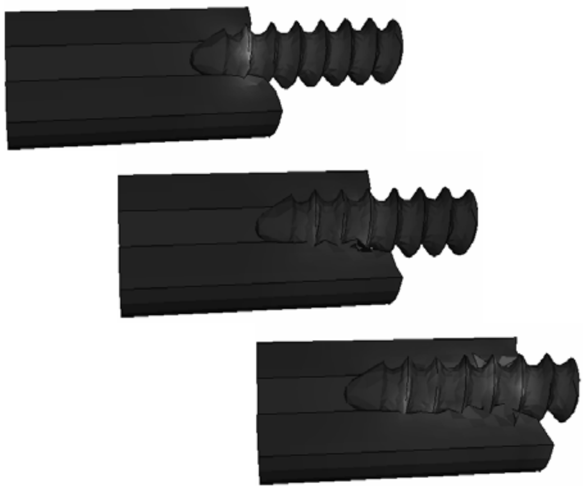

Fig. 4. Von Mises stresses in the tibia finite element model of the region around the screw (the screw was inserted into tunnel by combined axial/rotational movement).

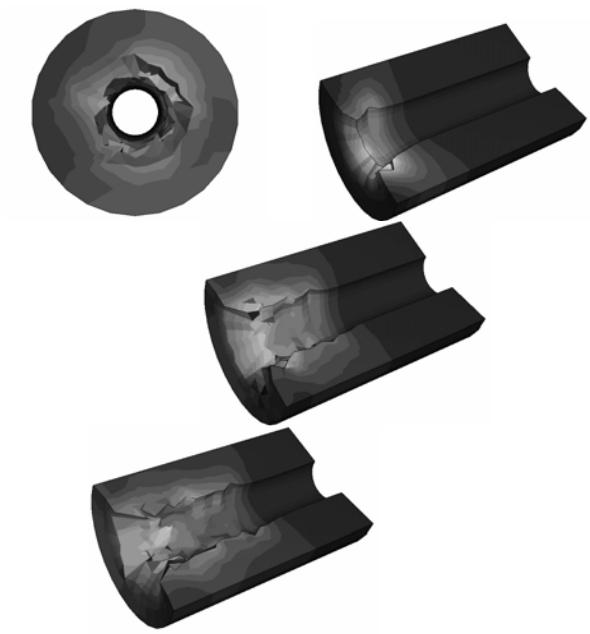

Fig. 5. Von Mises stress contour plots of transverse sections of tibia.

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A 9 mm diameter tunnel was assumed to be drill at locations reported by Fu et al. [11], i.e. at an angle of about 10^o to the midsagittal plane and 45^o to the midcoronal plane. The locations were within the boundaries of those used clinically.
The stresses in the bone and interface screw were examined at different stages of fixation (Fig. 4). The screw was loaded with a force of 200 N directed along the tunnel axis, which is an approximation of the graft tension at full extension of the knee during gait [12] and a rotational movement which was calculated based on the screw's pitch.
The cortical and cancellous bone properties of stiffness and shear modulus were obtained from the experimental data of current study and open literature [13, 14, 15].

III. RESULTS AND DISCUSSION

A. Simulation of Screw Advancement in the Tunnel

Fig. 4 shows the stress analysis of interference screw fixation in an ACL reconstruction (the colour scale indicates the range level of stress, green being the highest). The stress on the tunnel wall varies between 10-20 MPa. The maximum stress occurs in the interface between the sharp threads of the screw and the wall of the tunnel indicating the cutting of the threads in to the porous bone. The lowest stress is in the distal end of the tunnel. However the stress at the cortical bone interface is a controversial point, and we think it is a good point to achieve fixation.
The model enables assessment of bone tunnel deformation (Fig. 5), and the principal strain/stress patterns. The orders of magnitude in the interface of screw and tendon graft can be readily read out.

B. Cortical Bone Stress from Screw Fixation

It should be noted that cortical stress from screw fixation may be quite high even in normal gait. Stresses up to 100 MPa due to screw fixation loads were predicted at the tunnel aperture (Fig. 5). Note that with repeated stressing, micro-damages can develop and accumulate in the cortex [4] and potentially lead to its failure [1]. At 100 MPa, the fatigue of cortical bone is at approximately 10^6 cycles [16] while in reality, loading cycles of 10^7 are applied to bone tissue over a 10 year period [16]. This highlights the issue of stress caused micro-damages in local regions. However, one must note that the micro-damage in the bone is constantly being repaired by the body. Although it was assumed that a 200 N graft force occurs at full extension, this force could vary during gait and affect the stress results [1].

IV. CONCLUDING REMARKS

The agreement between the numerical model prediction and the experiment is good as shown in Fig. 2 considering the assumptions made on loading conditions, geometry and material properties and the exclusion of the mechanical influence of the muscles, ligaments, cartilage, and menisci of the knee. These assumptions clearly limit the possible clinical value of this model. Nevertheless, the model gives an insight into the type and magnitude of the forces acting on the bone in ACL reconstruction. At this stage of development, the model should be treated as being able to providing an accuracy of the order of magnitude rather than an exact analysis. Its usefulness remains in the prediction of patterns of the stress, strain and displacement in the regions around the bone tunnel.
In addition, this study also does not consider the effect of age in the model. This factor could be examined by including in the model the reported age-stiffness relations [17, 18] and stiffness-apparent density relations [6].
It should be noted that the function of the screw is to press the tendon against the tunnel wall in the tibia to allow biological fixation to occur between the tendon and the bone. At the same time, it provides a fixation to allow pretension of the tendon to be maintained to a certain degree for the functioning of the knee. Typical failure modes of the fixation procedure include loosening of the pre-tension in the tendon; breakage of the tendon due to cutting by the sharp screw edge, and failure at the screw/ligament. To overcome these problems, further research is needed into improved methods of fixation.

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