Journal of the Mechanical Behavior of Biomedical Materials 80 (2018) 27–32

Contents lists available at ScienceDirect

Journal of the Mechanical Behavior of Biomedical Materials

journal homepage: www.elsevier.com/locate/jmbbm

Positioning of the cross-stitch on the modified Kessler core tendon suture

L. Gil-Santos[a,b,c], M. Monleón-Pradas[b], F. Gomar-Sancho[d], J. Más-Estellés[b,⁎]

[a] Hospital Intermutual de Levante, CV-35, Km 11, Valencia, España
[b] Centro de Biomateriales e Ingeniería Tisular, Universitat Politécnica de València, Camino de vera s/n, 46022 Valencia, España
[c] Instituto Universitario de Investigación en Enfermedades Músculo-esqueléticas Universidad Católica de Valencia (San Vicente Mártir), C/Quevedo, 2, 46001 Valencia, España
[d] Departamento de Cirugía, Facultad de Medicina y Odontología. Universitat de València, Avda. Blasco Ibáñez, 15, 46010 Valencia, España

ARTICLE INFO

Keywords:
Biomechanics
Flexor tendon
Repair
Core and peripheral suture
Resistance

ABSTRACT

Cryopreserved human tendons were sutured with different variations of a modified Kessler-type grasping suture in a series of different designs in order to assess the influence of the distance between the cross-stitch on the core suture (5 and 10 mm from the cut tendon edge) on the peripheral suture. An original mathematical model was employed to explain the mechanical behavior (strength, deformation, and distribution of load) of the different suture designs. The effect of the peripheral epitendinous suture, combined with the distance of the core suture, was evaluated.

The variation of core suture distance had no relevant consequences on the overall resilience of the design. However, increasing the distance between the cross-stitches of the core suture reduces the deformation that is absorbed not only by the core suture itself but also by the peripheral suture.

Adding a peripheral epitendinous suture to a 10-mm design almost doubles the breaking load in absolute values. The mathematical model predicts that the peripheral suture will support a greater load when the distance of the core suture cross-stitches is increased. The evidence level is II.

1. Introduction

Outcomes after repairs of finger flexor tendons have certainly been improved by early motion programs (Ketchum et al., 1977; Silfverskiöld and Anderson, 1993; Wade et al., 1989). Obtaining optimal tendon excursion reduces the risk of adhesions or minimizes their effect and makes them compatible with normal finger functions (Trail et al., 1989, 1992). However, early motion requires a design capable of resisting the forces applied without altering normal tendon healing biology. This involves a combination of high initial resistance (Ketchum et al., 1977; Mason and Allen, 1941; Wade et al., 1989) (avoiding the occurrence of gapping and snagging) with the minimum of tissue strangulation, which impedes intrinsic vascularization and thus healing potential (Mason and Allen, 1941).

In the last years, several studies have been carried out about the materials and technique of tie (Gil et al., 2012; Ortillés et al., 2014, von Trotha et al., 2017). The Kessler type suture (Gil Santos, 1993; Kessler, 1973; Moriya et al., 2010) and its modifications have been the most frequently used in repairing flexor tendons in the hand. However, in our opinion, even though this type of suture is today widely employed, this design lacks systemization in several important aspects, such as in the distance of the cross-stitch from the cut tendon edge. The aim of this study was to apply an original mathematical model in order to assess the mechanical behavior of the Modified Kessler grasping tendon suture when the cross- stitch is placed at various distances from the edge and to quantify the influence of this distance on the peripheral epitendinous suture (also referred as “peripheral” or “epitendinous”).

2. Materials and methods

In the study 20 flexor tendons obtained from 10 human cadavers involved in violent deaths (10 in traffic accidents and one had been stabbed to death) with no history of organic pathology were used. The tendons were frozen by the Arnoczky method (Arnoczky et al., 1986) and after thawing in a saline ClNa 9‰ bath for 3 h at a constant temperature of 36°C in a thermostatically controlled Kowell® oven; tenorrhaphy was carried out on the different series for a total of 30 tests.

2.1. Tenorrhaphy. Experimental model

The tendon sutures were prepared by immersing specimens in a physiological solution under optical magnification by means of a Zeiss®

⁎ Correspondence to: Center for Biomaterials and Tissue Engineering, Camino de Vera s/n, 46022 Valencia, Spain.
E-mail address: jmas@fis.upv.es (J. Más-Estellés).

https://doi.org/10.1016/j.jmbbm.2018.01.018
Received 12 September 2017; Received in revised form 14 January 2018; Accepted 17 January 2018
1751-6161/

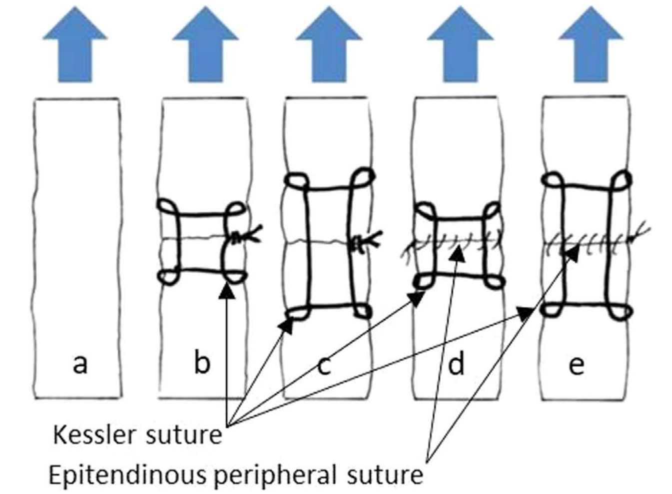

Fig. 1. The five series tested (left to right): a) intact tendons; b) Kessler at 5 mm; c) Kessler at 10 mm; d) Kessler at 5 mm plus epitendinous suture; e) Kessler at 10 mm plus epitendinous.

OPMI-1 surgical microscope and Keeler® type 2.5 × 300 magnifiers, using a millimeter grid on the bottom of the recipient as a guide to place the sutures in the required position.

All the core sutures had six simple knots, tied alternately towards right and left, with propylene monofilament non resorbable thread 4–0 sutures. The peripheral suture was also made with propylene monofilament non resorbable 6–0 single-stranded running epitendinous. Five different series of samples were tested (Fig. 1): a) intact tendons; b) Kessler at 5 mm from the cut edge; c) Kessler at 10 mm; d) Kessler at 5 mm plus epitendinous suture at 2 mm; e) Kessler at 10 mm plus epitendinous at 2 mm.

2.2. Tendon preconditioning and biomechanical study

Samples were tested on an Adamel Lhomargy DY-34® device (Adamel Lhomargy S.A., France) (Fig. 2). A stress-strain test was carried out by applying a constant strain rate until the suture material reached breaking point, while measuring the load throughout the test.

Preconditioning is considered necessary in in vitro experiments before accepting the registered values (Hooley, 1977; Monleón Pradas and Díaz Calleja, 1990). It simply consists of a preliminary loading and unloading of the specimen, in such way that after this loading and unloading, the results can be considered as repetitive (Fig. 3).

The experimental loading program itself consisted of two consecutive stretching cycles divided into three stages: *1st stage*, stretching at 5 mm/min to an absolute value of 5 mm; *2nd stage*, the tendon was allowed to return to its original position (null force) at a rate of 5 mm/min; 3rd stage, stretching at 5 mm/min until breakage of sutured specimens. This program was used in all the tests, the unsutured tendons being stretched between clamps. The device's associated software converts the experimentally measured load and displacement magnitudes into the values that appear on the graphs.

2.3. Mathematical model

A mechanical model is proposed in order to understand the experimental results. This model divides the sutured tendons into three mechanically different parts, consisting of: the tendon, the Kessler core suture, and the epitendinous peripheral suture. Each of these parts was characterized by essays on different samples, as indicated above.

In the stretching devicem a force F and displacement Δl are measured (Fig. 4a). If l_o is the initial length of the sample, $\varepsilon = \frac{\Delta l}{l_0}$ denotes the strain of the specimen. E is the ratio between load and strain, which we call "stretching modulus": $E = \frac{F}{\varepsilon}$. As the stretching modulus can only be

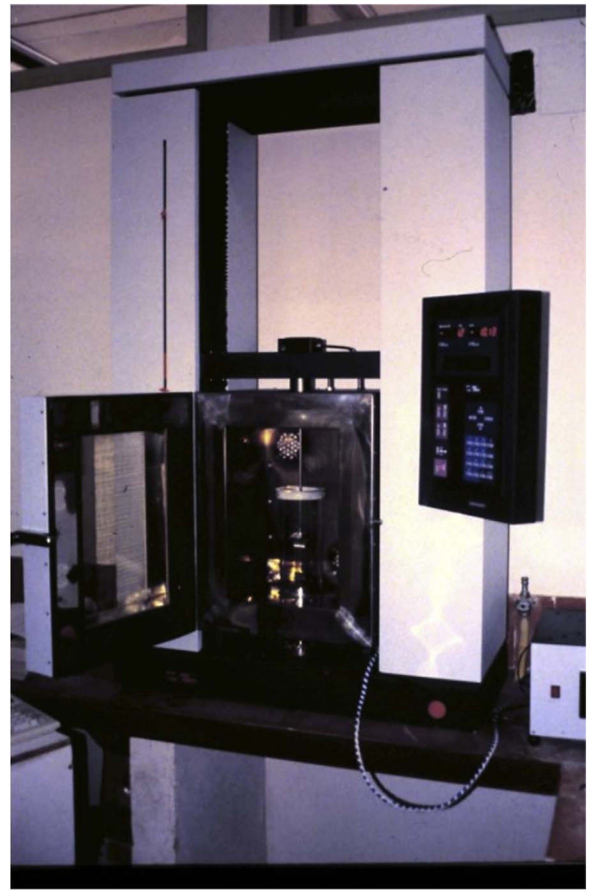

Fig. 2. DY.34 (Adamel Lhomargy S.A.) mechanical testing machine.

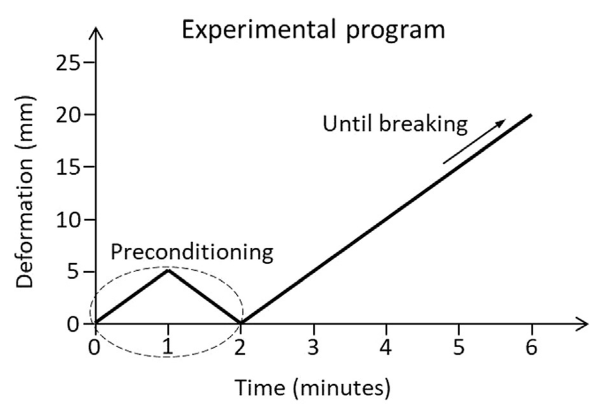

Fig. 3. Experimental program.

assumed to be approximately constant in the linear zone (in the non-linear zone it varies), only in this behavioral zone can a numerical comparison be carried out between intact and sutured tendons.

This definition of stretching modulus is not frequently used in material science. Instead stretching modulus is usually stated ~~as~~ the ratio between stress, $\sigma = \frac{F}{A}$ (A being the cross-section of the specimen), and strain, $\frac{\sigma}{\varepsilon}$. This magnitude is not significant for our purposes, since our specimens ~~has~~ have a badly-defined cross-sectional area, due both to the lack of uniformity of the tendinous bundles and, more importantly, to the fact that the effectively loaded section of the sutured tendons was not that of the tendon itself but that of the suture strands. Therefore, these issues justifie our definition of E.