Twisting of patellar tendon grafts does not reduce their mechanical properties

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Abstract

The purpose of this study was to investigate whether the twisting of a patellar tendon (PT) graft improves or reduces its mechanical properties. Twenty-seven pairs of 10 mm cadaveric PT grafts were tested at a strain rate of 10% min⁻¹. For each pair, the left specimen served as an unmanipulated control while the right specimen was either left untwisted, twisted +90°, or twisted −90°. All avulsion failures were excluded from the data analysis, focusing only on pure in-substance tendon ruptures. Higher ultimate load values than previously reported were obtained for both the twisted and untwisted specimens, without a statistical significant difference between the two. The values obtained for the left and right specimens from Group A were 4014 ± 319 and 3973 ± 245 N, from Group B 3613 ± 207 and 3891 ± 14 N, and from Group C 3997 ± 278 and 4415 ± 507 N, respectively. Stiffness and failure strain were not influenced by the twisting. Neither the presence of a twist, nor the direction of the twist were found to increase the ultimate load of the 10 mm cadaveric patellar tendon graft. Therefore, recommendation for twisting cannot be assessed to its mechanical properties.

Keywords: Patellar tendon; Pretwist; Mechanics; ACL reconstruction; Human

1. Introduction

The hamstring tendon graft and the patellar tendon bone graft are the two most common autografts used for the reconstruction of the ACL. Noyes et al. (1984) analyzed the mechanical properties of these grafts and noted that the 15 mm patellar tendon graft showed the greatest load to failure. Subsequent testing revealed that 10 mm wide patellar tendon grafts are able to withstand sufficient loads and with their decreased size offer a reduced risk of graft impingement and donor site complications (Cooper et al., 1993).

The fibers of the ACL have a natural rotation as they are traced from their proximal origin to their distal insertion. Attempts to reproduce this twist have been suggested in the literature. Pretwisting the graft is said to simulate the helicoid course of the ACL (Lambert and Cunningham, 1988), to facilitate fixation (Lambert, 1983; Wirth and Kohn, 1990), decrease the possibility of abrasive wear on the tunnel edge (Jackson and Lemos, 1993), and to improve the primary strength of the graft (Cooper et al., 1993). Cooper et al. (1993) showed that randomly twisting the graft 90° increased both the ultimate load and maximal stress in both 10 and 15 mm patellar tendon bone grafts. In a canine model, Munns et al. (1994) discovered that the direction of the twist played a role as to whether the mechanical properties of the patellar tendon grafts were improved or reduced. However, the patellar tendon graft is physiologically twisted during flexion of the knee. From that point of view, the pretwisting of the patellar tendon bone graft is not necessary to guarantee the possible benefit of a pretwist. But, in humans, it is still unclear if the twisting of a patellar tendon graft could reduce its mechanical properties, and whether the direction of the twist influences, for example, ultimate load.
The purpose of this study was to investigate whether the presence of a twist or the direction of the twist influences the mechanical characteristics in paired 10 mm cadaveric patellar tendon grafts.

2. Materials and methods

In accordance with Vienna University regulations and bylaws regarding cadaveric materials, intact patellar tendons with the whole patella proximally and the tibial tubercle distally were harvested from both knees in each cadaver within 12 h of death and immediately frozen at $-20^\circ$C. Fifty-four specimens were obtained from 27 cadavers (20 male and 7 female). The average age of the donors was 27 (Std.: ± 8) yr.

After thawing overnight at room temperature, the central portion of the patellar tendon was prepared, under firm manual tension, by one surgeon to a width of 10 mm using a custom-made two-bladed scalpel. Tendon lengths were measured to the nearest millimeter with a hand-held caliper. The patella and the tibial tubercle were then potted in a mold using an epoxy automotive filler (Aralite D Ciba-Geigy, Vienna, Austria).

In each of the twenty-seven pairs, the left graft served as a control and therefore did not undergo any manipulation. Three groups consisting of nine paired specimens (A, B, and C) each were formed. The right specimens in group A were also left untwisted to rule out any left-right differences. The nine right specimens in Group B were twisted 90° clockwise. The nine right specimens in Group C were twisted 90° counterclockwise. The direction of the twist was defined as relative to the longitudinal axis of the patellar tendon. An internal rotation was therefore defined as medial and an external rotation as lateral, from the median line of the patellar tendon.

Mechanical testing was performed using a Wolpert tensile testing machine (Otto Wolpert Werke G.m.b.H., Ludwigshafen am Rhein, Germany). The Aralite D blocks were inserted into stainless steel clamps shaped like the molds. These clamps were then attached to the testing apparatus using 1-inch threaded steel rods as couplers, so that the long axis of the patellar tendon was vertically orientated and placed in line with the applied force. After rotating the proximal clamp (tibial end), +90° (clockwise) in Group B and −90° (counterclockwise) in Group C, the setup condition was fixed. The reference length used was the distance between the grips when the load increased over the resolution of the load cell signal (Vidik, 1979). After that the right specimens from Groups B and C were twisted and the grip to grip distance was measured again. The tests were performed under strain control at a rate of 10% per minute until failure of the tendon was reached. Loads were measured using a 10 kN load cell and recorded with the X-T recorder of the testing apparatus. During testing the specimens were submerged in a phosphate-buffered saline bath at room temperature.

Failure of the tendon was defined as the point at which the bone–tendon–bone complex could no longer bear the applied load. Only specimens with an in-substance failure, defined as a rupture within the ligament substance without evidence of failure at the insertion sites, were included in the analysis. The specimens with an avulsion failure, in which the specimen failed at either of the insertion sites, were excluded from the analysis.

Ultimate load (N) was interpreted as the load at which the patellar tendon failed. The failure strain (%) was defined as the ratio of the failure displacement to the initial length of the patellar tendon. The failure displacement was derived from the grip-to-grip distance. Stiffness ($N \, mm^{-1}$) was defined as the tangent of the load–displacement curve at the point of maximum steepness. The energy to failure (kN mm) was defined as the area under the load displacement curve until failure occurred. The collagen content was calculated following the tensile testing from the dry-weight of the ligamentous portion of the specimen (Ellis, 1969; Hooley et al., 1980), in order to make a comparison between the specimens possible.

Statistical analysis was performed with a blocked analysis of variance with blocking on cadavers using the statistical software package SAS (1990). Hypotheses are tested with contrasts, including a paired analysis between the left and right knees and tests of differences between plus 90°, minus 90°-twisted, and untwisted patellar tendon grafts. Adjustments for multiple comparisons are made by the method of Tukey. In this blocked analysis of variance, the control values of the left knee are also considered for comparisons between plus and minus 90° twisted and untwisted groups, so that higher/lower measurements on the left knee can correct for higher/lower values on the right knee. The data are described by mean (± SD). $P$-values are two-sided and a $p$-value of less than 0.05 was considered statistically significant.

3. Results

The mean grip to grip distance of all patellar tendons left untwisted, was 30.7 mm (± 0.2 mm). Twisting the specimens did not significantly influence the grip to grip distance. After twisting, the mean grip to grip distances were 30.7 mm (± 0.2 mm) and 30.8 mm (± 0.2 mm) for Group B and Group C, respectively.

Seven avulsion failures occurred in the left control specimens and four avulsion failures occurred in the right patellar tendons grafts (Table 1). The eleven avulsion failures occurred in nine patellar tendon pairs, three in each group. The results obtained from these nine pairs were not included in the statistical analysis. Thus,
Table 1
Distribution of avulsion failures across test groups

<table>
<thead>
<tr>
<th>Side</th>
<th>Groups</th>
<th>A</th>
<th>B</th>
<th>C</th>
<th>Total</th>
</tr>
</thead>
<tbody>
<tr>
<td>Left</td>
<td></td>
<td>3</td>
<td>3</td>
<td>1</td>
<td>7</td>
</tr>
<tr>
<td>Right</td>
<td></td>
<td>2</td>
<td>0</td>
<td>2</td>
<td>4</td>
</tr>
</tbody>
</table>

Table 2
The mean collagen content of each test group

<table>
<thead>
<tr>
<th>Side</th>
<th>Groups</th>
<th>A</th>
<th>B</th>
<th>C</th>
</tr>
</thead>
<tbody>
<tr>
<td>Left</td>
<td></td>
<td>0.29 ± 0.03</td>
<td>0.29 ± 0.03</td>
<td>0.32 ± 0.02</td>
</tr>
<tr>
<td>Right</td>
<td></td>
<td>0.34 ± 0.04</td>
<td>0.35 ± 0.04</td>
<td>0.36 ± 0.05</td>
</tr>
</tbody>
</table>

Note: No significant differences found.

Table 3
Mechanical properties of the analysed thirty-six patellar tendon grafts

<table>
<thead>
<tr>
<th>Group A</th>
<th></th>
<th>Group B</th>
<th></th>
<th>Group C</th>
<th></th>
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<tbody>
<tr>
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<tr>
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<tr>
<td>Ultimate load (N)</td>
<td>4013.9 ± 318.8</td>
<td>3973.1 ± 245.3</td>
<td>3613.4 ± 207.1</td>
<td>3891.3 ± 147.2</td>
<td>3997.6 ± 278.0</td>
</tr>
<tr>
<td>Failure strain (%)</td>
<td>18.5 ± 1.3</td>
<td>18.0 ± 1.3</td>
<td>18.7 ± 1.9</td>
<td>20.0 ± 1.1</td>
<td>19.3 ± 1.1</td>
</tr>
<tr>
<td>Stiffness (N mm⁻¹)</td>
<td>526 ± 98</td>
<td>519 ± 85</td>
<td>597 ± 87</td>
<td>562 ± 85</td>
<td>564 ± 97</td>
</tr>
<tr>
<td>Energy to failure (kNmm)</td>
<td>22.1 ± 3.5</td>
<td>20.0 ± 1.8</td>
<td>21.5 ± 3.3</td>
<td>18.8 ± 2.3</td>
<td>21.8 ± 2.2</td>
</tr>
</tbody>
</table>

Note: No significant differences found.

4. Discussion

This study was undertaken to clarify both the effect of the presence of a 90° twist and the effect of the twist direction on the mechanical properties of a 10 mm cadaveric patellar tendon bone graft. While numerous authors suggest that the twisting of the patellar tendon bone graft is beneficial for technical reasons (Lambert, 1983; Lambert and Cunningham, 1988; Wirth and Kohn, 1990; Jackson and Lemos, 1993; Kurzweil and Jackson, 1994), the effect of the twist on the graft’s mechanical and structural properties is still not clearly understood.

Cooper et al. (1993) addressed the possible mechanical changes resulting from the addition of a randomly directed twist to a 10 mm cadaveric patellar tendon bone graft. They found a 30% increase both in ultimate load and stress when a 90° twist was added. Munns et al. (1994) utilizing a canine model found that an externally rotated twist produced significantly higher values for failure stress, elastic modulus, and strain energy density. No directional differences were found for failure strain.
The results of this study differed from previous work in that the addition of a 90° twist in either direction had no effect on either the mechanical properties of 10 mm cadaveric patellar tendon grafts. Cooper et al. (1993) tested 10 pairs of twisted 10 mm cadaveric patellar tendon grafts obtained from a tissue bank. We harvested and prepared twenty-seven cadaveric pairs of patellar tendon grafts. Cooper randomly twisted the graft. Our twists were not random. Within each of the twenty-seven pairs the left graft served as an unmanipulated control, while the right graft either remained untwisted or was twisted plus or minus 90°. Unlike Cooper, who included minor bony avulsions in their data analysis, we excluded all avulsions and only used in-substance ruptures to determine the mechanical properties of the tendon.

In this study, the values obtained for the ultimate load were higher than previously reported. The mean ultimate load was 3899 N (± 298 N) for all the untwisted patellar tendon grafts (Group A right and all left patellar tendons, n = 24). Cooper et al. (1993) explained their increased values relative to Noyes’s values as a product of improvements in clamp design. The clamp design in our study did not differ from that of Cooper’s, so this cannot account for our higher values. As previously stated, we excluded all avulsion failures (n = 11) from the main statistical analysis. The mean ultimate load obtained from these eleven specimens is similar to the values Cooper et al. obtained for their specimens and may account for the difference in ultimate load values (Table 4).

Our strain rate was lower than the strain rate used in previous studies. One could question whether our lower strain rate biased the results toward the higher ultimate loads. Danto and Woo (1993) found in a rabbit model, that an increased strain rate caused higher tensile properties, but did not affect the failure mode. The literature describes this same effect across different materials tested. Thus we would have expected lower ultimate loads.

As tendons are multiphasic materials containing a high amount of water, the collagen content was measured, which represents the load bearing component. According to Ellis (1969) the estimation of the structural properties can be derived from the collagen content per unit length. The water content does not contribute to, but influences the load bearing properties (Haut, 1997). The comparison of the structural properties between different investigators is sometimes difficult considering the method for the determination of the cross-sectional area such influence their results. In the present study only the mechanical properties were evaluated. The collagen content per mm was used in order to be able to compare the structural composition between the specimens. However, in the comparison of the collagen content for the twisted and untwisted specimens, no significant difference was displayed.

Arthroscopic ACL reconstruction requires the fixation of the femoral bone plug first, so that the tibial bone plug can be twisted as desired. For this reason the tibial insertion site was twisted. Our results suggest that the mechanical properties of the graft are not altered by this twist. This study and every mechanical study involving cadaveric tissues takes place in an artificial environment without the simulation of biological stresses. Therefore, the direct transfer of conclusions drawn at the benchtop to the clinical setting may be inappropriate. Nevertheless, such in vitro results should prompt us to re-evaluate our clinical practice.

The results of this study suggest that neither the presence of a twist nor the direction of the twist influence the mechanical properties of paired 10 mm patellar tendon grafts. The higher ultimate loads obtained in this study may reflect the inclusion of only pure in-substance ruptures under optimal fixation. The clinical significance of these higher ultimate load values has yet to be determined.

Acknowledgements

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References


