THE MECHANICAL PROPERTIES OF THE TWO BUNDLES OF THE HUMAN POSTERIOR CRUCIATE LIGAMENT

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Abstract—Successful reconstruction of ligaments requires knowledge of the properties of the intact ligament. This study examined the strength of the human posterior cruciate ligament (PCL), treating it as two separate fibre bundles. It was hypothesized (i) that the mechanical and material properties of the anterolateral (aPC) and the posteromedial (pPC) bundles of the PCL were significantly different and (ii) that previous studies have underestimated the strength of the whole PCL. The properties of the two bundles were measured in 10 donors (53–98 yr). The mechanical and material properties of the two bundles were found to be significantly different, the aPC was six times as strong as the pPC. The aPC had a mean strength of 1.6 kN. Allowing for age effects this study suggests that the strength of the PCL in young active people is 4 kN, which is higher than that suggested by previous studies. Because of the difference in the strengths of the two bundles, we conclude that the aPC is primarily responsible for the stabilising effect of the PCL. We therefore recommend that PCL reconstructions should be centered on the middle of the aPC bundle.

INTRODUCTION

This study is a preliminary to the design of a synthetic implant to replace the ruptured posterior cruciate ligament (PCL). The authors wanted to establish the function, structure and mechanical properties of the intact ligament as a basis for the design of the implant. A search of the literature revealed that there has been little data published on the mechanical and material properties of the PCL and that the results show much variation (see Table 1).

For the entire PCL mechanical properties have been published by Kennedy et al. (1976), Marinozii et al. (1983), Prietto et al. (1988) and Trent et al. (1976). Also, Butler et al. (1986) have published results for the material properties of the PCL based on individual fascicles of three donors. Individual fascicles were tested because it was thought that this would result in more consistent results for ligament material properties; initial length is easier to measure for a small specimen and even loading of the fibres is easier to achieve. It was anticipated that smaller specimens would yield superior material properties. Trent et al. (1976) tested bone block–ligament–bone block preparations and reported sequential failure with 'steps' in the load/extension curve. Also the low strain rate led to premature failure by bony avulsion: the effect of strain rate on failure mode has been demonstrated by Noyes et al. (1974). Kennedy et al. (1976) tested ligaments that were excised from their bony attachments and held in mechanical clamps, which can be expected to cause premature failure. Marinozii et al. (1983) tested bone block–ligament–bone block specimens. They did not report avulsion failure but may have subjected ligaments to uneven loading leading to sequential failure. Prietto et al. (1988) tested tibia–PCL–femur preparations and reported a higher maximum load than Trent et al. (1976) and Kennedy et al. (1976), but this could have been due to the lower age of donor (Hollis et al., 1988) rather than better loading conditions.

The studies of Trent et al. (1976), Kennedy et al. (1976), Marinozii et al. (1983) and Prietto et al. (1988) all suggested material properties for the PCL that appear to be significantly inferior to those published by Butler et al. (1986). We offer the following explanation for this difference. The PCL is generally considered to be made up of two bundles (Girgis et al., 1975; Hughston et al., 1980), the anterolateral (aPC) and the posteromedial (pPC) (see Fig. 1). Since the aPC and the pPC are load bearing at different angles of flexion and have different orientations, it is impossible to load the fibres of the whole PCL in parallel. Thus, any flexion angle chosen for testing of the whole ligament must compromise one or both of the bundles—leading to peeling failure; and result in uneven fibre loading—leading to sequential failure. These two effects result in underestimation of total PCL strength, maximum stress and modulus. It was anticipated that higher forces could be obtained if these effects were eliminated.

The mechanical and material properties of the two anatomical bundles of the PCL have not been measured previously. Since these two bundles appear to have different functions (indicated by reciprocal tightening and slackening) it was hypothesised that they may have different mechanical and material properties. Therefore, the purpose of this study was to measure and compare the mechanical and material properties of the two separate PCL bundles and to clarify the mechanical properties of the whole ligament.
Table 1. Data for the mechanical and material properties of the posterior cruciate ligament (PCL) previously published and data from the present study for the anterolateral (aPC) bundle of the PCL.

<table>
<thead>
<tr>
<th>Donors</th>
<th>Age (yr)</th>
<th>Length (mm)</th>
<th>Area (mm²)</th>
<th>Extension rate (mm min⁻¹)</th>
<th>Max. load (N)</th>
<th>Stiffness (N mm⁻¹)</th>
<th>Linear strain (%)</th>
<th>Linear stress (MPa)</th>
<th>Modulus (MPa)</th>
<th>Max. strain (%)</th>
<th>Max. stress (MPa)</th>
<th>Energy density at failure (J cm⁻³)</th>
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<tbody>
<tr>
<td>Trent</td>
<td>29-55</td>
<td>62 (20-75)</td>
<td>—</td>
<td>50 mm min⁻¹</td>
<td>739 ± 368</td>
<td>180 ± 58</td>
<td>18.8 ± 3.6</td>
<td>109 ± 50</td>
<td>—</td>
<td>—</td>
<td>—</td>
<td>—</td>
</tr>
<tr>
<td>Kennedy</td>
<td>62 (20-75)</td>
<td>1051 ± 237</td>
<td>—</td>
<td>500 mm min⁻¹</td>
<td>1051 ± 237</td>
<td>145 ± 66</td>
<td>17 ± 4</td>
<td>345 ± 107</td>
<td>—</td>
<td>—</td>
<td>—</td>
<td>—</td>
</tr>
<tr>
<td>Marinozzi</td>
<td>55-90</td>
<td>300 mm min⁻¹</td>
<td>—</td>
<td>100 mm min⁻¹</td>
<td>855 ± 225</td>
<td>204 ± 49</td>
<td>22.1 ± 4.3</td>
<td>345 ± 107</td>
<td>—</td>
<td>—</td>
<td>—</td>
<td>—</td>
</tr>
<tr>
<td>Prietto</td>
<td>22.5 ± 3</td>
<td>100% s⁻¹</td>
<td>—</td>
<td>100% s⁻¹</td>
<td>1627 ± 491</td>
<td>15.0 ± 3.8</td>
<td>26.4 ± 9.1</td>
<td>364 ± 12</td>
<td>—</td>
<td>—</td>
<td>—</td>
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</tr>
<tr>
<td>Butler</td>
<td>21,30,30</td>
<td>1000 mm min⁻¹</td>
<td>—</td>
<td>1000 mm min⁻¹</td>
<td>1620 ± 500*</td>
<td>1000 ± 500</td>
<td>28.7 ± 17.0</td>
<td>248 ± 119</td>
<td>—</td>
<td>—</td>
<td>—</td>
<td>—</td>
</tr>
<tr>
<td>This study</td>
<td>75 ± 14</td>
<td>35.3 ± 3.4</td>
<td>—</td>
<td>—</td>
<td>35.9 ± 15.2</td>
<td>35.9 ± 15.2</td>
<td>31.2 ± 1.4</td>
<td>—</td>
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<th>Mean ± S.D.</th>
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<td>*N = 10.</td>
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METHOD

Materials

Knee specimens were obtained at post-mortem from 10 donors within two days of death. Storage of knees at room temperature for up to four days has been shown not to significantly affect intraarticular ligament properties (Vidic et al., 1965). The knee joints were extracted complete with surrounding soft tissue and about 100 mm of the extracapsular femur and tibia. Specimens were stored until needed in sealed polyethylene bags, to prevent dehydration, at −20°C. This, too, has been shown not to affect ligament properties significantly (Tkaczuk, 1968). When required, the specimens were thawed in warm water. Knees that were found to have a damaged PCL or severe osteoporosis (leading to bone fracture rather than ligament failure) were not included in the experiment. The mean age of specimen donor was 75 yr (53-98).

Preparation

Each knee specimen was prepared as follows: the extracapsular tibia was stripped of all soft tissue and scored with a rasp. It was then held vertically in a cylindrical stainless-steel mounting pot by three grub screws. A rigid mounting was achieved by filling the pot with polymethylmethacrylate (pmma) ('Simplex rapid', Associated Dental Products, Kemdent Works, Purton, Wilts, England). The powder and liquid were stored at 4°C to increase the pot life of the mixture. The pmma was mixed by hand and after a curing time of 15 min. While the pmma was curing the knee specimen was covered by a wet paper towel to prevent dehydration.

The specimen was then clamped in a vice via the pot. Soft tissue was excised to expose the PCL, taking care not to damage the ligament itself. The lateral femoral condyle was then sawn off, taking care not to disrupt the ligament origin, to give a better view of the whole PCL. The synovial covering of the ligament was then carefully removed, by sharp dissection, along with any meniscofemoral ligaments to reveal the fascicular structure of the PCL. The anterolateral and posteromedial bundles were then identified and separated using blunt dissection. The bundles were always joined by synovial tissue but could be easily identified because of the differential tensions in flexion/extension: in extension the pPC was tight and the aPC was slack, and vice versa in flexion. This was true for all specimens (see Fig. 2). To separate the bundles the knee was held in extension while a MacDonald dissector was placed just anterior to the tight pPC. To check this location the knee was flexed, when the dissector was found to be just posterior to the tight aPC. The dissector was then pushed through the PCL, while repeating the above procedure, creating a gap between the bundles. The femoral origins of the two bundles were then separated using a chisel placed between the bundles, parallel to their fibres. Two femoral bone blocks, incorporating the origin of the bundles, were then cut with a hacksaw (see Fig. 3). The specimen was then held, inverted, in a retort stand and the bone blocks suspended, in turn, in small mounting pots. The bone blocks were carefully arranged, so that the bundle fibres were parallel to the cylindrical axis of the pots and so that there were no kinks in the fibres as they approached the bone. The bone blocks were then secured with pmma, which also embedded a crossbar that prevented slipping under axial load. The pmma was poured into the pots so that it covered the bone. After a few minutes, when it had partially set but was still soft, the pmma was pushed away from the ligament so that it did not restrict the fibres or present any sharp edges.
Fig. 1. (a) Posterior and lateral aspect of the extended knee with the lateral femoral condyle and lateral meniscus removed to reveal the PCL. The two bundles of the PCL have been separated by blunt dissection. (b) Posterior and lateral aspect of the extended knee with the lateral femoral condyle and lateral meniscus removed to reveal the PCL. The anterolateral and posteromedial bundles are labelled aPC and pPC, respectively.
Fig. 2(a, b).
Fig. 2. (a) Lateral aspect of the extended knee with the lateral femoral condyle and lateral meniscus removed. Note that, in extension, the anterolateral bundle is slack. (b) Lateral aspect of the flexed knee with the lateral femoral condyle and lateral meniscus removed. Note that, in flexion, the anterolateral bundle is tight. (c) Posterior and lateral aspect of the extended knee with the posterior capsule removed. Note that, in extension, the posteromedial bundle is tight. (d) Posterior and lateral aspect of the flexed knee with the posterior capsule removed. Note that, in flexion, the posteromedial bundle is slack.
Fig. 3. Posterior aspect of the tibia showing the anterolateral (top) and posteromedial (bottom) bundles of the PCL after cutting femoral bone blocks.

Fig. 4. Tibia inverted with the anterolateral bundle hanging below. The silicone rubber mould (bottom right) has just been removed.
Fig. 5. Typical section of a replica of the anterolateral bundle, with millimetre scale. Note the concavities.
**Measurement of dimensions**

The dimensions of bundles from seven knees were measured. The average fibre length of each ligament bundle was determined by taking the mean of the longest and shortest fibres. The average spread in fibre lengths was $\pm 1.5$ mm, which equates to $\pm 4.5\%$ of resting length. Measurements were made with callipers and a steel rule to the nearest millimetre, which equates to $\pm 1.5\%$. This was done after bone block separation and before potting. The bone blocks were held by hand so that the ligament fibres were just straight.

The cross-sectional areas of the bundles were measured as follows. Moulds of the bundles were taken after the bone blocks had been set in mounting pots. A strip of thin cardboard was taped around the mounting pot to form a tube (38 mm in diameter) enclosing the ligament bundle. The bone block, in its pot, was allowed to hang free with the ligament fibres parallel. This gave a 2.3 N load, which was sufficient to straighten the ligament fascicles. Quick setting silicone rubber was then poured around the ligament (see Fig. 4). The rubber used was 'Silcoset 105' (Ambersil, Bridgewater, Somerset, England), with rapid curing agent. It was mixed by hand to give a setting time of 15 min. When the rubber had set the cardboard was removed. The mould was removed by making a single radial cut with a scalpel, carefully avoiding damage to the ligament bundle. This cut allowed the mould to be easily peeled off.

Ligament bundle replicas were made by pouring pmma into the silicone rubber moulds. The moulds were held together with adhesive tape. When the replicas had set they were removed by cutting the tape.

The pmma replicas were scrubbed with disinfectant and detergent and then soaked in 'Hibiscrub' bactericidal cleansing solution. Sections 5 mm long were cut from the middle and the two faces filed perpendicular to the bundle axis. These faces were then stained with black ink and photographed with a steel rule in the same plane (see Fig. 5). The resulting slides were placed in a standard enlarger and the bundle replica cross-section traced onto graph paper, along with a 10 mm reference, magnified by a factor of 10. The cross-sectional areas were then determined by counting the enclosed squares and scaling according to the 10 mm reference. The smaller of the two areas for each replica was taken as the cross-sectional area of the bundle; careful dissection meant that ligament fibres were not removed, but some synovium might remain, causing overestimation.

**Tensile testing**

The pPC and aPC bundles were distracted to failure in an Instron 1122 materials testing machine with a 5 kN load cell. The tibia and femoral bone blocks (in their pots) were orientated so that the load acted along the axis of the ligament bundle. Thus, the femoral pot was mounted on the load axis, held in the Instron jaws suspended from the load cell on the moving cross-head. The tibial pot was then orientated so that the bundle fibres were also on the load axis and appeared to become taut simultaneously across the width of the bundle. We found that this procedure resulted in relative tibial orientations approximately to 0° flexion for the pPC (tibia vertical) and to 70° flexion for the aPC (tibia extended 45° from the vertical). The fixture holding the tibial pot was secured to the base of the Instron. The smaller pPC was ruptured first; this did not appear to compromise the tibial attachment of the aPC. A crosshead speed of 1000 mm min$^{-1}$ was used to give a high strain rate (approximately 50% s$^{-1}$). This increased the probability that ligament failure would occur rather than bony avulsion (Noyes et al., 1974). The load/crosshead-displacement data was captured on a Nicolet digital oscilloscope and then output to an X–Y plotter. The mechanical and material properties were then calculated from these graphs and the physical dimensions of the bundles.

**RESULTS**

The data shown in Table 2 resulted from interstitial fibre failure in all specimens. A typical graph of load/deflection is shown in Fig. 6. This graph is smooth, with no submaximal peaks; this indicates that gross sequential failure did not occur and suggests that the alignment of the bundles in the Instron resulted in even fibre loading. The properties of the two bundles of the PCL measured in this study are shown in Table 2 along with p values for paired Student's t-tests. The cross-sectional area of the aPC was considerably larger than that of the pPC. As would therefore be expected the mechanical properties of the bundles are different: the maximum load and stiffness of the aPC bundle were much higher than those of the pPC. The maximum strain for both bundles was the same but the maximum stress, modulus and energy density were significantly higher for the aPC; therefore the material properties of the two bundles were significantly different ($p<0.05$).

**DISCUSSION**

**Sources of error**

In this study the strain data is based on the grip-to-grip displacement of the Instron. This data is subject to systematic errors, the compliance of the clamping system, of the tibia and of the bone–ligament junction.

The stiffness of the clamping system plus tibia was tested by applying a load, via a wire, in the same manner as for the aPC test. The deflection of the tibia, relative to the Instron base plate, was measured with a dial gauge accurate to 0.01 mm. The stiffness of the clamp/tibia system was found to be 3.4 kN mm$^{-1}$. This figure was used as a correction factor in calculating the data in Table 2. The compliance of the bone–ligament interface is harder to establish. Some
Table 2. Mechanical and material properties of the two bundles of the PCL. ‘p’ values are for the significance level of differences between the anterolateral (aPC) and posteromedial (pPC) bundles in a paired t test

<table>
<thead>
<tr>
<th>Property</th>
<th>pPC</th>
<th>aPC</th>
<th>p (paired t)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Length (mm)</td>
<td>33.8 ± 3.2</td>
<td>35.3 ± 3.4</td>
<td>&gt;0.1</td>
</tr>
<tr>
<td>Area (mm²)</td>
<td>10.0 ± 1.3</td>
<td>43.0 ± 11.3</td>
<td>&lt;0.001</td>
</tr>
<tr>
<td>Max. load (N)</td>
<td>258 ± 83</td>
<td>1620 ± 500</td>
<td>&lt;0.001</td>
</tr>
<tr>
<td>Stiffness (N mm⁻¹)</td>
<td>77 ± 32</td>
<td>347 ± 140</td>
<td>&lt;0.001</td>
</tr>
<tr>
<td>Linear strain (%)</td>
<td>12.9 ± 2.9</td>
<td>13.2 ± 6</td>
<td>&gt;0.1</td>
</tr>
<tr>
<td>Linear stress (MPa)</td>
<td>18.7 ± 10.3</td>
<td>28.7 ± 17.0</td>
<td>0.086</td>
</tr>
<tr>
<td>Modulus (MPa)</td>
<td>145 ± 69</td>
<td>248 ± 119</td>
<td>0.011</td>
</tr>
<tr>
<td>Max. strain (%)</td>
<td>19.5 ± 5.4</td>
<td>18.0 ± 5.3</td>
<td>&gt;0.1</td>
</tr>
<tr>
<td>Max. stress (MPa)</td>
<td>24.4 ± 10.0</td>
<td>35.9 ± 15.2</td>
<td>0.027</td>
</tr>
<tr>
<td>Energy density at failure (J cc⁻¹)</td>
<td>2.06 ± 0.73</td>
<td>3.12 ± 1.4</td>
<td>0.045</td>
</tr>
</tbody>
</table>

Mean ± S.D.

Fig. 6. Typical load/deflection graph for the aPC: (a) beginning of linear region, (b) yield point, (c) max load.

workers have suggested that video-dimensional analysis (VDA) is the answer (Woo, 1982). However, this technique was rejected by the authors as, although precise, it did not promise significantly more accurate results—VDA is limited to surface strain measurement and since the bundles had a spread of fibre lengths of up to 3 mm it was not considered that the surface strain was necessarily representative of the whole bundle. Also, Butler et al. (1990) have shown that surface strain varies along a fascicle. VDA is also subject to error due to progressive fibre failure blurring the reference lines. A recent study has shown that VDA has an accuracy of about ±10% at 10% nominal strain (Lam et al., 1992). The authors considered that the grip-to-grip displacement gave a more representative estimate of the bone-to-bone extension of the ligament bundles as a whole. Also, Butler et al. (1990) have shown that surface strain varies along a fascicle. VDA is also subject to error due to progressive fibre failure blurring the reference lines. A recent study has shown that VDA has an accuracy of about ±10% at 10% nominal strain (Lam et al., 1992). The authors considered that the grip-to-grip displacement gave a more representative estimate of the bone-to-bone extension of the ligament bundles as a whole. (This method was also more convenient than accurately mounting a clip gauge on the bone attachments within the small pots.) The extension signal from the Instron was accurate to 0.01 mm; which gives an average error, for the extensions measured, of ±0.2%. The graphs of load/extension were read to an accuracy of 1%. The error in average rest length measurement was ±1.5%. Thus, the overall error in average strain was ±2.7% of the given value. Since there was a spread of fibre lengths the strain for fibres at the extremities was within ±4.5% of the average. This spread in fibre lengths must have caused some sequential failure. However, since no ‘steps’ were observed in the load/extension graphs the effect must have been small (see Fig. 6).

The accuracy of calculated stress is dependent on the error in measured load and cross-sectional area. The error in the Instron load signal was ±0.5% and the plotted graphs were read to ±1%, therefore the error in measured load was ±1.5%. The error in cross-sectional area measurement was ±2% (based on repeated tests on silver steel specimens of known dimensions). This is comparable to the accuracy reported for the laser micrometer (Lee and Woo, 1988). However, a laser micrometer could not be used on the pPCs because of obvious concavities. Even in the ‘rounder’ aPC our study indicates that assuming convexity would result in overestimations of areas of between 3 and 8% (see Fig. 5). By comparison, the area micrometer of Butler et al. (1986) deliberately applied a constant pressure of 0.12 MPa, which was needed to force the specimen to conform to a rectangular aperture in the apparatus. We estimate that this load regime caused the area micrometer to give areas approximately 20% lower than those found using the replica technique in this study. (Technical note submitted for publication.) The accurate determination of stress is also dependent, as is the maximum load, on the even loading of ligament fibres. Uneven fibre loading causes progressive failure and therefore underestimation of maximum load and stress. As stated earlier this type of error would have occurred if the whole ligament had been tested, as has been shown for the ACL by Woo et al. (1991). However, since the two bundles were tested separately and care was taken to align the bone blocks, the fibres across each bundle were loaded as near simultaneously as possible, given the spread of fibre lengths, and progressive tearing was limited, as can be seen from Fig. 6.

It is worth noting that several previous studies of ligament material properties have identified submaximal peaks and referred to them as the ‘yield point’
Mechanical properties of cruciate ligament

(Leauchamp et al., 1979, Cabaud, 1983, Kennedy et al., 1976, Marinozzi et al., 1983). The other studies in Table 1 did not include load/deformation graphs so their accuracy is not known. We think that 'yield point' is a misnomer, and that it actually refers to the onset of ligament fascicle ruptures. Noyes et al. (1974) and Noyes and Grood (1976) noted a submaximal peak, labelled as 'linear load'. In the study of Noyes et al. (1974) the text defines 'linear load' as the load at the first failure causing a drop in load of more than 2 kgf and goes on to say that this value is an approximation for the onset of failure. Subsequent studies seem to have taken this approximation to be the definition of the 'yield point'. The actual yield point, when the gradient of load/deformation graph decreases until failure, can only be accurately determined if sequential failure does not occur. The present study shows that this effect occurs immediately prior to simultaneous fibre failure. We believe that data published previously that includes 'yield points' is actually representative of tearing-type sequential failures, which will not lead to maximal loads.

Age effects

Noyes and Grood (1976) and Hollis et al. (1988) have shown that there is a correlation between ACL strength and age. Both studies show that the mean strength of the ACL in the younger population (under 26 yr) is greater, by a factor of about 2.5, than the mean strength in the older population (over 50 yr). The cause of the above correlation is not known, therefore comparisons with the present study based on age cannot be definitive. The validity of comparisons is dependant on the probability that the variable/s responsible for the changes found in the studies of Noyes and Grood (1976) and Hollis et al. (1988) show the same correlation with age in all the compared samples. Given the above caveat, the data in this study, which is of an older sample (53–98 yr), suggests that the mean strength of the PCL in young adults may be 1.6 \times 2.5 = 4 \text{kN}. Noyes and Grood (1976) also showed that between the younger and older populations the maximum stress decreased by a factor of 2.8; the modulus decreased by a factor of about 1.7; and the strain energy density decreased by a factor of about 3.3. Noyes and Grood suggest that the factors for maximum stress and strain energy are overestimates because the older ligaments failed prematurely by bony avulsion but that the aging effects were still significant.

Clinical relevance

If the age effects discussed above hold true for this study our results indicate that PCL reconstructions for young active people should be designed to match a natural structure with a strength of 4 kN.

This study supports the concept of a twin bundle PCL, as we found that the bundles could be separated by their reciprocal tightening and slackening in flexion and extension. It has also shown that the two bundles have different mechanical and material properties. The aPC is approximately six times as strong as the pPC. Therefore, we conclude that it is the aPC—which resists posterior draw in flexion (Girgis et al., 1975; Hughton et al., 1980)—which is primarily responsible for the stabilizing effect of the PCL. This strength difference and the superior material properties of the aPC suggest the hypothesis that the aPC is loaded either more frequently or to a higher stress than the pPC during normal activity, but data on this is not available.

Since the two PCL bundles have reciprocal actions, it is impossible to reconstruct the entire PCL with a single strand of material. To reproduce the natural PCL a double bundle, aPC and pPC, reconstruction is required. However, since the pPC has been shown in this study to contribute only one quarter of the bulk and one sixth of the strength of the aPC we do not consider the surgical complexity of a double bundle reconstruction to be justified. We recommend that PCL deficiency should be treated by reconstructing the aPC alone.

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REFERENCES


