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## Mechanical Analysis of Cartilage Graft Reinforced with PDS Plate

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### Abstract

**Objectives/Hypothesis**—This study attempts to characterize the biomechanical properties of a PDS-cartilage composite graft for use in septorhinoplasty.

**Study Design**—Experimental Study.

**Methods**—This study used a PDS analog, porcine cartilage cut to  $1 \times 5 \times 20$  mm, and a mechanical testing platform to measure flexure of a composite graft. Samples were assessed in four groups based on variations in suture pattern and orientation. The platform measured the force required to deflect the sample 2 mm in single cantilever beam geometry before and after the polymer was affixed to the specimen. Elastic Moduli were calculated before and after application of the polydioxanone polymer.

**Results**—The average modulus of the cartilage alone was  $17 \pm 0.9$  MPa. The modulus of the composite cartilage-polymer graft with 2 suture fixation was  $21.2 \pm 1.5$  MPa. The 3-suture configuration produced an increase to  $25.8 \pm 2.23$  MPa. The four-suture configuration produced  $23.1 \pm 3.19$  MPa. The five-suture configuration produced  $25.7 \pm 2.6$  MPa. The modulus of the analog alone was  $170 \pm 30$  MPa. The modulus of the 0.5 mm PDS was  $692 \pm 37.4$  MPa. The modulus of the 0.15 mm perforated PDS was  $447 \pm 34.8$  MPa.

**Conclusions**—The study found that suturing a polymer plate to cartilage resulted in enhanced stiffness of the composite. Under the conditions of the study, there was no significant difference in elastic moduli between suture configurations, making the two-suture linear configuration optimal in the one-plane cantilever deflection model.

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## Keywords

Basic research; facial plastic surgery; polydioxanone; cartilage; elastic modulus; rhinoplasty; septorhinoplasty; composite cartilage grafts; biomechanical properties of tissue; porcine cartilage

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## INTRODUCTION

Polydioxanone (PDS) plates or foils are sheets of a bioresorbable polymer used in the stabilization of autologous cartilage grafts in septorhinoplasty. PDS plates have been used for over 10 years in Europe for the stabilization of nasal cartilage framework structures.<sup>1</sup> Its primary and original use in nasal surgery was as a scaffold to assemble cartilage grafts and cartilage graft remnants for extracorporeal septoplasty. PDS differs from other graft-stabilizing materials in that it biodegrades and is completely resorbed by about 25 weeks. Nonetheless, the PDS plate remains structurally unchanged for at least 10 weeks. It is during this period that cartilage may remodel and be enveloped by connective tissue while supported and stabilized by the implant.<sup>1</sup>

Since FDA approval in late 2011, the use of PDS foil in nasal surgery has expanded. Most rhinoplasty surgeons view PDS foil as a material to enhance the intrinsic structural nature of the supporting cartilaginous framework of the nose, primarily along the dorsal and caudal L-strut augmentation after submucous resection of cartilage in septoplasty. PDS can also be used to straighten curved segments of cartilage for use in various roles, such as lateral crural strut and nonload-bearing settings, e.g. spreader grafts.

Though the usefulness of PDS plates in reconstructive procedures is well documented, to our knowledge no study has investigated the biomechanical properties of a PDS-cartilage composite graft. This study attempts to characterize the mechanical properties of a PDS-cartilage graft in flexure by calculating the elastic modulus before and after the application of the PDS plate. We thereby attempt to delineate increases in the strength of the graft.

We believe that PDS can function to enhance structural stability in a different role, beyond its application as a loadbearing structure that simply stiffens cartilage. PDS foil can be used as a means to stabilize tissue structure by resisting flexure. This comes from its ability to transmit tensile forces across a curved specimen, that is, resist curvature due to its ability to resist tensile deformation. Flexure and buckling are the most important modes of deformation in structural rhinoplasty, and these are the forces that PDS counteracts, in the same way that struts and ties in bridge or skyscraper construction balance loads and torsion without adding significant weight or construction costs. We believe PDS foil does primarily act as a reinforcement to stiffen tissue via the balancing and redistribution of forces.

Using materials as tensile elements to further stabilize structures is well known and these principles have been known by structural engineers for the last two millennia. Steel used in this setting carries a significant additional expense as excessively heavy structures can collapse under their own weight. Similarly, in rhinoplasty, the use of PDS comes at the expense of material costs, plus the potential for an extended foreign body reaction. The risk of this reaction increases proportionally with the mass of PDS used. As such, the thinnest,

most low-profile material possible should be selected for use in stabilization of the native structure. In this study, we provide information with respect to the use of PDS as a tensile element to stabilize nasal structure in septorhinoplasty.

## MATERIALS AND METHODS

### Tissue Preparation

Fresh porcine rib cartilage segments were obtained from a local abattoir, and  $1 \times 5 \times 20$  mm sections were excised using a precision-cutting device as previously described.<sup>2</sup> Briefly, the cartilage was prepared by removing connective tissue and perichondrium and placed in the cutting device. The device was set to cut a 1-mm thick specimen that was harvested from the center portion of the rib to minimize warping (balanced cross sections). The outer portions of the rib were discarded after cutting. Immediately after sectioning, samples were wrapped in gauze soaked in a phosphate buffer solution (PBS) (pH  $\approx$  7.4), vacuum-packed and refrigerated (4°C), unless immediately used. All samples were used within 1 week of harvest. Prior to use, the refrigerated specimens were allowed to re-equilibrate to ambient temperature and immersed in PBS to control hydration state and osmolarity. Figure 1 outlines the tissue preparation procedure used in this study.

### Mechanical Testing

Samples were analyzed using a fixed-end single-beam horizontal cantilever model similar to previous investigations.<sup>3,4</sup> Mechanical analysis was performed using a precision testing platform (Enduratec ELF 3200, Bose Corp., Eden Prairie, MN) with a custom designed jig constructed from Plexiglass (Fig. 2). A total of 49 specimens were tested. Each suture pattern was examined using 10 specimens, except for the two-suture pattern that had a sample size of 19. Specimens were secured and mounted on the jig with 20 mm of the tissue exposed. A stainless steel actuator deflected the specimen at the tip, 20 mm from the fulcrum (Fig. 2). This actuator was programmed to displace the specimen 2 mm at a frequency of 1Hz and a constant strain rate of 4 mm/s. The tip of the actuator was aligned so it formed a 90-degree angle with the surface of exposed tissue when a displacement of 1 mm was reached. Force of reaction to the mechanical displacement was measured by the load cell attached to the jig. Each specimen underwent 20 cycles of cyclic loading while force and displacement were recorded by the mechanical testing platform.

### Calculation of Elastic Modulus

Native cartilage tissue and composite PDS plate–cartilage grafts were tested and the stiffness was assessed using cantilever geometry approximation. The elastic modulus  $E$  was denoted for cartilage and  $E_c$  for composite cartilage grafts (PDS sutured to cartilage).  $E$  for a single beam cantilever is

$$E = \frac{PL^3}{3\delta I}$$

where  $P$  is the force required to displace the tissue,  $L$  is the length from the fulcrum to the location of probe,  $\delta$  is displacement distance, and  $I$  is the moment of Inertia. (Fig. 3) The

mechanical testing platform deflected the sample to a constant value of 2 mm for  $\delta$ . The moment of inertia for the cartilage sample  $I$  was constant across samples since the samples were cut to the same size. The length  $L$  was also constant across samples. The force  $P$  was measured by the mechanical testing platform for each sample, allowing the elastic modulus  $E$  to be calculated.

### PDS Plate Selection and Suturing Methods

As this study required mechanical analysis of numerous PDS–cartilage composite grafts in order to examine the impact of suture number and geometry on specimen stiffness, we opted to use a material with similar mechanical properties and identical thickness. After examining dozens of suitable materials obtained from art supply, household goods, hardware, and department stores, we identified a thin plastic sheet with similar mechanical properties to the PDS foil. The sheet found was the protective cover of a photo album (Kolo Lux Cortina, Kolo Retail LLC, Hartford CN). Pieces of the substitute polymer were first perforated using a 2.5 mm dermatologic punch that was identical to the measured 2.5 mm perforations of the 0.15 mm thick PDS plates. The mock PDS was then cut to the size of the cartilage specimens. The substitute plate was then sutured to the cartilage in one of four ways, as illustrated in Figure 1c: two sutures evenly spaced along the long axis of the cartilage sample; three sutures evenly spaced along the long axis of the sample; four sutures, each at one corner of the sample; and five sutures in a quincunx orientation. The use of this “mock” PDS plate material facilitated the execution of mechanical testing on numerous samples as using actual was cost-prohibitive as PDS plates cost \$299 per sheet (0.15 × 50 × 40 mm).

Mock PDS plate allowed the systematic examination of the impact of suture number on  $E_C$  in composite grafts. In this study, it was observed that  $E_C$  did not increase appreciably with an increase in suture number, at least not in the plane of deflection tested. Next, actual PDS–cartilage grafts were assessed, using the two thicknesses of PDS available for the study in limited quantities. The 0.15 mm perforated PDS and 0.5 mm unperforated PDS foils were both affixed to the cartilaginous specimen using the two suture configuration, as described above, and were then assessed using the mechanical testing platform.

### Statistical Analysis

The average elastic modulus of native and composite tissue ( $E$  and  $E_C$ ) was calculated for each suture style cohort. A one-way analysis of variance (ANOVA) was performed on the dataset to determine which suture configurations were statistically different, if any. If the result of the ANOVA analysis shows that means are different and statistically significant ( $p < 0.05$ ), a student  $t$ -test was used to determine what suture styles were significant ( $p < 0.05$ ) relative to other suture methods used in this study.

## RESULTS

Though the increase in  $E_C$  over native cartilage ( $E$ ) was significant for each suture in general, adding mock PDS stiffens the cartilage significantly (Fig. 4). The average elastic modulus of the native costal cartilage was  $17 \pm 0.9$  MPa. The modulus of the “Mock” PDS plate was  $170 \pm 30$  MPa. The modulus of the 0.5 mm PDS was  $692 \pm 37.4$  MPa. The

modulus of the 0.15 mm PDS was  $447 \pm 34.8$  MPa, lower than the 0.5 mm thick plates due to the difference in form factor (perforations). Figure 4 shows the moduli of the cartilage and mock-PDS samples in the four-suture geometries. Figure 5 depicts the moduli of a cartilage graft with 0.15 mm and 0.5 mm PDS sutured in the two-suture configuration. ANOVA has shown no statistically significant difference between suture techniques ( $p > 0.05$ ); therefore, paired  $t$ -test was not performed to see which suture techniques are statistically different.  $E_c$  is statistically different than  $E$  for all suture techniques ( $p < 0.05$ ).

## DISCUSSION

PDS is becoming more widely used in the United States after a decade of use in Europe. PDS has many clinical applications, including joining cartilage fragments, acting as a scaffold for cartilage remnants, and being used to bridge and brace segments in order to prevent and reduce flexure and torsion. To our knowledge, the mechanics of PDS foils and its impact on enhancing cartilage mechanical function has not been studied from a quantitative standpoint. PDS is available in 0.5 mm, 0.25 mm, and 0.15 mm thickness, and the indications for use of each have been left to the clinical judgment of the surgeon. A recent study compared the use of perforated and unperforated PDS foil with respect to functional and aesthetic sequelae following reconstruction of the nasal septum.<sup>6</sup> In their study, Tweedie et al. reported cases of saddling using unperforated PDS and advised against its use in this context.<sup>6</sup> They found no cases of saddling using the 0.15 mm perforated PDS and felt it was an excellent choice for septal reconstruction.<sup>6</sup>

The objective of this study was to investigate the mechanical behavior of cartilage–alloplast composite grafts. A horizontal flexure model was chosen for its simplicity and applicability: one of the most simple uses of PDS foils is to straighten and stabilize the L-strut dorsally and caudally. The clinical issue at hand is whether these subjective observations are valid, and what role suture geometry plays in the structural characteristics of a cartilage–alloplast composite graft. If the PDS-cartilage composite is viewed as a unified structural beam, the stability is achieved with increasing beam thickness/width: albeit at the expense of cost and foreign body reaction. In contrast, examining the composite structure from the vantage point of structural mechanics, the composite is akin to a bridge design where the composite structure forms a truss, and the PDS a “tie” with the cartilage forming a “strut” when subjected to flexural deformation. In this perspective, the thinnest plate with the least mass achieves the objective of resisting flexure. Simply put, the PDS resists tensile deformation and redistributes and balance forces.

The use of a substitute polymer to take the place of actual PDS in this project was a practical matter for a study of this scope. Using a proxy for PDS reduced the prohibitive cost of using actual PDS for each step of the experiment. Finding a suitable substitute for PDS presented an early challenge: it needed to approximate actual PDS in both thickness and stiffness. The substitute used in this study is of identical thickness to the 0.15 mm PDS, though somewhat softer, as evidenced by its lower elastic modulus.

Similar considerations were given to the choice of porcine costal cartilage as a proxy for its human counterpart. While there is no established animal tissue model for human septal

cartilage, rabbit, and porcine tissues have been studied in comparative analyses.<sup>7</sup> These tissues are readily available and are grossly similar in form factor and mechanical behavior. While it would be ideal to use fresh human septal cartilage, remnants left over from septorhinoplasty make it nearly impossible to obtain sufficient quantities of this tissue.

To simplify analysis, assumptions were made in this study regarding the conditions of mechanical analysis and the behavior of cartilage. Cartilage is not a linearly elastic material with isotropic mechanical properties. In reality, it is best described as anisotropic and a nonlinear viscoelastic material with triphasic behavior based upon matrix macromolecular mechanical properties, water flow through the matrix, and electrostatic factors.<sup>8</sup> Furthermore, the mechanical properties of cartilage differ in tension and in compression—both modes of deformation are encountered during cantilevered flexure on opposite sides of the graft during flexion.

A second major assumption was that the composite grafts created in the study behaved as a single mass with lumped material behavior. In reality, the PDS and the graft are fixed to one another at discrete points, and other regions are free to move independently. Mechanically, the equations of state describing the motion of these grafts even in simple deformation are complex, and best analyzed using finite element modeling that is beyond the scope of the present study, and only now just being used to describe the mechanics of facial cartilage.<sup>9,10</sup> The fixation of the PDS to the graft is best viewed from the vantage of structural engineering, where the alloplast in this case functions more like rebar in reinforced concrete, or trusses spanning a bridge.

With regard to suture configurations, we found no gains with additional sutures beyond two for the horizontal flexure deformation used in this experiment. The cartilage-PDS composites are not fused structures, such as in a true bilaminar composite material. The mechanical behavior is therefore an aggregate of the mechanical behavior of two separate structures. The behavior is predominantly a result of the rigidity of the plate, which limits deformation. This view, adopted by most surgeons, is analogous to a bridge where reinforcement comes with a thicker beam in parallel to and supporting the roadway. In contrast, PDS in rhinoplasty is more akin to a suspension beam construction where forces of tension and compression are balanced. It must be emphasized that the plate is secured to the cartilage only at discrete points, through which force is being transmitted between the two structures. This is analogous to the engineering concept of pin jointing and is similar to the lateral crural mattress suture used to correct lateral crural convexity.<sup>11</sup>

In this experiment, two attachment points were sufficient to resist the simple mode of horizontal deflection, and additional attachment points did not confer any significant further gains in modulus (Fig. 4). Since the composite cartilage graft and mock PDS plate did not comprise a solid structure but were attached to each other at fixed points, both the cartilage and the plate experienced linear stretching and compression along their respective convex and concave surfaces. Mechanical forces resisting these deformations are applied to the attachment points and are proportional to the distance between them. Introduction of additional attachment points will result in decrease in mechanical force applied to each point in proportion to the reduction in distance between them. However, the total mechanical

resistance (force) to deflection will not change as it is still the sum of all forces acting within cartilage-PDS composite.

In surgery, more complex forces acting on the graft may result in tension, shear, buckling, and torsional effects in three dimensions. Such complex deformations did not occur in this study, because only in-plane isolated flexure was investigated. In addition, if prolonged stress-creating deformations are expected, multiple sutures can be beneficial as they reduce deformation resisting force at the points of each suture and reduces the likelihood of “cheese-wiring,” that is, cutting of suture through cartilage.

Previous work provides information on what the minimum acceptable strengths is for select applications in rhinoplasty.<sup>5</sup> Further work needs to be performed using mechanical testing geometries and models that more closely reflect actual clinical conditions. We believe that for most applications, 0.15 mm-perforated PDS is of acceptable stiffness, and being thinner and perforated will allow it to absorb faster. Disadvantages of thicker plates are a greater potential for scar formation and reactive inflammation. Since cartilage is an avascular tissue that acquires nutrients through diffusion, a thicker PDS plate presents a greater barrier to diffusion of nutrients to the cartilage. In terms of stabilizing and reinforcing native cartilage frameworks, we believe less in plate size and thickness and more in the strategic placement of PDS to balance flexural and torsional deformation. We need to view the use of PDS plates in rhinoplasty, much in the way that structural engineers view stability in static and dynamic structures. The use of a thin, high modulus material can achieve dramatic changes in overall structural stability when strategically applied to redistribute force. As evidenced by the results of this study, PDS foil can fulfill this role in septorhinoplasty.

## CONCLUSION

The study found that suturing a polymer plate to cartilage resulted in significant gains in stiffness. In the study’s horizontal cantilever deflection model for measuring elastic modulus, there was no significant difference in elastic modulus between the suture configurations, making the simple two-suture linear configuration optimal in the plane of deflection tested. As such, PDS foil effectively redistributes tensile and compressive biomechanical forces to stabilize cartilaginous grafts in septorhinoplasty.

## Acknowledgments

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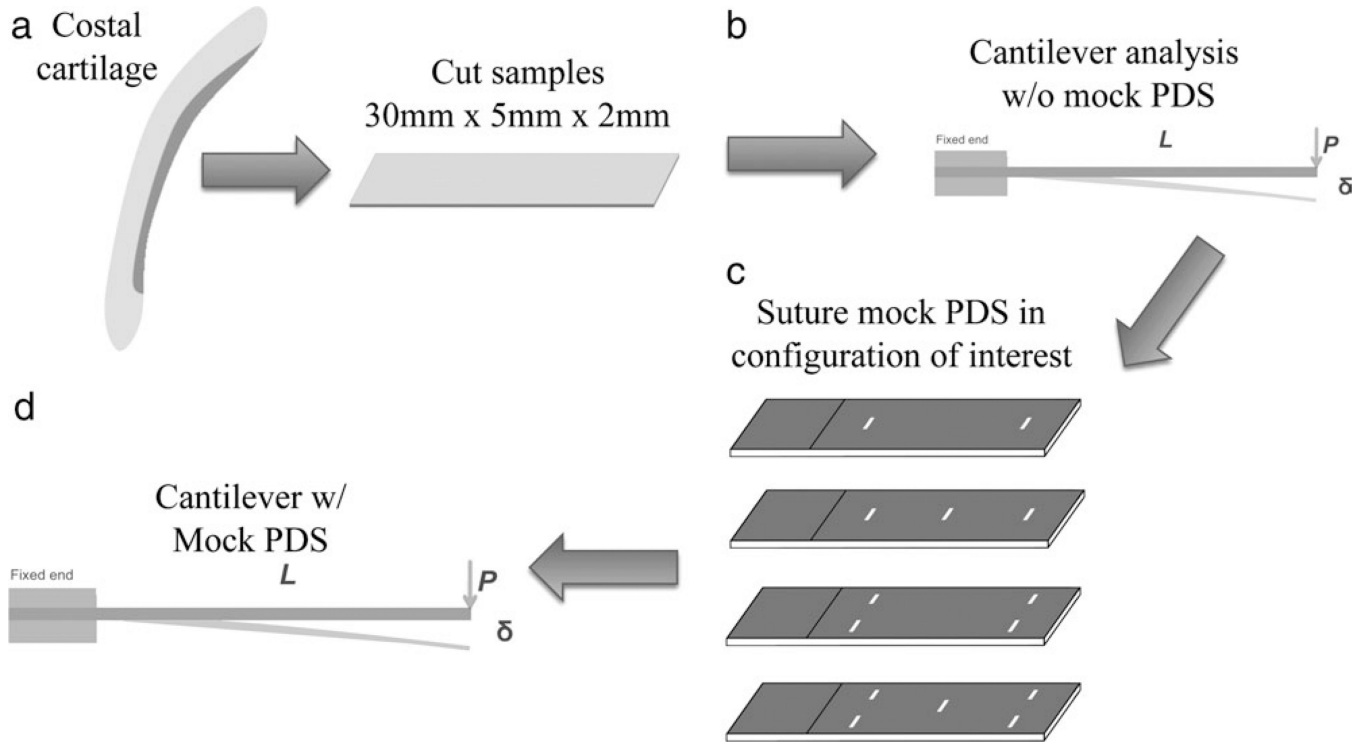
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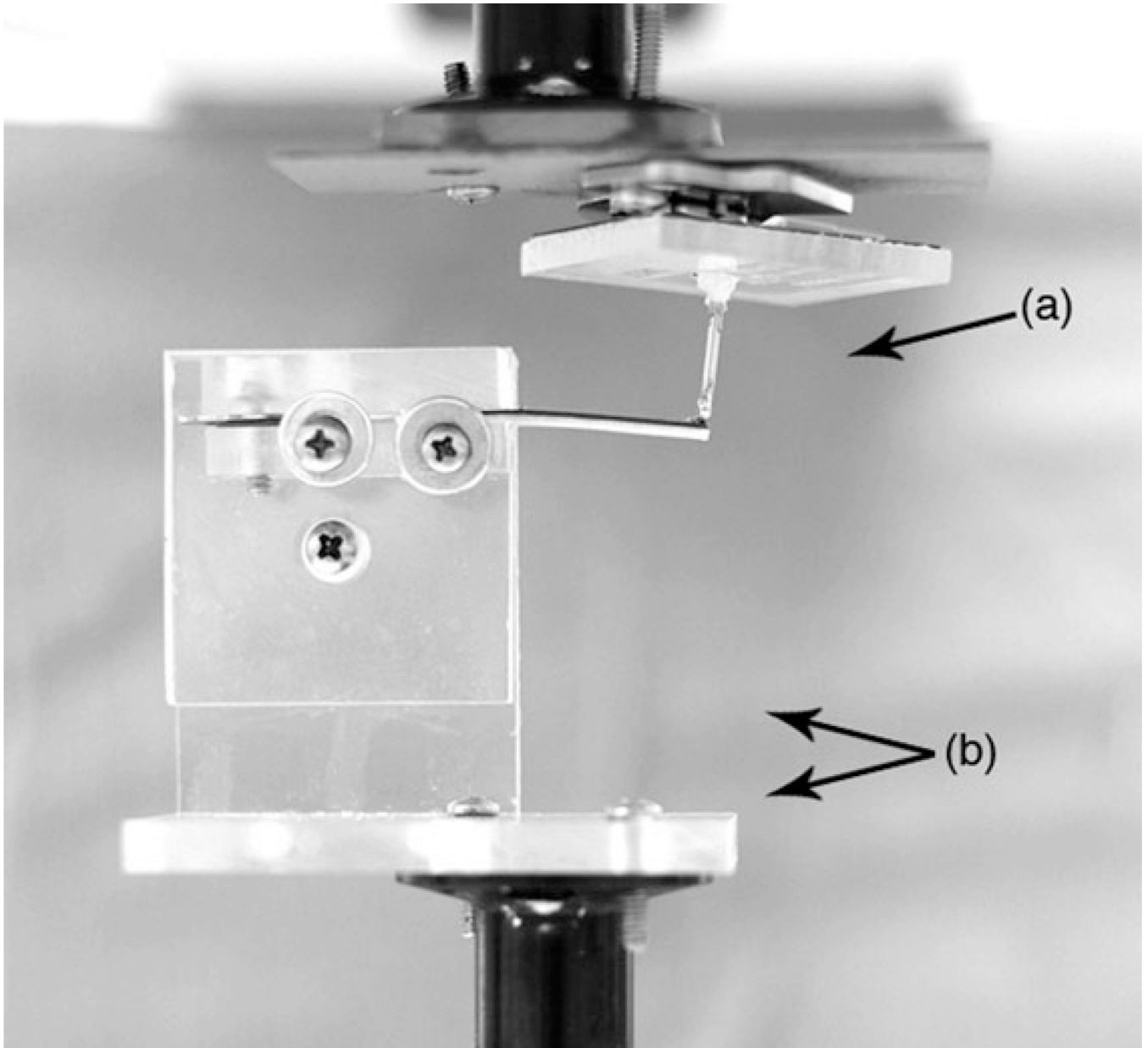
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**Fig. 1.**

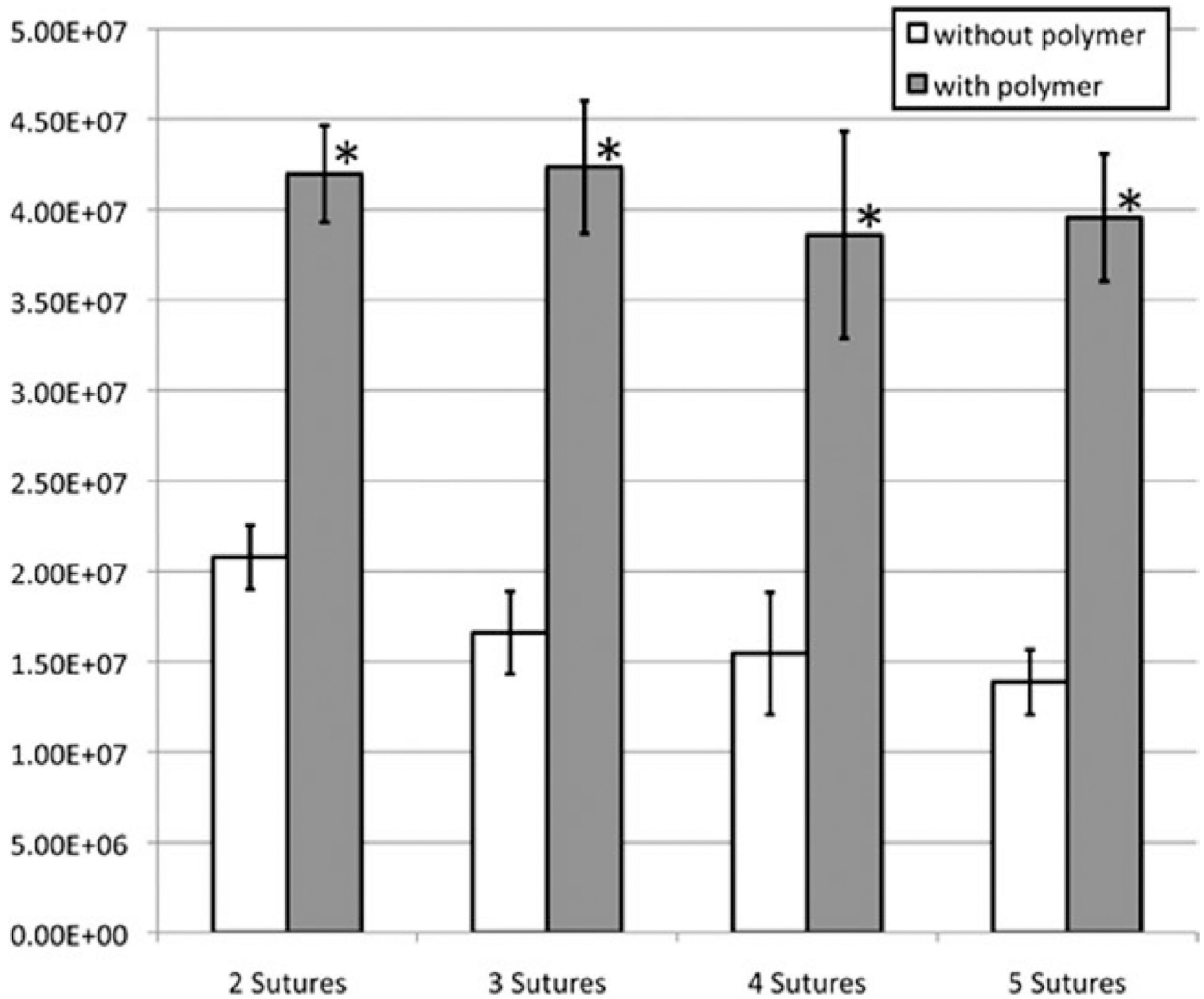
The experimental method for the first part of the experiment. (a) Porcine costal cartilage was cut to size in a precision cutting device.<sup>5</sup> (b) Specimen mounted on a custom jig using a cantilever beam geometry. (c) Mock PDS was affixed to the cartilage in one of the four configurations of interest: 2 and 3 sutures spaced evenly along the long axis of the sample, 4 sutures each in a corner of the sample, and 5 sutures in a quincunx. (d) The modulus of the cartilage-mock-PDS composite was assessed using the fixed-end horizontal cantilever model.



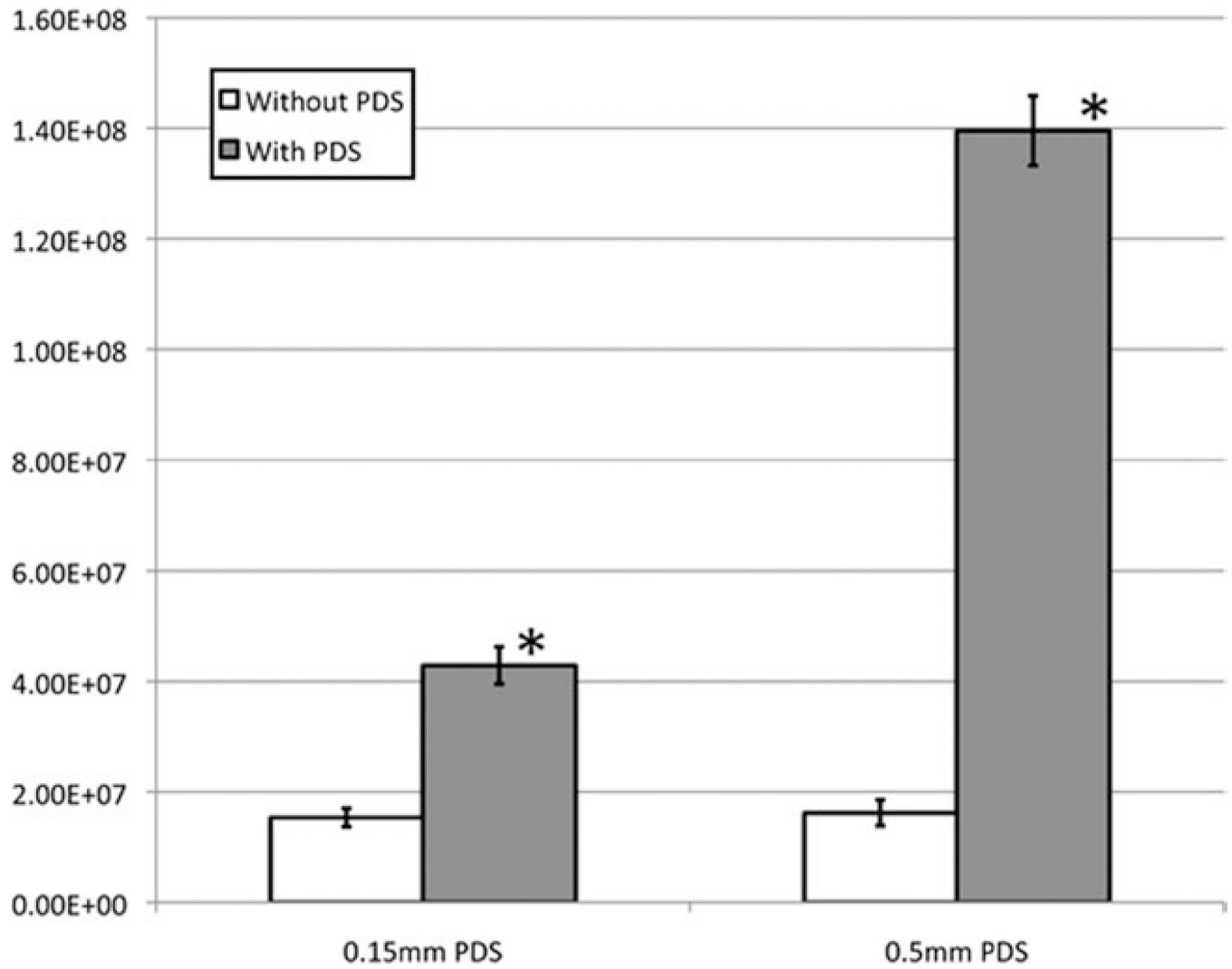
**Fig. 2.** The custom jig used to mount the cartilage, mounted on to the mechanical testing platform (Enduratec ELF 3200, Bose Corp., Eden Prairie, MN). (a) Linear actuator with stainless steel blunt tip. (b) Load cell with mounting apparatus.



**Fig. 3.** The fixed-end horizontal deflection cantilever model was used to assess the elastic modulus of the samples.  $L$  = length of cantilever,  $P$  = force applied to deflect cantilever,  $\delta$  = deflection of cantilever.



**Fig. 4.** The experimental data in bar graph form. Elastic modulus E of native cartilage compared to the elastic modulus, Ec of mock PDS. Asterisk denotes statistical significance ( $p < 0.05$ ).



**Fig. 5.** Elastic modulus  $E$  of native cartilage compared to the elastic modulus of PDS,  $E_c$  with the two-suture configuration. Asterisk denotes statistical significance ( $p < 0.05$ ).