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Journal

Facial Plastic Surgery, 29(06)

ISSN

0736-6825

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Publication Date

2013-12-01

DOI

10.1055/s-0033-1360593

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Peer reviewed



Published in final edited form as:

Facial Plast Surg. 2013 December ; 29(6): 502–505. doi:10.1055/s-0033-1360593.

Analysis of Cartilage-Polydioxanone Foil Composite Grafts

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Abstract

This study presents an analytical investigation into the mechanical behavior of a cartilage-polydioxanone (PDS) plate composite grafts. Numerical methods are used to provide a first-order, numerical model of the flexural stiffness of a cartilage-PDS graft. Flexural stiffness is a measure of resistance to bending and is inversely related to the amount of deformation a structure may experience when subjected to bending forces. The cartilage-PDS graft was modeled as a single composite beam. Using Bernoulli-Euler beam theory, a closed form equation for the theoretical flexural stiffness of the composite graft was developed. A parametric analysis was performed to see how the flexural properties of the composite model changed with varying thicknesses of PDS foil. The stiffness of the cartilage-PDS composite using 0.15-mm-thick PDS was four times higher than cartilage alone. The composite with a 0.5-mm-thick PDS graft was only 1.7 times stiffer than the composite with the 0.15-mm-thick PDS graft. Although a thicker graft material will yield higher flexural stiffness for the composite, the relationship between composite stiffness and PDS thickness is nonlinear. After a critical point, increments in graft thickness produce gradually smaller improvements in flexural stiffness. The small increase in stiffness when using the thicker PDS foils versus the 0.15 mm PDS foil may not be worth the potential complications (prolonged foreign body reaction, reduction in nutrient diffusion to cartilage) of using thicker artificial grafts.

Keywords

facial plastic surgery; polydioxanone; cartilage; elastic modulus; rhinoplasty

The use of polydioxanone (PDS) plate or foil as an alloplastic graft material during septoplasty, rhinoplasty, and nasal reconstructive surgery has become increasingly popular in the last decade.¹ PDS is a bioresorbable polymer that can be used as a scaffold to support and stabilize cartilaginous structures in the nose. Unlike other synthetic graft materials, PDS will completely biodegrade and resorb in approximately 25 weeks inside the body, providing structural support to the cartilage as it dissolves and allowing for alignment of scar tissue along favorable vector lines.¹

PDS grafts are commonly used in septal surgery. Several studies have shown that PDS foil can be effectively used as a batten or support for nasal cartilage during straightening or reconstruction of the septum.^{2,3} However, limited studies have investigated the mechanical behavior of a cartilage-PDS compound graft and how the different thicknesses of PDS affect cartilage-PDS grafts. Currently, PDS is available in 0.5, 0.25, and 0.15 mm thicknesses. The 0.15-mm-thick PDS sheets are perforated, whereas the 0.25- and 0.5-mm-thick sheets are unperforated. Fig. 1 shows images of 0.15 mm PDS foil being used to enhance the stability and straighten a curved caudal septum and L-strut during functional rhinoplasty surgery.

This study models the mechanical behavior of a cartilage-PDS graft by describing analytically how using PDS plates supports nasal cartilage and alters the stability and stiffness of the reconstructed septum. Numerical methods are used to provide a first-order approximation of the flexural stiffness of a cartilage-PDS graft. Flexural stiffness is a measure of a structure's resistance to bending and is inversely related to the amount of deformation a mass may experience when subjected to bending forces.

Methods

Cartilage-PDS composite grafts were modeled as a composite beam to allow closed form analytic description. The composite model assumes ideal bonding between the cartilage and PDS foil. The model also assumes that cartilage is a linearly elastic material with isotropic mechanical properties. The material properties of cartilage and PDS were obtained from a previous study that examined the effects of suture geometry on the flexural behavior of a cartilage-PDS composite graft.⁴⁻⁶ Table 1 displays the moduli values and constant values used in the numerical analysis.

Bernoulli-Euler beam theory and the “transformed-section method” were used to calculate a closed form equation for the theoretical flexural stiffness of the composite structure as a function of PDS thickness.⁷ Flexural stiffness is a measure of a beam's resistance to bending and is the product of the material's elastic modulus E and the beam's cross-sectional moment of inertia I . To calculate the stiffness of a two-material composite beam using the transformed-section method, the composite beam is converted into a beam made of a single material with a modified cross-sectional geometry that would yield a flexural stiffness equivalent to the original composite beam. In this study, the cartilage-PDS composite beam was converted into an equivalent beam made of only PDS. Fig. 2A illustrates the cross-sectional geometry of the original cartilage-PDS composite. Fig. 2B illustrates the cross-sectional geometry of the composite after it has been transformed to a beam made of only PDS that would yield the same flexural properties as the original composite. To transform the cross section, the width of the cartilage portion of the composite is adjusted from its original width b to a narrower width b' based on the ratio of the modulus of elasticity of cartilage to the modulus of PDS. The width of the original PDS portion remains the same. Once the cross-sectional geometry of the composite beam was transformed, a closed form formula for the theoretical bending stiffness of the composite was calculated as a function of PDS thickness, cartilage thickness, and beam width.

Numerical Analysis

The following equations were derived with Bernoulli-Euler beam theory and the “transformed-section method” and were used to determine the flexural properties of a cartilage-PDS graft:

$$b' = \frac{E_{Cart}}{E_{PDS}} \times b \quad (1)$$

$$\bar{y} = \frac{(0.5t_{PDS})(t_{PDS})(b) + (t_{PDS} + 0.5t_{Cart})(t_{Cart})(b')}{(t_{PDS})(b) + (t_{Cart})(b')} \quad (2)$$

$$\text{Flexural Stiffness } EI = E_{PDS} \times \left(\left[\frac{1}{12} b t_{PDS}^3 (0.5t_{PDS} - \bar{y})^2 \right] + \left[\frac{1}{12} b' t_{Cart}^3 + b' t_{Cart} (t_{PDS} + 0.5t_{Cart} - \bar{y}) \right] \right) \quad (3)$$

Where b is the original width of the composite, b' is the transformed width of the cartilage portion of the composite, \bar{y} is the location of the cross-sectional centroid of the composite, t_{PDS} is the thickness of the PDS portion of the composite, t_{Cart} is the thickness of the cartilage portion of the composite, and E_{PDS} and E_{Cart} are the elastic moduli for PDS and cartilage, respectively.

A parametric analysis was performed to see how the bending stiffness of the composite changed with varying thicknesses of the PDS graft.

Results

The theoretical flexural stiffness for cartilage-PDS composites was significantly higher than native cartilage alone. Fig. 3 displays the parametric analysis for flexural stiffness as a function of PDS thickness. Fig. 4A shows the theoretical cantilever tip deflection of the cartilage-PDS composite when subjected to a 0.001 N vertical force as illustrated in Fig. 4B. As seen in Fig. 3, for cartilage pieces varying from 1.5 to 2.5 mm thick, the bending stiffness of the composite initially increased very quickly with increasing thickness of PDS foil, specifically when the PDS thickness ranged between 0 and 0.1 mm. However, after a critical point, the increase in bending stiffness with increasing PDS thickness of the composite becomes more gradual. The same behavior is seen in Fig. 4 where the cantilever tip deflection of the composite, which is inversely related to flexural stiffness, initially decreases very quickly with increasing PDS thickness but eventually decreases more gradually after a critical point. Table 2 summarizes the flexural stiffness values for native cartilage and cartilage-PDS composites with commercially available thicknesses of PDS foil. The most dramatic increase in flexural stiffness is observed between the native cartilage and the cartilage-PDS composite with 0.15-mm-thick PDS foil. The stiffness of the cartilage-PDS composite using 0.15-mm-thick PDS was four times higher than cartilage alone. However, the composite with a 0.25-mm-thick PDS graft was only 1.15 times stiffer than the composite with the 0.15-mm-thick PDS graft.

Discussion

Certain assumptions were made to simplify the analysis in this study. First, cartilage was assumed to be a linearly elastic material with isotropic material properties when it is in fact an anisotropic, nonlinear viscoelastic material. Second, under the composite model, the cartilage and PDS were assumed to be perfectly bonded with infinite contact points between the two materials. In actual clinical applications, PDS foil is attached to cartilage at finite, discrete points using sutures. However, mechanical analysis of such a system is nonlinear and complex and requires the use of finite element modeling that is beyond the scope of this study. This study presents a simplified, first-order analysis of cartilage-PDS composite grafts, as a starting point to guide clinicians and a launch point for future detailed investigations.

With the composite model, adding PDS foil to support septal cartilage enhances the stability of the septum and increases the cartilage's resistance to bending. As expected, the overall flexural stiffness of the cartilage-PDS composite increases as the thickness of the PDS increases. However, while a thicker graft material will yield higher flexural stiffness for the composite, the relationship between flexural stiffness of the composite and PDS thickness is nonlinear. The parametric analysis seen in Figs. 3 and 4 showed that there is an optimum range of thicknesses (between 0 and 0.1 mm) in which the PDS foil has the greatest impact on the flexural stiffness of the compound graft. After a certain critical point, the increases in flexural stiffness of the composite with increasing PDS thickness are small.

Although using thicker PDS foil to enhance the structural properties of a septum will result in a stiffer compound, using larger amounts of artificial graft material increases the risk of potential complications such as reduced nutrient diffusion to the cartilage and prolonged foreign body reactions. Since the difference in flexural stiffness between a composite with 0.15 mm PDS foil and a composite with 0.25 mm foil is small, using the thicker PDS foil might not merit the potential complications of using thicker artificial grafts.

A recent study that reviewed 50 different cases of septal reconstructions using either 0.25-mm-thick, unperforated PDS foil or 0.15 mm perforated PDS foil and found that there was a much higher rate of complications using the 0.25 mm unperforated PDS foil versus the thinner 0.15 mm perforate PDS foil.⁸ Tweedie et al suggested that the poor surgical outcomes associated with the 0.25-mm-thick, unperforated PDS foil were likely due to the PDS foil obstructing the blood supply from the mucoperichondrial flaps to the septal cartilage. Unlike the 0.15-mm-thick PDS foil which has a quick resorption time and allows blood flow through the perforations in the foil, the 0.25-mm-thick, unperforated foil can obstruct blood flow between the mucoperichondrial flaps and the septal cartilage for weeks. Tweedie et al concluded that 0.15-mm-thick, perforated PDS foil is most suitable for septal reconstructions and have since abandoned the use of 0.25-mm-thick, unperforated foil due to the relatively high rate of postoperative complications associated with the thicker foil.⁸

Conclusion

Adding PDS foil to support septal cartilage enhances the stability of the septum and increases cartilage's resistance to bending. Thicker PDS material will yield higher flexural stiffness and resistance to bending, but the relationship between the flexural stiffness of the composite and PDS thickness is nonlinear. There is an optimum range in which the PDS foil has the greatest impact on the flexural stiffness of the compound graft. The difference in flexural stiffness between a composite with 0.15 mm PDS foil and a composite with 0.25 mm PDS foil is small, implying that the use of thicker PDS foil might not be worth the potential risks (prolonged foreign body reaction, reduction in nutrient diffusion to cartilage) of using thicker artificial grafts.

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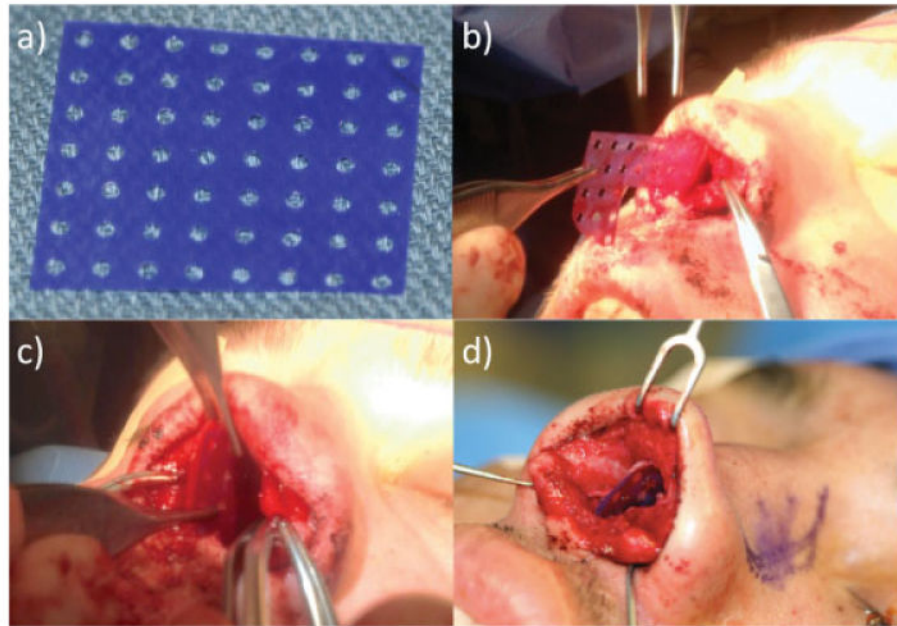


Fig. 1. (A) A 0.15 PDS foil. (B–D) PDS foil being used to straighten a curved caudal septum. PDS, polydioxanone.

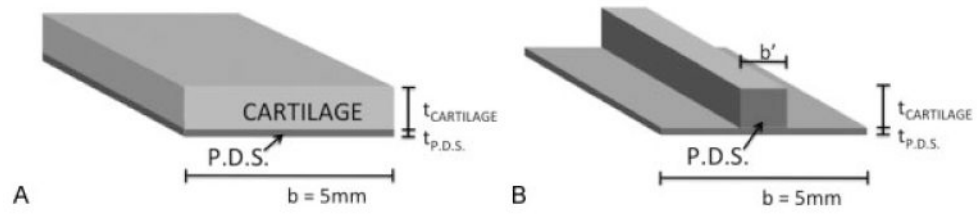


Fig. 2. (A) Cartilage-PDS compound graft modeled as a composite beam with two materials. (B) Homogenous beam made of just PDS with transformed cross-sectional geometry that yields a bending stiffness equivalent to the original composite beam. PDS, polydioxanone.

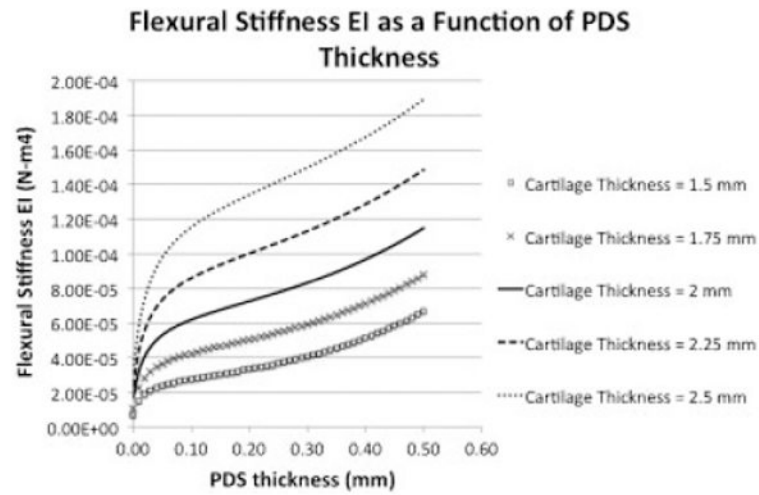


Fig. 3. Parametric analysis of flexural stiffness of the composite beam for varying thicknesses of PDS. Commercially available PDS thicknesses: 0.15 mm (perforated), 0.25 mm (unperforated), and 0.5 mm (unperforated). PDS, polydioxanone.

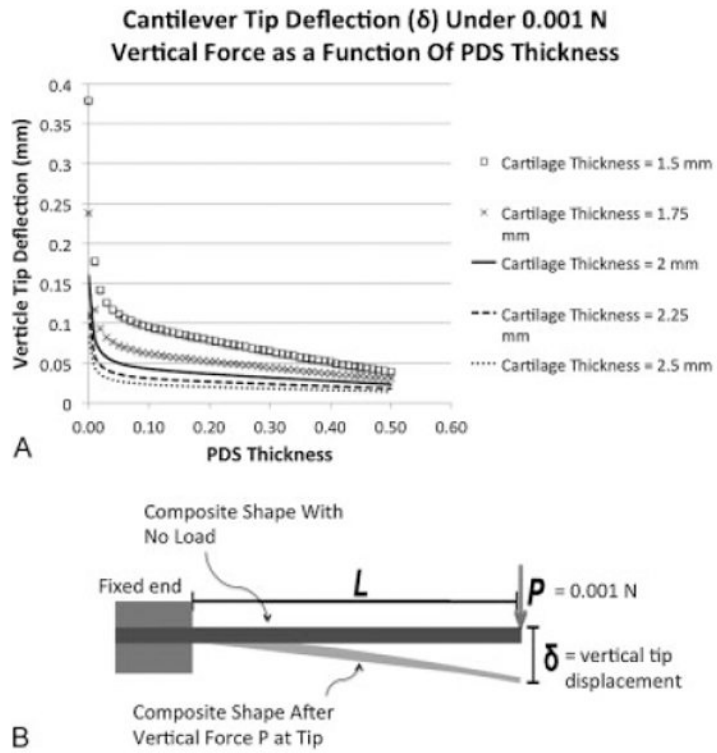


Fig. 4. (A) Parametric analysis of cantilever tip deflection for the composite beam with varying thicknesses of PDS. (B) Model of cantilever bending. PDS, polydioxanone.

Table 1
Material properties and constants used for the composite model

Elastic modulus of cartilage, $E_{\text{cartilage}}$	5 MPa
Elastic modulus of PDS, E_{PDS}	450 MPa
Composite width, b	5 mm

Abbreviation: PDS, polydioxanone.

Table 2
Theoretical flexural stiffness (EI) values for cartilage-PDS composite beams with 2-mm-thick cartilage and various PDS thicknesses

PDS thickness, t_{PDS} (mm)	Flexural stiffness, EI, of composite (N-m ²)	% Increase in flexural stiffness
0	1.67 E-05	–
0.15	6.76 E-05	305.75
0.25	7.77 E-05	14.92
0.5	1.15 E-04	47.86

Abbreviation: PDS, polydioxanone.