METHODS:  
and can employ the usual assumption of pure bending (elastic modulus was set to be 3MPa for SL in the direction perpendicular to the entire lateral surface (with the normal vectors parallel to x or y axis) to describe the observed intrinsic physical parameters. A triphasic, orthotropic model has been built to consider all these factors. Therefore, the objectives of this study were: 1) to develop a triphasic, orthotropic model to describe the observed mechanical and chemical properties of different layers; 2) to compare the predicted curvatures of thin 3-layer cartilage strips with previous experimental data [1]; and 3) to analyze the contributions of various intrinsic physical parameters.

RESULTS AND DISCUSSION:  
INTRODUCTION:  
An articular cartilage sample will curl and warp toward the articular surface after removal from the subchondral bone and the curvature is found to vary with the saline concentration ([Na+]) in the external solution [1]. It has been suggested that swelling effects and the layering inhomogeneities inside articular cartilage offer the potential motive forces for the observed curling behavior. The swelling effects are related to the highly negative charged glycosaminoglycans (GAGs) associated with the proteoglycan (PG) molecules of the cartilage solid matrix. This fixed negative charge density (FCD) generates a Donnan osmotic pressure inside the tissue. The inhomogeneities include non-uniform distribution of chemical contents, as well as the variation of mechanical properties and collagen fibril orientation with depth. The fibrils are aligned tangential to the surface in the superficial layer (SL), randomly in the middle zone (ML), and vertically in the deep layer (DL) (Fig 1). Until now, no theoretical model has been built to consider all these factors. 

The triphasic theory [3] was used to model the swelling behavior (A→B) of each layer. The total stress (σ) consists of elastic stresses inside the solid matrix (σs) and the osmotic pressure (p) as shown below:  

\[ \sigma = -p I + \sigma_s \]  

where \( p = \frac{\Delta \Pi}{V} \) and \( \Pi = \frac{\Delta \Pi}{V} \). Here \( \Pi \) is the universal gas constant, \( T \) is absolute temperature, \( \phi \) is the total ion concentration, \( \epsilon \) is the dilatation, and \( c^+ \) and \( c^- \) are the porosity and FCD at HRS, respectively. An orthotropic constitutive equation with tension-compression nonlinearity was used to describe the mechanical property of the solid matrix based on the collagen fibril direction. The typical parameters adopted are shown in Table 1, in which \( \mu \) is shear modulus of the solid matrix, and \( \lambda \) and \( \lambda_c \) are the elastic moduli at tension and compression, respectively. In particular, the elastic modulus was set to be 3MPa for SL in the direction perpendicular to the split planes [6]. 

Since the cartilage strip is assumed to be very thin, i.e., \( h \ll a \) and \( h \ll c \), the model can be based on a classical lamination theory (CLT), and can employ the usual assumption of pure bending (B→C) [7]. Over the entire lateral surface (with the normal vectors parallel to x or y axis) (Fig 1), the imposed stress resultant and bending moment must be zero. With these conditions, by virtue of the assumed linear variation of in-plane strains with depth, this problem is ready to be solved with simple linear algebra.

**Fig 1** The layer structure of articular cartilage and the predominant orientation of collagen fibrils

**Table 1.** Intrinsic parameters for cartilage, based on previous literatures [6, 8].

<table>
<thead>
<tr>
<th>( p_c )</th>
<th>( \phi^+ )</th>
<th>( \phi^- )</th>
<th>( c^+ )</th>
<th>( c^- )</th>
<th>( \lambda )</th>
<th>( \lambda_c )</th>
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<tr>
<td>SL</td>
<td>0.7</td>
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<td>0.12</td>
<td>0.2</td>
<td>0.4</td>
<td>19</td>
</tr>
<tr>
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<td>0.8</td>
<td>0.25</td>
<td>0.2</td>
<td>0.4</td>
<td>Isotropic</td>
</tr>
<tr>
<td>DL</td>
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<td>0.6</td>
<td>0.20</td>
<td>0.1</td>
<td>0.15</td>
<td>3.0</td>
</tr>
</tbody>
</table>

**RESULTS AND DISCUSSION:**  
**Fig 3** shows the curvature variation in the directions both parallel and perpendicular to split lines as a function of \( c^+ \) based on the parameters listed in Table 1. The curvature increases with the decrease of \( c^+ \), which agrees quantitatively with the previous experimental data by Setton et al [1]. It was also found that the curvature is always larger in the stiffer split line direction [6].

**Fig 2** shows the origin of curling behavior of articular cartilage. Three layers have different swelling potentials, which lead to curling behavior of cartilage strips. Therefore, the objectives of this study were: 1) to develop a triphasic, orthotropic model to describe the observed mechanical and chemical properties of different layers; 2) to compare the predicted curvatures of thin 3-layer cartilage strips with previous experimental data [1]; and 3) to analyze the contributions of various intrinsic physical parameters.

In particular, the elastic modulus was set to be 3MPa for SL in the direction perpendicular to the split planes [6].

Since the cartilage strip is assumed to be very thin, i.e., \( h \ll a \) and \( h \ll c \), the model can be based on a classical lamination theory (CLT), and can employ the usual assumption of pure bending (B→C) [7]. Over the entire lateral surface (with the normal vectors parallel to x or y axis) (Fig 1), the imposed stress resultant and bending moment must be zero. With these conditions, by virtue of the assumed linear variation of in-plane strains with depth, this problem is ready to be solved with simple linear algebra.