Modeling costal cartilage using local material properties with consideration for gross heterogeneities

Jason L. Forman a,b,*, Richard W. Kent a

a University of Virginia, Center for Applied Biomechanics, USA
b European Center for Injury Prevention, University of Navarra, School of Medicine, Irunlarrea 1 (ed. Los Castaños s230), 31008 Pamplona, Navarra, Spain

ARTICLE INFO

Article history:
Accepted 29 November 2010

Keywords:
Costal cartilage
Computer model
Indentation
Impact
Biomechanics

ABSTRACT

Contemporary computer models of the thorax designed to predict injury in automobile collisions model the costal cartilage as a homogeneous material using properties derived from local material characterization tests. No studies have validated the accuracy of these models in predicting the structural mechanics of costal cartilage. Two heterogeneities – the perichondrium and calcified regions – may affect the behavior of costal cartilage in a manner not accounted for by current models. This study sought to investigate the predictive ability of subject-specific models of whole costal cartilage segments, with the calcified regions modeled distinctly and with the perichondrium removed (from the physical specimens as well as from the simulations). Finite element models were developed in the case of five cadaveric costal cartilage segments. The properties of the cartilage were derived from indentation testing of each specimen, where the characteristic average instantaneous elastic moduli ranged from 8.7 to 12.6 MPa. Matched simulations and experiments were then performed, subjecting each specimen to cantilever-like loading with a dynamic posterior displacement of the sternal boundary (all other boundary degrees-of-freedom fixed). The models predicted the resulting peak anterior–posterior forces generated on the costal boundary with a minimum error of 1% and a maximum error of 36%. These results provide support to the previous implicit assumption that insight can be gained into the structural behavior of costal cartilage by observing the local material properties (when calcified regions are included and the perichondrium is removed). Future work includes the addition of the perichondrium, so as to model the whole costal cartilage composite structure.

1. Introduction

Prediction of automobile collision injuries with a finite element model requires an accurate reproduction of not only the individual tissues, but also of their interactions. In the study of thoracic injury prediction, considerable research has been focused on the study of mechanics of osseous components of the ribs and the sternum (e.g., Charpail et al., 2005; Li et al., 2010). Very little is known, however, of the cartilaginous components that connect them.

The costal cartilage consists of irregular cylinders of hyaline cartilage that connect the anterior ends of ribs 1–4 to the sternum and that connect ribs 5–10 to each other. In situations where loading is applied over a limited area of the anterior chest, the costal cartilage dictates the coupling of the ribcage components, affecting the propagation of stress and strain to the ribs. This dictates the overall stiffness and deformation of the ribcage and the probability of injury to each of the ribcage components (Murakami et al., 2006; Oyen et al., 2005).

Many current thorax models include homogeneous representations of costal cartilage, with material property definitions derived from local material characterizations (Lee and Yang, 2001; Iwamoto et al., 2002; Kimpara et al., 2006; Wang, 1995). There are, however, at least two types of heterogeneities in costal cartilage that may cause models developed with local material properties to erroneously predict its overall structural behavior. First, what is commonly termed the “costal cartilage” is actually a composite structure composed of an inner cylinder of hyaline cartilage surrounded by a thick layer of fibrous perichondrium. In a series of structural tests with the perichondrium intact and with the perichondrium removed, Forman et al. (2010) demonstrated that the perichondrium contributes an average of 50% of the structural stiffness of the costal cartilage in certain loading modes.

Second, costal cartilage’s mid-substance often undergoes a progressive calcification with age (Fig. 1; Teale et al., 1989; Rejtarova et al., 2004; Bahrami et al., 2001; McCormick, 1980; Linn and Sokoloff, 1965; Dearden et al., 1974; McCormick, 1980). Calcification has the potential to drastically change the local material properties of costal cartilage’s mid-substance, from a modulus in the order of 1–10 MPa in
healthy hyaline cartilage to a potential modulus in the order of 1–10 GPa (Lau et al., 2008). Dependent on the magnitude as well as on the pattern, calcification has the potential to affect the structural behavior of costal cartilage in a manner that cannot be predicted by the modeling of costal cartilage homogeneously with the material properties of healthy hyaline cartilage. This study sought to take the first step towards the investigation of the relationship between the local material properties and the overall structural behavior of whole costal cartilage segments. Specifically, the aim of this study was to determine if the structural behavior of individual costal cartilage segments (with the perichondrium removed) could be predicted with subject-specific finite-element models that incorporated both cartilage and calcification properties.

2. Methods

2.1. Specimens

Costal cartilage segments were harvested from the 4th ribs of human cadavers (Table 1). The 4th rib cartilage was chosen because it was relatively long (unlike the cartilages of ribs 6–10). The specimens consisted of a portion of the sternum, the entire length of the cartilage, and approximately 3 cm of attached rib. The perichondrium was removed, taking care to not damage the underlying cartilage. Prior to the collection of these specimens, the source cadavers were screened for HIV and hepatitis A, B, and C. All cadaver handling and test procedures were approved by the University of Virginia institutional review board.

<table>
<thead>
<tr>
<th>Specimen ID#</th>
<th>Location (rib#)</th>
<th>Age</th>
<th>Gender</th>
<th>#S</th>
<th>#T</th>
<th>$E_0$(MPa)</th>
</tr>
</thead>
<tbody>
<tr>
<td>403</td>
<td>4L</td>
<td>47</td>
<td>M</td>
<td>4</td>
<td>20</td>
<td>11.0</td>
</tr>
<tr>
<td>322</td>
<td>4L</td>
<td>40</td>
<td>M</td>
<td>4</td>
<td>19</td>
<td>10.5</td>
</tr>
<tr>
<td>400</td>
<td>4L</td>
<td>53</td>
<td>M</td>
<td>3</td>
<td>14</td>
<td>12.6</td>
</tr>
<tr>
<td>187</td>
<td>4R</td>
<td>54</td>
<td>M</td>
<td>4</td>
<td>16</td>
<td>9.5</td>
</tr>
<tr>
<td>197</td>
<td>4R</td>
<td>57</td>
<td>F</td>
<td>3</td>
<td>14</td>
<td>8.7</td>
</tr>
</tbody>
</table>

* #S is the total number of cross-sectional slices tested in indentation with that specimen and #T is the total number of indentation tests of that specimen.

* Characteristic average instantaneous elastic modulus derived from the linear visco-elastic models of the indentation tests with each specimen. $E_0$ used as the pseudo-elastic modulus in the subsequent finite element models.

### Table 1
Specimen information, test matrix, and characteristic average moduli.

Finite element models were developed from CT scans of each costal cartilage specimen. The resolution of the CT scans was 0.625 mm/pixel in-plane, with slices at 0.6 mm intervals. Contours describing the axial-circumference geometry of the specimens were digitized from the CT scans using WinSurf (SURFdrive Software, Kailua, Hawaii) 3-D digitization software. These contours were converted into a 3-D mesh of eight-node solid elements using custom software developed in MatLab (Version 7.1, The Mathworks, Inc.). For each specimen, the mesh was segmented into rib, cartilage, sternum, and calcification sections (Fig. 2). All simulations were performed with the dynamic finite element solver of LS-DYNA (version 970, Livermore Software Technology Corporation, Livermore, CA).

Fully integrated solid elements were used to eliminate hourglassing. A preliminary mesh sensitivity study was performed, where two models with different mesh densities (1.5 and 1 mm nominal element sizes) were constructed for each of the two specimens (187-4R and 403-4L). The 1 mm models each contained approximately 2.8 times the number of elements of their 1.5 mm counterparts. This increase in mesh density did not appreciably change the force–deflection responses targeted in this study (less than 4% change in peak anterior–posterior, x-axis force; 10% or less change in peak lateral, y-axis force). Because these models were observed to be insensitive to mesh density within this range, the 1.5 mm element size was deemed to be sufficient and was used to model the remainder of the specimens.

The cartilage substance was modeled as an isotropic, homogeneous, linear pseudo-elastic material, where the term pseudo-elastic is used here to indicate that the material models were developed with regard to a specific range of loading rates (Fung et al., 1979). In the finite element code, the cartilage was represented as a simple homogeneous, linear elastic material. The pseudo-elastic moduli of the cartilage of each specimen were determined from the indentation tests described below. All osseous regions (bone and calcified regions) were also modeled as homogeneous, isotropic, and linear elastic materials. An elastic modulus of 1 GPa...
Each of these specimens was potted in a plastic dish with Fast Cast (approximately 6 mm thick) by sectioning with a low-speed, diamond blade saw. The specimens were separated into 3–4 cross-sectional test specimens each (each spherical indentation material characterization tests (Lau et al., 2008). The cartilage of each specimen was soaked in a bath of physiologic saline at body temperature for approximately 30 min. Each specimen was then placed on the testing stage of a custom-built indentation machine. At initiation of testing, the indenter tip (2 mm diameter sphere) moved at approximately 0.5 mm/s (attaining this velocity prior to contact with specimen, so as to reduce inertial effects) till penetration into the cartilage to a depth of approximately 0.25 mm (approximately 4% of the 6 mm sample thickness). At this small relative depth of penetration, the tests can be regarded (and analyzed) to be an indentation into an infinite half-space (Oliver and Pharr, 1992). Previous studies have also demonstrated that cartilage visco-elastic material behavior is approximately linear under indentation deformations of this magnitude (Mattice et al., 2006).

The force generated by the indentation was measured by a 22.2 N uniaxial load cell (Honeywell Sensotec) mounted between the actuator of the indenter and the indenter tip; displacement was measured by means of the control instruments located within the indenter actuator. Data were sampled at 100 Hz. Each cross-section was indented at up to five locations. Indentation locations were chosen to avoid engaging calcified regions visible on the surface of the cross-sections.

The force–displacement results of the indentation tests were then used to calculate an instantaneous (short-time) elastic modulus for each specimen. First, obvious, presumed outliers were identified in the force–response datasets of each specimen and were removed from the analysis. Then a “characteristic average” force–response was calculated for each specimen using the method developed by Lessley et al. (2004). Using this characteristic average response, instantaneous elastic moduli were then determined using the analytical technique described by Lau et al. (2008). This analytical solution (Oyen, 2005; Matrice et al., 2006) assumes material linearity, isotropy, and homogeneity to determine a time-dependent (visco-elastic) description of the shear modulus of the material using a Hertzian contact solution. Assuming incompressibility, this approach yields an instantaneous (short-time) elastic modulus combined with a relaxation function. The pseudo-elastic cartilage moduli were then defined as the instantaneous elastic moduli derived from the characteristic average visco-elastic models of each specimen. The full analytical method is described by Lau et al. (2008) and Oyen et al. (2005).

2.4. Structural tests

Structural tests were performed on each costal cartilage specimen to evaluate the ability of the finite element models to predict their overall structural behavior. These were all performed prior to indentation testing (due to the destructive nature of those tests).

The goal of these tests was to study the structural behavior of the specimens under a simplified loading scenario that may occur as a result of chest loading in an automobile collision. The boundary conditions chosen for this study were the same as those used by Forman et al. (2010), and the impetus for these boundary conditions is described in detail therein. The whole cartilage segments were tested by displacing the sternal boundary posteriorly (in the x-direction), holding all other boundary degrees of freedom fixed (Fig. 4). Following the approximation of Forman et al. (2010), posterior displacement of the sternum was applied in the form of a 400 mm/s ramp. The maximum displacement was set to 15 mm, although subsequent analyses were conducted only up to the time of the first observance of failure in the specimen (as noted by examination of high-speed video).

The specimens were first collected from the ribcages of human cadavers and were cleaned of all superficial tissue (including the perichondrium). The capsular ligaments of the sterno-costal joints were left intact to preserve the integrity of the joint. The sternum and rib ends of the segments were potted in blocks of Fast Cast® (Goldenwest Manufacturing, Inc.) casting resin. These potting blocks were mounted in a test rig powered by an Instron (model 8874) material test machine. Forces in the anterior–posterior (x) direction those in the lateral (y) directions were measured using a custom, low-range load cell (Model 7931, Robert A. Denton, Inc., Rochester Hills, Michigan) attached to the costal end of the specimens (Fig. 4). All data were digitally sampled at a rate of 10 kHz with a dynamic data acquisition system (DEWE-2010 Series, Dewetron, Graz). All force signals were filtered using the Channel Frequency Class (CFC) 30 described in the Society of Automotive Engineers Standard J211 (1995); all displacement signals were filtered to CFC 1000. All specimens were soaked in body-temperature saline prior to testing.

These boundary conditions were then applied to the finite element models of the cartilage segments (Fig. 5). The model displacements were defined on the basis....
of the displacement time-histories measured in the physical tests. The forces on all the constrained nodes on the costal boundary were summed to determine the total reaction force. The resulting force–displacement behaviors of the models were then compared to the observed force–displacement behaviors of their physical counterparts.

3. Results

3.1. Indentation tests

A total of 83 indentation tests were performed on 18 cross-sectional slices of costal cartilage specimens. The peak forces resulting from the applied 0.25 mm indentations varied from approximately 0.25 to 4 N. The resulting characteristic average instantaneous elastic moduli parameters are shown in Table 1.

3.2. Models and structural experiments

Subject-specific finite element models were developed using the characteristic average instantaneous elastic moduli shown in Table 1. The meshes of these models are shown in Fig. 6. The resulting force–displacement responses of these models are compared to their corresponding experiments in Figs. 7 and 8.

4. Discussion

4.1. Model prediction fit

In three of the five cases, the models predicted the peak x-axis forces within 30% of those observed in the experiments. In all cases,
The peak x-axis model forces were within 36% of the forces observed in the experiments. While an error range of 36% is not ideal, these results are nonetheless encouraging given the simplicity of the finite element and material models used. This error is small compared to the effect of the perichondrium (which increases the stiffness of the total structure by an average of 100%; Forman et al., 2010). This error is also small when compared to the spread of costal cartilage modulus values in current whole-body finite models, which range from 9 (Ruan et al., 2003) to 49 MPa (Kimpara et al., 2006).

Most of the specimens exhibited softening non-linearities in their force–deflection responses, the types of which are usually associated with either sub-critical failure or plastic deformation (specimens 403-4L, 322-4L, 400-4L, and 197-4R). The models were able, however, to reproduce the general characteristics of these response curves, despite the fact that they were built with linear elastic material models. This indicates that those response features are at least partially the result of the geometry (and the geometry of deformation) of the specimens.

The errors observed with the models consistently took the form of an over-prediction of the peak forces generated. This suggests that the average local indentation material properties erred towards being too stiff. This could occur for several reasons. First, although the indentation sites were chosen to avoid visible areas of calcification, it is possible that calcified regions were present undetected below the surface of some indentation sites. Second, it is possible that the cartilage undergoes a nonlinear softening at strains greater than those produced in the indentation tests. This could be caused by a material plasticity, or it may be the result of micro-structural damage (for example, damage to the collagen fibril matrix in the cartilage). It may be possible to improve the prediction accuracy using a plastic or micro-structural damage model for the cartilage (e.g., Wren and Carter, 1998); however, care should be taken to ensure that such models reflect the actual material behavior and not some other unknown phenomenon.

The models predicted the y-axis force–displacement responses of the specimens less consistently than the x-axis responses. The y-axis responses appeared to match the experimental data very well for two
of the specimens (197-4R and 403-4L), but not very well for the other specimens. This may be indicative of material anisotropy or compressive instability in some of the specimens and may represent the limitation of representing the cartilage as a linearly elastic material.

4.2. Practical implications

Many investigations have studied the local material properties of costal cartilage (Feng et al., 2001; Guo et al., 2007; Roy et al., 2004; Abrahams and Duggan, 1964; Mattice et al., 2006; Lau et al., 2008). Up till now, there have been no validating data to indicate how these local material properties affect the structural function of costal cartilage. By showing that local material properties can be used to predict structural function, this study provides support to the previous implicit assumption that by studying local material properties we can gain an insight into the overall mechanical behavior of costal cartilage. In the future, the effects of changes in cartilage properties or those in calcification geometries could be studied using relatively simple finite element models such as those developed here.

4.3. Future work

The perichondrium contributes a substantial portion of the stiffness of the whole costal cartilage structure (Forman et al., 2010). Any effort to model the complete costal cartilage structure (e.g., in a whole-body model) must account for the effects of the perichondrium in some manner. Future work could include addition of the perichondrium to detailed models such as those developed here to study the mechanics of the complete composite structure of costal cartilage. Future work could also include study of the specific effects that calcified regions have on the structural behavior of the cartilage, including the effects of calcification growth with age.

Finally, future work also should include investigation into how the material behavior of costal cartilage, and the structural behavior of individual cartilage segments, affects the in situ or in vivo mechanical behavior of the ribcage as a whole. Previous computational studies have indicated that costal cartilage may play a significant role in the distribution of stress and strain throughout the ribcage when an external load is applied (Murakami et al., 2006; Oyen et al., 2005) due to its role as a mechanical coupler. The current study represents the first step towards the understanding of a link between the material properties of costal cartilage and the structural behavior of costal cartilage segments. The next step is to investigate how the behavior of costal cartilage (combined with the perichondrium) affects the structural behavior of the chest as a whole, possibly through in situ biomechanics experiments with cadaveric tissue.

5. Conclusion

Five subject-specific costal cartilage models (with the perichondrium removed) were developed and studied in cantilever-like loading with a dynamic posterior displacement of the sternum. The material properties of the cartilage in the models were derived from local indentation tests—the characteristic average instantaneous elastic moduli of these specimens ranged from 8.7 to 12.6 MPa. The resulting models predicted the peak x-axis forces generated on the costal border during cantilever-like loading with a minimum error of approximately 1% and a maximum error of 36%. The models also predicted x-axis force-versus-displacement response curves similar in shape to their experimental counterparts. These results suggest that when calcified regions are included in the models (and the perichondrium is removed from

Fig. 8. Plots (continued) of the force–displacement responses of the models developed with their perichondrium removed compared to their respective structural tests. The data are shown until the time of the first signs of failure of the physical specimens (as determined from high-speed video).
the tests), models of costal cartilage developed with subject-specific material properties can reproduce the x-axis structural behavior of costal cartilage reasonably well (under the type of loading studied here). Topics for future study include the investigation of methods to model the effect of the perichondrium and the investigation of different loading modes and rates.

Conflict of interest statement
None of the authors have any conflicts of interest to report.

Acknowledgements
All experiments and computer modeling were performed at the University of Virginia Center for Applied Biomechanics. This manuscript was prepared with support from a Whitaker International Scholars Grant. All analyses and opinions expressed here are solely those of the authors.

References