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Characterization of craniofacial sutures using the finite element method

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ABSTRACT

Characterizing the biomechanical behavior of sutures in the human craniofacial skeleton (CFS) is essential to understand the global impact of these articulations on load transmission, but is challenging due to the complexity of their interdigitated morphology, the multidirectional loading they are exposed to and the lack of well-defined suture material properties. This study aimed to quantify the impact of morphological features, direction of loading and suture material properties on the mechanical behavior of sutures and surrounding bone in the CFS. Thirty-six idealized finite element (FE) models were developed. One additional specimen-specific FE model was developed based on the morphology obtained from a μCT scan to represent the morphological complexity inherent in CFS sutures. Outcome variables of strain energy (SE) and von Mises stress (σ_{vm}) were evaluated to characterize the sutures' biomechanical behavior. Loading direction was found to impact the relationship between SE and interdigitation index and yielded varied patterns of σ_{vm} in both the suture and surrounding bone. Adding bone connectivity reduced suture strain energy and altered the σ_{vm} distribution. Incorporating transversely isotropic material properties was found to reduce SE, but had little impact on stress patterns. High-resolution μCT scanning of the suture revealed a complex morphology with areas of high and low interdigitations. The specimen specific suture model results were reflective of SE absorption and σ_{vm} distribution patterns consistent with the simplified FE results. Suture mechanical behavior is impacted by morphologic factors (interdigitation and connectivity), which may be optimized for regional loading within the CFS.

1. Introduction

Sutures are articulations (fibrous joints) present in the craniofacial skeleton (CFS). The bones that make up the suture are generally of intramembranous origin and grow by ossification at the suture margin (Rice, 2008). Sutures function to hold the bones of the skull together while allowing for mechanical stress transmission and deformation (i.e. distortion during childbirth, cyclic loading from muscle activity, forces from therapeutic mechanical devices and traumatic impacts) (Mao et al., 2003). The primary function of CFS sutures changes with age. In postnatal stages and early development, sutures provide high flexibility to allow for enlargement of the head around the developing organs. Calvarial sutures undergo most of their growth during these early stages of development. In contrast, facial sutures are most active during adolescence. In adulthood, sutures are believed to function primarily as shock absorbers to dissipate stresses transmitted through the skull (Buckland-Wright, 1978; Byron et al., 2004; Herring and Teng, 2000; Jaslow, 1990; Jaslow and Biewener, 1995; Pritchard et al., 1956; Rafferty et al., 2003; Rayfield, 2004, 2005a).

Sutures can be classified into three types based on morphology: (1) flat or butt-ended sutures, (2) overlapping sutures with rough or ridged contact surfaces and (3) interdigitating sutures with interlocking bony processes. Suture morphology changes from a simple butt joint in early life (which must stay patent to function) to a joint with differing degrees of interdigitation and interlocking projections in adulthood (Rice, 2008). Under normal conditions, sutures in the human skeleton have been reported to be fully fused by late adulthood (Rice, 2008). However, micro-computed tomography (μCT) imaging has shown that sutures remain partially open beyond the seventh decade, with the degree of connectivity varying in different CFS sutures (Maloul et al., 2010). Thus, suture morphology cannot be limited to the shape of the adjacent bones

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but must include the degree of bony bridging across the suture gap.

The human CFS is subjected to three main types of loading. Quasi-static tensile loading due to the growth of internal organs occurs mainly during the first two decades of life. Cyclic loading applied through muscle contraction and bite force during mastication that can result in both compressive and tensile forces (Herring and Teng, 2000; Szwedowski et al., 2010a,b; Szwedowski, 2007). Impact loading can also occur at high magnitudes and rates of loading (i.e. due to falls, fighting, foreign objects, motor vehicle accidents, etc.). The sutures participate in the mechanical response of the skull to loading by modulating the load transmitted between adjacent bones – this can include direct load transmission, reorientation of loading and damping (Behrents et al., 1978; Buckland-Wright, 1978; Herring et al., 1996; Hylander et al., 1987; Rayfield et al., 2001; Rayfield, 2004). Large variations in suture morphologies (shape, complexity and stiffness) and fiber arrangements, have led to speculation that sutures may adapt to their environment (Herring, 2008). Studies have shown that suture morphology can be linked to compression, tensile or bending loading (Herring and Teng, 2000; Herring et al., 2001; Rafferty and Herring, 1999), but controversy remains as to how suture morphology impacts the global biomechanics of the adult CFS during loading.

Experimental strain gauge studies and computational modeling (finite element (FE) and multibody dynamic analyses) have been used to evaluate the mechanical behavior of sutures in the CFS (Behrents et al., 1978; Herring and Teng, 2000; Popowics and Herring, 2007; Rayfield, 2005b; Wang et al., 2010). FE analyses have been successful in developing a better understanding of the biomechanics of the human skeleton, characterizing stresses and strain patterns under physiological or non-physiological loadings (Chalk et al., 2011; Curtis et al., 2008; Dumont et al., 2005; Ross et al., 2005; Szwedowski et al., 2010a; Wang et al., 2010). In spite of efforts to understand the global functional and mechanical properties of the CFS, little attention has been directed at understanding the mechanical consequences of variation in suture morphology (i.e. number of interdigitations and connectivity) and direction of loading (i.e. perpendicular or parallel to the suture) at the individual suture level (Behrents et al., 1978; Herring and Mucci, 1991; Herring and Teng, 2000; Jaslow and Biewener, 1995; Rafferty and Herring, 1999; Rafferty et al., 2000, 2003; Smith and Hylander, 1985; Sun et al., 2004). For example, previous investigation on the mechanical behavior of individual sutures has focused on parallel loading alone (Borke et al., 2003; Jasinoski et al., 2010; Yu et al., 2004; Zhang et al., 2002). This study aims to investigate how morphological features (number of interdigitations and bony connectivity), direction of loading (parallel, perpendicular and pressure loading), and suture material properties (isotropic or transversely isotropic) influence the mechanical behavior of the suture and surrounding bone. It is hypothesized that the design of a single suture is adapted for regional functional specialization within the CFS.

2. Materials and methods

Idealized FE models of the bone–suture–bone complex were developed in Abaqus CAE (Simulia, USA) using 3D tetrahedral elements ($\approx$300 K elements). The dimensions of the full complex were 9 mm $\times$ 20 mm $\times$ 3 mm with a 0.2 mm suture width. Differing suture morphologies were modeled by varying the number of interdigitations. The interdigitation index (I.I.), defined as the ratio between the entire length of the suture divided by the straight distance between the suture ends (Rafferty and Herring, 1999), was modeled as low (I.I.=2.3), moderate (I.I.=3.4) and complex (I.I.=4.3). To account for connectivity in the suture gap, the bone surfaces were bridged with bony connections that represented 17% connectivity for each of the three interdigitations indexes (Maloul et al., 2010).

In all FE models the bone was treated as an isotropic material (Young's modulus E=6 GPa, Poisson's ratio ν=0.27). Young's modulus and Poisson's ratio for cranial and facial sutures were assigned based on average values determined by Jasinoski et al. with isotropic material properties (E=50 MPa, ν=0.30) or transverse isotropic material properties (E_1=80 MPa (aligned with the direction of the fibers), E_2=20 MPa, E_3=20 MPa, ν_{12}=0.4, ν_{13}=0.1, ν_{23}=0.4, G_{12}=20 MPa, G_{13}=26.6 MPa and G_{23}=20 MPa) (Jasinoski et al., 2010). Two bone–suture complexes (coronal and zygomaticotemporal) were excised from a fresh frozen human CFS (85 years of age) obtained from the Department of Anatomy at the University of Toronto, decalcified in formic acid and embedded in paraffin wax. The sutures were sectioned in the sagittal plane and stained using Hematoxylin and Eosin. The orientation used in the transversely isotropic models was determined based on the histological sections, which demonstrated fibers arranged in straight lines between adjacent surfaces at a 40° offset angle (Figs. 1 and 2).

A section of the coronal suture was excised from a fresh frozen cadaver head and μCT scanned at an isotropic voxel size of 14 μm^3 (GE Explore Locus, General

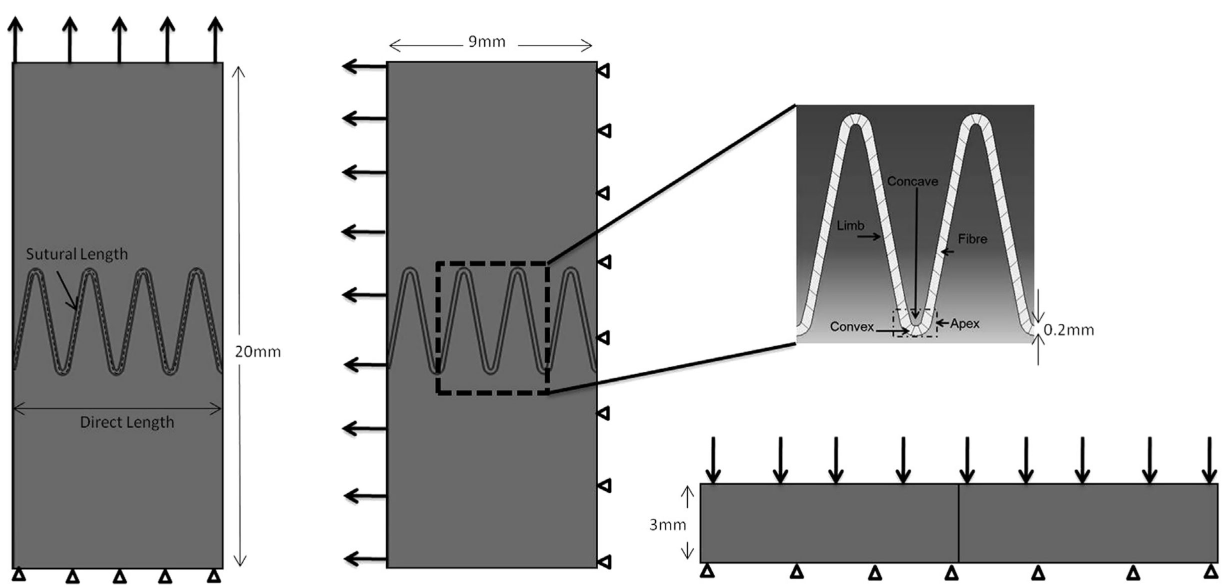

Fig. 1. Three different loading directions simulated in Abaqus/CAE: (A) parallel – a tensile load applied to the upper surface of the bone (parallel to the suture), lower edge constrained; (B) perpendicular – load applied to the left side surface of the complex (perpendicular to the suture), right edge fully constrained in the same direction as the loading and one center point constrained in all directions and (C) pressure – loading applied as a full surface pressure to represent impact loading – with fiber orientation within the limbs and apices of the suture. *Note*: Interdigitation index is determined by measuring the total sutural length, by tracing the entire path length as shown cross hatched in (A), and dividing this length by the shortest distance between the suture ends measured along the surface of the bone.