ANNALS of the ORADEA UNIVERSITY.

Fascicle of Management and Technological Engineering, Volume VII (XVII), 2008

THE BIOMECHANICAL MODELING OF TEMPOROMANDIBULAR JOINT DISC

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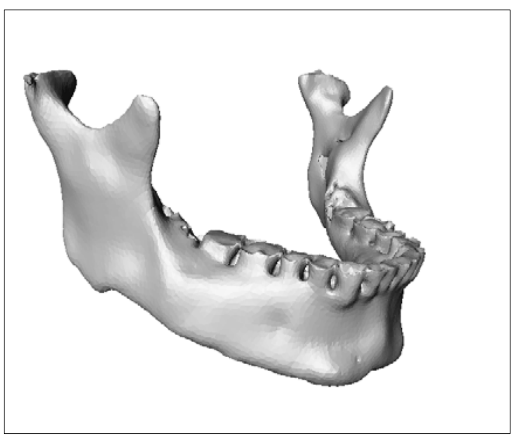
Key words: Stress and strain, viscoelastic, biomedical modeling, elastic modulus

Abstract

The temporo-mandibular joint (TMJ) disc construction results in a viscoelastic response of the disc to loading and enables the disc to play an important role as a stress absorber during function. The viscoelastic properties depend on the direction (tension, compression, and shear) and the type of the applied loading (static and dynamic). The compressive elastic modulus of the disc is smaller than its tensile one because the elasticity of the disc is more dependent on the collagen fibers than on the proteoglycans.

Recent developments in technology and software are providing better data and methods to facilitate research in biomedical modeling and simulation. In the area of segmentation, original methods involved the time-consuming task of manually tracing structures from slice to slice. This process is now possible with significantly less interaction from the user. Programs such as 3D Slicer [6], Mimics (Materialise N.V., Heverlee, Belgium) and Amira 2.3 (TGS) provide semi-automatic image-processing-based segmentation and modeling from CT images. These programs use the Generalized Marching Cubes algorithm to create a polygonal wireframe mesh of the segmented area and export a model in STL or other formats. Using the Amira 2.3 program a 3D polygonal mesh (STL model) was created with the Generalized Marching Cubes Algorithm and decimated for later use (Figure 1).



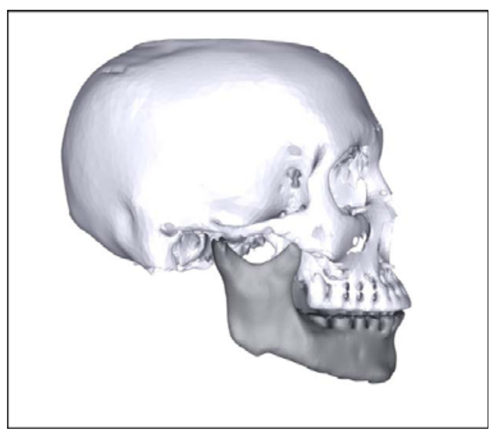


Figure 1: Polygonal 3D STL mesh reconstruction of (a) the mandible and (b) skull using Amira 2.3 software.

The Jaw Motion Analyzer tracks the position in space of three points defining a triangular plane (Figure 2). Left and right condylar points are selected on the patient with a pointer that is attached to the ultrasonic emitter array. The 3D mandibular model is then aligned in space as shown in Figure 2 such that the condylar points coincide with the corresponding points on the triangle and the base center of the JMA T-attachment is at a distance of 5 cm from the jig.

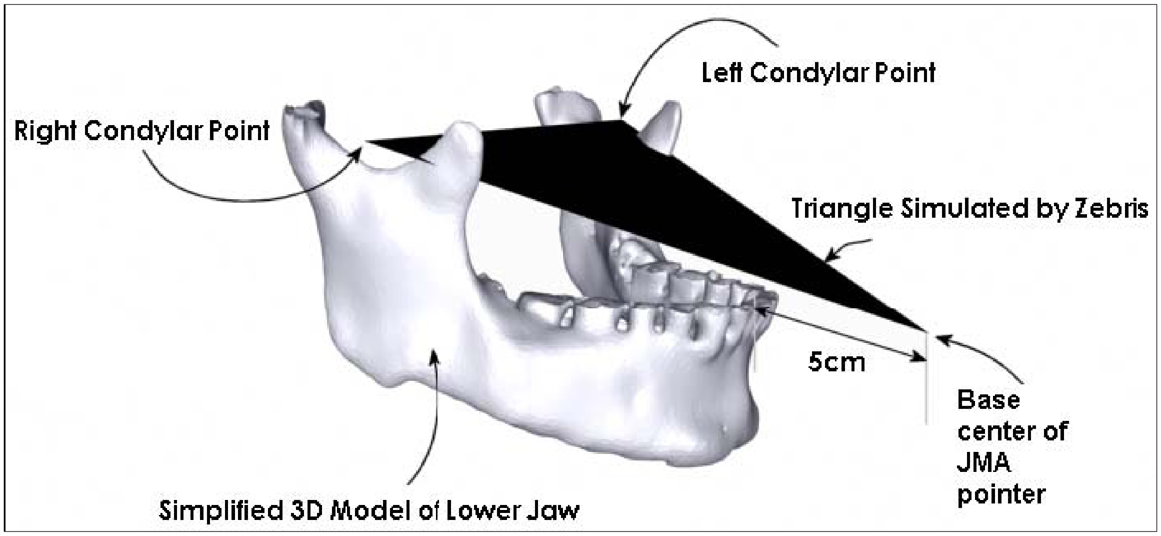


Figure 2: Alignment of the 3D jaw model and the jaw motion tracking data.

The disc is composed of variable amounts of cells and an extracellular matrix. The matrix consists of macromolecules and fluid. The macromolecules constitute about 15-35% of the wet weight of the disc, while the tissue fluid constitutes about 65-85%. These macromolecules consist mainly of collagen (85-90%) and proteoglycans (10-15%). The mechanical properties of the disc are largely dependent on its collagen fiber and proteoglycan composition and organization and on their interaction with the tissue fluid.

The articular surfaces of the TMJ are highly incongruent. Due to this incongruence, the contact areas of the opposing articular surfaces are very small. When joint loading occurs, this may lead to large peak loads, which may cause damage to the cartilage layers on the articular surfaces. The presence of a fibro-cartilaginous disc in the joint is believed to prevent these peak loads, since it is capable of deforming and adapting its shape to that of the articular surfaces. These deformations ensure that loads are absorbed and spread over larger contact areas. In addition, the shape of the disc and the area and location of its contact areas with the articular surfaces change continuously during jaw movement to adapt to the changing geometry of the articular surfaces of the mandible and temporal bone. As a result, there will be a continuous change in the magnitude and location of the deformations that occur. For example, according to the work of [5], when loading occurs in the jaw-closed position, the deformations in the disc are spread throughout the entire intermediate zone, while translation of the condyle in the forward direction to obtain a protrusive or open jaw position leads to a concentration of the deformation in the lateral part of the disc.

The magnitude of the deformation and resulting stress of the disc is primarily determined by the nature of the applied loads and by the biomechanical properties of the disc, such as stiffness and strength. An understanding of these properties is important for several reasons. First, they determine the role of the disc as a stress-distributing and load-absorbing structure [6]. Therefore, the properties of the disc will also influence the stresses and strains that occur in the cartilage layers on the bone surfaces. These stresses and strains are of critical importance for adaptation and wear. For example, mechanical stress affects the proteoglycan synthesis in the disc [7], resulting in an adaptation of stiffness. Second, precise information on the biomechanical properties of the disc is required to develop suitable joint simulation models, with which the distribution of stress and strain in the structures of the joint can be estimated. In the last decade, several three-dimensional finite element models of the joint have been developed [8]. However, thus far, the available models do not include all relevant properties, such as the shock-absorbing capabilities of the disc.

Many studies have been conducted on the elastic properties of the disc since Fontenot's initial investigation (1985). In general, the elastic modulus has been measured on small specimens by means of static tests [8]. However, due to the large interspecies variations and different experimental protocols, the results of the various studies cannot be easily compared. Differences in experimental techniques include the size, hydration fluid and holding of the specimens, and different testing machines and protocols. Due to this variation, the reported moduli show a large range (from 1 to 100 MPa).

Another problem when the results of various studies are compared is that different tests (for example, compression and tension) have not been performed on the same specimen. Also, as mentioned before, the relationship between stress and strain for the disc is non-linear and time-dependent.

The above-mentioned *quasi*-static experiments have provided valuable information on how the behavior of the disc changes over time. With *quasi*-static experimental set-ups, however, only the linear viscoelastic behavior of the disc can be studied. The disc should essentially be approached as a structure with non-linear behavior, and thus its dynamic viscoelastic properties need to be examined, although the mechanisms responsible for stress distribution, energy dissipation, and stress absorption are the same as those for *quasi*-static loading. Therefore, dynamic experiments have recently been performed (compression,[9]; tension, [8]. In general, the dynamic properties are determined in cyclic tests at a physiologic strain range.

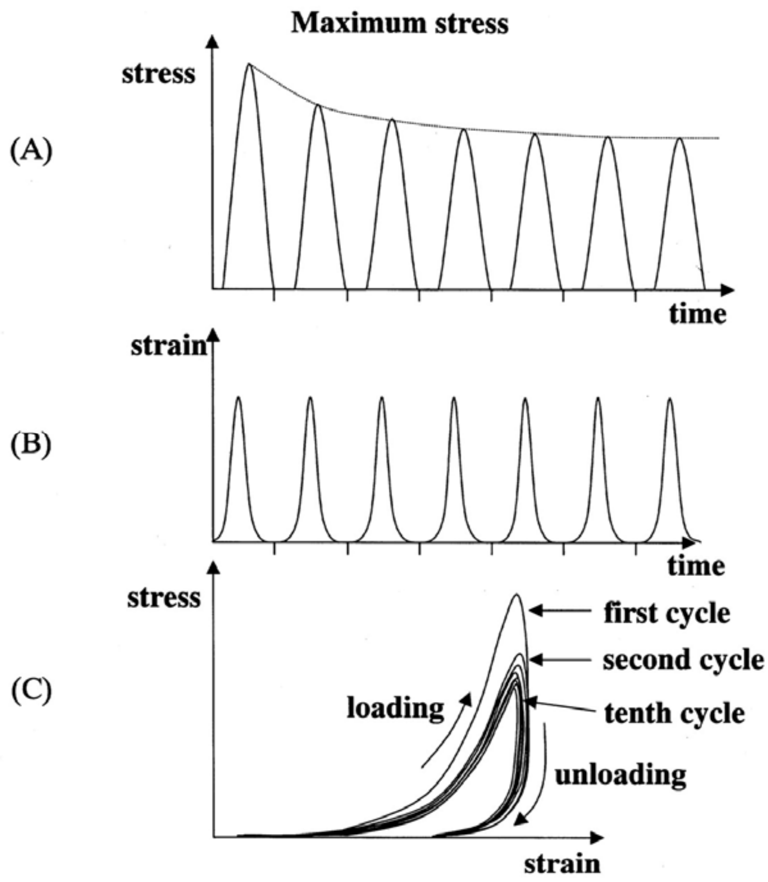


Figure 3. Measurement signals of a cyclic test with a constant frequency. Data from Beek et al. (2001a). (A) Stress vs. time. (B) Strain vs. time. (C) Stress vs. strain.

When a sinusoidal oscillating strain is applied to the disc with a constant amplitude, a small phase shift is present between the strain and stress signals; the stress reaches its maximum earlier than the strain [9]. The phase shift between the strain and stress depends on the difference between the viscous and elastic properties of the disc. In subsequent cycles, the value of the maximum stress decreases asymptotically, and the stress shows an almost steady level after 7 to 10 cycles. Therefore, the hysteresis loop during loading and unloading at a constant rate after 7 to 10 cycles will remain essentially unchanged. The hysteresis loops show that energy is dissipated inside the disc. As a result, dynamic cycle tests enable us to comprehend the viscous and elastic properties of the disc, and its energy dissipation function.

The dynamic properties of the disc are dependent on the frequency and strain of loading. For example, in dynamic compression tests on the human disc, the indentation amplitude (strain: 0.25, 0.30, and 0.35) and frequency (0.02, 0.05, and 0.1 Hz) had a proportional effect on the value of the maximal stress and the amount of energy dissipation [7]. Furthermore, the maximal stress and the energy dissipation are significantly larger in the intermediate zone than in the anterior and posterior bands [7]. Cyclic testing of articular cartilage specimens at a constant strain amplitude and at moderate strain levels showed repeatable stress-strain relationships within 7 to 10 cycles of pre-conditioning. Furthermore, it is known that the curves showing the experimental peak and valley stresses in cyclic testing match well with those obtained by the theoretical load-relaxation curves according to the *quasi*-linear viscoelastic theory. Therefore, the dynamic properties are likely to exhibit smaller values than the static ones, and present some resemblance to the static modulus after load-relaxation rather than the elastic modulus.

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