Biomechanical response of the temporomandibular joint disc complex to mechanical loads

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ABSTRACT:

The temporomandibular joint (TMJ), one of the load-bearing organs in human body, is composed of bone components and soft tissues. Of the soft tissues, the TMJ disc consists mainly of collagen fibers and proteoglycans constrained in the interstices of the collagen network. This construction results in a viscoelastic response 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). For instance, upon dynamic loading the disc is likely to behave less stiff than under static loading because of the difference of fluid flow through and out of the tissue during loading. Furthermore, the retrodiscal tissue adjacent to the TMJ disc has an important joint-stabilizing function during mouth opening although this tissue exhibits less stiffness than the TMJ disc and therefore has no or less function as a stress absorber. Information about the viscoelastic behavior of the soft tissues in the TMJ is required to understand its function, which is a requirement, for instance, to develop a suitable replacement and tissue engineering of the TMJ. In the present paper, the biomechanical properties of the TMJ disc and retrodiscal tissue in response to various loading conditions are discussed.

INTRODUCTION

During mandibular movements, the temporomandibular joint (TMJ) undergoes various loadings as a load-bearing organ (Koolstra and van Eijden, 2005; Tanaka *et al.*, 2004). The modulation of the TMJ loading is dependent on the biomechanical response of the TMJ soft tissues located between the mandibular condyle and glenoid fossa. The articular surfaces of the TMJ are highly incongruent. Due to this incongruency, the contact areas of the opposing articular surfaces are very small. Upon joint loading this may lead to large peak loads, which may cause damage to the cartilage layers on the articular surfaces. The presence of fibrocartilaginous disc is believed to prevent these peak loads (Tanne *et al.*, 1991; Scapino *et al.*, 1996), as it is capable to deform and to adapt its shape to that of the articular surfaces.

The magnitude of the deformation and resulting stress in the disc is primarily determined by the nature of the applied loads and by its biomechanical properties. Therefore, a better understanding of these properties is of critical importance for several reasons. First, they determine the role of the disc as a stress distributing and load absorbing structure (Nickel and McLachlan, 1994; Beek *et al.*, 2001). 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. Finally, information on the biomechanical properties of the disc is important to develop TMJ replacement and tissue engineering of the TMJ.

The retrodiscal tissue of the TMJ contains collagen and elastic fibers, and numerous blood vessels and nerves (Kino *et al.*, 1993). Of the components, collagen fibers are responsible for resistance to tensile forces and for maintaining the shape of the tissue, while elastin may function in the restoration of shape after load removal (Zhu *et al.*, 1994). The retrodiscal tissue consists of a temporal, condylar, and intermediate part (Scapino, 1997). The temporal and condylar part are directly attached to the posterior band of the disc and are supposed to play an important role in controlling the disc position during jaw opening and closing (Scapino, 1997). During jaw opening, the anteroposterior length of the temporal part increases, while the condylar part is folded beneath the posterior band (Scapino, 1997). In contrast, during jaw closing the length of the temporal part decreases, while that of the condylar part increases. This suggests that the temporal and condylar part of the retrodiscal tissue contribute to maintain the position of the disc relative to the condyle during, respectively, jaw opening and closing. Therefore, insight into the viscoelastic properties of the retrodiscal tissue respond to various loading is also important to unravel the possible nature of secondary tissue changes.

In this paper, we provide a thorough review of biomechanical studies of the TMJ disc and retrodiscal tissue. The biomechanical properties of the disc and retrodiscal tissue, including elastic modulus and viscoelasticity, are summarized.

BIOMECHANICAL BEHAVIOR OF THE DISC AND THE RETRODISCAL TISSUE

Elastic constants

The relationship between stress and strain of an elastic material can be described by a stress-strain curve (Figure 1). The curve has both elastic and plastic deformation regions. In the elastic region a toe region and a transition zone can be distinguished (Fung, 1981), which are more or less arbitrarily divided by the so-called critical point (Tanne *et al.*, 1991; Tanaka *et al.*, 2000). In order to evaluate the basic biomechanical characteristic of a tissue, the elastic modulus E is commonly calculated. This modulus is defined as the slope of the elastic region of the stress-strain curve. The tensile and compressive moduli are a measure of the ability of the tissue to resist deformation in the direction of the applied load. The shear modulus G is a measure of the ability of the tissue to resist shear stress in a particular plane.

Tensile modulus

The tensile modulus is mainly dependent on the orientation of collagen fibers, because they can resist tension only in the direction parallel to their orientation. The intermediate zone of the disc consists mainly of anteroposteriorly oriented fibers (Detamore and Athanasiou, 2003). Therefore, the tensile modulus of the intermediate zone is larger in anteroposterior direction than in mediolateral direction (Teng *et al.*, 1991; Beatty *et al.*, 2001). For example, the tensile modulus of the porcine disc was 76.4 MPa in the anteroposterior direction, whereas it was 3.2 MPa in the mediolateral direction (strain rate: 500 mm/min; Beatty *et al.*, 2001). As in the anterior and posterior bands of the disc the collagen fibers run mainly mediolaterally they have a relatively large tensile modulus in mediolateral direction.

With respect to the retrodiscal tissue, the first attempt to investigate the elastic properties of the retrodiscal tissue under tension was reported in 2003 (Tanaka *et al.*, 2003). In the bovine retrodiscal tissue, the stress-strain relationships were also nonlinear in nature and were best represented as a power function ($\sigma = A \varepsilon^{B}$). Then, the non-linear stress-strain relationship could be represented by a bilinear relation of two line segments with a certain critical stress. The obtained critical stress was around 0.5 MPa, and the elastic moduli were 2.08 MPa for the low (0-0.5 MPa) and 4.30 MPa for the high (0.5-2.0 MPa) stress region, respectively. The both moduli are approximately 10-20% of the bovine disc elastic modulus (Tanaka *et al.*, 2001). This implies that the retrodiscal tissue has not sufficient strength to pull the disc back although it can support and control the disc position slightly.

Compressive modulus

The resistance to compression is mainly dependent on the density of proteoglycans, especially of the large chondroitin sulfated proteoglycans. As the distribution of the proteoglycans is different in various regions of the disc, regional differences in its compressive modulus can be expected. In the anterior and posterior bands, and in the central region of the intermediate zone the compressive modulus is higher than in the medial and lateral regions of the intermediate zone (del Pozo *et al.*, 2002). The possible explanation for this regional difference is that the large chondroitin-sulfate proteoglycans and the related chondroitin sulfate are preferentially localized in the central region of the intermediate zone and in the anterior and posterior bands (Mizoguchi *et al.*, 1998).

The retrodiscal tissue is subjected to tension during normal jaw opening (Scapino, 1997). However, this tissue seems also subjected to compression. A previous study suggested that when the condyle is nearing the fully closed position, the condylar part of the retrodiscal tissue may be under compressive load (Scapino, 1997). Furthermore, with an abnormal anterior disc displacement, the retodiscal tissue is subjected to continuous compression (Scapino and Mills, 1997). The compressive modulus of the retrodiscal tissue was 1.54 MPa (Tanaka *et al.*, 2002a), which is almost 1/20 of that in the disc. This indicates that the retrodiscal tissue has no or less function as a stress absorber.

Shear modulus

Investigation of shear properties in synovial joints is of particular interest, because shear stress can result in fatigue, damage and deformation of cartilage (Zhu *et al.*, 1993, 1994). Therefore, data on the shear modulus might contribute to a better understanding of secondary tissue damage. Shear stress is likely to occur during loading of the disc, because the articular surfaces that compress the disc do not run parallel. This causes that not all areas of the disc are deformed in the same direction leading to local shear. Another reason why shear stress occurs in the disc is its inhomogeneous structure. Its inner layer consists mainly of anteroposterior running collagen fibers and the 'leaflet-like' proteoglycans (Nakano and Scott, 1996), whereas the superior and inferior surface layers mainly consist of anteroposteriorly and mediolaterally running collagen fibers and small proteoglycans (Nakano and Scott, 1996). Therefore, these layers are considered to have different biomechanical properties (Nakano and Scott, 1996; Mizoguchi *et al.*, 1998) which might lead to shear stress. This is supported by the results of a finite element study, in which a relative large shear stress was predicted in a disc consisting of three layers (Tanaka *et al.*, 1994).

With respect to the shear modulus of the disc, Lai *et al.* (1998) reported that the shear modulus of the intermediate zone of the human disc (strain rate: 0.02 mm/s) was 1.0-1.75 MPa. Furthermore, it appeared that in the central region the shear modulus (about 1.0 MPa) was lower than in the lateral and medial regions (about 1.75 MPa).

Viscoelastic properties

Quasi-static behavior-stress relaxation

How the viscoelastic properties of the disc change over time during constant loading can be characterized by stress-relaxation testes, creep tests, and restoration tests (Figure 2). The parameters obtained from these tests provide valuable information on the tissue behavior as a function of time and are of great importance for a better understanding of mechanical properties of the disc such as energy dissipation and stress absorption. In a stress-relaxation test, a stepwise deformation with a specific strain level is applied to a specimen and this strain level is kept constant until the stress reaches an almost steady level. From this test, the relaxation time and the relaxed modulus are obtained.

The relaxation time of the disc ranges from 3 to 45 s and the relaxed modulus ranges between approximately 5 and 50% of the initial modulus (Tanaka *et al.*, 1999; del Pozo *et al.*, 2002). This indicates that the movement of fluid out of the disc under loading is relatively slow and not proportional to the fluid pressure. Because of this relatively long relaxation time, the loaded disc becomes relatively stiff when it is cyclically loaded, during for example chewing and speaking (Beek *et al.*, 2001). After stress relaxation, a biomechanical equilibrium will occur eventually which implies a balance between the applied stress and the resistance to this stress in the disc. More than 50% of the initial stress is dissipated. This behavior implies that the disc functions as a stress absorber and a stress distribution material. Without the dissipation of strain energy, storage of excessive strain energy can lead to breakage of the disc and other components of the TMJ (Teng *et al.*, 1991; Nickel and McLachlan, 1994; Tanaka *et al.*, 1999; del Pozo *et al.*, 2002).

The retrodiscal tissue exhibited an almost similar stress relaxation response as the disc. In the case that the stress-strain curve was described by a two-segment piecewise linear function with a threshold stress of 0.5 MPa, the relaxation times τ_ε for the constant strain were 30.1 s in the low stress region and 39.1 s in the high stress region (Tanaka *et al.*, 2003). The stress relaxation ratio was approximately 55% after 5 min relaxation. This indicates that there exists an energy-dissipation mechanism in the retrodiscal tissue. The remaining 50-60% of the tensile stress after stress-relaxation works for tethering the disc during jaw closing. It can be concluded that the retrodiscal tissue has a great capacity for energy-dissipation and resistance to tensile forces, and thus might have a function to maintain the position of the disc relative to the condyle during jaw closing.

Dynamic behavior

The above-mentioned quasi-static experiments have provided valuable information on how the tissue 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 and retrodiscal tissue should essentially be approached as structures with a non-linear behavior and thus their dynamic viscoelastic properties need to be examined, although the mechanisms responsible for stress distribution, energy dissipation and stress absorption are the same as for quasi-static loading.

In dynamic tests (Figure 3), a complex dynamic modulus E* can be determined experimentally by applying a sinusoidal strain (Tanaka *et al.*, 2002b; Tanaka and van Eijden, 2003). It consists of a real part, the storage modulus E', and an imaginary part, the loss modulus E''. E' describes the elastic deformation under stress and is directly proportional to the energy storage in a cycle of deformation. E'' describes the viscous deformation and is proportional to the average dissipation or loss of energy as heat in a cycle of deformation. The values of dynamic moduli ($|E^*|$, E' and E'') increase as the frequency increased from 0.1 to 100 Hz (Tanaka *et al.*, 2002b). In a dynamic tensile test, the dynamic viscoelastic E-moduli were about 2 times larger at 100 Hz than those at 1 Hz (Tanaka *et al.*, 2002b). This non-linear dependency on the frequency is due to fluid flow and squeezing within the matrix of the disc. Below a certain frequency, the fluid flow may match up to the applied frequency. At higher frequencies, the proteoglycans occupying the interfibrilar spaces interfere with smooth fluid flow, which leads to strain energy dissipation, resulting in a higher stiffness.

CONCLUSIONS

With this in mind, a proper understanding of the viscoelastic properties of the disc and retrodiscal tissue under simulated loading conditions can presumably give us a better focus in the selection and development of mechanically compatible synthetic or autogenous substitutes for TMJ replacement and tissue engineering.

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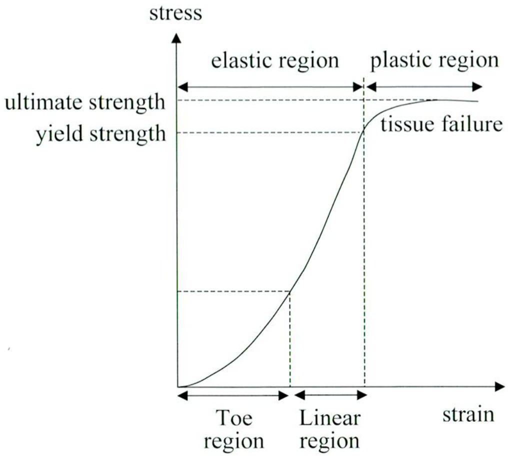

Figure 1 Stress-strain curve for soft tissue.
The curve has both elastic and plastic deformation regions. If the structure is not loaded beyond the elastic region, it will return to its original shape once the load is released. If the structure is loaded up to its plastic region it will not return to its original shape when the load is released. After plastic deformation, the stress will cause permanent damage of the tissue. The elastic region, a toe region and transition zone can be distinguished, which are more or less arbitrarily divided by the so-called critical point.

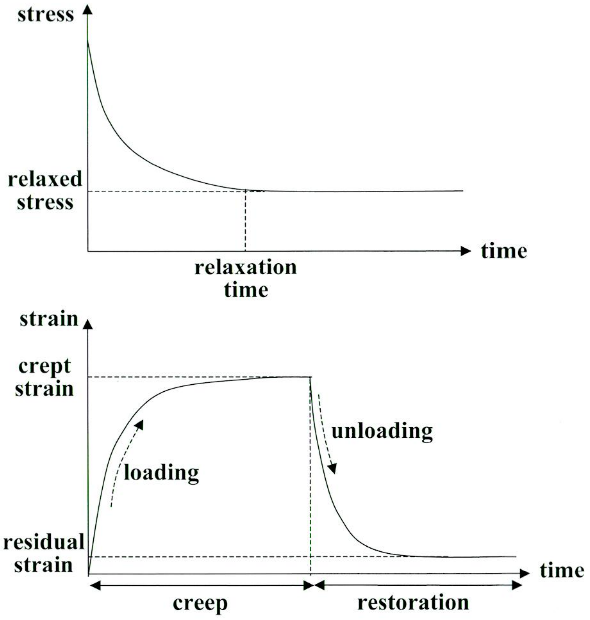

Figure 2 Stress relaxation curve (1) and creep and restoration curve (2).

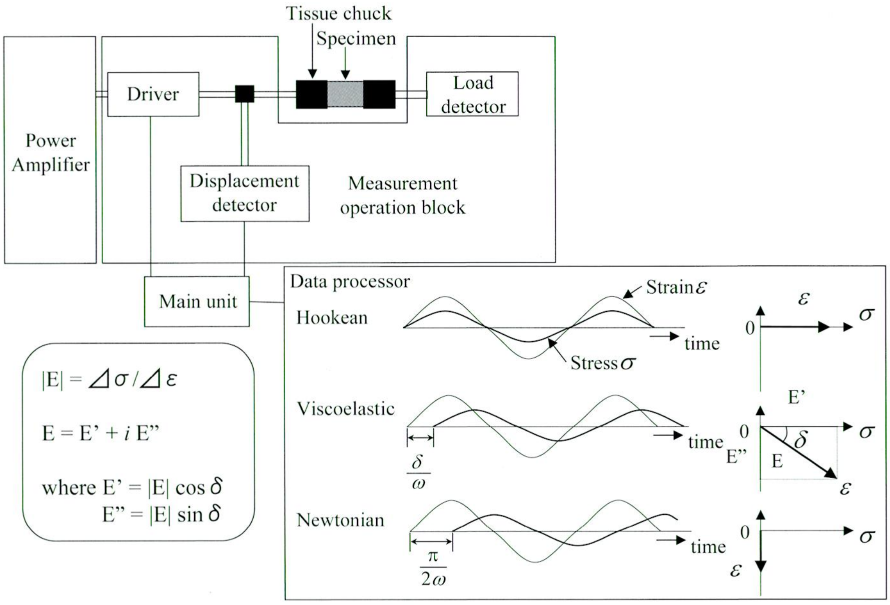

Figure 3 Block diagram of dynamic viscoelastometer and schematic representation of relationship between stress and strain of perfectly elastic solid (Hookean body), viscoelastic material and perfectly viscous liquid (Newtonian fluid) with sinusoidal varying stress.
The sinusoidal strain is produced by a tension control motor in the driver and the stress and strain are measured by means of load and displacement detectors and transmitted to a data processor. In a viscoelastic material, the phase difference between stress and strain is somewhere in between ($\pi/2 > \delta > 0$), and the complex modulus E* is resolved into two components, i.e., the storage modulus E' and the loss modulus E', shown vectorially. Furthermore, the tangent of the phase angle (δ) between stress and strain is a measure of the ratio of energy loss to energy stored during cyclic deformation.