REAR-IMPACT INFLICTED TEMPOROMANDIBULAR JOINT INJURY

C.G.Lyons, C.L.Brady, C.K.Simms
Department of Mechanical & Manufacturing Engineering
Trinity College Dublin. Ireland

ABSTRACT:

Indirect injuries to the jaw are not life threatening. However, they often lead to a range of painful symptoms, which can have serious consequences for the sufferer. These injuries are confirmed by clinical evidence but the injury mechanism is not understood. This work is part of an ongoing investigation into the nature of the injury process.

We have reported on the testing of a mechanistic model of a human head/neck/mandible. Results from these tests indicate high angular velocities and accelerations of the mandible but forces reacted upon the joint tissues were not significant. Human cadaveric samples of the temporomandibular joint (TMJ) were tested to quantify some material properties of the joint. The results from this and data from a mathematical model of the TMJ are reported.

IN OUR PREVIOUS work, Lyons et al (1997), we noted the prevalence of TMJ trauma among the victims of automobile collisions, particularly rear-end MVA's. Further research reinforces these data, Goddard (1993), O'Shaughnessy (1995), Goss & Bosanquet (1993). As with previous works, we see a range of injuries to soft tissue elements of the joint.

Goddard reports from a larger study which includes four MVA traumas. He notes the presence of myofascial pain and joint inflammation for all these cases and for two of them anterior disc displacement, confirmed by MRI. All patients also had accompanying head, ear and neck-ache. In one case, where pre- and post-trauma evidence could be compared, an increase of maximum incisal opening was recorded. Goddard hypothesised that this could be explained by the trauma causing tearing of the TMJ tissues, although no evidential proof is supplied.

O'Shaughnessy reports on a new form of computed tomographic technique devised specifically for the investigation of TMJ injury. Whilst the article does not present data in a scientific manner, he does note the occurrence of injuries to the sphenomandibular, the stylo-hyoid and stylomandibular ligaments, along with other more normally implicated cranio-mandibular muscles and ligaments. He further states that although radio-lucent evidence of injury fades with time, some residual evidence indelibly marks the injury site.

Goss & Bosanquet present data from a study of 30 patients, five of which are due to MVA's. Unfortunately, they do not specify if the MVA injuries were due to "direct" or "indirect " traumas. However, they do note temporomandibular disc tearing or shredding. They also comment on the remarkable propensity for TMJ injuries to resolve spontaneously, but state that some patients do present years later with internal derangement of the joint. Athroscopic investigations of these deranged joints commonly showed adhesions between disc and glenoid fossa.

So, as with previous reports, we see significant evidence of the pathology of joint injury but no reporting on injury mechanisms. Our work is an attempt to determine the mechanics of these "low level", but painful, TMJ injuries.

ANATOMY OF THE TEMPOROMANDIBULAR JOINT:

We can define the TMJ as a synovial, moveable hinge joint between the condyle of the mandible and the glenoid fossa on the squamous part of the temporal bone. Situated between these two bony elements is a fibro-cartilaginous disc which permits easy movement between mandible and cranium, Figure.1. The joint is capable of a range of motions in three space - both gross and subtle, and is under the control and restraint of a number of muscles and ligaments. The masseter, lateral & medial pterygoid and the temporalis muscle, along with the temporomandibular ligament play major roles in this control function. The joint is enclosed within a capsule, which consists of a thin sheet of collagen bound down to the neck of the condyle inferiorly, and to the rim of the glenoid fossa superiorly, Figure.2.

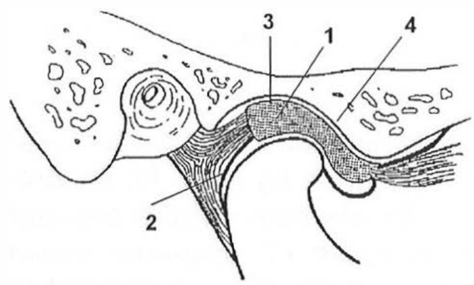


Figure.1. Sagittal View of the TMJ
1) Disc of the Condyle
2) Posterior attachment of Condyle
3) Glenoid Fossa
4) Articular Eminence

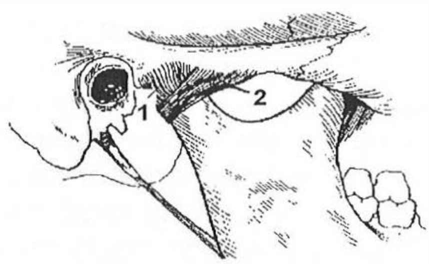


Figure.2. Sagittal View of TMJ of
1) Joint Capsule
2) Temporomandibular Ligament

The lateral ligament of the TMJ consists of a thickening within the joint capsule. This is attached from the tubercule of the articular eminence superiorly and to the posterior aspect of the condylar neck inferiorly. The lateral ligament has a further attachment at the lateral pole of the condyle, where it is contiguous with the attachment of the disc to the condyle in that area. The disc also has a number of attachments. In particular we may note;

- The Posterior attachment, which connects the posterior aspect of the disc to the posterior wall of the fossa and to the posterior aspect of the condyle.
- A Muscular attachment to the superior lateral pterygoid, anteriorly
- The Collateral Ligaments,
  a) The Medial Collateral Ligament attaches the medial pole of the condyle to the medial aspect of the articular eminence.
  b) The Lateral Collateral Ligament which attaches the lateral pole of the condyle to tubercule on the lateral aspect of the articular eminence.

The ligaments and capsule of the joint are composed of stiff collagenous fibres and may be prone to damage during involuntary motions of the mandible. The joint is further complicated by having the superior belly of the lateral pterygoid muscle inserted into the antero/medial portion of the TMJ meniscus. The inferior belly of the same muscle has attachments onto the neck of the condyle.

The joint is therefore complex, with a great many elements which may be, either singularly or in combination, subject to trauma.

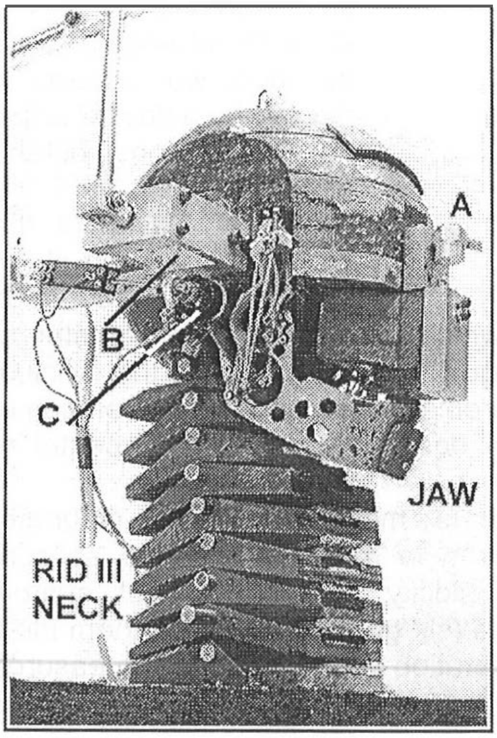


Figure 3. The Physical Model

THE PHYSICAL MODEL & TESTING:

As previously reported, Lyons et al (1997), we have built a mechanistic model of the human cranium, mandible, and cervical spine. The cranium is modelled on the anthropometric data for a 50th percentile male, and is an accurate, but simplified, model of the human cranium. The neck of this model is based upon the RID III neck proposed by Svensson et al(1993). The body of the mandible is made from steel with condyles of PTFE (Fluon), its geometry and that of the condyles were derived from measurements taken on a skeletal specimen of a 50th percentile male, Figure.3. The mass (0.162 kg) of the condyle and centre of gravity (0.073 m) from its condylar axis are taken from data given by Schneider et al (1989) and are consistent with the authors' measurements. The bony elements of the TMJ were constructed from wax impressions of the skeletal specimen. The glenoid fossae and articular eminences were cast in "cold cure" acrylic. These were embodied into the model cranium. Under test conditions the interface between these surfaces and the condyles were lubricated with petroleum jelly. At present the articular disc and associated tissues are not included.

Mechanical analogues of the Masseter, Temporalis and Medial Pterygoid muscles are incorporated as prime jaw closing muscles in the model. The stress strain properties of these muscle models are based on the experimental results of Hawkins and Bey (1994). The "muscles" were produced from combinations of elastic material and nylon thread. These have intrinsically linear properties in the elastic range but when arranged to become taut at varying extensions combine to provide a good approximation of the non-linear properties of passive muscle. For scaling purposes muscle cross sectional areas were taken from the data of Osborn and Baragar (1985). The data of Hawkins & Bey and the slopes of the muscle analogues (100% - passive) used by the authors are shown in Figure.4.

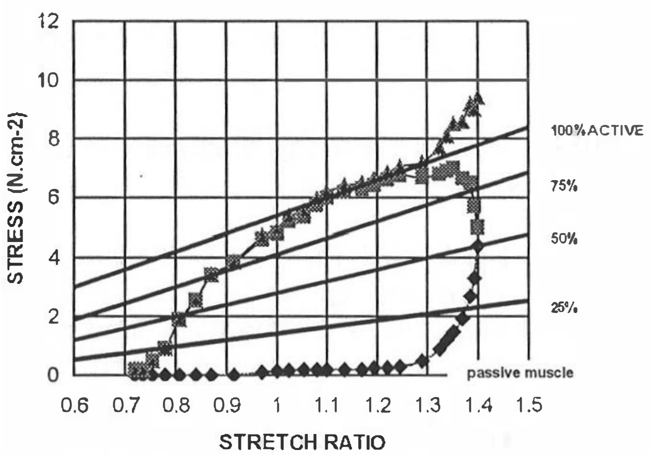


FIGURE.4. Properties of Human Muscle

EXPERIMENTAL: Rear impacts were simulated by means of a "mini sled" system driven by a pneumatic impact cylinder. An impact velocity of 9.2 km/h was used for all our tests. Head Angular displacement was measured directly by a device which uses a rotational potentiometer fixed at the centre of gravity of the cranium. The shaft of the potentiometer is connected to an almost frictionless mechanism which allows the potentiometer to translate freely whilst responding only to angular displacements of the cranium to which it is attached. This overcomes the problem of measuring the rotation of a body about a fixed point when that body does not have a fixed centre of rotation.

The angular velocity of the cranium is measured by a magneto-hydrodynamic (MHD) sensor (ATA Sensors type ARS-04E). These sensors yield a voltage proportional to angular velocity and are almost entirely insensitive to linear velocities. Their low mass (six grammes) makes them ideal transducers for our purposes. Angular acceleration of the head was measured by means of accelerometers.

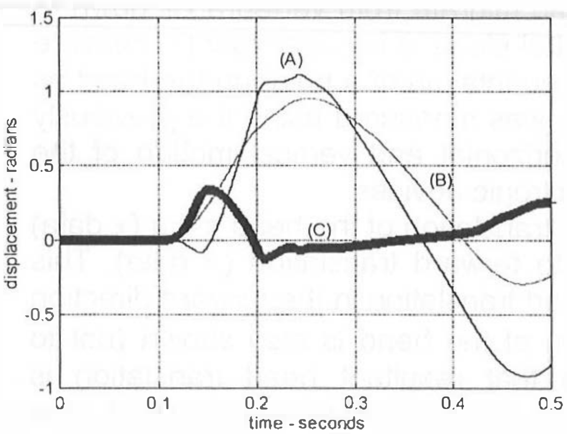


Figure.5. Angular Displacements Vs Time.
(A) Jaw rotation, (B) Head Rotation, (C) Mouth opening

The angular velocity of the mandible was measured with an MHD sensor mounted coaxial with the inter-condylar axis of the mandible. In this position the transducer measured the mandibles rotational velocity in the sagittal plane. Because of its low mass, it did not appreciably affect the behaviour of the mandible. The angular displacement of the mandible was derived from numerical integration of the measured angular velocity. Data logging and manip--ulation are described in detail in our previous paper

RESULTS OF TESTS: The bio-fidelity of our physical head/neck model has been validated by comparison with the results of Svennson (1992) and also with retroflexion data from human volunteer tests (McConnell et al. 1995). For equivalent Δv's close agreement was found for both displacements, velocities and accelerations. These data and the results of our physical tests have been reported, Lyons et al (1997). Of primary interest we note that for purely passive and for 25% active muscles, mouth opening was a feature of the retroflexion phase of whiplash. In particular for a Δv of 9.2km/h and purely passive elevator muscles, a maximum mouth opening angle of ~20 degrees was measured, see Figure.5.

HEAD ICR: The instantaneous centre of rotation of the head is important for the TMJ because a point on a rotating body experiences forces proportional to the

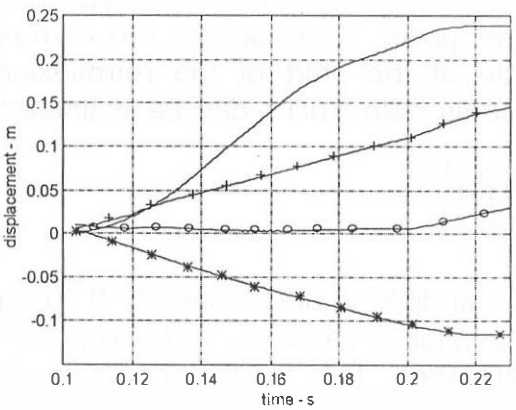


Figure.6 Displacement Vs Time for;
+ Sled (absolute), ∗ Head (relative to sled)
o Head (absolute) − Head Angle

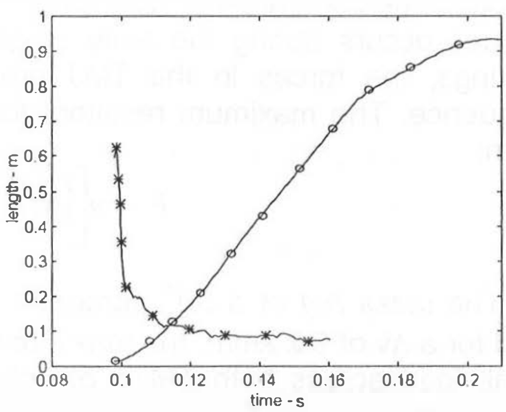


Figure.7. ∗-Radius of Rotation of Head & o-Head Retroflexion Vs Time
(impact at t=0.1s)

distance between the point and the centre of rotation. The head does not have a unique pivot point as the axis of rotation can migrate from vertebra C_0 down to C_7. In order to track this motion in the sagittal plane, it is necessary to measure the horizontal and vertical position and the orientation of a point on the head as it rotates. The orientation of the head c.o.g was measured using the previously described rotational potentiometer. The horizontal and vertical motion of the head's c.o.g. was measured using opto-electronic devices.

Fig.6. shows the time series of rearward translation of the head c.o.g (∗ data) relative to the sled and the sled's absolute forward translation (+ data). This figure also shows their resultant, the net head translation in the forward direction of travel (∘ data). The angle of retroflexion of the head is also shown (not to scale). From these data, we may note that resultant head translation is insignificant and its motion is predominantly rotational throughout the retroflexion sequence. This is because as the head commences retroflexion, the head and shoulders are both travelling forward, and the resulting linear displacement of the head is minimal.

The radial ($-r\dot{\theta}^2$) and tangential ($r\ddot{\theta}$) accelerations are the two orthogonal components of acceleration at a point on a rotating body. Using these and the mass of the mandible the resultant force acting across both TMJ's can be estimated. However, the radius *r* is a function of the instantaneous centre of rotation (icr). This is highly sensitive to error because the instrumentation fails to register the very small magnitudes of angular displacement at the onset of head movement, and this results in an apparently infinite radius of rotation. The disadvantage is that $\dot{\theta}^2$ and $\ddot{\theta}$ are largest when θ is still very small, and this makes forces calculations in this region unreliable. However, for the majority of the retroflexion phase, the distance from the head icr to the TMJ is ca. 0.1m, see Figure.7.

As linear head displacement is negligible in the retroflexion stage the forces at the TMJ are primarily the resultant of the tangential and radial components of head rotation. This is simply the result of the inertial properties of the head and disputes the contention of Howard *et al.* (1995) that compression of the TMJ soft tissues occurs during the early stages of whiplash. According to the current findings, the forces in the TMJ are tensile at the start of the retroflexion sequence. The maximum resultant force across both TMJ's can be estimated from;

$$F = m\left[\left(r\ddot{\theta}\right)^2 + \left(r\dot{\theta}^2\right)^2\right]^{0.5}$$

The mass (m) of a 50th percentile male mandible is approximately 0.16 kg and for a Δv of 9.2 km/h, the maximum acceleration is ca. 50ms^{-2}. These yield a peak load across both TMJ's of only 10 Newtons. This force appears to be within physiological limits, but to assess its likely effect on a TMJ it is necessary to have data from tensile tests of soft tissue from cadaveric specimens.

MATHEMATICAL MODELLING

Motion of the jaw during retro-flexion of the head is primarily a problem in dynamics, and therefore a mathematical model is a good auxiliary method of investigation. There are some distinct advantages of such a model over a conventional physical one. A mathematically based formulation lends a stronger understanding to the purely empirical results gained from sled tests. Additionally, it readily allows a parametric analysis and a variation of inputs beyond that which can be achieved in a physical model. However, strict simplifications are necessary to allow the basic governing equations to be derived. A number of techniques are possible, and the approach used here is that of differentiating the energy equations according to the Lagrange system. Once the various components of energy in the system have been correctly identified standard routines can yield the differential equations of motion. These were then numerically integrated by use of a low order Runge-Kutta routine.

The literature contains one previous attempt at modelling the dynamics of the head, neck and mandible, Schneider et al, (1989). Their model predicted very large angles of mouth opening for simulated impact speeds of 6.71m/s and 13.41 m/s., 1.17 rad and 1.11 rad respectively. It did not include jaw muscles, rather jaw motion was investigated only in terms of inertial motion. The relatively low mass of the jaw and the high strength of its elevator muscles would lead one to question strongly the validity of this approach. Additionally, they do not comment on the fact that a higher velocity of impact produced a smaller peak jaw-opening angle.

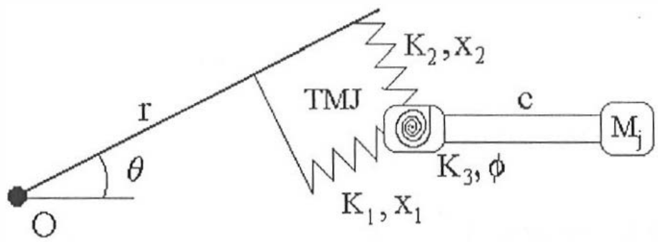


Figure.8.Schematic of Mathematical Model.

In our current mathematical model the head and jaw are defined as rigid bodies, Figure.8. Motion is two dimen--sional and constrained to the sagittal plane. The soft tissues of the TMJ are approximated as having a linear stiffness in both the radial and the tangential directions. A rotational stiffness proportional to angle of mouth opening is used to represent the passive elevator muscle forces. The linear spring parameters, K_1 & K_2 are based upon the results of quasi-static force/elongation tests on cadaveric human TMJ's (see below). These are taken as 5000 N/m. The rotational spring coefficient is estimated by simplifying the stress/stretch-ratio data of Hawkins & Bey (1993) used to construct the muscle bundles for the physical model. In the physical model maximum mouth opening was ~20 degrees. This will produce a stretch in each muscle of 10mm, giving stretch ratios varying from 1.18 - 1.27. From Fig.4., it can be seen that the passive response is linear in the stretch range 1-

1.27. Thus, mouth opening of ~20 degrees will produce a torque of 0.22Nm and this is used to calculate a rotational stiffness of 0.78 Nm/rad for the model, Table.I. Presently no damping is supplied.

Table I. Calculations For Torque Due To Elevator Muscles

Muscle	Masseter	Med. Pterygoid	Temporalis
c.s.a /muscle pair	6.8 (cm2)	3.8	8.4
Natural length (mm)	60	55	40
Stretch length (mm)	70.9	65.9	50.9
Stretch ratio *	1.18	1.198	1.2725
σ (N/cm^2)	0.3844	0.4028	0.5031
Force (N)	2.6139	1.53	4.226
Radius (mm)	30	30	35
Torque (Nm)	0.078	0.0459	0.1479
Total Torque (Nm)		0.2718	

* at mouth opening of 10 degrees

The Kinetic energy *(T)* of the system shown in fig.8. is ;

$$T = \frac{M}{2}[(r+x_1)^2\dot{\theta}^2 + \dot{x}_2 - 2(r+x_1)\dot{x}_2\dot{\theta} + 2c(r+x_1)\dot{\theta}(\dot{\theta}-\dot{\phi})\cos\phi$$
$$-2c\dot{x}_2(\dot{\theta}-\dot{\phi})\cos\phi + c^2(\dot{\theta}-\dot{\phi})^2 + \dot{x}_1{}^2 + \dot{x}_2{}^2\dot{\theta}^2 + 2\dot{x}_1 x_2\dot{\theta}$$
$$+2c\dot{x}_1(\dot{\theta}-\dot{\phi})\sin\phi + 2cx_2\dot{\theta}(\dot{\theta}-\dot{\phi})\sin\phi + \frac{1}{3}Mc_2\dot{\phi}^2$$

and its Potential energy (*V*), ignoring gravity, is

$$V = \frac{1}{2}K_1 x_1{}^2 + \frac{1}{2}K_2 x_2{}^2 + \frac{1}{2}K_3\phi^2$$

where
M = mass of jaw (kg)
r = distance from centre of rotation of the head to TMJ (m)
x_1 = displacement within the joint in the radial direction (m)
x_2 = displacement within the joint in the tangential direction (m)
c = distance from the TMJ to the centre of gravity of the jaw (m)
θ = angular displacement of the head (rad)
ϕ = angular displacement of the jaw with respect to the head (rad)
K_1 = linear spring coefficient of the TMJ in radial direction (N/m)
K_2 = linear spring coefficient of the TMJ in tangential direction (N/m)
$K\phi$ = rotational spring coefficient for the jaw (Nm/rad)

The 'dot' operator represents the derivative with respect to time

From these we can derive the Lagrangian expressions for x_1, x_2 and ϕ

$$Lx_1 = M\ddot{x}_1 + Mx_2\ddot{\theta} + 2M\dot{x}_2\dot{\theta} + Mc(\ddot{\theta} - \ddot{\phi})\sin\phi + 2Mc\dot{\theta}\dot{\phi}\cos\phi$$
$$- Mc(\dot{\theta}^2 + \dot{\phi}^2)\cos\phi - M(r + x_1)\dot{\theta}^2 + K_1x_1 = 0$$

and

$$Lx_2 = M\ddot{x}_2 - M(r + x_1)\ddot{\theta} - 2M\dot{x}_1\dot{\theta} + 2Mc\dot{\theta}\dot{\phi}\sin\phi + Mc(\ddot{\phi} - \ddot{\theta})\cos\phi$$
$$- Mc(\dot{\phi}^2 + \dot{\theta}^2)\sin\phi - Mx_2\dot{\theta}^2 + K_2x_2 = 0$$

and

$$L\phi = Mc^2(\ddot{\phi} - \ddot{\theta}) - 2Mc\,\dot{x}_1\,\dot{\theta}\cos(\phi) - Mc\,(r + x_1)\,\ddot{\theta}\cos(\phi)$$
$$+ Mc\,\ddot{x}_2\cos(\phi) + Mc\,(r + x_1)\,\dot{\theta}^2\sin(\phi)$$
$$- Mc\,(\ddot{x}_1 + x_2\,\ddot{\theta} + 2\,\dot{x}_2\,\dot{\theta})\sin(\phi) - Mcx_2\,\dot{\theta}^2\cos(\phi) + k_3\phi = 0$$

The input to the Lagrangian model is the rotational movement of the head measured from the experimental rig. Thus angular displacement, velocity and acceleration are specified at point 0, Fig.8. The resultant movement of x_1,x_2 and ϕ are measured. The results show that the condyles move anterior and inferior relative to the articular eminence as predicted in the acceleration diagram, Fig.9., but the amplitude of this motion is small. In the radial direction, the maximum displacement is 2mm @ 115 ms after impact, and in the tangential direction the maximum is 1.4mm after 75 ms.

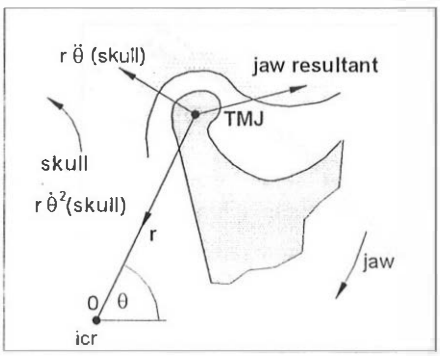


Figure.9. Head & Jaw accelerations

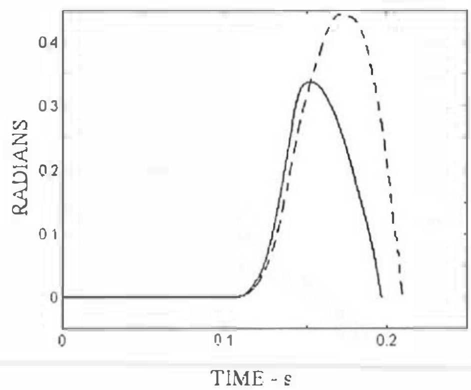


Figure.10. Mouth Opening
(solid - physical model, dashed -Lagrange)

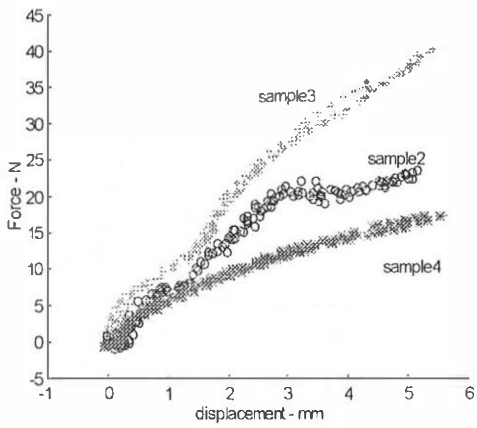
The peak stretch rate is 0.1m/s. The spring parameters and head input values were varied over a wide range of values, and it was found that in its present form the model is very stable. No qualitative changes occur and quantitative changes produce a broadly linear response.

The time series of mouth opening agrees quite well with that measured on the physical model, Figure.10. This is a preliminary validation of the mathematical model. Further, it should be noted that the ratio of the peaks is 1:0.75. Clearly, neither of these exceeds physiological limits.

CADAVERIC SPECIMEN TESTING:

The literature contains no data directly concerning the mechanical properties of the soft tissues of the TMJ. The variable collagen content, fibre orientation and geometry of the joint preclude the adaptation of existing data from tests on other ligamentous materials. It was therefore necessary to experimentally determine some of the mechanical properties of the joint. Volunteer tests are clearly not possible and so human cadaveric specimens preserved in 10% formalin were subjected to tensile loading. Again, the literature gives no guidance for the effect of formalin preservation on the dynamic mechanical properties of soft tissue. Yamada (1970) notes that static bulk properties, such as stiffness, decrease by some 50% and then stabilise after preservation. In the absence of any other evidence, we must assume that the dynamic properties behave in a like fashion. If this is so, it simply reinforces our conclusions.

The cadaveric specimens consisted of a section of the cranium including the glenoid fossa, the intact capsule of the joint, and a section of the mandible. The connections of both the masseter and pterygoid muscles into the joint capsule were excised on its exterior. The protruding bony sections of both cranium and mandible were fixed in acrylic blocks, which functioned as grips for the specimen. The mandibular aspect was affixed to a "slider" system which allowed sufficient freedom for the condyle to "glide" over the eminence of the fossa as it would in normal, *in vivo*, translation.



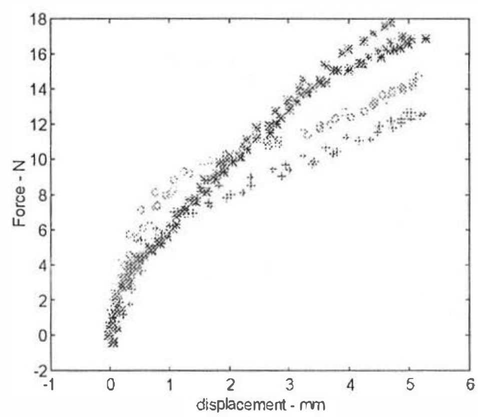


Figure.11.(a & b). Force Vs displacement for Cadaveric Specimens

(3 specimens, mouth closed) (spec 4; ★ 0,5; o 10 + 15 degrees mouth open)

The tensile loading machine was not capable of applying a high strain rate, and was used instead at very low strain rates to measure the elastic component of stiffness. The Force - elongation curve, Figure.11(a), shows the behaviour of three of the samples at a stretch velocity of 0.16mm/s. The results show quite close agreement within each sample, and a relatively high variability between samples. The latter is probably a consequence of the considerable geometric variation even between the two sides in the same subject (McDevitt 1989).

The initial load jump of ca. 5N after almost zero displacement in every case is the gravitational weight of the upper half of the joint once it has been lifted off the lower joint half. After this artefact, the curve behaves in a surprisingly linear manner. One of the difficulties with this mode of testing is that only lacerations or disruptions of the tissue can be readily detected. It is possible that *in-vivo* pain may be produced without recognisable trauma to the tissues. However, inspection of the samples after these tests showed no exterior signs of injury, and a protrusion of 5mm can be easily achieved by a healthy in vivo subject.

In the course of retroflexion of the head, the mandible undergoes both rotation and translation. To accurately reproduce this motion in the tensile testing machine would be difficult. Instead the force required to protrude/distract the mandible for various fixed angles of mouth opening was measured. The angles chosen were 0, 5, 10 and 15 degrees. The load required to produce 5mm protrusion of the condyle for these conditions is shown for a typical specimen, Figure.11(b). It can be seen that the effect of increasing mouth opening is to increase stiffness at low displacements and decrease it at higher displacements. This is due to geometric changes within the joint, but it is clear that the magnitude of the load is not greatly altered. Mouth opening does not greatly affect the internal joint stiffness.

It is also necessary to investigate the strain rate dependency of joint stiffness, as this may be the key to the cause of injury. This dependency is a consequence of the visco-elastic behaviour of biological tissues, where the load produced by stretching depends not only on the magnitude of stretch, but also on the rate of stretch. A number of empirical laws have been developed that predict the probability of injury for a given strain rate. The viscous criterion (VC) developed by Lau and Viano (1986) is one of the most advanced of these. It states that the critical compression velocity for strain rate type injuries is 3m/s. Our Lagrangian model shows a maximum relative jaw velocity of 0.14m/s. Although this lies considerably below the stated VC limit, it is possible that because the VC was designed to predict serious injuries to vital organs (such as heart and lungs), it may not be a suitable predictor of soft tissue trauma in the TMJ.

The maximum stretch speed of the testing machine is 0.0083m/s. It was found that at this speed the load deflection curve had a higher slope than at 0.00016m/s, but this could not be extrapolated up to the desired speed of ca. 0.1m/s. Strain rate dependency needed to be investigated by other means. A dynamic materials testing analyser (Rheometer Ltd. DMTA Mk3) was used to provide a measure of the damping within the joint. This precision instrument is used in the polymer industry to provide dynamic stiffness and damping measurements over a range of temperatures and frequencies. For our requirements, the temperature was set to 37deg Celsius and the frequency was varied between 1Hz and 30Hz. Retroflexion of the head occurs over a period of 130ms, which approximates to a frequency of 2Hz. The corresponding velocity and acceleration pulses are in the range 3Hz to 10 Hertz.

Strips of ligamentous material dissected from the capsule of a human TMJ were prepared for the DMTA. These were clamped at either end and an actuator is then used to apply a predefined cyclic strain. The resultant load is measured and the software provided produces a value for both the elastic (storage) modulus and $tan(\delta)$, the ratio of viscous (loss) modulus to storage modulus, Figure.12(a & b).

The viscous modulus may readily be recovered from the equations.

Elastic Modulus: $E' = \cos\delta(\sigma/\varepsilon)$. Storage Modulus: $E'' = \sin\delta(\sigma/\varepsilon)$

Complex Modulus $E^* = \sqrt{(E')^2 + (E'')^2}$ Phase Angle $\tan\delta = (E'/E'')$

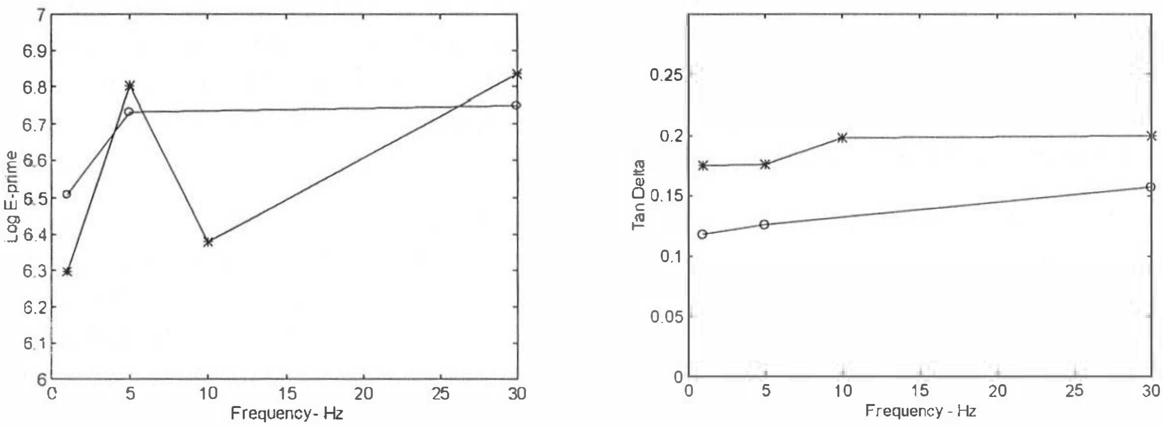


Figure.12.(a & b). Variation of the Log of Storage Modulus and tan(δ) of TMJ ligamentous tissue with Excitation Frequency. (* sample 1 o sample 2)

The Elastic Modulus calculated from this mode of testing is comparable with that derived from the force/elongation tests measured on other cadaveric specimens. From Figure.12(a), E'=3.16MPa. The force/elongation tests, Fig.10. yield a load of ca. 15 Newtons (excluding initial 5N jump) for a stretch of 5mm. It is difficult to measure the cross-sectional area of soft tissue in the joint, but it can be estimated at 15mm^2. Thus, at 5mm elongation there is a stress of 1MPa. The strain in the fibres is also difficult to estimate, but an elongation of 5mm represents ~30% strain. Elastic Modulus is the ratio of stress/strain. From the quasi-static force/elongation tests the value is equal to ~3Mpa. The similarity of these readings is striking, but there is considerable error in the latter calculation.

From Figure.12(a & b), it can be seen that over the Frequency range measured, the ratio of loss modulus to storage modulus is, at worst, 1:5. Therefore, the viscous effects are only ~20% of the elastic stiffness of the material tested. The implication of this is that although there will be some strain dependent load developed, this is not critical.

CONCLUSIONS:

Our physical and mathematical models indicate clearly that for low velocity, zero incident-angle rear-end collisions, the forces reacted within the TMJ lie within the envelope of physiological limits and so physical injury will not occur. However, the considerable body of clinical evidence of injury cannot be ignored, we must explain them in some other way. Non zero incident angle automobile impacts, or initial head positions which are turned about the coronal plane will introduce additional moments and the forces reacted on the mandible may be changed considerably. Our physical model is constrained to head motion in the sagittal plane and so is insufficiently sophisticated to simulate this more complex behaviour. This will be examined by developing the Lagrangian model to include a full mandible and both temporomandibular joints. The effect of three dimensional head rotation on the TMJ will then be fully quantified.

Some of the mechanical properties of the *in vitro* joint have been quantified. The visco-elastic properties are such that they will not significantly influence joint stiffness. The high inertia of the head results in relatively low head angular velocities (15 rad/s). At higher velocities the visco-elastic properties would be more significant, these too will be incorporated into the Lagrangian model.

ACKNOWLEDGEMENTS:

To Prof M O'Brien,. Head of the Dept of Anatomy, Trinity College Dublin, for many helpful discussions and for the use of facilities. To "The Polymer Development Centre" of Forbairt Ireland and particularly Dr Fiona Coyle for help with, and the use of, testing facilities.

REFERENCES:

Goddard.G, *Articular Disk Displacement of the TMJ Due to Trauma,* Journal of Craniomandib. Prac. Vol 11, No 3, 1993, pp 221-223.

Goss A.N, & Bosanquet A G, *The Arthroscopic Appearance of Acute temporomandibular Joint Trauma*, J Oral Maxillofac Surg., 48: 1990, pp 780-783.

Hawkins.D, & Bey. M. *A Comprehensive Approach for Studying Muscle-Tendon Mechanics.* J. Biomechanical Engineering, vol. 116, pp. 51-55, 1994

Howard P R, Hatsell C P & Guzman H M, *Temporomandibular Joint Injury Potential Imposed by the Low-Velocity Extension-Flexion Maneuver.* J Oral Maxillofac Surg, 53; 1995, pp 256-262

Lau. V, Viano. D, *The Viscous Criterion - Bases and Applications of an Injury Severity Index for Soft Tissues.* 30th Stapp Car Crash Conference, SAE Trans Vol. 95, Section 5, 1986

Lyons C.G, Brady C.L & Simms C, *An Investigation of Damage to the Human Temporomandibular Joint*, in the 41st Annual Proc of AAAM. 1997, pp-315-330.

McConnell W.E, Howard R.P, vanPoppel J, Krause R, Guzman H, Bomar J, Radin J, Benedict J, Hatsell C. *Human Head and Neck Kinematics after Low Velocity Rear-End Impacts- Understanding "Whiplash".* SAE Trans. Vol 104 Section 6. 215-238, 1995

McDevitt W. E, *Functional Anatomy of then Masticatory System.* Butterworth and Co., 1989

O'Shaugnessy T, *Tomographic Proof of Trauma-Induced Injury to the TMJoint & Other Sites in the Body.* The Functional Orthodontist, Nov/Dec 1995, pp 20-28.

Osborn. J.W, & Baragar. F.A, *Predicted Patterns of Human Muscle Activity During Clenching Derived From A Computer Assisted Model: Symmetric Vertical Bite Forces.* J.Biomechanics, vol.18, no, 8, 757-767. 1984

Schneider. K, Zernicke, R.F. & Clark, G. *Modeling of Jaw-Head-Neck Dynamics During Whiplash.* J Dent Res. 68(9): 1989. pp1360-65.

Svennson M.Y & Loevsund A, *A Dummy for Rear End Collisions- Development and Validation of a New Dummy Neck.* Proc 1992 Intl IRCOBI Conf, Verona, Italy. 1992.

Yamada H., *Strength of Biological Materials.* Williams and Wilkins Press. Baltimore. US. 1970.