BITE FORCE AND EMG STUDIES ON THE JAW-CLOSING MUSCLES

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To my Parents for their continual Love and Support

Στους γονείς μου για τη συνεχή τους Αγάπη και Στήριξη

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This thesis is the original work of the author

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SUMMARY

The jaw-closing muscles perform movements of the mandible, provide force for a variety of natural functions, and are subject to disorders often manifested as jaw muscle pain. Bite force measurement and surface electromyography have been used in the present investigation to assess jaw-closing muscle activity and strength in young healthy adult subjects and in older edentulous patients.

Human bite forces have been previously studied with several types of equipment, and the maximum values reported have varied greatly. Apart from anatomical and physiological characteristics of the subjects, bite force measurement depends on a number of other factors, such as design and comfort of the transducer and the position of the transducer in the mouth.

Therefore, the first experiment was carried out to investigate the effect of measuring bite force in three different transducer locations within the dental arch, on different occasions and to determine the reliability of these measurements. Maximum voluntary bite force was measured in a group of fully dentate participants in three different positions, using three different patterns of strain-gauge force transducer. The unilateral transducer was placed between the second premolar and first molar teeth on one side, the anterior transducer was placed between the anterior teeth, from canine to canine, and the bilateral transducer was placed between the second premolars and first molars on both sides.

The highest bite forces were measured with the bilateral transducer and the lowest on the anterior transducer. Maximum bite force was most reproducible (i.e. showed least within-subject variability) when measured between the first molar and second premolar teeth on one side only. In addition, there was little difference in bite force between sessions when measured in the same position within the dental arch, whichever of the three positions that may be; maximum bite force is relatively consistent.

Electromyography (EMG) is a widely used method of monitoring skeletal muscle activity. In many muscles EMG amplitude increases linearly with force and changes in the EMG/force ratio or shifts of the EMG power spectrum median frequency provide measures of fatigue. Previous studies have found both linear and non-linear relations between increasing bite force and integrated EMG of the jaw-closing muscles. Acoustic myography (AMG) is a non-invasive method which also may be used as an indicator of skeletal muscle activity but is still at an early stage of development for use with the jaw-closing muscles.

The second experiment was carried out to examine the relationships between the EMG, AMG and bite force in masseter muscles of a group of healthy male subjects and to assess these two indirect measures (i.e. EMG and AMG) of activation of the masseter muscle.

A linear relationship was found between rectified integrated EMG, AMG and force at the four different submaximum clenching levels tested. It seems that AMG, under control conditions, may be useful in situations where EMG is difficult or presents problems, although it is not quite such a good monitor of force as EMG.

It is well known that bite force is greatly reduced in edentulous elderly subjects, due to disuse atrophy of their jaw muscles. However, the susceptibility of the jaw-closing muscles to localised fatigue in edentulous patients is less certain. This information is less clear for edentulous subjects with jaw muscle pain and history of TMD.

Thus, the third experiment was carried out to investigate bite force and endurance time - as an indicator of the resistance to muscle fatigue - in healthy complete denture wearers and in denture wearers with TMD, using a comfortable strain-gauge transducer. Further, to assess the shift in the median frequency of the EMG power spectrum and the changes in the relaxation rate before and after a fatiguing task, as indicators of the fatiguability of the masseter muscles in these two edentulous groups.

Maximum voluntary bite force was found to be low in elderly edentulous subjects and was further significantantly reduced in edentulous TMD patients. The mean endurance time in healthy complete denture wearers was relatively similar to those reported in published figures for subjects with natural dentition, but it was considerably reduced in TMD complete denture wearers. The shorter endurance time was consistent with the greater median frequency decrease and the slower rate of relaxation observed in edentulous TMD patients, indicating that their masseter muscles were more susceptible to fatigue than in healthy edentulous subjects.

LIST OF ABBREVIATIONS

AMG	Acoustic Myography
ANOVA	Analysis of Variance
ATP	Adenosine Triphosphate
EMG	Electromyography
g	Gram
HP	Hewlett-Packard Microphone
Hz	Hertz
IAMG	Integrated Acoustic Myography
IEMG	Integrated Electromyography
kg	kilogram (1Kg = 9.8N = 2.21 lb.)
MF	Median Frequency
MuAP	Motor Unit Action Potential
MVC	Maximum Voluntary Contraction
μm	Micrometre ($\mu m = 10^{-6}$ m)
N	Newton
RMS	Root Mean Square
PC	Personal Computer
SD	Standard Deviation
SEM	Standard Error of the Mean
s	Second
TMD	Temporomandibular Disorders
V	Volt

GENERAL INTRODUCTION

The human jaw-closing muscles are essential to mastication and play an important role in the natural function and functional disorders of the masticatory system (Bakke, 1993). The measurement of bite force is often necessary to assess masticatory muscle function and also to compare jaw muscle activity between subjects in a wide variety of experimental situations.

The measurement of maximum voluntary bite force presents several problems; first, psychological factors (i.e. anxiety, motivation, fatigue) which limit the force which is generated, and secondly, the design and comfort of measuring instruments.

Maximum bite force has been defined as the force applied by the subject's jaw-closing muscles voluntarily between antagonist teeth without any food being present (Carr and Laney, 1987). However, when a volunteer is requested to clench maximally on a bite force meter, it is notoriously difficult to be sure that the force is the maximum possible because of discomfort, fear of breaking cusps of teeth and dental restorations, and the inevitable varying motivation of individuals (Klaffendach, 1936; Van Steenberghe and De Vries, 1978a; Mericske-Stern et al, 1995). There is also an inhibitory response from the pain receptors in the periodontal membrane which supports each tooth (Van Steenberghe and De Vries, 1978a,b). These problems could lead to an underestimation of the bite force, and this requires consideration.

Bite force transducers have usually been based on strain gauges or piezoelectric crystals, and their unilateral or bilateral sensors have varied in thickness, as well as in biting area (Hagberg, 1987). The jaw separation that is caused by the intra-oral insertion of the force transducers can affect bite force values (Manns, Miralles and Palazzi, 1979). It has been suggested that the rest position may not be the optimal position for producing a maximum bite force (Manns et al, 1979; Lindaeur, Gay and Rendell, 1993). The positioning of the bite force transducer within the dental arch is also important. The degree to which different muscles are involved in the force production, the root area and the favourable position of the teeth close to the jaw closing muscles are factors affecting bite force development in different sites within the dental arch.

In addition to problems with the measuring instruments, there are also other factors, such as the anatomical and physiological characteristics of the subjects, influencing bite force (Carlsson, 1974). Therefore, variability in bite force between subjects of the same population could be expected with different positioning of the force meter in the mouth.

Although there is inadequate support for the use of surface electromyography (EMG) as a diagnostic tool, it has been used to assess function and dysfunction of the masticatory muscles (Dahlstrom, 1989; Lund and Widmer, 1989; Widmer, Lund and Fein, 1990). EMG studies have been carried out to relate muscular force and EMG of the jaw closing and other muscles, and most investigators reported this relationship as linear although some have found nonlinearity (Ahlgren, 1966; Moller, 1966; Haraldson et al, 1985; Lawrence and De Luca, 1983). Because the EMG amplitude to muscle force relation has been considered to be linear in a non-fatigued state, the EMG amplitude has been used as a direct measure of force (Van Boxtel et al, 1983; Neill et al, 1989). However, the EMG amplitude at a given force increases as fatigue occurs and it is difficult to know when someone is observing a fatigue effect or simply an increase in force output, and it can be seen that problems exist with this approach.

Acoustic myography (AMG), the recording of low frequency sounds on the surface of a contracting muscle, is a method of assessing muscle force although requires development (L'Estrange, Rowell and Stokes, 1993; Orizio, Perini and Veicsteinas, 1989a). The relationship between AMG and the level of force production appears to vary in different muscles, but it is generally agreed that AMG shows a positive correlation with increasing force (Stokes and Dalton, 1991b).

Patients with temporomandibular disorders (TMD) seem to have significantly lower bite force and electromyographic activity (Cooper, 1996), indicating weaker levels of masticatory muscle strength compared to healthy subjects (Bakke and Moller, 1992). The function of the jaw-closing muscles in the pathophysiology of TMD and certain types of headache, requires investigation.

The TMD include a number of different but related clinical disorders that affect the masticatory system, and produce symptoms of pain and dysfunction in the muscles of mastication and the temporomandibular joint (Dworkin, 1996). According to several epidemiological studies TMD are found in all age groups and almost equally in both genders (Agerberg, 1988). Also, TMD seems to be almost as prevalent in complete denture wearers as in dentate individuals, (Choy and Smith, 1980; Zissis, Karkazis and

Polyzois, 1988), although the symptoms appear to be of low intensity (Lundeen et al, 1990).

The relevance of the process of fatigue in the jaw-closing muscles lies in the fact that it is considered to be of significance in the aetiology of TMD (Laskin, 1995). Hyperactivity of the jaw closing muscles and parafunctions such as clenching and grinding have often been related to symptoms of TMD (Laskin, 1969; Yemm, 1985). It is probable that both could cause jaw muscle fatigue, which, in turn could result in symptoms of muscle pain and tenderness (Mao, Stein and Osborn, 1993).

Muscle fatigue has been defined as 'any reduction in the force-generating capacity of the entire neuromuscular system, regardless of the force expected' (Bigland-Ritchie & Woods, 1984). On the other hand, endurance is described as the length of time a given muscular contraction can be sustained (Clark and Carter, 1985). The jaw-closing muscles present special difficulties for the investigation of fatigue because it is not possible to measure the individual muscle force output, but only their collective output.

The presence or absence of force decay is not the only method of evaluating neuromuscular fatigue. Several studies have examined changes in the EMG frequency spectrum of the masticatory and peripheral limb muscles during a fatiguing task (Naeije and Zorn, 1981; Bigland-Ritchie, Donovan and Roussos, 1981; Lyons, Rouse and Baxendale, 1993). They have all demonstrated a progressive shift of the mean and median power frequency to a lower frequency during a sustained isometric contraction , which can be used to identify muscle fatigue.

It has been reported that muscle fatigue during voluntary sustained contractions is characterised not only by loss of force but also by a slowing of the rate of relaxation (Jones, 1981). This slowing of relaxation may be seen in the jaw-closing muscles also, although the reasons for this slowing on fatigue are still not completely understood. However, it has been found that jaw-closing muscles are more fatigue resistant than limb muscles (Van Steenberghe, De Vries and Hollander, 1978).

The functioning of the jaw-closing muscles in edentulous subjects has been assessed previously with bite force (Haraldson, Karlsson and Carlsson, 1979) and EMG studies but the examination of their susceptibility to fatigue has received relatively little attention (Jacobs and Van Steenberghe, 1993). This information is even less certain for edentulous subjects who have TMD.

CHAPTER 1

REVIEW OF LITERATURE

AND AIMS OF INVESTIGATION

1.1 BITE FORCE

1.1.1 Evaluation of jaw-closing muscle function

The jaw-closing muscles perform movements and provide force for a variety of natural functions, including mastication, biting, swallowing, speaking, elevation of the mandible and control of jaw posture. Parafunctional activities such as grinding or clenching the teeth are often undertaken.

Masticatory muscle function has been evaluated: 1) by measuring masticatory performance (i.e. the particle size distribution of particular food chewed for a given number of strokes), 2) masticatory efficiency (i.e. the number of masticatory strokes required to reduce food to a certain particle size) (Bates, Stafford and Harrison, 1976), 3) bite force recording (Black, 1895; Klaffenbach, 1936; Worner, 1939; Carlsson, 1974) and 4) by electromyographic recordings during chewing and maximum biting (Moller, 1966; Miralles et al, 1989). Furthermore, the endurance time while maintaining a constant force during isometric contraction (Christensen, 1979; Clark, Beemsterboer and Jacobson, 1984), and to a certain extent by determining the state of muscle fatigue (Palla and Ash, 1981a; Christensen, 1981a) have also been used as objective measures of masticatory muscle function. Subjectively, an individual's masticatory function has been assessed be means of questionnaires and interviews related to chewing efficiency in breaking up a certain test material (Carlsson, 1974; Haraldson, Carlsson and Ingervall, 1979; Carlsson, 1984; Harle and Anderson, 1993).

1.1.2 History of bite force measurement

Bite force has been defined as 'the result of muscular force applied on opposing teeth; the force created by the dynamic action of the muscles during the physiologic act of mastication; the result of muscular activity applied to opposing teeth' (Glossary of Prosthodontic Terms, 1994). Van Steenberghe and De Vries (1978a) make a distinction between the theoretical and the practical maximum clenching force. The theoretical maximum clenching force is the fc a subject could develop between both jaws by maximum contraction of his jaw closing muscles at their optimum length. This has been calculated with mathematical models based on cross-sectional areas and estimated intrinsic strength of jaw closing muscles and skull shape. The theoretical maximum bite force values have varied in the molar region from 753N to 2000N and in the incisal region from 687N to 740N (Mainland and Hiltz, 1934; Carlsoo, 1952; Pruim, Jongh and Ten Bosch, 1980; Koolstra et al, 1988).

The practical maximum bite force is the force developed between the jaws during maximum voluntary contraction of the jaw closing muscles. The measured maximum bite force has, in general, been notably smaller. In Western populations, average maximum bite forces between the molar teeth are usually reported to be in the range of 600-750N (Hagberg, 1987). The greatest maximum biting force was recorded by a male Eskimo (1500N), using unilateral recording (Waugh, 1937) and by a man from Florida (4356N) with bruxing-clenching habits, using a bilateral instrument (Gibbs et al, 1986).

Since this first published work of Borelli (1681), a number of methods have been described in the literature depending on the principles of construction of the force measuring device (Carr and Laney, 1987). These principles have included the use of springs (Black, 1895; Boos, 1940), manometers and combinations of levers and springs (Brawley, Sedwick and Rochester, 1938), hydrostatic gnathodynamometer (Brekhus, Armstrong and Simon, 1941), psychophysical methods (Wennstrom, 1971a, b), electronic strain gauge transducers between antagonist teeth or into crowns, bridges and prosthetic appliances (Linderholm and Wennstrom, 1970; Helkimo, Carlsson and Garmeli, 1975), piezoelectric transducers (Ahlgren and Öwall, 1970), telemetric devices (De Boever et al, 1978) and sound transmission (Gibbs et al, 1981). During the past two decades most bite force transducers have been based on strain gauges or piezoelectric crystals.

It is thus understandable that the maximal bite force values recorded using these different measuring devices, under varying test conditions and from natural and artificial dentitions, have varied greatly (Hagberg, 1987), and have been subject to widely different interpretations.

1.1.3 Factors Influencing Bite Force

The variability of bite force reported in the literature is probably due to differences in force transducers, in placing and stabilising the transducers, in mental attitude of volunteers and fear of pain or breaking cusps of teeth (Hagberg, 1987). When the force transducer is made of metal, people may be hesitant to clench forcefully because of the risk of enamel fractures, or damage of dental restorations and prostheses (O'Rourke, 1949; Carlsson, 1974; Mericske-Stern & Zarb, 1996).

In the bite force studies by Hagberg (1985, 1986) 1-mm thick acrylic splints were used both to protect the teeth and to stabilise the bite fork. Gutta percha and self curing acrylic resin coated on the biting surfaces of the force transducers has also been used to protect the teeth against the metal bite plates (Hagberg, 1987).

1.1.3.1 Muscle Strength and periodontal receptors

The maximum bite force depends primarily upon two major factors, muscle strength and the periodontal receptors. The greatest force generated by the jaw closing muscles is during clenching (Moller, 1966).

According to the sliding filament theory of muscle contraction, force is developed by the myosin heads, or cross-bridges, in the region of overlap between the thick filaments, composed of the contractile protein myosin, and thin filaments of muscle fibres, composed of the protein actin (Huxley, 1969; Huxley, 1974). A key to the sliding mechanism is the calcium ion, which turns the contractile activity on and off. Under resting conditions, the sliding forces between the actin and myosin filaments are inhibited, but when an action potential travels over the muscle fibre membrane (sarcolemma), it causes the release of calcium ions from the sarcoplasmic reticulum into the sarcoplasm surrounding the myofibrils. These calcium ions initiate attractive forces between the actin and myosin filaments, causing them to slide together, and contraction begins. The energy for muscle contraction is provided by the breakdown (hydrolysis) of high energy bonds of myosin adenosine triphosphate (ATP) (Huxley, 1974).

The major determinant of strength is the muscle size. The magnitude of bite force is proportional to the cross-sectional area of the muscle (Weijs and Hillen, 1984; Sasaki, Hannam and Woods, 1989), rather than its length. The cross-sectional areas of the masseter and medial pterygoid muscles could explain up to 50% of the intra-individual variation of the maximum bite force in the molar region (Van Spronsen et al, 1989; Sasaki et al 1989).

Bite force is transmitted through the teeth to the periodontium. The periodontium is innervated from the maxillary and the mandibular branches of the trigeminal nerve and the axons reach the periodontal ligament either from the apical end or through the alveolar process (Van Steenberghe, 1979). In the periodontal ligament (periodontal membrane) there are mechanosensitive free nerve endings of sensory fibres, known as periodontal mechanoreceptors, divided into rapidly - adapting and slowly - adapting types, that respond when a force is applied to the teeth. It is also known that the periodontal ligament mechanoreceptors are situated evenly around the roots of the teeth, with more found closer to the apex of the tooth (Kizior et al, 1968). In order to protect the teeth, the periodontal receptors show negative feedback to the force developed by the elevator muscles during clenching efforts (Black, 1895; Worner and Anderson, 1944; Hannam and Matthews, 1968; Van Steeberghe and De Vries, 1978a,b). It has been suggested by Hannam, Matthews and Yemm (1970) that these mechanoreceptors do not play a major role in producing reflex changes in the elevator muscles. However, following tooth contact, they may play a part in limiting the maximum force developed during mastication. This would have a protective effect from overloading by inhibitory influences on the motoneurons of the elevator muscles and excitation of the depressor muscles.

If this is the case, then local anaesthesia of the tooth should cause an increase in the maximum voluntary biting force. O'Rourke (1949) observed that there is an average increase of 36 % in biting force under nitrous oxide analgesia. It has also been shown that greater bite forces can be developed after blocking of periodontal and intradental receptors by local anaesthesia (Van Steenberghe and De Vries, 1978b).

Other experimental evidence, however, suggest that periodontal mechanoreceptors have a positive feedback on jaw closing muscle contraction and that this is supplemented by input from other receptors, probably muscle spindles. (Lund and Lamarre, 1973; Lavigne et al, 1987).

A likely hypothesis is that the low threshold receptors provide positive feedback to the jaw closing muscles and therefore an increase of the biting force (Lavigne et al, 1987; Ottenhoff et al, 1992), but as bite force reaches a certain magnitude, higher threshold receptors are activated and limit bite force (Thexton, 1976; Van Steenberghe and De Vries, 1978b).

1.1.3.2 Intra-oral Variables

Several investigations have shown that bite force varies from one part of the oral cavity to another and that it is greatest in the region of the first molars and only about one-third to one-quarter in the region of the incisors (Worner, 1939; Carlsson, 1974; Devlin and Wastell, 1986; Hagberg, 1987). Maximum bite force among Europeans and Americans are reported to be in the range of 600-750N between the molar teeth (Hagberg, 1987).

Bakke et al (1990), in a large sample of 63 females and 59 males, found the mean maximum bite force unilaterally in the molar region to be 522N in men and 441N in women. Proffit et al (1983) reported a mean maximum unilateral first molar bite force of 310N in a sample of 21 adults of a normal population. Van Eijden (1990) found the mean maximum bite force at the second premolars to be 587N when measured unilaterally, and 606N when measured bilaterally. Unilateral measurement of the maximal bite force in subjects with natural teeth in the molar region has shown a bite force of about 300-600N (Bakke et al 1989; Bakke et al, 1990), and the force with bilateral measurements in the same region was considerably higher (Bakke et al, 1989;

Pruim et al, 1980). Measurements of maximum biting forces with the mandible in lateral excursions or in protrusion and retrusion have shown a lower bite force values compared to the intercuspal position (Leff, 1966; Molin, 1972).

The forces at the incisors have been found to vary between 140N to 200N (Hellsing, 1980), and from 120N to 350N between the canine teeth (Lyons and Baxendale, 1990).

It is suggested that the reasons for greater force between the molars than between the anterior teeth is as followings:

- Lever action. Mansour and Reynik (1975) interpreted the behaviour of the mandible as a class III lever system with the fulcrum located at the centre of the condyle and the muscles of mastication applying the force.

  In this model, because load and effort are on the same side of the fulcrum, moving load distally will reduce each moment about the condyle. This implies that less muscular effort will be necessary to produce a given bite force further back in the mouth.

- Root surface area. It is probable that the greater bite force capacity of the posterior teeth is partly due the larger area of their roots (Worner, 1939; Waltimo and Kononen, 1994).

- Neurophysiological considerations. The anterior teeth are reported to have more proprioceptive nerve endings that the posterior teeth. Therefore, the nerve endings serve as a greater protector of the anterior teeth by possibly inhibiting the moto-neurons serving the jaw closing muscles.

The largest maximum voluntary bite force is developed in the vertical direction (i.e. perpendicular to the occlusal plane of the upper teeth). In oblique directions, the larger force is exerting with posteriorly directed effort corresponding to about 90% of the maximum bite force developed vertically; medial and lateral efforts correspond to about 80% and anterior to about 70% (Van Eijden et al, 1990). Therefore, in most cases the magnitude of bite forces has been assessed by force transducers which measured only the vertical component of bite force (Linderholm and Wennstrom, 1970; Helkimo et al, 1975; Floystrand, Kleven and Oilo, 1982).

1.1.3.3 The Vertical Jaw Separation

Another important factor that contributes to bite force is the jaw separation required by the measuring device (Manns et al, 1979). In 1940, Boos used an intra-oral pressure gauge (Bimeter) to record maximum bite force in edentulous patients and he suggested that the vertical dimension around the rest position of the mandible may be the optimal position for exerting bite force. O'Rourke (1949) and Boucher, Zwemer and Pflughoeft (1959) have criticised these findings on the grounds that the bite force was measured in edentulous patients and uncontrolled variables such as pain, apprehension and low tolerance might influenced the force measurement.

Manns et al (1979) reported that 15 to 20mm of jaw opening, measured from the distal borders of canines, is the optimal muscle length, and at this length the highest bite force is produced. Furthermore, MacKenna and Turker (1983) suggested that the maximum incising force in dentate subjects was greatest at around 17mm interincisal distance.

Fields et al (1986) studied the progressive increase in bite force by increasing the vertical jaw opening and found the maximum force at about 20mm, followed by a decrease and then a second increase to near maximum force at about 40mm, in young adults. Lindauer et al (1993) reported that the maximum increases in bite force associated with minimum increase in EMG activity occurred between 9 and 11mm of jaw opening measured at the first molar region. The range of jaw opening corresponds to an interincisal distance of 15 and 20mm. Moreover, the effect of altering the vertical dimension of occlusion on the biting force was studied in edentulous subjects, and the greatest bite force was found at a jaw opening of 15mm (Prombonas, Vlissidis and Molyvdas, 1994).

Variation in bite force with vertical dimension is due to the length-tension relationship of the muscle fibres (Storey, 1962; Manns et al, 1979). The optimal jaw-closing muscle length has been suggested to be with the mandible at rest position (Moller, 1966; Storey, 1962).

However, it has become increasingly apparent that the bite force is exerted most efficiently when the vertical jaw separation is 9-20mm, measured at the canine-molar region (Manns et al, 1979; MacKenna and Turker, 1983; Lindauer et al, 1993). The contractile unit of a skeletal muscle fibre is a sarcomere, and above and below the optimum sarcomere length, the tension developed during a contraction declines. Since the muscle length-tension relationship is ascribed to number of cross-bridge formation (Gordon, Huxley and Julian, 1964; 1966), the optimal muscle length corresponds to the optimal sarcomere length with the maximum cross-bridge formation. Histological findings in the rat demonstrated that the sarcomere length of the masseter and temporalis muscle increased as the mouth was opened, but the magnitude of the increase differed between the two muscles (Nordstrom and Yemm, 1972; Nordstrom, Bishop and Yemm, 1974). It was also found that at jaw opening of 8 to 12mm between the incisors, the maximum tension was recorded (Nordstrom and Yemm, 1974). Furthermore, histological findings of Van Eijden and Raadsheer (1992) have shown that the sarcomere length of the human masseter muscle is suboptimum at a closed jaw position, ranging between 2.27-2.55 μm. The sarcomere of the anterior and posterior portions of the masseter muscle is evaluated to be about 3.3 and 2.7μm, respectively, at about 10 degrees of mouth opening, which corresponds approximately to an interincisal distance of 15-20mm (Pullinger et al, 1987). These values are similar to the optimal sarcomere length of about 3μm determined in human leg muscle (Walker and Schrodt, 1974). In addition, it is known that the greater the number of cross bridges, the greater the force of contraction, and so the decrease in bite force measured in smaller or larger jaw openings could possibly be due to a reduced number of cross bridges (Manns et al, 1979).

Changes in the functional characteristics of a particular jaw closing muscle resulting from alterations of jaw opening may be due to changes in the relative contribution of the synergist muscles and/or changes in the biomechanical advantage of the muscle (Lindauer et al, 1993).

1.1.3.4 State of the Dentition

Bite forces measured in subjects with natural teeth seem to depend on the state of the dentition and on the type of prostheses that replace missing teeth (Helkimo, Carlsson and Helkimo, 1977). Local pathological conditions of the teeth and the supporting tissues, including caries, pulpitis, periodontitis, tooth mobility and malocclusion often cause a reduction of the maximum bite force (O'Rourke, 1949; Carlsson, 1974).

Laurel (1985) measured masticatory force in patients who were treated for periodontal disease. He concluded that dentitions restored with cross-arch bridges have a tendency to lead to less chewing efficiency and bite force, related to the reduced amount periodontal tissue supporting the abutment teeth.

Most studies of the bite force of complete denture wearers have shown that they have maximum bite force which is about one-third to one-sixth of those with a natural dentition (Worner, 1939; Helkimo et al, 1977; Hellsing, 1980; Haraldson, Karlsson and Carlsson 1979; Michael et al, 1990) (table 1.1). Lundqvist, Carlsson and Hedegard (1986) studied the effects of new or optimally adjusted complete dentures on bite force measurements in two groups of 49 edentulous patients, after two and six months of the treatment. They showed that in the first group there is an increase in maximum bite force values (64N before and 74N after) two months after being given the complete dentures, but in the second group there is a decrease in bite force (75N before and 63N after) when checked after six months. Haraldson, Karlsson and Carlsson (1979) found similar results in a group of denture wearers who received new dentures, and no significant difference between patients with satisfactory and unsatisfactory dentures. They concluded that even clinically satisfactory complete dentures are a poor substitute for natural teeth. Limitations in maximal bite force of denture wearers may be ascribed to masticatory muscle weakness, to the greater reduction of the cross sectional size of the jaw closing muscles in the edentulous subjects, and to the pain of the denture-bearing soft tissue or to tilting of the dentures.

Similarly the maximum bite force produced with removable partial dentures is less than with dentate subjects, but more that those with complete dentures (Hagberg, 1987). Maximum occlusal forces measured on patients with fixed partial dentures were found to be almost the same as in subjects with natural dentitions (Yurkstas et al, 1951; Carlsson, 1974; Lundgren et al, 1975).

Various studies show increased occlusal forces, accompanied by improved function, when edentulous patients were restored with implant-supported overdentures in the mandible (Jemt and Stalblad, 1986; Haraldson et al, 1988). Patients wearing fixed prosthesis supported on osseointegrated implants in the edentulous maxilla showed a significant increase in the maximum bite force during a three years observation period (Lundqvist and Haraldson, 1992). It seems that patients with fixed prostheses supported by implants had similar bite forces to fully dentate subjects (Haraldson, Carlsson and Ingervall, 1979; Haraldson and Zarb, 1988). Regardless of the type of measuring device, forces with implant supported prostheses were increased by a factor 2 to 5 when compared with measurements in complete denture wearers (Haraldson and Carlsson, 1977; Haraldson and Carlsson, 1979).

Investigators	**Number of Subjects (M:male, F:female)**	**State of Dentition**	**Maximal Bite Force (N)**
Linderholm & Wennstrom, 1970	58 M 14 F	Natural teeth (molars)	490 430
Ringqvist, 1973	29 F	Natural teeth (molars)	477
Helkimo, Carlsson & Helkimo, 1977	57 M 68 F	Natural teeth (molars) or complete dentures	382 216
Haraldson, Karlsson and Carlsson, 1979	20 M & F 10 M & F	Complete dentures Natural teeth (premolars)	69 383
Floystrand, Kleven & Oilo, 1982	8 M & 8 F	Natural teeth (molars)	500
Lundqvist, Carlsson and Hedegard, 1986	49 M & F	Complete dentures (old) (group1) (group 2)	 64 75
Bakke et al, 1990	59 M 63 F	Natural teeth (molars)	522 441
Michael et al, 1990	5	Complete dentures	160
Waltimo and Kononen, 1993	22 M 24 F	Natural teeth (molars)	847 597
Waltimo and Kononen, 1995	56 M 73 F	Natural teeth (molars)	909 777

Table 1.1: Maximal bite force values measured in Newtons (N) with strain-gauge or piezoelectric force transducers in subjects with natural teeth or complete dentures.

1.1.3.5. Facial morphology.

A relationship has been established between bite force and facial morphology, as maximum biting force increases with decreasing gonial angle (Moller, 1966; Ringqvist, 1973; Proffit, Fields, and Nixon, 1983; Bakke and Michler, 1991). Proffit et al (1983) found that the maximum biting force in 21 subjects with normal facial morphology was two to three times higher than in 19 long face individuals. Furthermore, Kiliaridis et al (1993) investigated the relation between facial morphology and bite force at different ages during growth in six groups of 136 healthy individuals, and found that the vertical proportions of the anterior face are related to the maximum biting force; growing individuals with a proportionally smaller lower facial height had the highest maximum bite force between the incisors. This is in agreement with the findings of Garner and Kotwal (1973) in a sample of 150 individuals that a positive correlation exists between the incisal bite force and overbite, since individuals with proportionally small lower facial height have deeper overbite.

Whether these observations can be attributed to weaker masticatory muscles of the long face individuals, (Moller, 1966; Proffit et al, 1983) or to mechanical disadvantage (Throckmorton, Finn and Bell, 1980), is still unclear. It has been reported that in long face subjects, the cross sectional area of the masseter, medial pterygoid and temporalis muscles are significantly smaller than the normal subjects (Van Spronsen et al, 1992).

It has been also suggested that there is a greater mechanical advantage for the elevator muscles in subjects with less vertical height of the maxilla and smaller mandibular plane and gonial angle than normal subjects (Throckmorton et al, 1980).

It seems likely that the differences between long face and short face subjects in developing bite force could relate to (1) the cross sectional area of the muscle, (2) the proportion of the different types of muscle fibres; and (3) the mechanical advantage of the elevator muscles.

1.1.3.6 Temporomandibular Disorders (TMD)

The term temporomandibular disorder (TMD) has been defined by McNeil (1993) as "a collective term involving a number of clinical problems related to the masticatory muscles, the temporomandibular joints (TMJ) and associated structures or both". Many alternative names for this condition have been proposed in the past, including temporomandibular joint dysfunction syndrome (Schwarts, 1955), myofascial pain dysfunction syndrome (Laskin, 1969),), masticatory pain dysfunction (Bell, 1969) and craniomandibular disorders (CMD) (McNeil, 1983). Although TMD was previously viewed as one syndrome, current research supports the view that TMD is a cluster of different but related disorders in the masticatory system, and not one disorder (Griffiths, 1983; Bell, 1990). It has been pointed out that TMD may be primarily of muscle origin (myogenous) or primarily joint origin (arthrogenous), perhaps with secondary muscle involvement (Hansson, 1988; De Leeuw et al, 1994).

The most frequent presenting symptom of TMD is pain, usually localised in the muscles of mastication, the preauricular area, and/or the temporomandibular joint (Bell, 1990). In most cases the pain is reported to be unilateral (Christensen, 1981b), although bilateral pain is very common (Weinberg, 1980). The masseter is the most frequent jaw muscle involved (Laskin, 1995).

Epidemiological studies have shown that approximately 50% of the general (non-patient) population are aware of at least one symptom of TMD, with 5% requiring treatment because the condition is a significant problem (Rugh & Solberg, 1985; Agerberg, 1988). The frequency of these symptoms seems to increase with age, and some correlation has been found between symptoms of TMD and the wearing of complete dentures; nevertheless the older complete denture wearers are not always sufficiently disturbed by such problems to seek help (Szenpetery, Fazekas and Mari, 1987; Wilding and Owen, 1987; Mercado and Faulkner, 1991).

The aetiology of TMD is multifactorial, including factors such as parafunction, stress, external trauma and sustained adverse loading (Kopp, 1982; Ash, 1986). Following the introductory work of Schwartz (1955) there was a growing acceptance of the concept of neuromuscular inco-ordination which suggests that there may be an underlying muscle problem (Bell, 1969) for which muscle hyperactivity is the suggested cause (Naeije and Zorn, 1981; Kopp, 1982; Yemm, 1985).

Laskin (1969) ascribed the symptoms of this disorder directly to the muscles and advocated the name "myofascial pain dysfunction syndrome" (the psychophysiologic theory). He proposed that the most common cause was masticatory muscle spasm, the spasm being initiated in one of three ways: muscular overextension, muscular overcontraction, or muscle hyperactivity and the consequent muscle fatigue. Fatigue was thought to be the most important factor, resulting from chronic oral habits such as clenching or grinding the teeth, which generated from psychological stress. It has also been found that after clenching and grinding their teeth and jaws for 30 minutes, previously symptom - free subjects developed pain, stiffness and limitation of jaw movement, symptoms similar to TMD (Christensen, 1971).

For a number of years the major hypothesis for the cause of TMD was that continued hyperactivity led to muscle spasm which resulted in constant muscle fatigue, impairment of blood flow and ischemic pain (Laskin, 1969; Miles, 1978; Laskin and Block, 1986). A notion which is better supported by experimental evidence is that of local mechanical micro-trauma following hyperactivity (Yemm, 1985). Yemm (1985) has concluded that muscle hyperactivity, brought either by central factors (stress) or by peripheral mechanisms (e.g. defective occlusions), is the main cause of TMD. Evidence for the presence of inflammation is provided by the increase in skin surface temperature (Berry and Yemm, 1974), and increase in tissue fluid pressure (Christensen, 1971) which occurs following voluntary tooth clenching.

Patients with TMD have been reported to have lower maximal bite force values than healthy subjects (Molin, 1972; 1973; Markland and Molin, 1972; Helkimo, Carlsson and Carmeli, 1975; Kroon and Naeije, 1992), and the weakness of masticatory muscles has been suggested to be a factor predisposing to TMD (Sheikholeslam, Moller and

Lous, 1980). These weaker muscles would not be functionally or anatomically capable of handling normal loads and certainly not hyperactivity.

An increase in bite force up to normal levels has been reported to follow successful treatment of TMD (Helkimo, Carlsson and Carmeli, 1975), but there are also results contradictory to this (Ow, Carlsson, Jemt, 1989). Subjects with TMD demonstrate no significant differences in bite force between the affected side with muscle pain and the non-affected side (Molin, 1972).

However, in a study by Hagberg, Agerberg and Hagberg (1986), no significant difference could be found between maximum bite force values for a control group of ten healthy women and thirty women patients with pain in the masseter muscles. One reason for this discrepancy in results could be that TMD patients with muscular pain alone participated in this study, in contrast with others investigations (Molin, 1972; Sheikholeslam et al, 1982). Pain in the temporomandibular joints could have a more inhibiting effect on bite force that muscular pain only. Lyons and Baxendale (1995) also found similar bite force values between myogenous TMD patients and controls. A likely explanation for this similarity in bite force was that only a small part of the masseter muscle was affected by the disorder and this small part was the source of the pain. During maximum voluntary effort each subject was possibly able to overcome the protective inhibition of activity and produce a near normal maximal force output; this supports the idea of localised mechanical micro-trauma.

However, it seems reasonable that severe pain should have the effect of inhibiting the bite force. Molin (1973) found that patients with TMD had significantly lower pain threshold compared with healthy individuals. Furthermore, Mallow et al (1980) found that myogenous-TMD patients, have a lower pain threshold and a greater tendency to report pain in reaction to experimental pain stimulation. In fact, the pain does not even have to be associated with muscles or joints to cause a fall in bite force (High et al, 1988).

1.1.3.7. Bruxism.

Bruxism has been defined as the grinding or clenching of teeth during non-functional movements of the mandible, thereby being regarded as parafunctional behaviour (Ramfjord, 1961; Nadler, 1957; Faulkner, 1990). Parafunctional activities such as biting of the tongue, lips or cheeks, or on foreign objects such as pencils, pipe stems, and nails, can also occur while the individual is awake; this behaviour is termed diurnal parafunction (Agerberg and Carlsson, 1973).

A number of clinical signs and symptoms of bruxism have been observed (Ramfjord, 1961; Pavone, 1985):

- non-functional gnashing or grinding of the teeth in the daytime or during sleep
- occlusal sounds during sleep
- masticatory muscle hyperactivity, fatigue and hypertrophy of the masseter muscles,
- increased mobility of the teeth, pulpal hyperaemia and tooth sensitivity,

- functional tooth surface wear, i.e. attrition facets, periodontal disease and occlusal trauma,
- soreness of oral mucosa beneath dentures

Oral parafunctional habits have also been proposed as one of the etiologic factors of TMD and headaches (Ingervall, Mohlin, and Thilander, 1980; Seligman et al, 1988).

Bruxists have been shown to have increased bite force (Helkimo and Ingervall, 1978; Lyons and Baxendale, 1990). Helkimo and Ingervall (1978) reported that subjects with bruxism had higher bite forces at the incisors but not at the molars. They suggested that muscular training due to parafunctional habits could have been performed in an eccentric mandibular position which was reflected as increased biting force only at the incisors.

This possible training of the muscles for biting in specific mandibular positions may explain the contradictory results of a study in a group of twelve year old children by Lindqvist and Rinqvist (1973). They found no significant difference in bite force at the molars between the bruxists and the control groups.

Clarke, Towensend and Carey (1984) also described that the forces of clenching in bruxers during sleep were higher than maximum conscious clenches, while Gibbs et al (1986) reported biting forces in bruxers as six times higher than those of non-bruxers.

It seems reasonable to expect that in cases where extensive attrition has been caused by occlusal parafunction, especially the grinding type of bruxism, the jaw-closing muscles might have been strengthened through repeated dynamic exercise. Waltimo, Nystrom and Kononen (1994) found significantly higher incisal bite force and a high incisal / molar bite force ratio of 63% in a group of patients with rectangular facial morphology and severe dental attrition, caused mainly by nocturnal bruxism. The high biting forces of these patients, especially in the incisal area, could probably be explained by strong masticatory muscles and a mechanically favourable skull morphology which in its turn has been influenced by the surrounding muscles.

1.1.3.8 General Muscle Force, Muscle Training, Gender and Age.

Bite force does not seem to be closely related to general muscle force or body build (Linderholm and Wennstrom, 1970; Linderholm et al, 1971).
Nevertheless, it seems well established that bite force can be considerably increased by jaw closing muscle training and chewing exercise (Carlsson, 1974). A muscle increases in size, strength and endurance in response to exercise.

An experiment by Brekhus, Armstrong and Simon (1941) in which two groups of fifty subjects chewed a cube of paraffin wax for an hour a day for 50 days, showed an increase of 20 - 25% in bite force of both groups after 30 days.
During a one year experimental period, Ingervall and Bitsanis (1987) also found that there was a significant increase of bite force and maximum muscle activity in subjects who chewed a tough material consisting of resin from a pine tree (mastic from the Greek island of Chios). Other evidence of the effect of exercise is provided by the study

from Corruccini et al (1985) about the rural people in North India who eat hard foods, use their teeth as tools, have much less professional dental care and have higher bite force values than people from urban areas.

Many researchers have found a significant correlation between maximum bite force and age and gender, with higher bite force values in men than in women (Garner and Kotwall, 1973; Helkimo et al, 1975; Bakke et al, 1990; Waltimo and Kononen, 1993). Bakke et al (1990) have also demonstrated that age, gender, tooth contact and body height could account up to 30% of the intraindividual variation of maximum bite force. It has been shown that the maximum bite force increases with age from childhood, stays fairly constant from 20 to 40 yr. of age and then declines (Helkimo et al, 1977; Bakke et al, 1990). It seems that the decreased maximum bite force associated with advancing age could be due to age-dependent deterioration of the dentition (Helkimo et al, 1977) and the reduction in cross-sectional area and density of the jaw muscles (Newton et al , 1987).

1.2 EMG AND MUSCLE FORCE

1.2.1 Electromyography

The functional unit of skeletal muscle is the motor unit, which includes a single motor neuron, its axon and the muscle fibres innervated by branches of the axon. Electromyography (EMG) is a method of studying the electrical activity of contracting muscle by recording a summation of motor unit action potentials within range of the recording electrodes (Moller, 1969).

The surface membrane of a resting skeletal muscle fibre displays no differences in electrical potential. However, the inside of the fibre is polarised and it is maintained at a negative potential of about -90mV in relation to the outside. Under normal conditions, a nerve impulse (action potential) propagating in a motor axon activates all the branches of the axon; these in turn activate all the muscle fibres of a motor unit (Paton and Wand, 1967). This action potential causes the release of a neurotransmitter substance called acetylcholine from the axon terminals at the neuromuscular junction. Acetylcholine increases the permeability of the motor end plate to sodium and potassium ions, producing an end-plate potential. The end-plate potential depolarises the muscle fibre membrane, generating a muscle action potential which is initiated over the membrane surface (Hof, 1984).

Following the sarcotubular system, depolarisation triggers the release of

the calcium ions necessary to activate the contractile process (sliding of filaments) and the source of energy (hydrolysis of adenosine triphosphate). Hence the electrical activity of muscle initiates the interaction of the actin and myosin and force development.

The electrical activity produced by muscle action potentials may be detected by pairs of electrodes, over the surface of a muscle or intramuscularly. The raw EMG signal must always be amplified and filtered to display or permanently record it on paper, magnetic tape, or in a computer as the signal is originally generated in microvolts or a few millivolts. Most EMG studies treat the signal waveform by taking the basic positive and negative peaks and rectifying them to one polarity and by calculating the area under the waveform by integration (Miller et al, 1985).

The amplitude of the motor unit action potentials is dependent on the diameter of the muscle fibre, the distance between the active muscle fibre and the detection site, the size of the motor unit and the number of active motor units and the filtering properties of the electrode (Basmajian and De Luca, 1985a). The EMG signal picked up from surface electrodes is critically dependent on their position on the skin relative to the muscle and on the impedance of the skin and underlying tissues (Kramer et al, 1972; Garnick and Ramfjort, 1962). The surface electrodes are best placed approximately 2 cm's apart, from the centre of each electrode, and in the line with the main direction of the muscle fibres (Yemm, 1977b).

Factors such as age, gender, facial morphology, skin thickness, the muscle under study, history of TMD and bruxism also influence the levels of muscle activity recorded by surface electrodes (Carlson, Alston and Feldman,1964; Visser and De Rijke, 1974;

Throckmorton et al, 1980; Sherman, 1985; Poffit et al, 1983; Visser et al, 1995). The EMG activity generated during an isometric contraction has been shown to decrease in amplitude with increasing age, probably due to progressive muscle atrophy (Carlson et al, 1964). The surface EMG amplitude of females has been found higher than that of males producing the same contractile force, implying that women must recruit a larger number of motor units to produce the same force as men (Visser and De Rijke, 1974). Moreover, in studies of isometric contractions of jaw closing muscles, the maximum EMG level tended to be lower in TMD patients than in healthy subjects (Sheikholeslam et al, 1980; Naeije & Hansson, 1986, Visser et al, 1995).

1.2.2 History of electromyography

The history of electromyography was reviewed by Basmajian and De Luca (1985b), where they report that the relationship between muscle contraction and electricity was first observed by Galvani in 1791. However, methods of measuring electrical activity from human muscles remained uninvestigated until the introduction of the metal surface electrode by Piper in 1907, and the needle electrodes by Adrian and Bronk in 1929.

The electromyography has been a valuable method in analysis of the actions of jaw closing muscles for many years (Moyers, 1949; Moller, 1969; Sheikholeslam, Moller and Lous, 1982), and in the study of the physiology of the motor unit activity (Milner-Brown, Stein and Yemm, 1973a; Yemm,1977a, b). The first EMG studies of the muscles of mastication may be attributed to the orthodontist Moyers (1949). The first researchers who used EMG for investigation of patients with temporomandibular disorders were Jarabak in 1956 and Lous, Sheikholeslam and Moller (1970).

Electromyographic investigations of the orofacial muscles were briefly reviewed by Frame, Rothwell and Duxbury (1973). For assessing the activity of masticatory muscles, the researchers have utilised both surface electrode techniques, which find particular application for recording from accessible surface muscles such as masseter and temporalis (Moller, 1966; Besette et al, 1973), and needle electrode methods, which are essential when monitoring activity in deeper muscles such as the lateral pterygoid (Moller, 1966; Molin, 1973).

Cecere, Sabine & Pancherz (1996) studied the effect of relocation of surface electrodes on the reproducibility of the EMG. They found that there was no significant influence of electrodes repositioning upon the reproducibility, which supports the finding of Visser et al (1992) and Pancherz and Winnberg (1981). On the other hand, it has been reported that replacement of the electrodes on the masseter muscle was a significant source of error (Frame et al, 1973; Nouri et al, 1976). It seems that with providing suitable protocols for the site of electrodes and accuracy in the interelectrode distance the reproducibility of EMG recordings is good (Burdette & Gale, 1990; Ferrario et al, 1991).

Electromyographic investigations have been carried out to relate biting force to muscle activity (Carlsoo, 1952; Garrett et al, 1964), to observe muscle activity during mastication (Moller, 1966), to relate mandibular movement to electromyograms

(Ahlgren, 1967), to investigate the load carried by the TMJ (Barbenel, 1969) and to study muscle fatigue (Edwards and Lippold, 1956; Palla and Ash, 1981b; Naeije, 1984) and muscle spasms (Ramfjord, 1961). Furthermore, Yemm (1969 a, b; 1971) was one of the first to investigate the effects of emotional stress on masseter muscle function, and later established a technique for recording single motor unit potentials from the first dorsal interosseous muscle (Milner-Brown, Stein and Yemm, 1973a).

In the last two decades, EMG of the masticatory muscles has been widely used in the investigation of temporomandibular disorders (TMD) and to assess muscle function and dysfunction during rest, biting and mastication (Naeije and Hansson, 1986; Dahlstrom, 1989; Kroon and Naeije, 1992; Cooper, 1996; Cooper, 1997).

1.2.3 Fibre Composition of Masticatory Muscles

During the last twenty years, the traditional anatomic classification of red (slow) and white (fast) muscle fibres (Edgerton and Simpson, 1969) has been extended. Various systems have been proposed based on: histochemical methods in which fibres are classified as types I (slow), IIA and IIB (fast) and IIC (transitional) according to the pH sensitivity profile of their myosin ATPase (adenosine triphosphatase) activity (Brooke and Kaiser, 1970), and on their oxidative capacity and glycolytic activity (Peter et al, 1972). Muscle fibres also may be classified into physiological categories of slow contracting and fatigue resistant (S), fast contracting and fatigue resistant (FR), fast contracting and fatigue susceptible (FF) (Burke et al, 1971).

The histochemical and physiological classification of type I, IIA and IIB, or correspondingly S, FR and FF can be broadly correlated with the anatomic classification of red, intermediate, and white fibres. Type I produce sustained low-level force in tasks such as maintaining posture and have a better developed blood supply, have more mitochondria, and are richer in mitochondria enzymes.

Type IIB fibres are larger in diameter than the type I fibres and lack red myoglobin, are anaerobic, have fewer mitochondria, and quickly fatigue. The fast-twitch muscle fibres tend to contract rapidly for a shorter time and fatigue relatively quickly. The intermediate (type IIA) fibres are a distinctly separate group with intermediate properties and are almost absent from human jaw muscles (Eriksson and Thornell, 1983).

A summary of these classifications has been presented (Van Boxtel et al, 1983):

Type I: Slow-twitch, low ATPase activity, oxidative. These fibres have a greater endurance that other types, i.e. they are fatigue resistant.

Type IIA: Fast-twitch, high ATPase activity, oxidative-glycolitic. Intermediate fatigability.

Type IIB: Fast-twitch, high ATPase activity, glycolitic. High fatigue susceptible.

The jaw-closing muscles have heterogeneous fibre composition, probably reflecting their complicated activity pattern (Eriksson and Thornell, 1983).

Type I fibres predominate, most markedly in almost all parts of the young, adult human masseter, where they make up 62 to 72% of the muscle's total fibre content (Eriksson and Thornell, 1983; Hannam and McMillan, 1994). More that 70% of the fibres in each anterior deep part are type I, but there are about equal proportions of type I and II fibres (mostly type IIB) in the posterior superficial part.

In general, type I fibres predominate in the anterior parts of all the jaw elevator muscles and relatively similar proportions of type I and type II fibres are present in the posterior parts (Eriksson and Thornel, 1983; Hannam and McMillan, 1994).

The sizes of muscle fibres, the fibre types and the ratios of the various types of motor units in masticatory muscles differs from that in limb and trunk muscles (Ringqvist, 1971; Ringqvist, 1974a, b; Vignon, Pellisier and Serratrice, 1980; Eriksson and Thornell, 1983; Hannam and McMillan, 1994). In jaw elevators, there is a marked difference between type I and type II fibre diameters, type II muscle fibres have considerably smaller diameters (and cross-sectional areas) than type I fibres, (Ringqvist, 1974b; Eriksson et al, 1981; Eriksson et al, 1982) but both are small. Large number of ATPase intermediate (IM) fibres (with moderate staining intensity and ATPase activity intermediate to that of type I and type II fibres), and type IIC fibres are found in the jaws, but they are scarce in the limbs (Ringqvist, 1974b; Brooke and Kaiser, 1970; Eriksson and Thornell, 1983). These differences have been found significant mainly in human studies and have often been related to specific functional demands of masticatory muscles (Ringqvist, 1974a, b; Eriksson et al, 1982; Eriksson and Thornell, 1983).

The fibre size and distribution seem to influence masticatory muscle strength during biting and chewing, as strong positive correlations have been shown between the size of type II fibres in the masseter muscle and bite force (Ringqvist, 1974b) and between the area and diameter of type I fibres and the amplitude of chewing activity (Bakke, Stoltze and Tuxen, 1993). It has been suggested that type II fibres are designed for powerful contractions and are activated mainly for strong biting efforts and that the size of type II fibres could be a measure of force (Ringqist, 1974b). It seems that the human masseter shows heterogeneous activation during controlled biting performance (Blanksma, Van Eijden and Weijs, 1992; Blanksma and Van Eijden, 1995). Mao, Stein and Osborn (1992) hypothesise that the preponderance of type I fibres in the anterior part of the masseter permits this part of the muscle, which is close to the molar teeth to have more precise control over dental forces and the maintenance of the jaw posture than could be achieved by the posterior part. The latter, because it contains relatively fewer type I fibres, is supposed to contribute to more forceful , faster but coarser functional acts.

The predominance of type I fibres in the masticatory muscles may explain their higher resistance to fatigue than the limb muscles (Van Steenberghe, De Vries and Hollander, 1978; Clark and Carter, 1985), with recruitment at low force levels (Goldberg and Derfler, 1977). Van Steenberghe et al (1978) suggested that the resistance of the jaw closing muscles to fatigue may be also due to better oxidative capacity of their fibres. However, the near absence of type IIA fibres from human jaw muscles (Eriksson and Thornell, 1983) seems to suggest that they would be readily susceptible to fatigue during sustained effort at stronger force levels because fast, fatigue-susceptible (type IIB) fibres in the posterior parts of the masseter and medial pterygoid muscles must be recruited at these levels.

1.2.4 EMG characteristics During Non-Fatiguing Muscle Contractions

The force developed by a masticatory muscle varies with the length of the muscle, the frequency of recruitment and number of motor units, and the velocity of shortening. The closest correlation between force and electrical muscle activity occurs when the muscle does not change length, for example on intercuspal clenching or biting on the incisors.

A linear relationship has been shown between the EMG activity of human limb muscles, contracting under constant length, and the output force they produce (Lippold, 1952; Milner-Brown and Stein, 1975). In a non fatigued state, a linear relation has also been found between bite force and amplitude of integrated EMG of jaw closing muscles (Ahlgren, 1966; Moller, 1966; Garrett, Angelone and Allen, 1964; Kawazoe, Kotani and Hamada, 1979; Manns et al, 1979; Bakke et al, 1989; Lindauer, Gay and Rendell, 1991). It has been shown, using bipolar intramuscular electrodes, that the integrated EMG activity increases in proportion to the activity level in the anterior, middle and posterior regions of the temporalis muscle (Ahlgren, Sonesson and Blitz, 1985).

However, non-linear EMG-force relations have been observed for both limb muscles (Lawrence and De Luca, 1983; Woods and Bigland-Ritchie, 1983) and jaw-closing muscles (Hagberg, Agerberg and Hagberg, 1985; Wastell and Devlin, 1987). Moreover, using bipolar hook electrodes, the relation has been reported to be linear for the anterior temporalis but not for the masseter muscle (Haraldson et al, 1985). During conditions of changing force it has been found that the relationship between EMG and force in the masseter muscle is not simply linear, but depends on the rate of change of force (Devlin and Wastell, 1985).

Although all these investigations suggest a complex EMG-force relation, there seem to be a linear relationship between integrated EMG and isometric force at submaximum levels, (Pruim, Bosch and De Longh, 1978; Hosman and Naeije, 1979). At high contraction levels a deviation of this linear behaviour is observed showing a faster increase of the EMG activity as a function of the force (Pruim et al, 1978; Hagberg et al, 1985).

The presence of complex relationships between EMG and muscle force can be due to physiological and/or anatomical differences of motor unit organisation (Woods and Bigland-Ritchie, 1983) rather than differences in the experimental conditions (Moritani and De Vries, 1978). The physiological phenomena contributing to the EMG/force relationship are the distribution and quantity of slow twitch and fast twitch fibres within the muscle, electrical cross-talk from adjacent muscles, or co-contraction of agonist and antagonist muscles (Lawrence and De Luca, 1983). Motor unit recruitment patterns and firing rate properties within a muscle can also alter the EMG/force relationship.

There are two ways of controlling muscle force: changing the number of active motor units (recruitment) and changing the firing frequency of the active motor units (firing rate). Generally, at the beginning of a contraction up to 30% MVC, motor unit recruitment is the dominant factor, progressively larger motor units being recruited as the force increases. For force levels above 30% MVC the dominant factor is the increase in firing rate; above 75% little recruitment occurs (De Luca, 1979). During maximal voluntary contraction, all motor units tend to be active and to respond with fully fused tetanus (Bigland-Ritchie, 1981a).

Jaw closing muscles contain different proportions of fast and slow fibres. Henneman's size principle states that when a muscle contracts with increasing force, slow motor units are recruited first, followed by fast motor units (Henneman, 1957). This orderly recruitment pattern has been shown to exist in jaw muscles (Yemm, 1977a; Goldberg and Derfler, 1977). The fast fibres in jaw muscles are usually recruited for stronger forces. Goldberg and Derfler (1977) studied the masseter muscles of 11 adult subjects using fine stainless steel wires inserted as pairs into the muscle. The smallest amplitude spikes were recruited with the jaw at rest and with low bite forces, and spikes with larger amplitudes were recruited as the biting force increased.

As assessed from electromyographic and bite force recordings, the relative contribution to the clenching force from each of the masticatory muscles is about 35% for the masseter, 30-50% for the temporalis and about 20-40% for the medial pterygoid muscle. The temporalis appears to be most predominant in slow contractions, and the medial pterygoid in brisk contractions (Desmedt and Godaux, 1979).

1.2.5. Masticatory Muscle Fatigue

Muscle fatigue, has been defined as a 'decreased force-generating capacity or the inability to maintain the required or expected force' (Edwards, 1981; Hainaut and Duchateau, 1989). According to Maton et al (1992), muscle fatigue is 'a physiological and biochemical process of the neuromuscular system that is usually defined ergonomically in terms of the point in time (failure point) when a given muscle or group of muscles is unable to maintain a constant force'. Generally, muscle activity leads to a decline of force production and speed of contraction which is known as fatigue (Westerbland et al., 1991). Additionally, localised muscular fatigue has been defined by Chaffin (1973) as an inability to maintain a desired force output, with augmented muscular tremor and localised pain. Endurance, on the other hand, is the ability of muscles to withstand prolonged strain; that is, to resist muscle fatigue.

Two types of muscle fatigue are commonly described; central fatigue and peripheral fatigue. Central fatigue involves a reduction of motor volleys from the motor cortex, accompanied by a loss of concentration or effort by the subject. Peripheral fatigue is due to a failure of transmission at the neuromuscular junction, or of the muscle action potential or of the force generation capacity of the fibre (Westerbland et al., 1991). During prolonged periods of muscle activity, a submaximum force can only be sustained with increased effort. This is usually accompanied by an increased EMG/force ratio and is commonly regarded as an early indicator of fatigue.

A muscle can be fatigued by voluntary contractions or by electrical stimulation of the motor nerve or the muscle itself. Electrical stimulation can produce two types of peripheral fatigue; high frequency fatigue, which corresponds to synaptic-transmission fatigue and is the result of stimulation above approximately 80 Hz, and low-frequency fatigue, which corresponds to contraction fatigue and is the result of stimulation below approximately 20 Hz (Mao, Stein and Osborn, 1993). At high frequencies (80-100Hz in man) force declines rapidly within 30 seconds, and the fatigue is quickly induced and recovery is rapid. Transmission fatigue is present when a parallel decline in both force output and EMG amplitude occurs (Bigland-Ritchie, 1981b). During low frequency stimuli (20Hz) the force can be maintained longer, fatigue takes a longer time to develop and recovery is slow. Contractile (or neuromuscular) fatigue is present when the amplitude of the EMG activity is unchanged while the force declines, or the EMG activity is increased in amplitude in order to maintain a given force level unchanged in a sustained isometric contraction (Edwards and Lippold, 1956; Merletti et al, 1990).

When unilateral bite force was measured with a transducer, the force gradually declined and EMG activity increased in the ipsilateral masseter muscle and decreased in the contralateral (Haraldson et al, 1985). When 40% incisal bite force was sustained the EMG activity decreased in the masseter muscle and increased in the temporalis (Hellsing and Lindstrom, 1983).

However, it has been also shown that during a sustained contraction of the masseter muscle at 25-100% force level, the average EMG amplitude and bite force did not decrease for any force level (Clark and Carter, 1985). Additionally, Clark et al (1988) found that the sum of the EMG amplitude of masseter and anterior temporalis muscles divided by the bite force (EMG/force ratio) during sustained contractions at various levels of force, remained constant. They suggested that the masticatory muscles may not be susceptible to contraction failure. The study of Van Steenberghe et al (1978) showed that the jaw closing muscles were more fatigue resistant than were the muscles of the upper limbs in repeated transient maximal isometric contractions.

Christensen (1981a), investigated the relationship between experimental tooth clenching and the resulting muscle fatigue and facial pain in healthy subjects. He concluded that maximum voluntary contractions of the jaw closing muscles, with the mandible in the position of maximum intecuspation of the teeth, can induce fatigue after 0.5 minute of clenching while onset of pain in the jaw muscle occurred after about 1.0 minute of clenching. It has been found that extensive function and hyperactivity, by maximal voluntary teeth clenching, lead to pain and dysfunctional symptoms in healthy subjects, probably due to fatigue (Christensen, 1981a, b; 1989; Bowley and Gale, 1987; Clark, Jow and Lee, 1989).

Another method of demonstrating neuromuscular fatigue is to measure any shift in the EMG frequency spectrum (Naeije and Zorn, 1981; Palla and Ash, 1981b; Clark et al, 1988). The median frequency is defined as the frequency at which power spectrum is divided into two regions containing equal power. The mean frequency is the average frequency and the mode frequency is the frequency at which peak energy is found in the spectrum. The median frequency has been shown to be a theoretically more reliable estimator than the other convenient parameters, as it is less susceptible to noise (Stylen and De Luca, 1981). Changes in the EMG frequency spectrum in the jaw closing muscles during a fatiguing process have been studied during sustained isometric contractions, and demonstrate a progressive shift of the mean or median power frequency (MF) to a lower level (Palla and Ash, 1981b; Lindstrom and Hellsing, 1983; Clark et al., 1988; Kroon and Naeije, 1992; Lyons et al, 1993). In addition, the MF shifted to lower frequencies more rapidly at stronger contraction levels (Kroon, Naeije and Hansson, 1986). As a consequence, the mean and median frequency of the power density spectrum provide a reliable and consistent measure of fatigue (Stylen and De Luca, 1981) and have been used as an objective index of local fatigue.

Frequency decrease and EMG amplitude increase as fatigue occurs may be due to a recruitment of additional motor units (Edwards and Lippold, 1956), or synchronisation of motor unit action potentials (Milner-Brown, Stein and Lee, 1975; Naeije and Zorn, 1982), or a decrease in conduction velocity of the muscle fibre action potentials along the membrane (Lindstron, Magnuson and Petersen, 1970; Lindstrom and Hellsing, 1983). The two explanations most widely accepted are muscle fibre conduction velocity reduction and hence a longer time duration of the motor unit waveforms, and synchronisation of motor units (De Luca, 1985).

Moreover, when skeletal muscle fatigues, the amount of force that can be developed from an isometric contraction decreases and the rates of force development and of relaxation are slowed. The rate of relaxation from an isometric contraction has long been recognised to decrease with fatigue (Jewell & Wilkie, 1960; Edwards, Hill & Jones, 1972; 1975a; Bigland-Ritchie et al, 1983). It is well established that during relaxation, force decays exponentially during the period following its maximum rate of change (Edwards, Hill & Jones, 1972; 1975a; Bigland-Ritchie et al, 1983). The half time of the later part of this exponential phase was found to increase two to threefold as a result of fatiguing voluntary contractions (Jones & Round, 1990).

The reasons that a slowing of relaxation occurs are still not completely understood. It is likely that two processes are involved: a reduced rate of dissociation of cross-bridges after the removal of the activating calcium back into the sarcoplasmatic reticulum (Edwards, Hill & Jones, 1975b; Cady et al, 1989) or a reduced rate of calcium pumping by the sarcoplasmatic reticulum (Dawson, Gadian & Wilkie, 1980; Cady et al, 1989).

It has also been shown that there is a positive correlation between endurance time at 50% of the maximal contraction and the percentage of the slow twitch, type I fibres in the human leg muscles (Viitasalo and Komi, 1978; Hulten et al, 1975). Since it is known that the jaw closing muscles are predominated from type I, fatigue resistant fibres, the endurance time has been used as an indicator for the resistance to fatigue of the jaw muscles (Naeije, 1984; Dahlstrom, Tzakis and Haraldson, 1988). During a sustained voluntary clenching at 50% of MVC, the mean endurance time have been found ranging from 47 to 270 seconds, in healthy dentate subjects (Naeije, 1984; Dahlstrom et al, 1988; Lyons & Baxendale, 1990). In a normal population, the endurance time is inversely related to bite force level; a decrease in bite force is accompanied by an increase in endurance time (Maton et al, 1992).

1.3 ACOUSTIC MYOGRAPHY (AMG)

The existence of a detectable, low frequency sound on the skin surface over a contracting muscle is a well known phenomenon. Herroun and Yeo, in 1885, noted that the sounds, produced by voluntary and electrically stimulated contraction of muscles were identical. Individual muscle fibre contractions were detected by Gordon and Holbourn (1948), using a small piezo-electric microphone placed on orbicularis oculi.

Using an electronic stethoscope, Oster and Jaffe (1980) found that the dominant acoustic frequency was 25±2.5Hz. They also demonstrated that human muscle sounds were from muscular activity and not due to blood flow or artefacts such as tremor on the microphone scraping against the skin. The low frequency range of the AMG signal has been also confirmed by other studies (Rhatigan et al, 1986; Wee and Ashley, 1989; Dalton and Stokes, 1993).

The precise aetiology of muscle sounds is still unclear, although it seems that it is due to the activity of single muscle fibres, particularly fast twitch fibres (Oster, 1984). A number of suggestions have been made, including that the sounds were produced by the thickening of muscle fibres during contraction (Gordon & Holbourn, 1948), by the elastic connective tissue at every individual contraction (Rhatigan et al, 1986) and a gross lateral movement of the central regions of the muscle (Frangioni et al, 1987). Recently, there is increasing evidence that lateral movements of muscle fibres during contraction produce the low frequency (under 100Hz) sounds (Barry, 1987; 1990; Wee & Ashley, 1989).

Many different types of transducers have been used to record muscle sounds. The most important characteristic of the recording apparatus is the frequency response. This has to be sensitive to frequencies between 1 and 100Hz since almost all the AMG signal is in this range (Bolton et al, 1989). Large piezoelectric contact transducers such as the Hewlett-Packard 21050-A, weighing 44 grams, have been used over muscles with greater mass (Orizio, Perini, Veicsteinas, 1989a, ; Wee and Ashley, 1989).

Under certain conditions, AMG activity reflects changes in muscle force but this relationship still requires investigation. Recent studies of AMG indicated that the relationship between the amplitude of AMG and the level of force production, during isometric contractions, appears to vary in different muscles. However, it is generally agreed that AMG shows a positive correlation with increasing force (Barry et al. 1985; Orizio et al, 1989a; Dalton & Stokes, 1991; Stokes & Dalton, 1991b). Studies also have shown that the relationship between force and AMG does not alter in fatigued muscle (Barry et al, 1985; Stokes & Dalton, 1991a).

The reproducibility of AMG of the masseter muscles within the same session has been shown to be good and the technique practical (L' Estrange, Rowell & Stokes, 1993). Acoustic myography may therefore be of increasing interest and may be potentially useful for examining muscle function, but the use of AMG on the jaw closing muscles has not been fully tested.

1.4 SUMMARY OF REVIEW OF LITERATURE

Jaw closing muscle function is often evaluated by measurement of bite force. However, previous studies relating to bite strength with natural and artificial dentition have shown a wide range of forces (Carlsson, 1974; Helkimo et al, 1977; Hagberg, 1987).

The great variation in values reveals that recording of maximum bite force is dependent on many factors related to the anatomical and physiological characteristics of the volunteers, to the different measuring methods and to different location of measurements on the natural teeth or the prostheses (Carr and Laney, 1987). Psychological affects, pain threshold, increased jaw separation, replacement of natural teeth with complete dentures, long facial morphology, age related muscle atrophy, jaw muscle pain and symptoms of TMD have mostly been considered to be factors of limitation of maximum bite force generation capacity.

A variable which also should be considered when measuring bite force is the position of the force transducer within the dental arch (Leff, 1966; Waltimo and Kononen, 1994). Different positions i.e. unilateral, bilateral, anterior, posterior will influence which muscles involved in the force production, the protective responses which may arise in periodontal receptors, the supportive structure, the morphology and the number of teeth loaded during the biting action, and this requires further consideration.

EMG has been widely used as a method of monitoring jaw muscle activity (Dahlstrom, 1989), although EMG measurement reliability was found to be dependent on the effects of electrodes placement, the impedance of the skin, subcutaneous fat and the depth of the muscle under study (Mohl et al, 1990; Widmer et al, 1990). It has been shown that the EMG activity of the jaw closing muscles, under constant length, increases proportionally as bite force increases (Bakke et al, 1989), although deviation from this linearity was observed on the masseter muscle (Haraldson et al, 1985). Several authors have also found that AMG during isometric muscle contraction may be used as an indicator of muscle activity, alternatively to EMG (Barry et al, 1985; Stokes & Dalton, 1991a). However, AMG on the jaw closing muscles is a more recent development (L' Estrange et al, 1993).

In spite of the fact that the jaw closing muscles have a higher resistance to fatigue than the limb muscles (Van Steenberghe et al, 1978; Clark and Carter, 1985), clinical problems thought to be related to fatigue, i.e. TMD, are relatively common (Rugh & Solberg, 1985; Mao et al, 1993). Although the aetiology of TMD is considered to be multifactorial, it is not well understood. Epidemiological studies on the prevalence of TMD have shown that jaw muscle pain and fatigue are frequently reported in all age groups (Salonen, Helden and Carlsson, 1990).

Bite force and several EMG parameters have been measured to investigate the functioning of jaw closing muscles in a search for neuromuscular factors involved in the aetiology of TMD. Power spectral frequency changes clearly occur during a fatiguing task, and are mainly due to changes in muscle fibre conduction velocity. The median frequency is a suitable estimator of the power spectrum.

Slowing of relaxation from a sustained contraction is probably due to reduced rate of calcium ions uptake by the sarcoplasmatic reticulum or to reduced activity of the actomyosin cross-bridges, although the precise mechanism is not completely understood.

It has been shown that bite force and EMG activity are considerably reduced in completely edentulous denture-wearing subjects but the susceptibility of their jaw closing muscles to localised fatigue is less certain. This information is even less clear for complete denture wearers with history of jaw muscle pain and symptoms of TMD.

1.5 AIMS OF INVESTIGATION

The aims of this investigation were to evaluate jaw closing muscle function by measuring bite force and electromyographic parameters in young healthy dentate subjects and in older edentulous patients, during isometric contractions and during fatiguing tasks.

The specific aims of this study were as follows:

1. To investigate the variability of maximum bite force measurements in three different force transducers positions in the mouth, on different occasions and to determine the reliability of these measurements.

2. To investigate the relationships between the EMG, AMG amplitude in masseter muscles and bite force, and to compare these two indirect measures (electromyography and acoustic myography) of activation of the masseter muscle.

3. To determine maximum bite force and endurance time in healthy complete denture wearers and complete denture wearers with history of TMD. Further, to evaluate the shift in EMG median frequency and the changes in the relaxation rate before and after a fatiguing task as indicators of the fatiguability of the masseter muscles in these two edentulous groups.

CHAPTER 2

GENERAL MATERIALS AND METHODS

2.1 Ethical Committee Approval

Approval was obtained from the Area Dental Ethics Committee of the Greater Glasgow Health Board prior to the experiments.

2.2 Bite force transducers

Bite force was measured using four different patterns of stainless steel bite force transducer (see chapter 3 and 5). The force transducers were based on a design which has been described in detail previously by Lyons and Baxendale (1990).

The unilateral posterior transducer (70mm long, 16mm wide, 8mm thick) was made from two parallel stainless steel beams and separated by a stainless steel spacer. The three parts were held together by two stainless steel bolts. In order to measure forces applied to the bite fork, two strain gauges were attached to each side of one of the beams with flexible, rapid setting epoxy resin (Araldite Rapid) and wired to form a Wheatstone bridge circuit. A Wheatstone bridge circuit is commonly used for the rapid and precise measurement of resistance. The strain gauges were compensated for mild steel, 8mm long, and with a resistance of 120Ω (RS components Ltd, Corby, Northants, U.K.). The part of the beam with the strain gauges attached was coated with a silicone rubber compound in order to effect a watertight seal. The strain gauges have to be effectively sealed from the oral cavity to prevent electrical short circuits which occur when these gauges are exposed to saliva (Stafford and Glantz, 1991). When a load was applied to the beams, the mechanical deformation altered the resistance of the strain gauges and the change in signal voltage was used to provide a measure of force.

The anterior transducer (63mm long, 19mm wide, 9mm thick) was constructed in the same manner but one end of both beams were extended to provide a large biting area. This enabled contact of all incisor and canine teeth with the transducer beams.

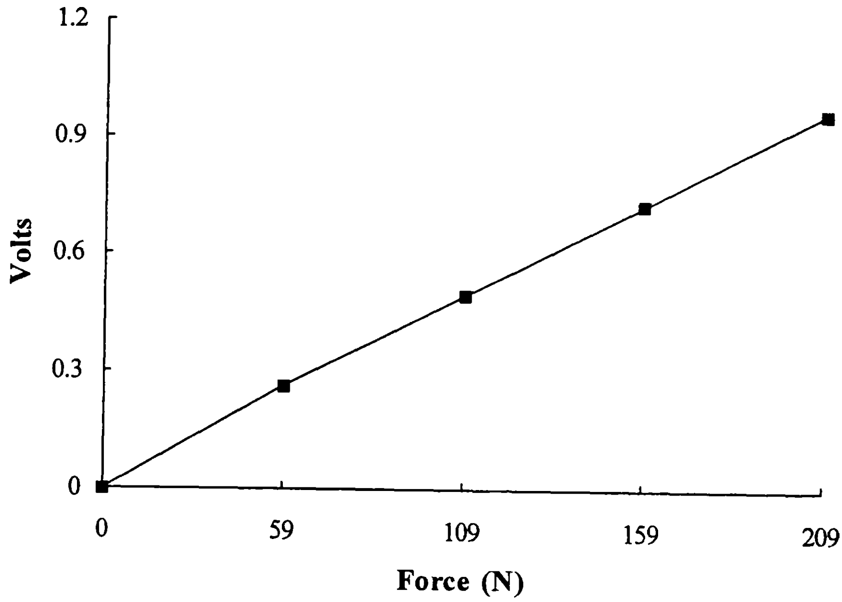
The bilateral posterior transducer (65mm long, 20mm wide, 10mm thick) consisted of two stainless steel beams with only two strain gauges attached to one beam and the other two strain gauges were replaced by two resistors to balance the bridge circuit. These resistors were located away from the transducer beams.

The T-shape bilateral transducer (80mm long, 20mm wide, 8mm thick), specially designed for edentulous patients, consisted of two stainless steel beams cut to a T-shape. Two foil strain gauges were cemented on each side of the long arm of one of the beams and wired to form a Wheatstone bridge circuit.

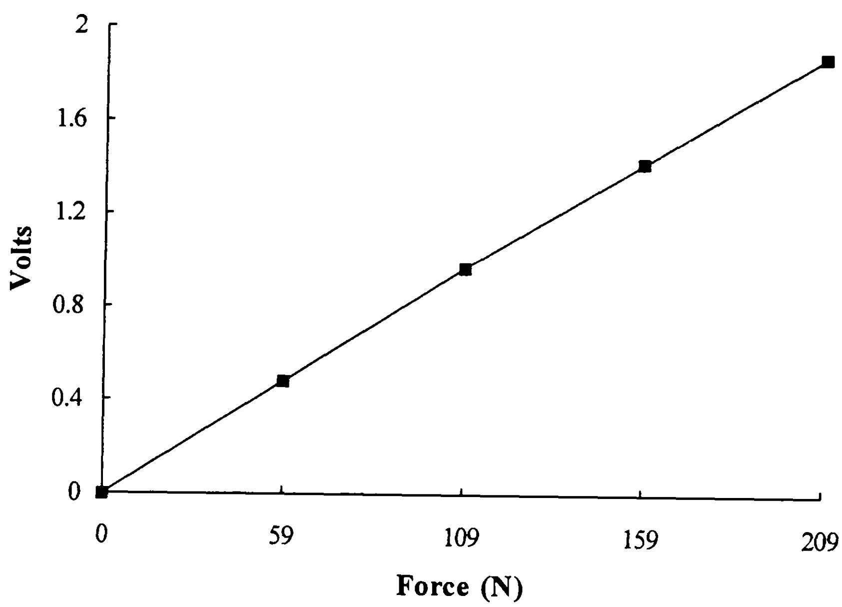
In each transducer the strain gauges were connected to a 2m four-core braided cable (NL 953, Digitimer Ltd). This cable is supplied with a male connector on one end (Lemo F00304) which mates with the input socket of the recorder amplifier (NL107, Digitimer Ltd, UK). This differential amplifier was used as a bridge amplifier and had an integral power supply.

The bite force transducers were calibrated with standard weights before each experiment. The calibration was calculated in Newton ($1kg \cong 9.81N$). The response of the instruments was found to be linear in the range (0-209N) tested (Figure 2.1), and consistent between sessions.

The bite force level was displayed on an oscilloscope screen (Tektronix 5103N, Tektronix Inc., Beaverton, OR 97077, USA) for visual feedback and stored on video tape cassettes with the use of a PCM-8 A/D video recorder adapter (Medical Systems Corp., Greenvale, NY 11548, USA) for later analysis.



(A)



(B)

Figure 2.1 The calibration curves of the strain gauge (A) anterior transducer, and (B) T-shape bilateral transducer.

2.3 Electromyogram

2.3.1 Type and location of electrodes

EMG was recorded using surface metal foil electrodes (Littman 2325 VP 3M Ltd). These are disposable single use, pre-gelled adhesive electrodes for diagnostic purposes. A bipolar electrode configuration was used in every case, with a ground electrode being placed on the ear lobe or forehead. It was necessary to ground, or earth, subjects when recording EMG in order to reduce unwanted noise. To reduce skin impedance the skin was vigorously prepared using gauze soaked in surgical spirit.

EMG recordings were made for the masseter muscle bilaterally or unilaterally. The masseter is not only easily accessible for surface EMG but is also one of the most powerful of the masticatory muscles. It is also believed to be a common muscle involved in myogenous TMD (Laskin, 1969; 1995).

The area of the masseter muscle from which recordings were taken was the lower anterior part of the main belly of the muscle determined by palpation (Greenfield & Wyke, 1956). Roy, De Luca and Schneider (1986) reported that the relatively high impedance of the tendon tissue truncates the action potential; the superficial layer of the masseter has tendon tissue in the upper half of the muscle. Therefore, the masseter has more actual muscle fibre in its lower half. The electrodes were also positioned parallel to the main direction of the muscle fibres since Ahlgren (1966) found that in the masseter muscle more electrical activity was registered when the electrode pairs were orientated parallel than when orientated across the muscle fibres.

2.3.2 Amplifier specifications

The amplifiers used in the experiments were components of the Neurolog system (Digitimer Ltd), and consisted of the NL824 4--channel AC pre-amplifier and NL 820 isolator amplifier. The general layout of equipment may be seen in Figure 2.2.

The raw EMG signal was led from the electrodes by paired wires to the four channel, low noise, differential AC preamplifier (NL 824, Digitimer Ltd) for noise elimination, where it was amplified at a gain of at least x2000, depending on the subject. The lower cut-off frequency was set to 3 Hz and the higher cut-off frequency was >10kHz. The preamplifier was placed close to the volunteer so as to keep the wires from the electrodes as short as possible. This was desirable to minimise electromagnetic interference.

An isolator amplifier (NL 820, Digitimer Ltd) was also used as it provided both amplification of the signal (at least x2000) and power supply isolation to human subjects. The input impedance was 10KΩ, noise 4mV at 150 kHz.

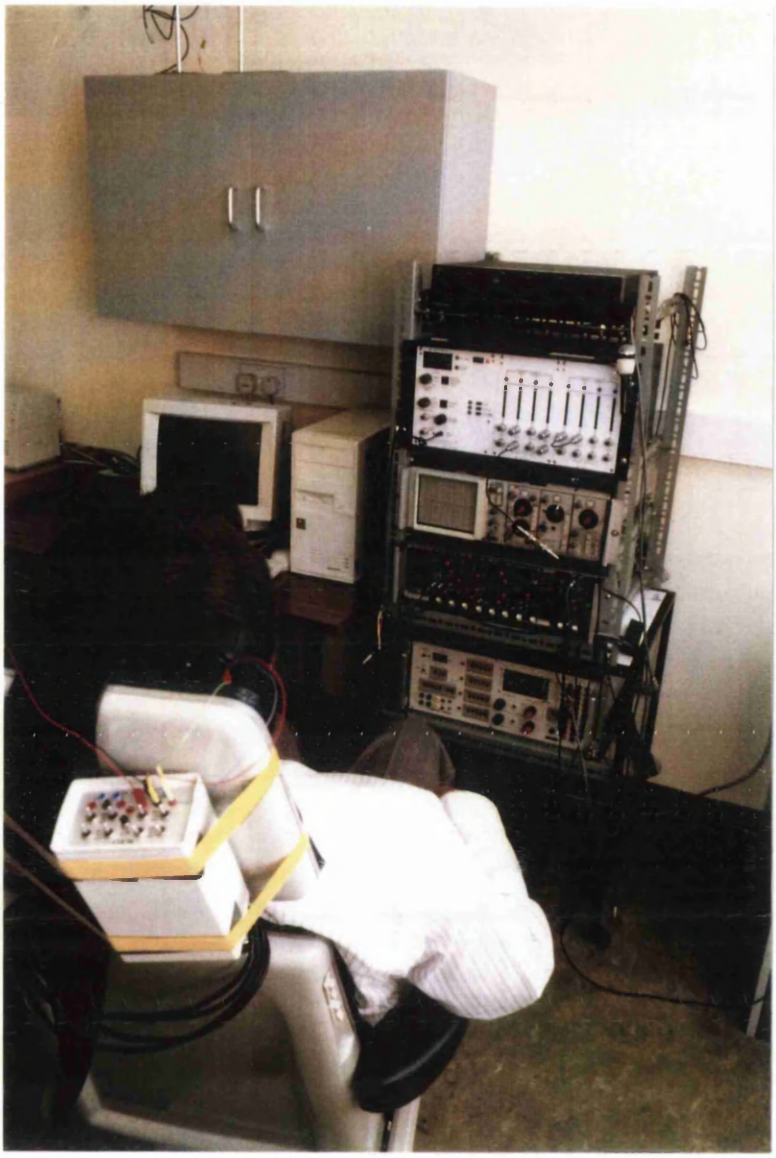


Figure 2.2 General layout of equipment in the research lab. The PCM-8 adapter may be seen second from top in the rack.

2.3.3 Signal Storage

Signal storage was obtained by the use of a PCM-8 recorder adapter (described previously in page 74). This enabled up to eight channels of data to be stored on high quality video tape cassettes. The PCM-8 operated by multiplexing analogue inputs through a single analog-to-digital converter into a digital data stream. This digital data was then modulated by a video carrier in a format compatible with a standard VHS video and the encoded data was fed to the video input of a video recorder. Retrieval of data was obtained by playing the video tape back through the PCM-8. The signal was decoded, D\A converted, demultiplexed, and presented at the analogue outputs of the PCM-8 at the same amplitude as recorded. There was also an audio channel to record and replay comments made during the experiment.

CHAPTER 3

THE VARIABILITY OF BITE FORCE MEASUREMENT

3.1 SUMMARY

The effect of measuring bite force with different patterns of transducer on different occasions was studied. Maximum voluntary bite force was measured in 8 healthy volunteers. Three transducer positions, each with a different pattern of transducer, were used; between the anterior teeth, between the second premolar and the first molar on one side and between the second premolars and first molars bilaterally. Visual feedback of force was provided. Two sets of five maximum clenches were recorded with a rest period in between. This sequence was repeated for each transducer and the experiment was repeated on three different days.

The highest forces were measured with the bilateral posterior transducer (mean 579 N, SEM± 83.2) and the lowest on the anterior transducer (mean 286 N, SEM±58.1).
The standard deviations of the bite force mean values were used as an indication of the variability and were subjected to a non-parametric ANOVA (Kruskal-Wallis). The forces recorded with each transducer position were significantly different between the transducers ($P < 0.01$) and the maximum bite force showed least variability when measured between the posterior teeth on one side only. There was little difference in bite force between the three different sessions ($P \geq 0.05$) when measured in the same position within the dental arch, whichever of the three positions that may be.

3.2 INTRODUCTION

It is well known that the jaw closing muscles are required to provide force for regular and repetitive chewing movements, occasional heavy biting, and fine positioning of the mandible. The maximum voluntary force output of the jaw-closing muscles is frequently measured for research purposes, either as measure of maximum force output *per se* or to compare percentage muscle activity between individuals. However, bite force measurement is notoriously difficult and the reliability of the result depends on a number of factors (discussed in Chapter 1), such as lack of motivation of the subjects, fear of breaking cusps of teeth and dental restorations, the design and comfort of the transducer and the position of the transducer within the dental arch.

Previous studies relating to bite strength with natural dentitions and/or fixed or removable prostheses, using different measuring instruments under varying test conditions have shown a wide range of forces (Hagberg, 1987). In a study of ten young females the mean maximum bite force between the molar teeth was 396N (Hagberg, Agerberg and Hagberg, 1985). Dahlstrom et al (1988), using a strain gauge transducer, reported that the mean maximum bite force recorded in the premolar/molar region was 661N, in a healthy group of men and women. In another study of Waltimo and Kononen (1993), using a quartz force transducer, the mean maximum occlusal forces for both men and women were remarkably high for both molar (847N for males and 597N for females) and incisal regions (287N for males and 243N for females). Similarly, considerably higher bite forces were measured in the molar region (909N for men, 777N for women) than in the incisal area (382N for men 325N for women) in a healthy large sample of 129 young adults (Waltimo and Kononen, 1995). Some of these variations may be due to the different physical attributes of the different populations and some to the measuring instruments and techniques.

It seems well established that the bite force varies with region in the oral cavity, being greatest in the first molar area and only about one-third to one-fourth of that when measured between the incisors (Carlsson, 1974; Helkimo et al, 1977; Waltimo & Kononen, 1993).

It is also likely that the degree of jaw opening, hence muscle length, is important in influencing maximum bite force. It has been suggested that the highest bite force is achieved when the inter-occlusal space in the canine-molar region is 9-20mm (Manns et al, 1979; Mackenna & Turker, 1983; Lindauer et al, 1993). Moreover, Fields et al (1986) have reported that at a given jaw opening, changes in head posture did not significantly influence the bite force values, since head posture does not directly affect the jaw closing muscles, but might affect the activity and orientation of the depressors muscles of the mandible. Therefore, they have suggested that the head posture must be controlled due to its interaction with the degree of jaw separation, during bite force measurement.

With some of these influencing factors in mind, it is understandable that even with similar methods and measuring devices, variability in bite force between subjects of the same population could be expected.

Therefore the aims of the present study were:

- to measure maximum bite force in three different force-transducer locations within the dental arch, on different occasions, and
- to determine the variability of these bite force measurements.

3.3 MATERIALS AND METHOD

3.3.1 Experimental subjects

Participants in the study consisted of 8 male volunteers with ages ranging from 25 to 32 years, mean age 29 years. All the subjects had a full complement of natural teeth, and no pain or clicking sounds were present in the temporomandibular joints or associated muscles. All of the subjects were fully informed of the procedure prior to the experiment, and they gave informed consent.

3.3.2 Recording protocol

The maximum voluntary bite force was measured in three different positions using three different patterns of stainless steel bite force transducer which has been previously described in Chapter 2 (Figure 3.1). The unilateral transducer was placed between the second premolar and first molar teeth on one side, the bilateral transducer was placed between the second premolars and first molars of both sides. The anterior transducer was placed in the midline between the upper and lower incisors and canine teeth with the mandible in a protrusive position.

Small acrylic resin indices were made for each subject with the teeth just in contact with the metal faces of the transducers to minimise the risk of fracturing teeth and restorations and to make heavy biting on the transducer as comfortable as possible. These indices were removed from the transducers after each experiment and kept for the next experiment. In this way the position of the transducer was exactly the same for each individual, for each session. The total thickness of the transducers was between 8-10 mm, which gave a jaw separation of 11mm when measured from the distal borders of the canines.

The bite force transducer was calibrated with known weights before each experiment. The response of the instruments was found to be linear in the range tested, and consistent between sessions. The bite force level was displayed on an oscilloscope screen for visual

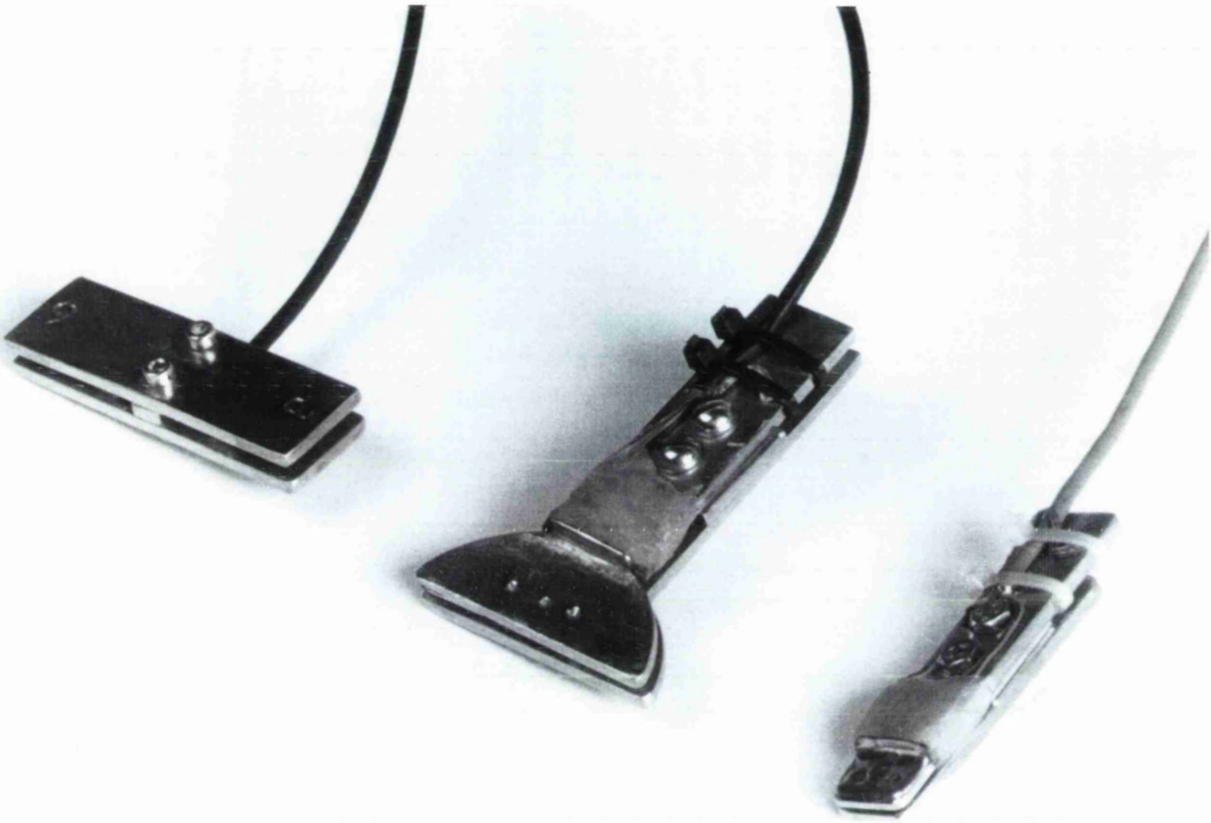


Figure 3.1 The three patterns of bite force transducers which were used A) Posterior Bilateral, B) Anterior and C) Posterior Unilateral.

feedback and stored on video tape cassettes with the use of a PCM-8 video recorder adapter for later analysis (see Chapter 2).

The procedure was explained to the volunteers and they were allowed to become familiar with the force transducer and the visual feedback on the oscilloscope screen. The subjects were seated upright in a dental chair and with head support. Each volunteer was asked - and verbally encouraged - to produce five maximum clenches. A five minute rest was taken and then this sequence was repeated; the same procedure was followed for each of the three different force transducers. One recording session was carried out for each subject on the same day each week for 3 consecutive weeks.

3.3.3 Signal processing and statistical analysis

The bite force signals from the three recording sessions were subsequently played back through the PCM-8 A/D adapter. A computer interface (1401+, Cambridge Electronic Design, Cambridge, UK) was used to digitise and transfer the data to a PC (Viglen Ltd, London, UK). Signal processing was carried out using a signal analysis software package (Signal Averager, Cambridge Electronic Design, Cambridge, UK).

In order to measure the variability of bite force between the three transducer positions and between the different sessions, a non-parametric ANOVA was performed. Non-parametric tests (Kruskal-Wallis one way analysis of variance by ranks) were used because of the small number of participants and the doubt about the constancy of variance and the existence of a normal distribution. The Kruskal - Wallis test is based on ranks of measurements (ordinal data and qualitative data that are ranked) and is one of the most powerful of the non-parametric tests.

3.4 RESULTS

A) It may be seen that the maximum voluntary clenches were relatively consistent (Figure 3.2). The maximum bite force was highest when measured with the bilateral transducer between the posterior teeth (mean 579 N, SEM±83.2), lower when biting on the unilateral transducer between the posterior teeth on one side only (mean 428 N, SEM±46.8) and least when measured with the anterior transducer between the anterior teeth (mean 286 N, SEM±58.1). (See Table 3.1).

Subject	Unilateral	Anterior	Bilateral	Mean
1	551.3	555.5	702.1	603.0
2	599.5	268.3	1039.3	611.3
3	348.5	222.9	365.6	311.4
4	241.8	185.8	269.4	231.6
5	432.9	170.0	587.2	403.0
6	573.7	529.8	646.9	584.9
7	338.3	120.8	457.3	308.5
8	341.4	240.2	566.4	382.7
Mean:	428.4	286.7	579.3	
SEM	46.8	58.1	83.2	

Table 3.1. Mean maximum bite force values (N) over all three sessions for the unilateral, anterior and bilateral transducer in all eight subjects.

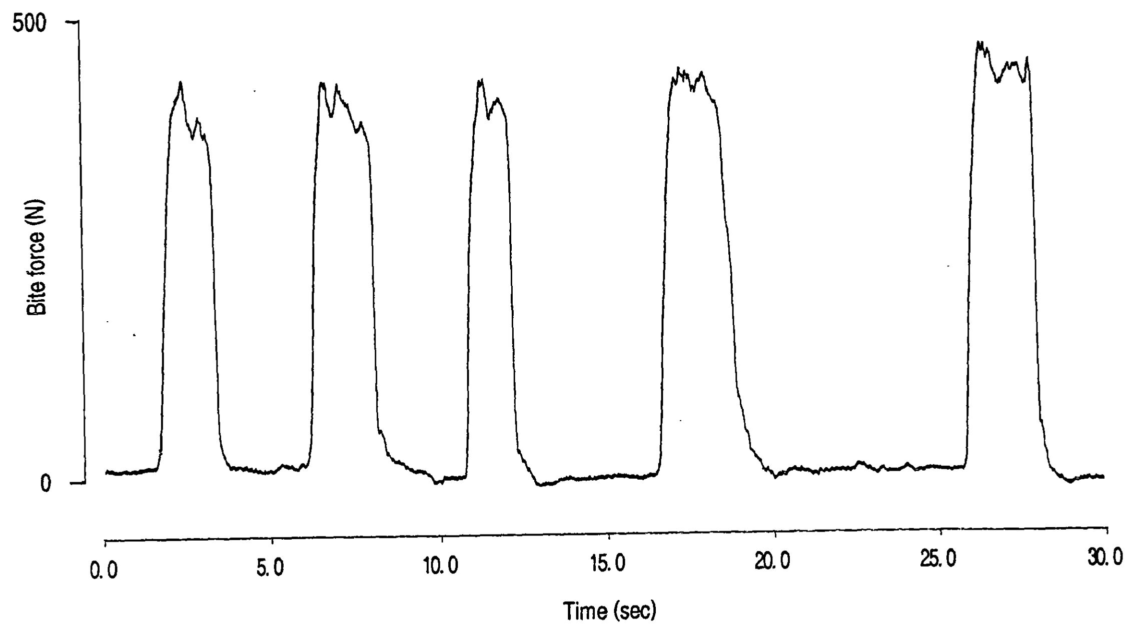


Figure 3.2 An example of maximum bite force recordings with the unilateral transducer from one subject.

B) The standard deviations of the mean of the ten maximum bite force values (N) obtained during each of the three sessions were used as an indicator of the variability (Table 3.2).

	SESSION 1			SESSION 2			SESSION 3		
Subject	**Uni.**	**Anter.**	**Bil.**	**Uni.**	**Anter.**	**Bil.**	**Uni.**	**Anter.**	**Bil.**
1	15.0	54.5	46.4	13.0	35.6	26.2	14.5	46.7	27.2
2	48.9	79.8	226.6	25.1	48.5	111.7	13.7	59.7	72.6
3	20.8	22.1	33.8	17.6	26.3	59.7	24.6	28.7	74.8
4	22.2	23.0	76.5	9.0	15.0	25.9	20.5	16.5	20.8
5	48.8	55.3	36.4	40.1	43.7	63.4	45.3	31.9	55.6
6	19.1	65.8	17.5	17.4	52.0	53.7	37.2	18.2	44.9
7	53.9	70.9	54.5	107.4	27.0	103.7	42.7	21.9	69.4
8	31.7	40.1	78.5	20.3	14.4	80.2	10.5	13.7	30.2

Table 3.2 Within-subject standard deviations of the mean of ten maximum bite force measurements during the 3 sessions for the eight subjects, with the three transducers: Unilateral (Uni.), Anterior (Anter.), Bilateral (Bil.).

As a measure of variability of bite force between the three positions within the dental arch, these within-subject standard deviations of the maximum bite force values for each subject were subjected to a Kruskal-Wallis test and the difference between the within-subject standard deviations for the three different transducer positions was found to be highly significant ($P \leq 0.01$). This was confirmed by the dotplots of the within-subject standard deviations of maximum bite force values for individual subjects (Figure 3.3). The bite force measured between the posterior teeth on one side only with the unilateral transducer produced the lowest within-subject variability and the force measured between the posterior teeth of both sides with the bilateral transducer produced the highest variability. The data presented in Figure 3.3 as dotplots for clarity is the same as that presented in Table 3.2

As a measure of variability of bite force between the three different sessions, wherever the recording position in the dental arch may be, the standard deviations of the maximum bite forces were also subjected to a Kruskal-Wallis test at the session level. There was no statistically significant difference ($P \geq 0.05$) between the three different occasions and this may be clearly seen in the illustration (Figure 3.4).

An example of the variation of maximum bite force values (N) over all three sessions for the unilateral, anterior and bilateral transducer in one subject may be seen in the figure 3.5. The transducers were used in the same order for each volunteer and on each occasion.

```
    .
  .::.
  :::: ...::.            .
  -------+---------+---------+---------+---------+--------- Unilateral
   .  .
  .::::....:.:.. . .
  -------+---------+---------+---------+---------+--------- Anterior

    ...::. ...:....::      . .                        .
  -------+---------+---------+---------+---------+--------- Bilateral
        40        80        120       160       200       240
                           Standard deviation
```

Figure 3.3 Dotplots of the within-subject standard deviations of maximum bite force values for individual subjects for the three different transducer positions.

```
      .     . .
  ::: :.. :.: . ..:                                  .
  -------+---------+---------+---------+---------+--------- 1st session
  :  .
  .:::: ....: ..   .     ...
  -------+---------+---------+---------+---------+--------- 2nd session
  . :     .
  :::.::. :. .. ...
  -------+---------+---------+---------+---------+--------- 3rd session
        40        80        120       160       200       240
                           Standard deviation
```

Figure 3.4 Dotplots of the within-subject standard deviations of maximum bite force values for individual subjects, for the three different transducers and the three different sessions.

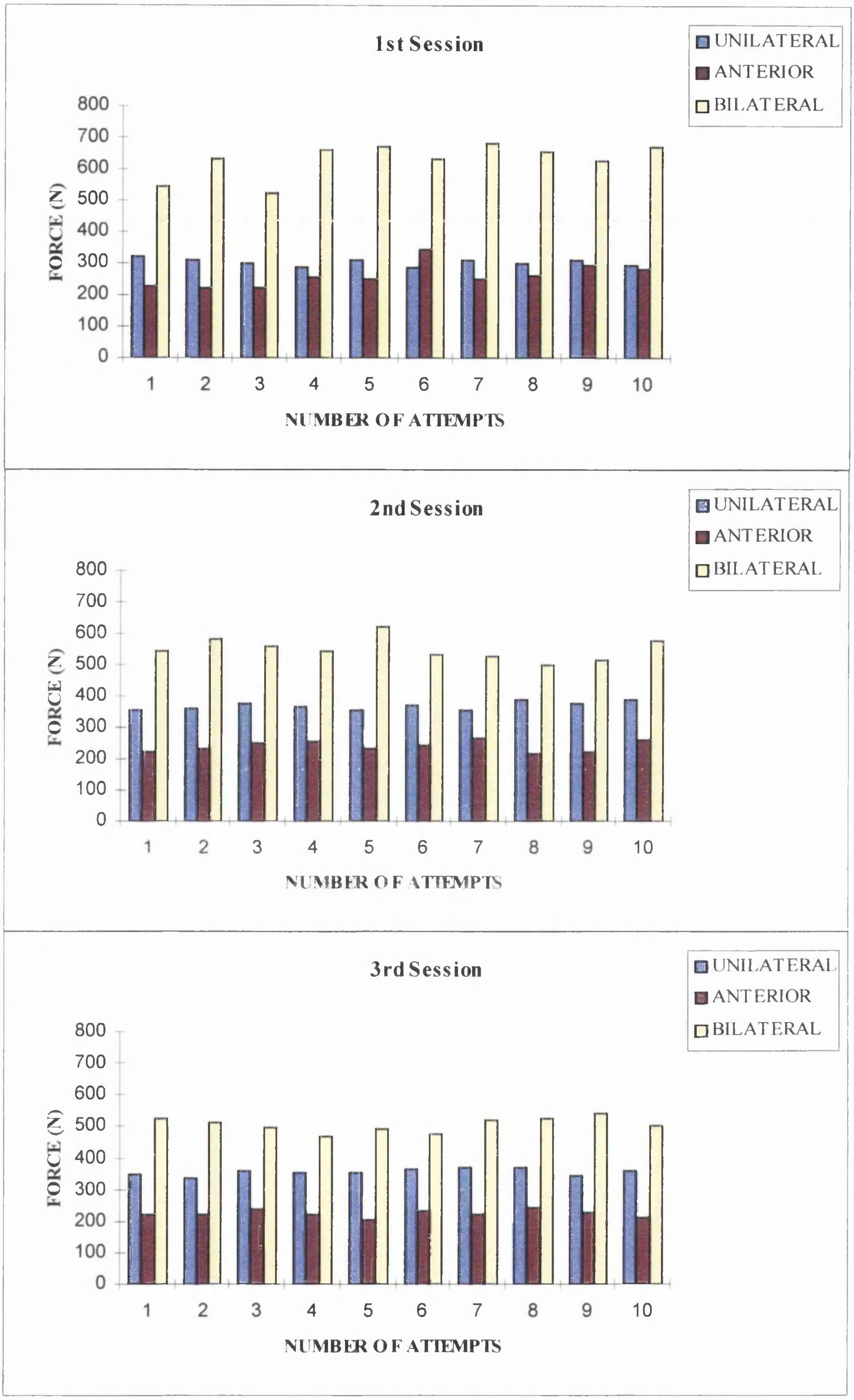


Figure 3.5 Maximum bite force values (N) over all three sessions for the unilateral, anterior and bilateral transducer in one subject.

3.5 DISCUSSION AND CONCLUSIONS

The consistency of the maximum voluntary clenches suggested that these were indeed the maximum of which the subjects were capable of producing. The maximum bite forces recorded posteriorly, both unilateral and bilateral, were notably higher than those recorded anteriorly. This is in agreement with previous studies (Leff, 1966; Carlsson, 1974; Van Steenberghe & De Vries, 1978b; Lyons and Baxendale, 1990). The more posteriorly the force transducer is placed within the dental arch the greater the bite force, partly because of the lever effect of the mandible and partly because there is a larger area of tooth root and therefore a larger area of periodontal ligament around posterior teeth. This larger area of support is likely to reduce the inhibitory effect of nociceptive afferent volleys on force output. Different positions will also influence the degree to which different muscles are involved in the force production. If the transducer is placed anteriorly between the incisor teeth, with a resultant mandibular protrusion, the masseter muscle will produce most of the force together with the medial pterygoid muscle. If the bite force meter is more posteriorly placed, then the anterior fibres of the temporalis muscle will become more active and hence make a greater contribution to the effort (Carlsoo, 1952; Hellsing and Lindstrom, 1983).

In the present study, maximum biting forces were higher when measured between the molars bilaterally than when measured between the molars on one side only which is in agreement with other investigators (Van Steenberghe & De Vries, 1978b; Pruim et al., 1980; Bakke et al., 1989). It is likely that the maximum force achieved was highly dependent on the number of posterior teeth loaded during the biting action i.e. the more teeth, the higher the force.

It was also the case that the within-subject variability of bite force values were generally smallest for the unilateral transducer and greatest for the bilateral transducer; this has also been found by others (Bakke et al., 1989, 1992). A possible explanation for the high variability in bite force values from the bilateral transducer was that it was more difficult to construct the four acrylic indices and therefore the transducer was less comfortable in this case. In contrast, the anterior and unilateral transducers required only two acrylic indices and therefore they were easier to construct and hence more likely to be comfortable. It is possible that the results with the bilateral transducer could have been less variable if study casts had been mounted on an articulator and indices had been made on those study casts. It seems reasonable to say that the design of the measuring device is important in the consistent measurement of bite force; aspects of design which are of particular relevance are the facility to stabilise the transducer on the teeth, to be able to replace the force transducer in the same position within the oral cavity and to provide a comfortable surface against which the subject can bite.

In conclusion, maximum bite force was most reproducible (i.e. showed least within-subject variability) when measured between the first molar and second premolar teeth on one side only. The variability of the maximum bite force is very small when measured in the same position on different occasions, wherever that position may be in the dental arch; maximum bite force is relatively consistent. The position of the transducer is important when measuring bite force, as the more posteriorly the bite force is recorded, the greater the maximum force achieved.

CHAPTER 4

ACOUSTIC MYOGRAPHY, ELECTROMYOGRAPHY AND BITE FORCE

4.1 SUMMARY

Previous studies have shown both linear and non-linear relations between increasing bite force and integrated electromyogram (EMG) of the jaw closing muscles.

This study examined the relationship between the acoustic myogram, the electromyogram (EMG) and bite force in the masseter muscles of nine healthy male subjects, at four different submaximum clenching levels. The acoustic myogram (AMG) offers some advantages over electromyography in certain circumstances, but the use of AMG on the jaw-closing muscles has not been fully tested. AMG was recorded using a piezoelectric crystal microphone and EMG was recorded with surface electrodes. Force was recorded between the anterior teeth with a strain-gauge transducer.

Bite force was normalised to % MVC to make valid comparisons between subjects. Analysis showed that Pearson's correlation coefficient was ≥ 0.913 for force/AMG, and ≥ 0.973 for force/EMG in all subjects, indicating a linear relationship between force, AMG and EMG at the four different force levels tested (25%-75% of maximum).

It is apparent that AMG may be used as an accurate monitor of masseter muscle force production, although some care is required in the technique.

4.2 INTRODUCTION

Various methods have been developed for the measurement of bite force, commonly using a force transducer (Hagberg, 1987). Indirect methods of measuring force production have also been used, including electromyography (Moller, 1966; Ahlgren, 1966), transmitted sound vibrations (Gibbs et al., 1981) and psychophysical measurement (Wennstom, 1971a, b).

It has been shown that the EMG activity of the jaw closing muscles increases proportionally as voluntary bite force increases (Ahlgren et al, 1985; Bakke et al., 1989). Although the measurement of EMG has been used as an indicator of bite force (Van Boxtel et al, 1983; Naeije, 1984), it is well known that when localised fatigue occurs, either the amplitude of the EMG activity remains unchanged while force declines, or the EMG activity increases while force remains unchanged (Edwards & Lippold, 1956; Bigland-Ritchie, 1981b). In addition, there are some methodological problems in recording surface EMG from the jaw-closing muscles (Lund & Widmer, 1989). Parameters such as age, gender, skin thickness, electrode position over the muscle belly, type of electrode, the electronic equipment used to amplify and record the potential change, and any symptoms of TMD should be controlled before the recording of EMG activity (Lund, Widmer and Feine, 1995). Surface EMG becomes also a slightly less reliable index of muscle force when sweating occurs, due to problems of short circuiting, or when the muscle surface is covered with hair-bearing skin.

Acoustic myography (AMG) is the recording of low frequency sounds produced during a skeletal muscle contraction (Oster & Jaffe, 1980; Barry, Geiringer, and Ball, 1985; Wee and Ashley, 1989; Stokes and Dalton, 1991b). The amplitude and frequency of these sounds have been shown to vary systematically with the force developed.

Various theories have been suggested to explain the origin of muscle sounds and there is increasing acceptance of the view that lateral movements of muscle fibres during contraction produce these low frequency sounds (Barry, 1987; Wee and Ashley, 1989; Frangioni et al, 1987; Dalton and Stokes, 1993). AMG activity therefore may reflect the mechanical component of muscle contraction while EMG represents the electrical activity.

It has been shown that the relationship between AMG and increasing isometric force is linear for submaximum isometric contractions of the biceps brachii muscle (Barry et al, 1985; Orizio et al, 1989a). However, for contractions up to 100% of the MVC, the relationship has been found to be either curvilinear (Maton et al, 1990) or to increase up to 80% MVC followed by a sharp decline to 100% MVC (Orizio et al, 1989b). Orizio et al (1989a, b) explained the reduction in AMG above 80% as being due to controll of force output by increasing motor unit firing rate since no new motor units remain to be recruited in this range. Increasing force also increases the stiffness of the muscle which may reduce or eliminate muscle sounds. Alternatively, the reduction in the AMG might be due to the contact transducer characteristics, since it is commonly found that the AMG recorded by heart sounds microphones falls in amplitude at higher force levels. (Orizio et al, 1989b; Smith and Stokes, 1993).

Examination of the masseter muscles, during maximum voluntary contractions, demonstrated that AMG recordings within the same session were repeatable under controlled conditions and that the frequency range of the AMG signal was similar to that seen in other skeletal muscles (L' Estrange et al, 1993). The fact that the relationship between AMG and force varies between muscles and for different types of contraction indicates that the relationship needs to be established for the jaw-closing muscles.

Therefore, in this study masseter muscle function was investigated using two different indirect methods of measuring force production, i.e. EMG and AMG. The masseter muscle was chosen for this investigation because although it is not possible to isolate the force output of individual jaw closing muscles, in a protrusive closure most of the force is produced by the masseter muscles with small contribution of the temporalis muscles (Carlsoo, 1952). Furthermore, the masseter is also a muscle which appears to be very often involved in myogenous TMD (Laskin, 1995).

The specific aims of this study were:

- to investigate the relationships between the EMG, AMG amplitude and bite force in masseter muscles, during submaximum clenching levels, and
- to assess these two indirect measures (EMG and AMG) of activation of the masseter muscle.

4.3 MATERIALS AND METHOD

4.3.1 Experimental subjects.

Nine healthy, male volunteer subjects, fully dentate with no crowns or large composite restorations on their incisor teeth and with no history or symptoms of TMD, participated in this study. The subjects' ages ranged from 25 to 35 years (mean 30 years) and informed consent was obtained from each volunteer before the experiment.

4.3.2 Recording protocol

Throughout all of the experiments, EMG and AMG from the right masseter muscle and bite force were recorded simultaneously. The subjects were seated upright in a dental chair and were given time to become familiar with the bite force meter and the oscilloscope screen. Each volunteer was able to control his clenching level through visual feedback of force.

The bite force was measured between the anterior teeth, from canine to canine, with the mandible in a protrusive position using an anterior bite force transducer which has been described in detail previously (see Chapter 2). Several layers of gauze were used on the area of tooth contact instead of acrylic indices, as it was found that gauze was more comfortable for the teeth than acrylic resin on maximal biting. The total thickness of the transducer and the gauze was approximately 9 mm. Calibration of the transducer was carried out with known weights prior to each experiment and the response was found to be linear (see Chapter 2).

The subjects were instructed to clench as hard as possible on the force transducer for at least 2 seconds, and this was done three times with adequate rest periods. The highest value of force was used as the maximum voluntary activity for that trial. Then a series of four clenches, sustained for 6 seconds, were performed at each of 25%, 50%, 60%, 75% of maximum voluntary bite force.

4.3.3 Electromyography

EMG was recorded from the right masseter muscle, using surface, self-adhesive electrodes in a bipolar configuration, having a centre to centre distance of 2cm. The earth electrode was attached to the ear lobe. Skin preparation was thoroughly performed using gauze soaked in surgical spirit.

The EMG signals were amplified x 5000, filtered 5 - 800 Hz, and stored on video tape cassettes with the use of a PCM-8 video recorder adapter (see Chapter 2). The sampling frequency of the adapter was 22 kHz and the frequency response of each channel was DC - 7 kHz.

4.3.4 Acoustic myography

The AMG was recorded simultaneously with EMG, a piezoelectric crystal microphone (HP 21050 A, Hewlett-Packard, USA) being placed over the belly of the masseter muscle. This microphone was designed to record heart sounds and so has an appropriate sensitivity in the 0-200Hz frequency range. The microphone was adapted to achieve a small circular skin contact surface of 5 mm diameter via an acrylic resin piston and this was placed between the EMG electrodes (Figure 4.1).

The microphone was secured by an adjustable elastic band. Skin contact force was measured and adjusted to 200 g with the use of a Correx force meter. The force with which the transducer contacts the skin surface is important to the operation of the microphone, and therefore needs to be considered when recording muscle sounds (Bolton et al, 1989; Orizio et al, 1989a, b; Smith and Stokes, 1993).

The AMG signal was amplified x 2000 and was recorded on video tape cassettes with the use of the PCM-8 video recorder adapter, along with force and EMG records (Figure 4.2). A filter setting at 2 Hz to 160Hz was chosen for AMG in order to eliminate low frequency unwanted noise related to the external environment, or to cut off any audible, scraping sounds probably caused by movement of the microphone over the skin surface (Wee and Ashley, 1989).

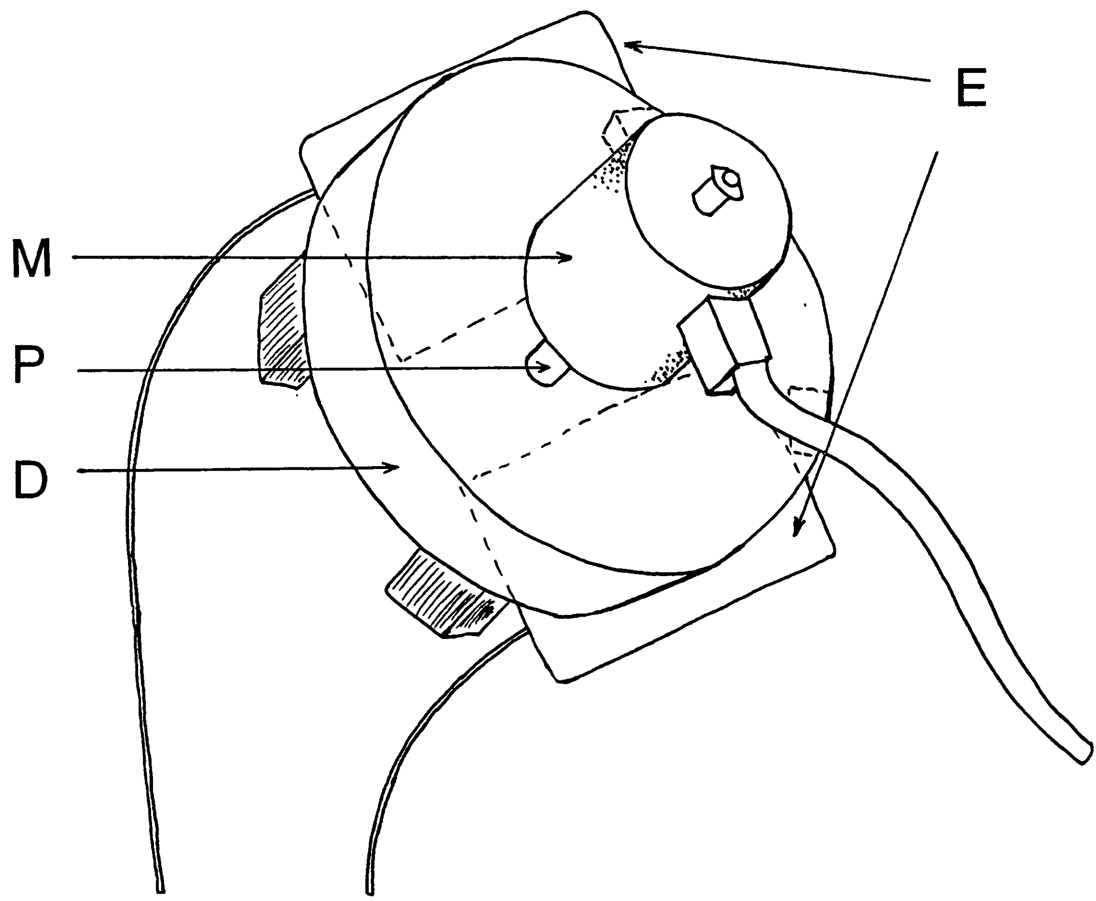


Figure 4.1 The arrangement of the electrodes and the microphone over the skin surface.
E: electrodes, M: microphone, P: skin-contact piston, D: perspex disc to support microphone.

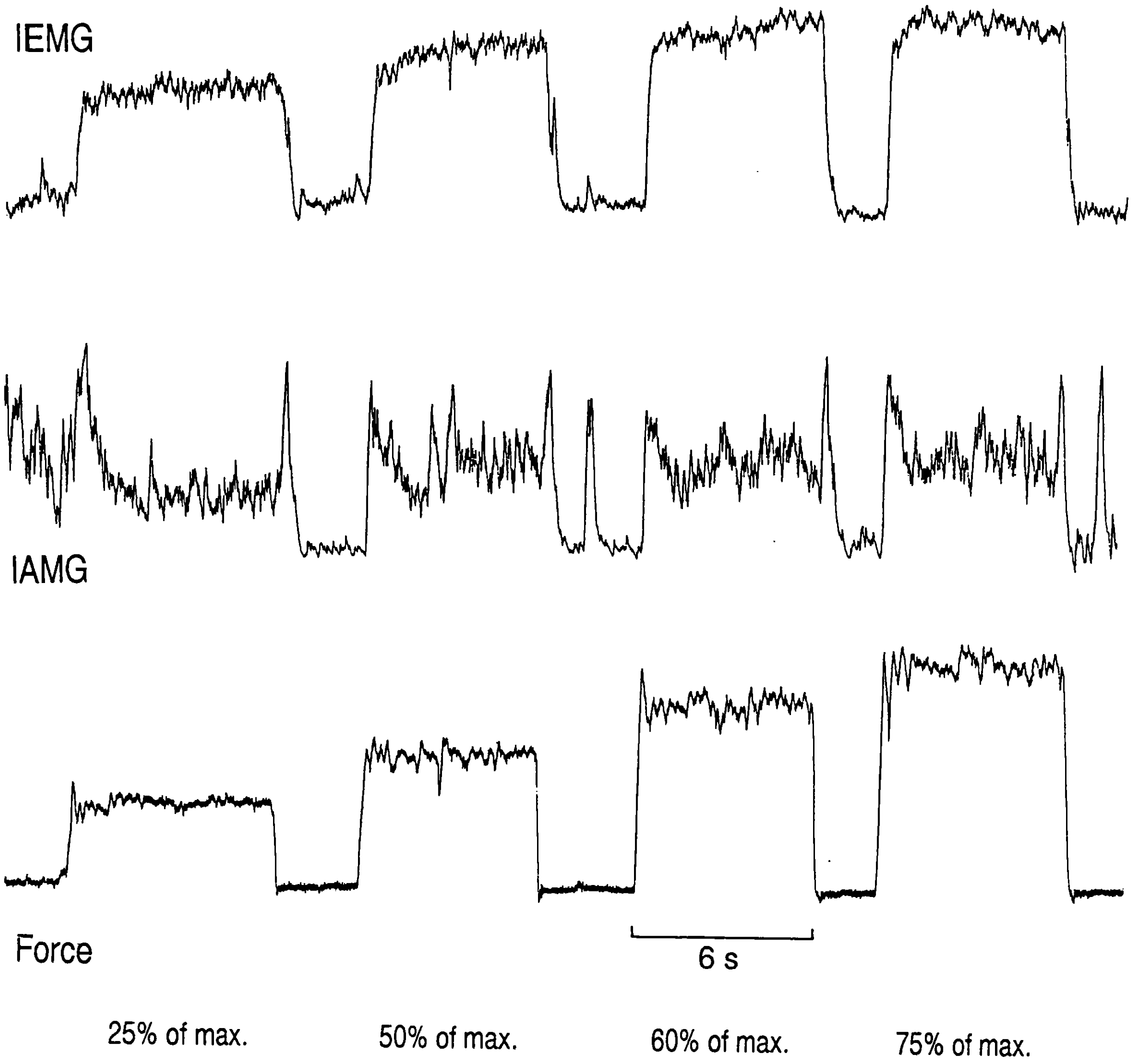


Figure 4.2 Recordings of IEMG, IAMG and bite force during a series of 6s sustained clenches at 25%, 50%, 60% and 75% of maximum.

4.3.5 Signal analysis and processing

The raw signals of force, EMG and AMG were subsequently played back through the adapter. A computer interface (CED1401+, Cambridge Electronic Design, Cambridge, UK) was used to digitise the data in three separate channels, and signal processing and analysis was carried out using a signal analysis software package (Signal Averager, Cambridge Electronic Design, Cambridge, UK) on a PC.

EMG was band-pass filtered 3Hz to 1kHz and AMG 2Hz to 160 Hz, using Neurolog NL106 and NL 125 modules. Both AMG and EMG were full-wave-rectified and RMS integrated, using a Neurolog NL 705 integrator unit and then being referred to as IAMG and IEMG respectively. The time constant was set to 200 ms for EMG and to 500 ms for AMG. The amplitudes of the RMS signals and the bite force were measured at the same time precisely during a 2 s window in the middle of a 6 s sustained contraction, when the muscle force was stable.

In order to compare the relationships of force/IEMG and force/IAMG between nine different individuals, the force values were normalised as a percentage of maximum voluntary contraction (% of MVC). The IEMG and IAMG signal amplitudes were normalised by the value of the IEMG and IAMG signal corresponding to 75% MVC to avoid the inconsistent variations of the EMG and AMG signal amplitude that occur at higher force levels.

A Pearson's correlation analysis and a linear regression analysis were carried out to test the strength of the relationship between the IAMG and bite force, and IEMG and bite force.

4.4 RESULTS

The maximum bite force between the anterior teeth varied considerably among the 9 subjects, ranging from 274 N to 440 N (mean 339 N, SD± 60.8).

The normalised amplitude values of IEMG and IAMG were plotted against force and linear relationships were seen (Figure 4.3 and 4.4). Pearson's correlation coefficients were 0.973 for force/IEMG and 0.913 for force/IAMG in all subjects, indicating a strongly linear relationship between force, IEMG and IAMG amplitudes at the four different force levels tested (25% - 75% of maximum).

For each subject the IEMG and IAMG rise progressively up to 75% MVC with increasing force as may be seen in figure 4.5.

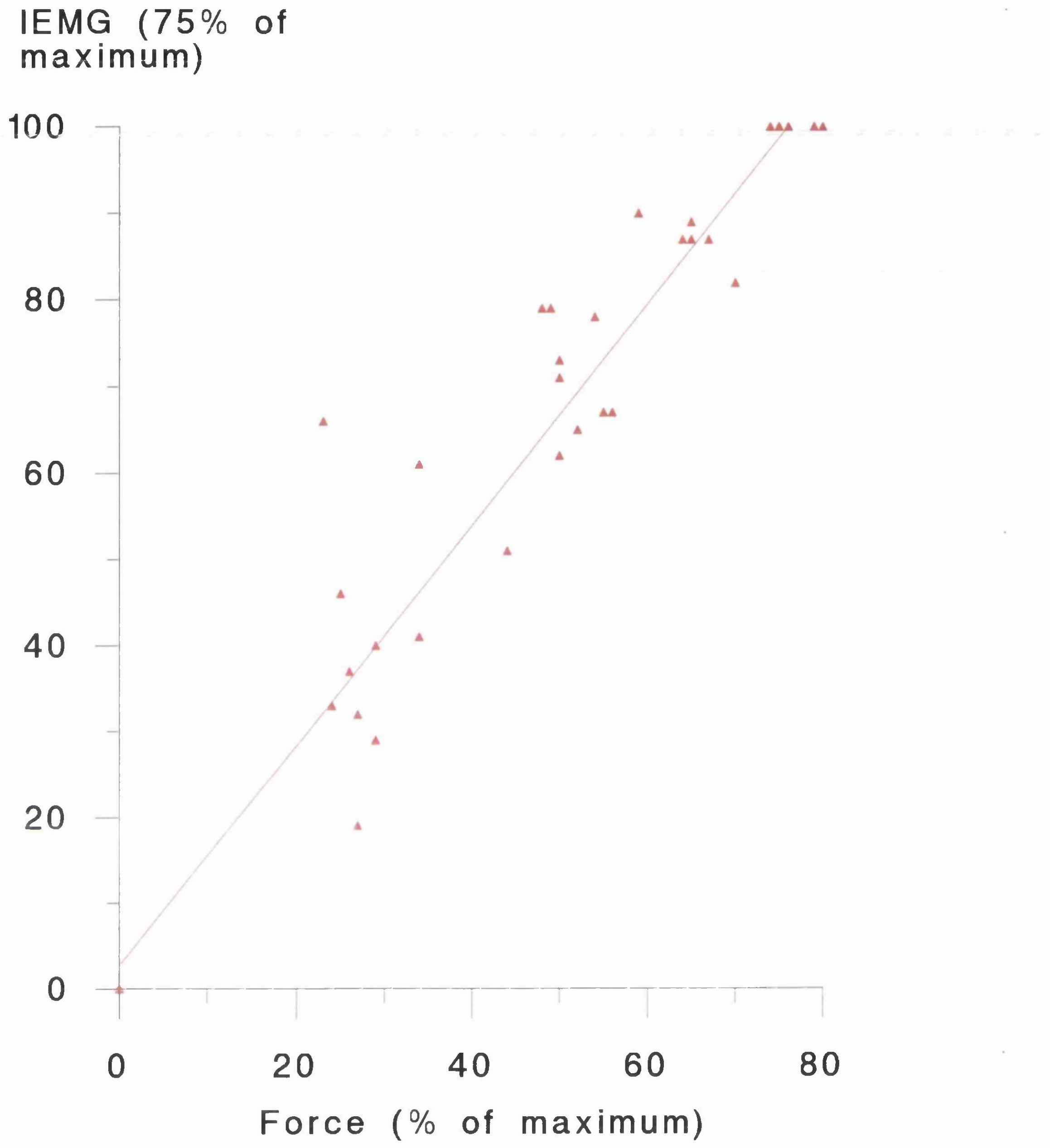


Figure 4.3 Amplitudes of IEMG, normalised to their respective values at 75% MVC, plotted against normalised force (expressed as a percentage of maximum bite force).

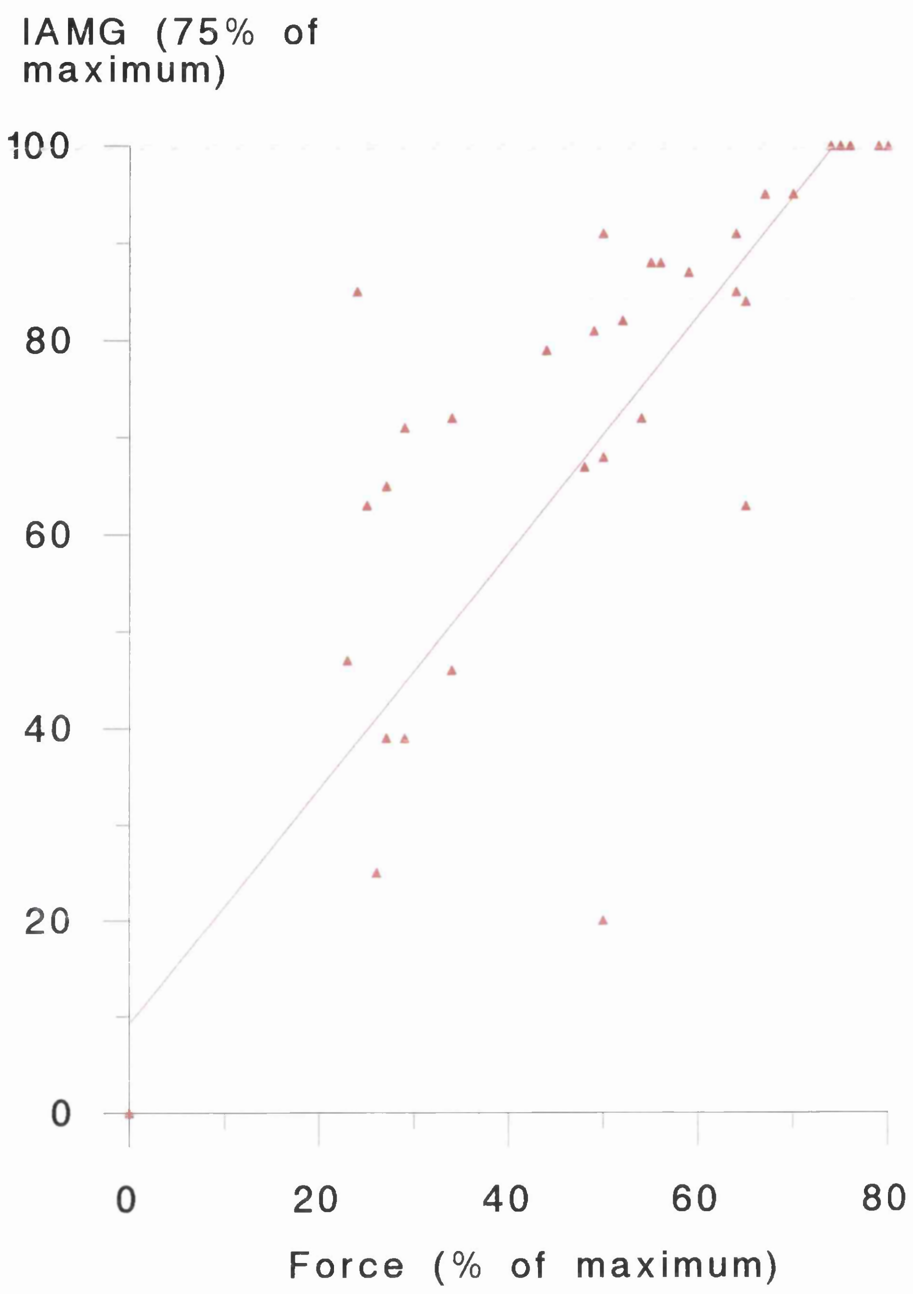
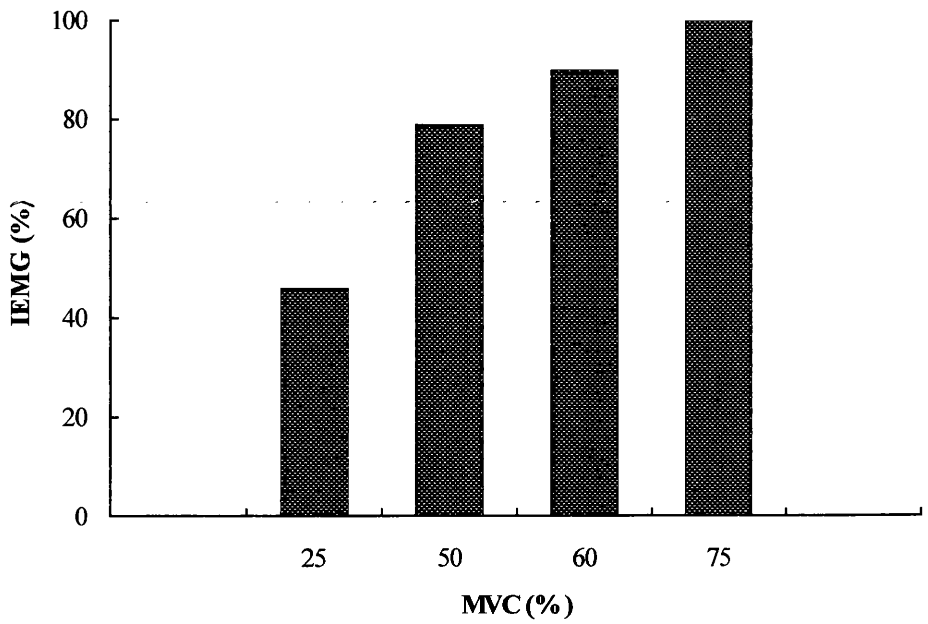


Figure 4.4 Amplitudes of IAMG, normalised to their respective values at 75% MVC, plotted against normalised force (expressed as a percentage of maximum bite force).



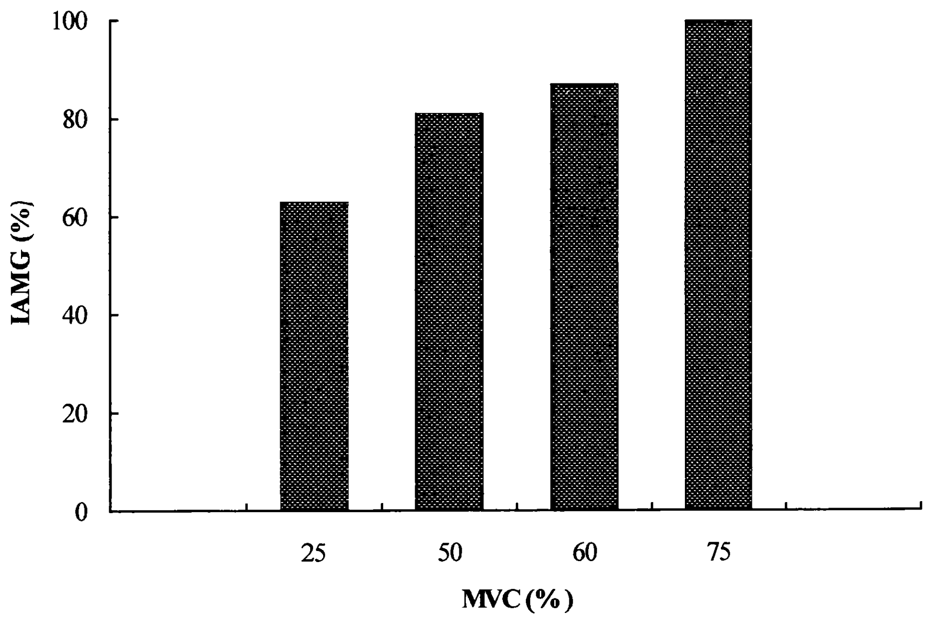


Figure 4.5. The rectified integrated EMG and AMG against MVC in four submaximum clenching levels in one subject. The amplitude of IEMG and IAMG rise progressively up to 75 % of MVC.

4.5 DISCUSSION AND CONCLUSIONS.

The raw EMG and AMG were full wave rectified and integrated to overcome possible transient movement artefacts of raw signals. During investigations of submaximum clenching, the experimental set-up permitted certain amount of movement of the masseter muscle at the beginning and end of the contractions. These movement artefacts at both ends of contraction were the result of the muscle shortening to pull the tendon taught from slack state (tendon snap). This can be seen in Figure 4.2.

The results of this study have shown that there is a linear relationship between bite force and IEMG in a non-fatigued state of masseter muscle at sub-maximal clenching levels. This is in agreement with most previous investigations on the jaw-closing and other muscles in man, the majority of which have shown that the relationship is linear (Kawazoe et al, 1979; Bakke et al, 1989; Lindauer, Gay & Rendell, 1991). However, Haraldson et al. (1985) found the relationship to be linear for the anterior temporalis muscle but not for the masseter muscle. A non-linear relationship has been also observed in the biceps brachii and deltoid muscles (Komi and Buskirk, 1970; Lawrence and De Luca, 1983) and in the jaw muscles (Devlin and Wastell,1985; Wastell and Devlin,1987). Probably the most widely held view is that a linear relationship exists at submaximal contraction levels (Pruim et al, 1978; Hosman and Naeije, 1979), while a deviation from linearity exists at higher force levels.

These discrepancies can be explained by the fact that the relationship depends on the physiological characteristics of the particular muscle (Woods and Bigland-Ritchie, 1983). Some of these characteristics are fibre-type composition, spread of signals from adjacent muscles, action of antagonist and synergist muscles and recruitment patterns and firing rate properties (Lawrence & De Luca, 1983) (This point is discussed more fully in Chapter 6).

The results of the present study also show that there was a linear relationship between bite force and IAMG at submaximum clenching levels. When recording AMG from the masseter muscle, care was taken when the microphone was strapped over the belly of the muscle, using an elastic band. Secure attachment and firm pressure is needed to minimise any slight movement of the sensor relative to the muscle surface (Bolton et al., 1989). However, the contact pressure of the microphone to the skin considerably affects the signal amplitude, an increase in pressure tending to increase the amplitude (Bolton et al, 1989). This probably reflects the efficiency of transmission of mechanical signals to the microphone. It cannot affect the underlying nature of the relationship between AMG and force since the signal amplitudes were all normalised. Care was also taken to position the sensor over the belly of the muscle, as the sound wave has been shown to be at maximum amplitude at this position (Bolton et al, 1989; Stokes and Dalton, 1991b).

The results are also in agreement with the findings of others that the relationship between IAMG and force was linear up to 75% MVC (Oster and Jaffe, 1980; Barry et al, 1985;

Zwarts and Keidel, 1991; Rouse and Baxendale, 1991). In support of this, Orizio et al. (1989a, b) and Takamjani and Baxendale (1993) have found that IAMG increased with force up to 80% MVC and then decreased. Other investigators have described this relationship as linear up to maximum voluntary force (Stokes and Dalton, 1991a, b), non-linear (Smith and Stokes, 1993) or parabolic (Orizio et al, 1989a). In addition, Stile and Pham (1991) reported that during 0-30% MVC, the AMG amplitude of masseter and temporalis muscles tended to increase to a maximum value at 5 or 10% MVC and then remained nearly constant or decreased at higher forces. However, technical considerations in their study such as the method of attaching the AMG device to the skin and the failure of the foil electret microphone which has been used to detect representative activity of the whole muscle might explain the contradictory results.

The observed differences in IAMG/force relationship are probably due to variations in experimental conditions such as different types of transducer, different amount of contact pressure on the skin, different muscles, and differences in the range of forces under examination.

The combination of surface EMG and AMG provides a measure of electromechanical coupling in muscle, i.e. EMG is a measure of electrical activity and AMG is a measure of the mechanical effect of this electrical activity. This combination of EMG and AMG has been used on the paraspinal muscles to assess electromechanical uncoupling during fatigue (Cooper et al, 1991) and it is likely that this combination would also be useful in fatigue studies of the jaw-closing muscles in order to differentiate between central and peripheral fatigue. AMG would be a useful monitor of clenching force, with the teeth together, in fatigue studies.

Moreover, AMG may be useful for examining muscle function, and under certain circumstances offers some advantages over surface EMG, principally in that it has a very narrow bandwidth (therefore it is easy to filter out noise) and the signal amplitude is unaffected by fatigue (Barry et al, 1985; Stokes and Dalton, 1991a). In contrast, EMG amplitude is known to become dissociated from force during isometric contractions of fatigued muscle.

In summary the present investigation showed a direct relationship between force, IEMG amplitude, and IAMG amplitude in the masseter muscle at the four different levels of bite force tested (25-75% of maximum). It is apparent that AMG may be used as a monitor of masseter muscle force production, although some care is required in the technique and it is not quite as good a monitor of force as EMG. AMG may be useful in situations where EMG is difficult or presents problems, and in fatigue studies.

CHAPTER 5

BITE FORCE, ENDURANCE AND FATIGUE IN EDENTULOUS PATIENTS

5.1 SUMMARY

It is well known that bite force and EMG activity are considerably reduced in edentulous patients, but the susceptibility of their jaw-closing muscles to localised fatigue is less certain. This information is even less clear for edentulous subjects who have TMD.

Eleven healthy edentulous subjects and ten edentulous subjects with TMD participated in this study. Maximum bite force was measured first, with the transducer placed on the canine-first premolar region bilaterally. Then two rapid relaxations were made from a brief voluntary clench to 50% MVC. A sustained voluntary clench of 50% MVC was then maintained and endurance time was noted. EMG was recorded from both masseter muscles and the median frequency of the power spectrum of the EMG from 2*s* at the beginning of the sustained clench and 2*s* at the end was subsequently calculated. Two more rapid relaxations from brief clenches were performed immediately after the sustained clench.

The mean maximum bite force in the healthy group was 115N (SD±41) and in the TMD group was 75N (SD±22), this difference being significant (P=0.013). The mean endurance time in the healthy group was 86*s* (SD±51) and in the TMD group was 63*s* (SD±20). The percentage change in the median frequency in the healthy group as a result of the sustained contraction was 6% (left) and 8.6% (right) and in the TMD group was 13.9% (left) and 12.8% (right). The percentage change in the median frequency for the masseter muscle in healthy group was 10.23%, and for the painful muscle in TMD

group was 15.06%, a non-significant difference (P=0.18). The percentage change in the mean relaxation half time for the healthy group was 28.5% and for the TMD group was 72%, a significant difference (P=0.046).

It was apparent that (1) the maximum bite force was low in edentulous subjects and was further reduced in edentulous TMD subjects (2) endurance time was reduced in TMD subjects (3) fatigue resistance of the masseter muscles was reduced in TMD subjects.

5.2 INTRODUCTION

It is well known that maximum bite force and jaw-closing muscle EMG activity are considerably reduced in complete denture wearers compared to dentate individuals (Helkimo et al, 1977; Michael et al, 1990; Miralles et al, 1989). The main reason for the greatly reduced bite force is thought to be that the jaw closing muscles atrophy because they are unable to function as vigorously as when natural teeth are present. Moreover, Ingervall and Hedegard (1980) pointed out that complete denture wearers are handicapped, judging from the low EMG activity during maximum biting and swallowing.

TMD seem to be as prevalent in complete denture wearers as in the dentate population, varying from 15% to 20%, although the symptoms have been noted to be of low intensity in denture wearers (Agerberg, 1988; Zissis et al, 1988; Lundeen et al, 1990). It has been also reported that non-denture wearing edentulous subjects have a lower prevalence of TMD symptoms than the complete denture wearers (Wilding and Owen, 1987).

Although the development of TMD has a multifactorial aetiology, posterior occlusal wear and incorrect occlusal vertical dimension may contribute to muscle pain in complete denture patients (Monteith, 1984; Wilding & Owen, 1987; Agerberg, 1988; Zissis et al, 1988). Parafunctional habits (i.e. clenching and grinding, dislodging the dentures), which have been found prevalent among complete denture wearers may also contribute to the development of TMD (Mercado & Faulkner, 1991). It has been noted that jaw-closing muscle pain and fatigue may possibly be attributed to hyperactivity of these muscles (Laskin, 1969; Yemm, 1985; Mao et al, 1993).

Fatigue has been defined as a failure to maintain the required or expected force (Edwards, 1981). It has been suggested that the endurance time or time to muscle pain tolerance, defined as the length of time a given muscular contraction can be sustained, may be an indicator of the resistance of the jaw closing muscles to fatigue (Dahlstrom et al, 1988; Clark & Carter, 1985; Christensen, 1981a). During a sustained voluntary clench at 50% of MVC, the endurance time has been found to range from 47 to 270 seconds, in healthy dentate subjects (Naeije, 1984; Dahlstrom et al, 1988; Lyons & Baxendale, 1990; Lyons et al, 1993). The subject's motivation and the ability to endure discomfort in the fatiguing task are important psychological factors influencing the endurance tests and probably interpreting the large variation in endurance time values.

During sustained isometric contraction a shift in the EMG power spectrum towards lower frequencies occurs due to an increase in power in the low frequency range and decrease in the high frequency range. Therefore, the rate of shift of the mean or the median frequency of the power spectrum during sustained clenching can be used to identify muscle fatigue (Palla & Ash, 1981b; Naeije & Zorn, 1981; Lindstrom & Hellsing, 1983; Clark et al, 1988; Lyons et al, 1993). Edentulous subjects have been found to have a lower mean power frequency than dentate individuals, which was interpreted as representing a reduced number of type II fibres probably due to disuse atrophy (Wastell, Barker & Devlin, 1987).

It is well known that muscle fatigue during voluntary sustained contraction is characterised not only by loss of force but also by slowing of the rate of relaxation (Edwards, Hill & Jones, 1972; 1975a; Bigland-Ritchie et al, 1983). During relaxation, force decays exponentially after the period of maximum rate of change (Edwards et al, 1972, 1975a; Bigland-Ritchie et al, 1983; Jewell & Wilkie, 1960). The half time ($t_{0.5}$) of the later part of this relaxation phase, which follows exponential time course, was found to increase two to threefold as a result of fatiguing voluntary contractions.

In patients with TMD it appears from clinical observation that the muscle soreness often leads to rather slow and deliberate mandibular movements; this slowing of movement is probably a defensive reaction to avoid pain and is not due to a prolonged relaxation rate.

The aims of this study were:

- ♦ to determine maximum bite force and endurance time in healthy complete denture wearers and complete denture wearers with TMD, using a comfortable design of bite force transducer and further,
- ♦ to evaluate the shift in the median frequency of the EMG power spectrum and the changes in the relaxation rate before and after a sustained contraction as indicators of the fatiguability of the masseter muscles in these two edentulous groups.

5.3 MATERIALS AND METHOD

5.3.1.Experimental subjects

Eleven healthy edentulous subjects and ten edentulous patients with history of TMD participated in this study, their ages ranging from 64 to 75 years (mean age 67 years). The chief complaint of the patients with TMD was a unilateral jaw muscle pain and reduced mobility of the mandible. All the participants (10 females and 11 males) gave written informed consent and Ethical Committee approval was obtained.

5.3.2. Recording protocol.

The experiment was carried out over two sessions with an interval of one week between each. In the first session upper and lower alginate impressions of the existing dentures and a jaw registration using a special acrylic jig (with the same shape as the bite force transducer) were taken. The casts were mounted on a simple hinge articulator in the position recorded and then stored (Figure 5.1).

Small self-curing acrylic indexes were made with the teeth of the casts just in contact with the metal faces of the bite force transducer before the second appointment. The subjects were seated in a dental chair in an upright position and with a head support. Each patient was given time to become familiar with the force transducer and the oscilloscope screen; visual feedback was provided by the force display.

A series of three maximum voluntary clenches (MVC) were recorded with a strain gauge transducer for each subject. EMG was recorded simultaneously from the masseter muscles bilaterally, using self-adhesive surface electrodes (see Chapter 2). The patient was then asked to clench to 50% of the maximum and immediately completely relax as quickly as possible; this sequence was then repeated once more. Each subject was then required - and verbally encouraged - to maintain a steady force of 50% of MVC for as long as possible and endurance time was noted. Endurance time was defined as the length of time a subject could maintain bite force at 50% MVC until he or she was unable to continue this task because of intolerable muscle pain, joint pain or fatigue. Immediately after the cessation of the sustained clench, each subject was asked to quickly clench to 50% MVC and then relax completely; this rapid clench was immediately repeated a second time.

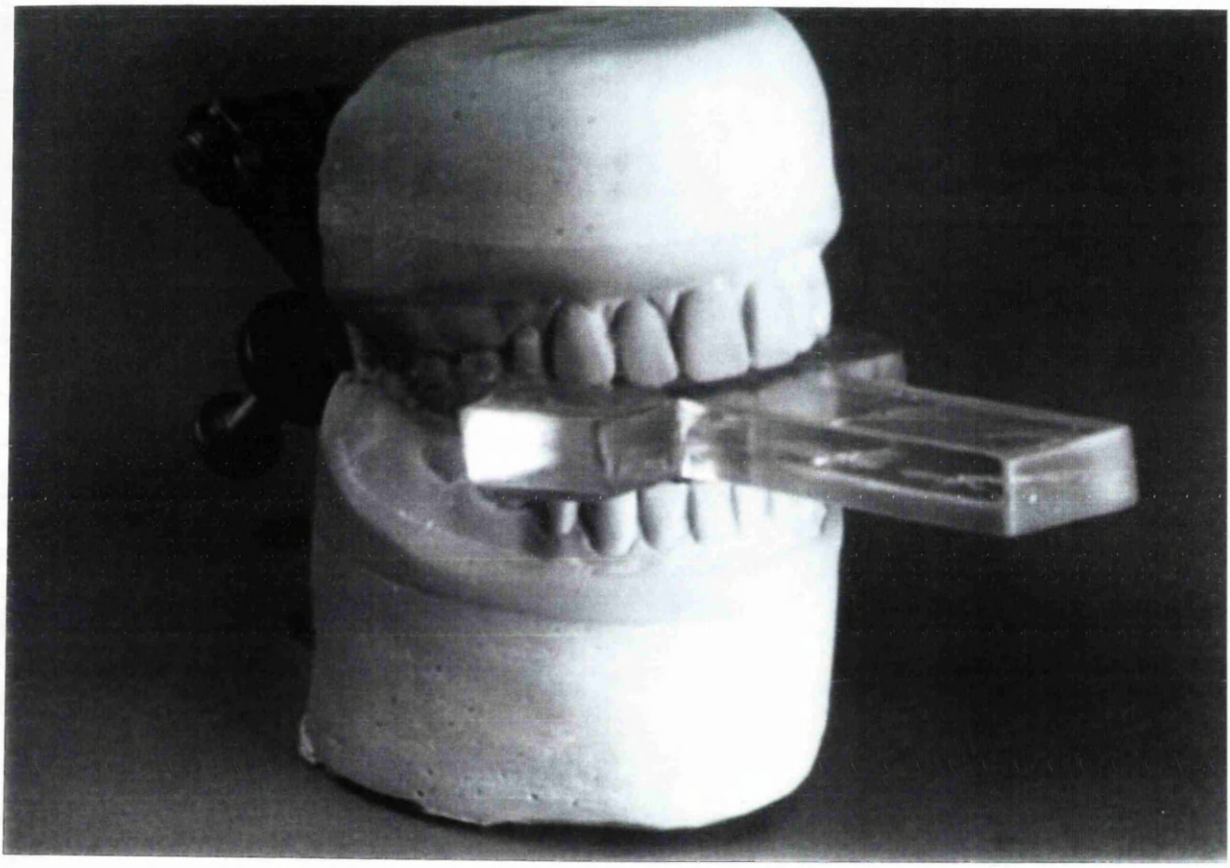


Figure 5.1 The casts mounted on a simple hinge articulator with a jaw registration using a special acrylic jig.

5.3.3 Bite force measurement.

Bite force was measured bilaterally in the region of the canines and the first premolars with a specially designed bite force transducer which consists of two stainless steel beams cut to a T-shape (figure 5.2) (see Chapter 2). It should be noted that the maximum bite force is exerted in the premolar region by a denture wearer, while in the dentate subjects it is greatest in the molar region (Worner, 1939).

The biting surfaces of the force transducer were coated with self-curing acrylic indices which provided a stable and repeatable contact area for the upper and lower denture teeth. The total thickness of the transducer, including indexes, was 8mm. The transducer was calibrated known weights before each experiment and the response of the instrument was found to be linear in the range tested and consistent between sessions (see Chapter 2).

Figure 5.2 The acrylic jig used to record the jaw relationship together with the T-shape bilateral bite force transducer.

5.3.4 Electromyography.

EMG recordings were obtained from the right and left masseter muscles using self-adhesive electrodes placed at an interelectrode distance of 2 cm. An earth electrode attached to the forehead. The skin was thoroughly cleaned with gauze soaked in surgical spirit before applying the electrodes in order to reduce the skin impedance.

The EMG signals were amplified X 5000, filtered 5-800 Hz, and stored on video tape cassettes, along with the force record in separate channels, with the use of a PCM-8 video tape recorder adapter (see Chapter 2).

5.3.5 Signal processing and analysis.

The force and EMG signals were played back from the video tape through the PCM-8 A/D adapter. The EMG signal was band pass filtered 3Hz to 1Khz to remove any DC offset and high frequency noise using a Neurolog Filter NL 125.

The EMG and force data were acquired by a 486 PC through an A\D board. This board was a 12-bit successive approximation analog to digital converter (ADC) with integral sample and hold. The highest force value obtained was considered to be the maximum voluntary bite force.

Sections of EMG signal of approximately 2s duration were sampled from the right and left masseter muscles at the beginning of the sustained contraction at 50% MVC level and again at the end. The median frequency of the power spectrum at the beginning and at the end of the sustained clench was then calculated using data analysis and display software (DADisp/Win, DSP Development Corp., Cambridge, Massachusetts, USA) (Figure 5.3).

The median power frequency was chosen as the parameter to describe the power spectrum as it is said to be less sensitive to noise than other commonly used parameters (Stylen & De Luca, 1981). The endurance time was measured from the beginning of the sustained clench to the end, using a stopwatch.

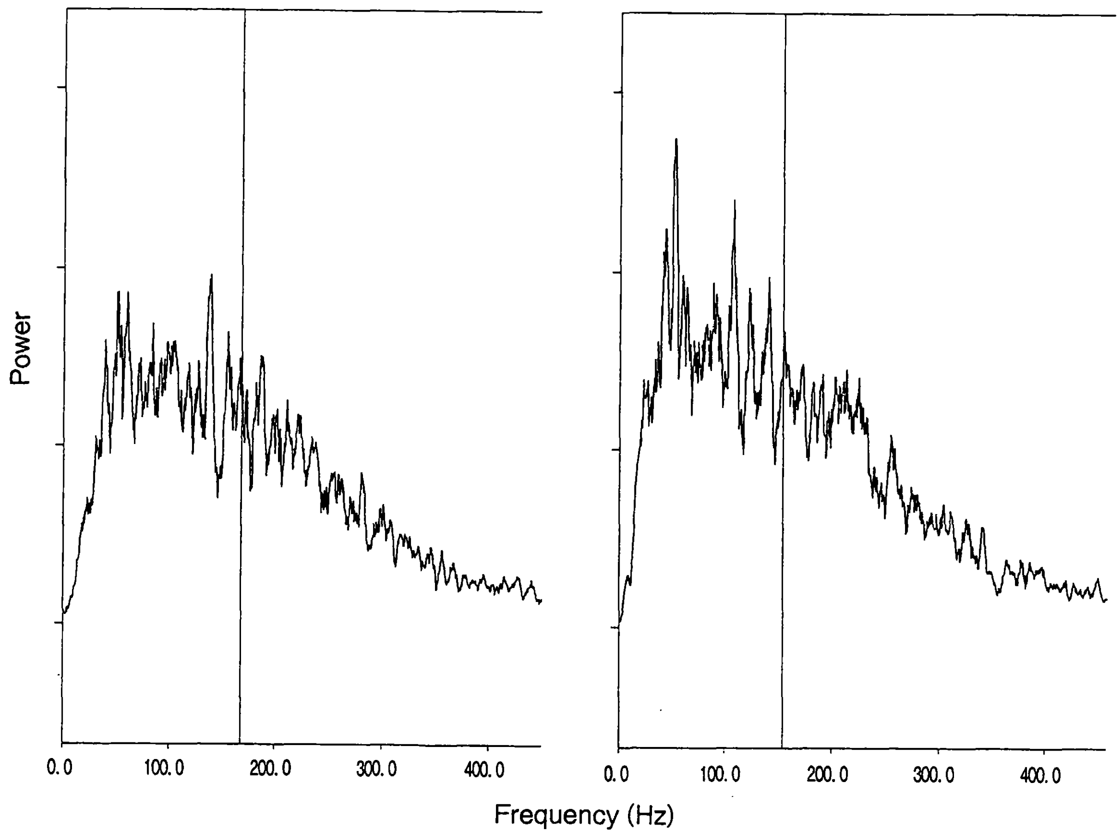


Figure 5.3 Power spectra of 2s of EMG (a) before and (b) after the sustained contraction. The vertical line represents the median frequency (MF). Note the increase in amplitude and decrease in MF after the contraction.

The relaxation rate was calculated as the time required for the force to fall from its value at the beginning of the exponential phase of relaxation to half that value ($t_{0.5}$); this was expressed as a relaxation rate constant ($k = 0.693/t_{0.5}$) (Edwards et al, 1972, 1975a, 1975b; Dawson, Gadian and Wilkie, 1980; Cady et al 1989). The relaxation half time ($t_{0.5}$) is given by the slope of the linear portion of the semi-log plot of force against time (Jewell and Wilkie, 1960; Edwards et al, 1975). It was necessary to assess the rate of the relaxation in this way, rather than as a half relaxation time, because the relaxations were from voluntary rather than electrically stimulated contractions (Wiles et al, 1979).
In the present study the relaxation rate constant (k) and the relaxation half time ($t_{0.5} = 0.693/k$) were calculated from the averaged signals of the two rapid clenches and relaxations before and after the sustained clench using a data analysis and display software system (DADisp/WIN, version 3.01 D, 1994) (Figures 5.4 and 5.5).

In order to determine whether there was any statistically significant difference between the healthy and the TMD group in bite force, endurance time, EMG median frequency and relaxation rate before and after the sustained clench, independent t-tests were used. Probability levels at $p<0.05$ were considered statistically significant.

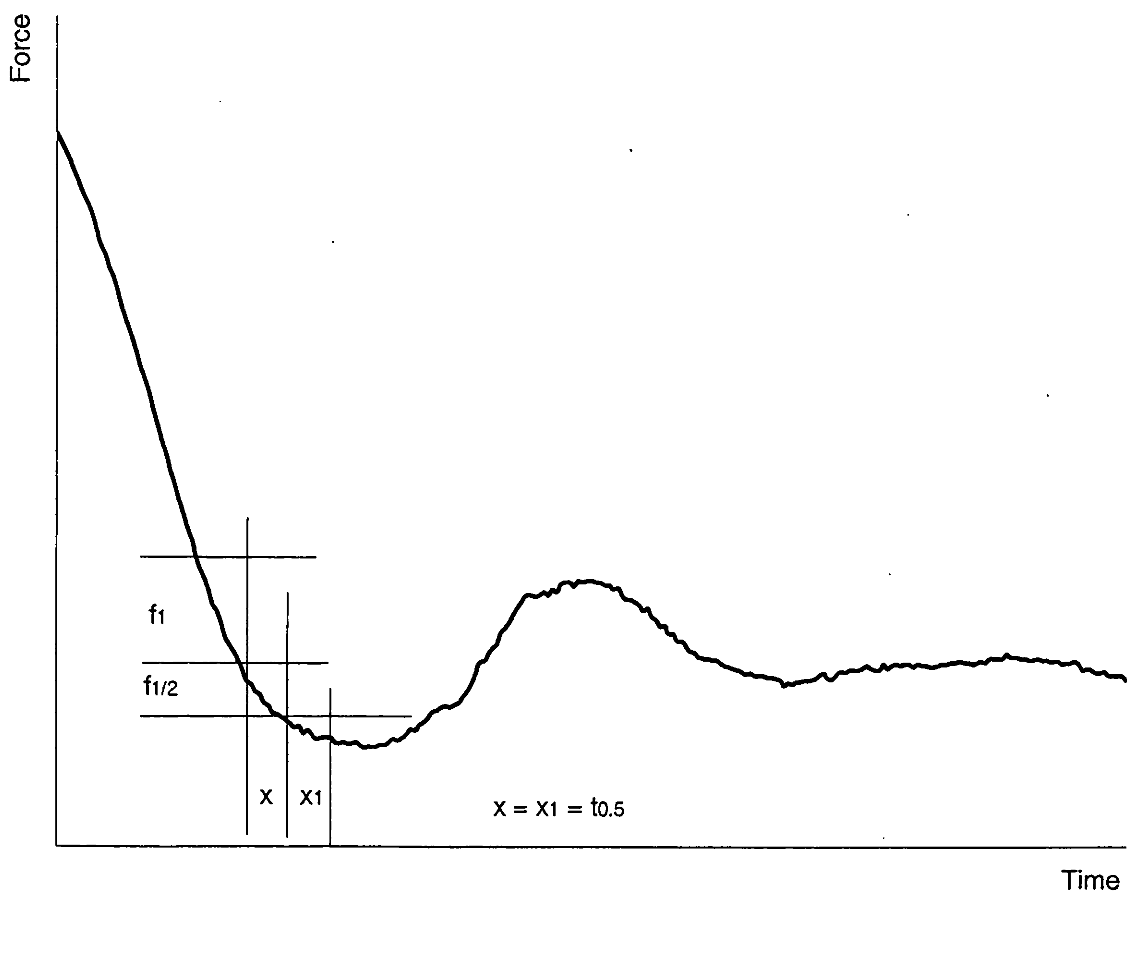


Figure 5.4 An average of two forces records from the same subject showing the relaxation of force. The exponential nature of the late phase may be seen by noting the halving of the decrease in force in the subsequent same time period (the half-time of the curve).

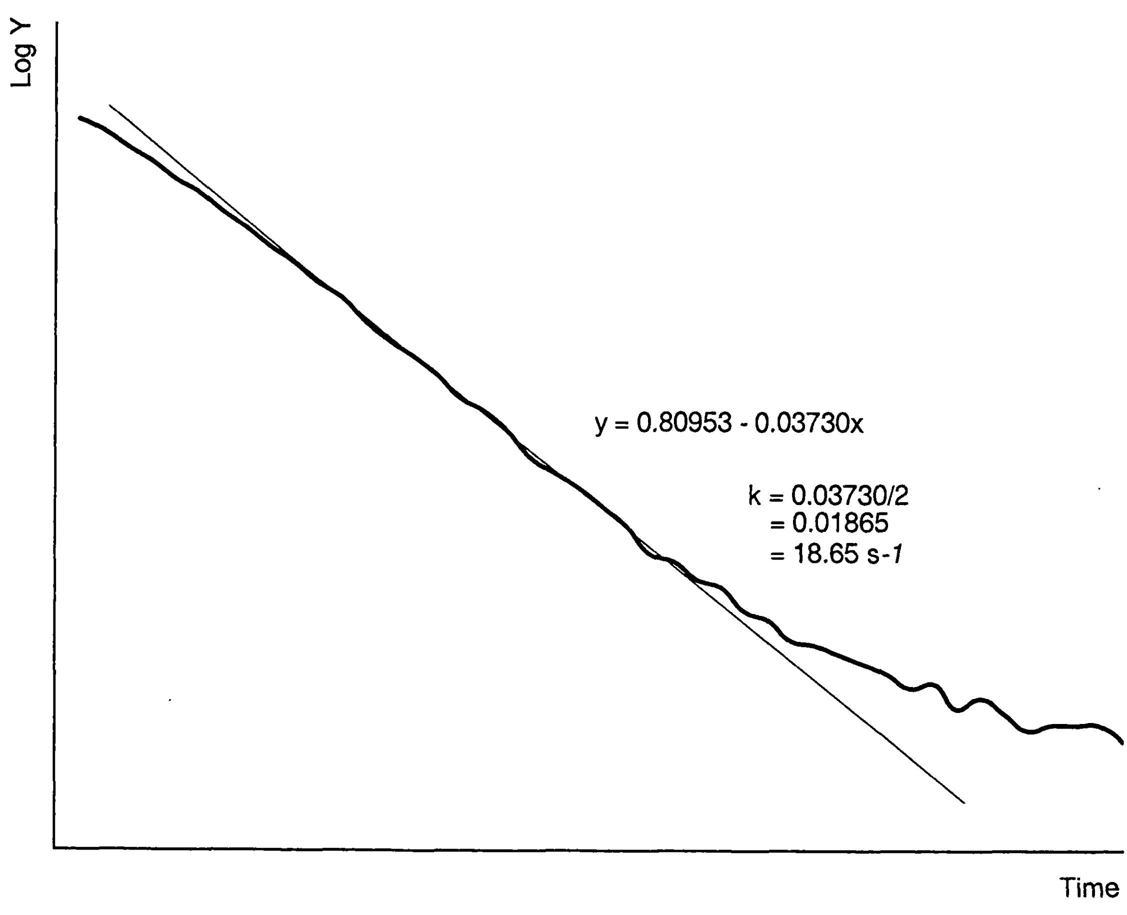
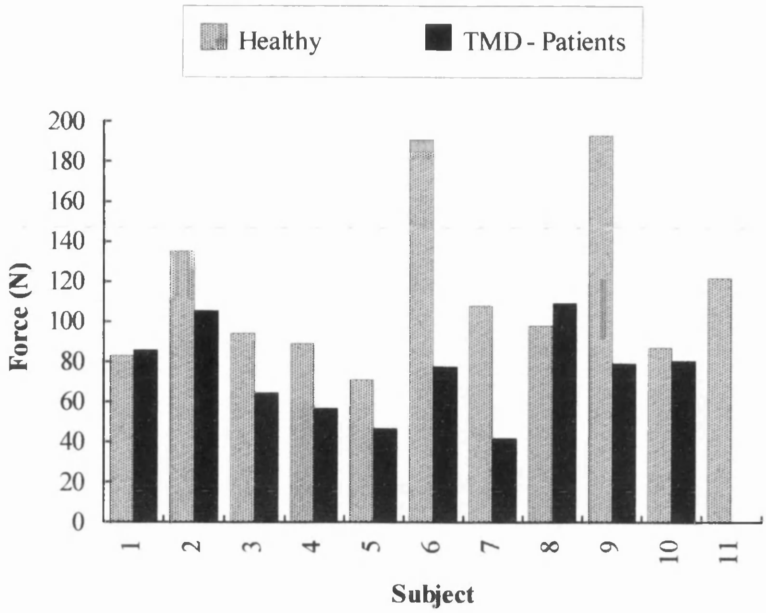


Figure 5.5 The exponential phase of relaxation is shown as a semi-log plot. The slope of this line (obtained from the regression equation) is the relaxation half time ($t_{0.5}$).

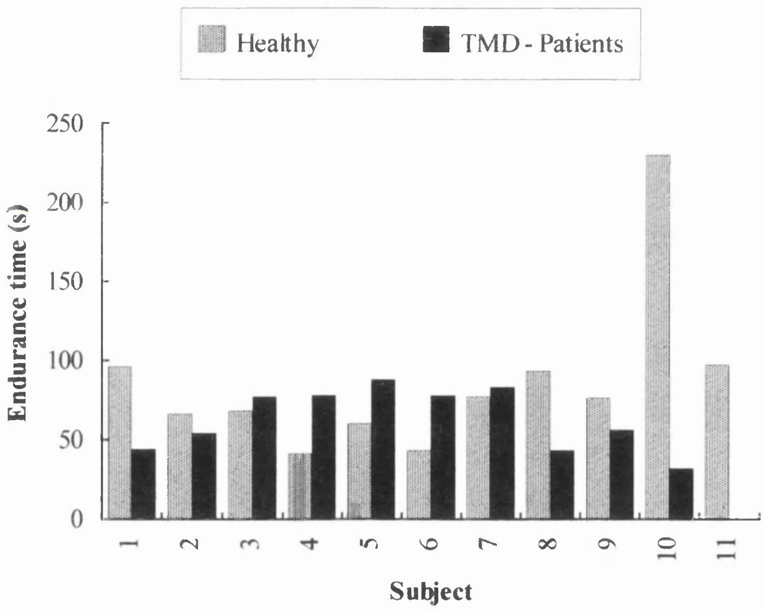
5.4 RESULTS

The mean maximum bite force value for the healthy group of 11 edentulous subjects was 115 N (SD ±41) and for the group of 10 TMD edentulous patients 75 N (SD ±22). The independent t-test showed a significant difference in maximum bite force values between the healthy and the TMD groups (P = 0.0013) (Figure 5.6).

The endurance time varied from 41 to 230*s* (mean 86*s*, SD ±51) in the healthy group, and was considerably reduced in the TMD group ranging from 32 to 88*s* (mean 63*s*, SD ±19). There was no significant difference between the endurance times of the two groups (P=0.20) (Figure 5.6).



[A]



[B]

Figure 5.6 A) Maximum bite force values B) endurance times in 11 edentulous healthy subjects and in 10 edentulous TMD-patients.

In the healthy subjects, the mean median frequency (MF) at the beginning of the sustained clench was 151.36Hz (SD±25.36) for the left masseter and 162.36Hz (SD ±22.54) for the right. At the end of the contraction the mean MF was 142.5Hz (SD ±28.92) for the left and 147.73Hz (SD ±26.24) for the right. The percentage change in the mean MF in the healthy group as a result of the sustained contraction was 6% (left) and 8.6% (right).

In the TMD patients, the mean MF value at the beginning of the sustained clench was 144.4Hz (SD ±40.1) for the left masseter muscle and 163.3Hz (SD ±29.76) for the right. At the end of the task the mean MF was 124.3Hz (SD ±27.71) for the left masseter and 142.45Hz (SD ±19.55) for the right. The reduction in mean MF for the right masseter was 8.9% and for the left masseter it was 5.8%. The differences in percentage changes in MF (from beginning to failure) between the healthy subjects and the TMD patients were not significant (P>0.05).

In the TMD patients, the mean MF value at the beginning of the sustained contraction was 153.8Hz (SD± 42.5) for the painful masseter muscle and 154.9Hz (SD± 23.02) for the non-painful. At the end of the sustained clench the mean MF was 128.4Hz (SD± 32.3) for the painful muscle, and 138.3Hz (SD±15.2) for the non-painful. The reduction in mean MF for the painful muscle was 15.06% and for the non-painful was 9.31%; the differences in percentage changes were not significant (Table 5.1).

TMD patients	MF (Hz) Beginning	End	% change
Painful	153.8 (SD 42.5)	128.4 (SD 32.3)	15.06
Non - painful	154.9 (SD 23.02)	138.3 (SD 15.28)	9.31

Table 5.1 Mean values and SD, and percentage changes of the MF values (Hz) in the painful and non-painful muscles of the TMD patients at the beginning and the end of the sustained clench.

The percentage change in the mean MF, as a result of sustained contraction, was 10.23% for the masseter muscle (randomly selected) of healthy group and 15.06% for the painful muscle of TMD group; the differences in percentage changes were not significant.

In healthy edentulous subjects, the mean relaxation half time was 41.75ms (SD ±28.37) before the sustained clench and 53.8ms (SD ±34.9) afterwards. The mean relaxation rate constant was found $27.33s^{-1}$ (SD ±19.70) before the fatiguing clenching and $19.87s^{-1}$ (SD ±14.91) afterwards.

In TMD edentulous patients before the sustained contraction, the mean relaxation half time was found 61.5ms (SD ±43.4) and the relaxation rate constant $19.23s^{-1}$ (SD ±19.16). Afterwards the mean relaxation half time was 105.7ms (SD ±57.4) and the relaxation rate constant $10.33s^{-1}$ (SD ±9.81).

The percentage change in the mean relaxation half time for the healthy group was 28.5% and for the TMD group was 72%, a significant difference (P=0.0046).

5.5 DISCUSSION AND CONCLUSIONS

Previous studies have shown that there is no significant difference in bite force when subjects wear either new or old dentures (Haraldson et al, 1979; Lundqvist et al, 1986), therefore biting force was measured with the old dentures of each subject for convenience.

In the present study, mean maximum bite force was low in elderly edentulous subjects and was further significantly reduced in edentulous TMD subjects. The low bite forces in the two groups of complete denture wearers are similar to earlier findings that bite force decreased greatly with the replacement of missing natural teeth by complete dentures and with advancing age (Haraldson et al, 1979; Lundquist et al, 1986; Bakke et al, 1990; Helkimo et al, 1977).

Cross sectional area is an important indicator of the force-generating capacity of a jaw closing muscle in dentate and edentulous subjects (Sasaki et al, 1989; Raustia, Salonen and Pyntinen, 1996) and there is a reduction of the cross-sectional area of the masseter and medial pterygoid muscles with advancing age (Newton et al, 1987). The reduction of the cross-sectional area of the masseter and medial pterygoid muscles in edentulous subjects when compared to dentate, may be the major contributing factor to the reduction of bite force (Newton et al , 1993) i.e. the muscles have become smaller. The low bite force in the edentulous groups might also be a result of a reduction of general muscle strength, as a significant decrease in isometric and dynamic muscle strength in older subjects has been found (Campbell, McComas and Petito, 1973; Larsson, 1978).

The results of this study demonstrated lower maximum bite force and reduced endurance time in the edentulous TMD group than in the healthy group. A reduced bite force has been reported in dentate TMD patients (Molin, 1972; Helkimo et al, 1975; Kroon and Naeije, 1992) although this has not been found by others (Hagberg et al, 1986; Lyons and Baxendale, 1995). Shorter endurance times have also been measured in TMD patients when compared to healthy subjects (Clark et al, 1984; Stegenga et al, 1992; Gay et al, 1994;). Furthermore, bite force and endurance time have been found to increase after treatment in TMD patients (Helkimo et al, 1975; Choy and Kydd, 1988).

For the edentulous patients with TMD, two explanations for their reduced bite force and shorter endurance time are possible. First, pain in the masticatory muscles of the group with TMD affects their fatigue tolerance, making these muscles less fit to endure hyperactivity (Kroon and Naeije, 1992). Second, the group with TMD may have weaker masticatory muscles; subjects with low muscle strength are more susceptible to fatigue and pain (Sheikholeslam et al, 1980; Bakke and Moller, 1992). Additionally, Christensen (1976) has suggested that TMD symptoms may therefore arise in subjects with "less fit" jaw closing muscles (presumably those which are less able to adjust to the demands of muscle hyperactivity).

The results of this study also show that the decrease in median frequency during the sustained clench at 50% MVC was small in both groups. The median frequency decrease was greater for the TMD group than the healthy group, but the differences were not statistically different.

The lack of statistical significant difference of MF shift between two groups was possibly due to the large standard deviations of the mean values. One of the problems with using spectral anaylsis, as a measure of the level of fatigue, is the high variability between subjects (Palla and Ash, 1981b). Others have found relatively large standard deviations of the mean power frequency (MPF) of the masseter in asymptomatic subjects biting at 50% MVC e.g. 239± 31Hz (Naeije and Zorn, 1981), 188±35Hz (Naeije and Hansson, 1986), 133±47Hz (Palla and Ash, 1981a) and also in the median frequency, 158±32Hz (Lyons et al, 1993).

Naeije and Hansson (1986) failed to find differences in MPF of the masseter muscles between myogenous, arthrogenous TMD patients and healthy group biting at 50% of MVC. However, Gay et al (1994) found that the decreases in MPF of the masseter and anterior temporalis muscles, during sustained clenches at an absolute level of 10N, were significantly greater in the myogenous TMD group than the healthy group. They suggested that the jaw closing muscles in myogenous TMD patients fatigue at a faster rate than those of healthy subjects.

The slower rate of relaxation observed in TMD patients after the sustained contraction is consistent with the shorter endurance time and the greater median frequency shifts towards lower frequencies, indicating masseter muscles more susceptible to fatigue.

The magnitude of the relaxation half time has been reported previously as: 31.2msec for the quadriceps muscle, 30msec for the first dorsal interosseous (Edwards et al, 1972), 40ms for the first dorsal interosseous (Cady et al, 1989), 28.4ms for the adductor policis (Bigland-Richie, 1983).

The small percentage changes in the median frequency and the relaxation half time of the masseter muscles following the sustained contraction at 50% MVC were surprisingly small in both groups, indicating a small reduction in the state of fatigue.
A likely explanation is that during the sustained contraction a constant force was not maintained by the masseter muscles; there may be a migration of activity from the masseter to the synergistic temporalis and medial pterygoid muscle. However, is is known that the jaw closing muscles are fatigue resistant (Van Steenberghe et al, 1978).

Previous studies have shown that maximum bite force does not fall after either repetitive maximum contractions or a sustained clenching in dentate subjects, indicating a high resistance of jaw closing muscles to contractile fatigue (Van Steenberghe et al, 1978; Clark et al, 1984; Clark & Carter, 1985). Furthermore, Clark et al (1988) showed that the sum of the EMG amplitude of masseter and anterior temporalis muscles divided by bite force (EMG to force ratio) did not change during sustained isometric contractions at various force levels. When fatigue occurs the EMG signal amplitude increases to sustain a constant force level (Edwards & Lippold, 1956; Merletti et al, 1990), The absence of change suggested that jaw closing muscles are not susceptible to neuromuscular fatigue and they are more fatigue resistance than the limb muscles (Van Steenberghe et al, 1978).

The reasons for this fatigue resistance have been suggested : 1) high blood flow and thus better oxygenation and 2) high percentage of type I fibres.

Histochemical studies have been shown that type I, fatigue resistant fibres predominate most markedly in almost all regions of the masseter muscle where they make up 62 to 72% of the total fibre content (Eriksson & Thornell, 1983; Hannam and McMillan, 1994). Additionally it has been reported that there is a reduction in type II, fatigue susceptible fibres in edentulous subjects with complete dentures (Ringqvist, 1974b). It has been suggested that this change maybe attributed to altered functional demands with progressive atrophy of these fibres as a result of tooth loss.

It was apparent that the edentulous TMD patients show a higher degree of masseter muscle fatigability and lower maximum bite force than the healthy edentulous subjects.

CHAPTER 6

GENERAL DISCUSSION AND CONCLUSIONS

6.1 GENERAL DISCUSSION

6.1.1 Bite force in dentate subjects and edentulous patients.

The measurement of voluntary maximum bite force is notoriously difficult and the reliability of the results depend on several variables, one of which is that most direct methods require a certain thickness of bite force transducer. Since the greatest bite force was previously reported in dentate and edentulous subjects at a vertical jaw opening of 9-20mm (Manns et al, 1979; Lindauer et al, 1993; Prombonas et al, 1994), the thickness of the different patterns of force transducer used in the present investigation was of the order of 8-10mm (see Chapter 2). However, the use of a large vertical jaw opening for measurement may have prevented obtaining full muscle force. It was difficult to reduce the thickness of the transducer to less than 8mm as the beams become too weak to resist the forces involved. Floystrand et al (1982) developed a miniature force transducer for measuring bite force between the first molars, which results in a reduced jaw separation of 3mm (Merischke-Stern et al, 1993), but this is not widely available.

A variable which should be considered when measuring bite force is the varying motivation of volunteers or patients because of discomfort, fear of damaging the teeth and dental restorations. In the present investigation, in order to make heavy biting on the transducers more comfortable and to protect the teeth against the possibility of enamel fracture, either small acrylic indices (Chapter 3, 5) or several layers of gauze (Chapter 4) were used on the metal beams. Visual feedback was provided and considerable verbal encouragement was given to produce a maximum effort. Although it is commonly held that most normal subjects can fully activate the majority of their muscles voluntarily (Jones and Rutherford, 1987), the application of the twitch interpolation technique to the jaw-closing muscles might be useful to determine the maximum potential bite forces (Lyons et al, 1996). The twitch interpolation technique was based on the principle that if all the muscle fibres are fully activated, and hence the contraction is the maximal, there will be no increase in force output when a muscle is electrically stimulated.

The design, the comfort and the placement of the force transducer within the dental arch, i.e. unilateral, bilateral, posterior or anterior, have been shown to influence the magnitude and the variability of bite force measurement in the present investigation.

The highest forces have been measured with bilateral, posterior placement of the transducer (Chapter 3). This may be because of the greater periodontal support of posterior teeth which distribute the force more widely, thus causing less reaction from periodontal pain receptors than measurement on one side only. It would appear quite reasonable to assume that the larger the area of distribution of a force on the periodontal membrane, the less the discomfort. Moreover, the more posteriorly the force transducer was placed within the dental arch the higher the bite force obtained, partly because of the favourable position of the first molar-premolar teeth close to the jaw-closing muscles and partly because of their larger roots and a larger area of periodontal ligament.

In the present study the maximum bite force remained relatively constant between the three different sessions and between the three different transducer positions (Chapter 3); this constancy of maximum biting force between sessions has also been found by others (Van Steenberghe & De Vries, 1978a; Hosman & Naieje, 1979). The constancy of the maximum bite forces suggested that these were indeed the maximum of which the volunteers were capable of producing. The bite force might be more constant if a force transducer was not used, because the limiting effect of the periodontal mechanoreceptors would be reduced.

Maximum bite force was most reproducible between sessions (i.e. showed least within subject variability) when measured with the unilateral posterior transducer, probably because it was more comfortable. The bilateral posterior transducer was more bulky, and it was more difficult to produce accurate acrylic indices.

In order to measure bite forces on the canine-first premolar region bilaterally in edentulous patients special attention was given to comfort and to prevent the dentures from tipping (Chapter 5). The T-shape design of bilateral transducer, with the main bulk of the instrument outside the mouth and the facility to stabilise the transducer with the pre-prepared acrylic indices on the mounted casts was more effective and comfortable. Therefore, it could be used conveniently to measure maximum bite force and to assess the jaw-closing muscle strength as a whole.

Masticatory muscle atrophy, wearing of complete dentures, jaw muscle pain and history of TMD are also important variables which limit the maximum force generation capacity (see Chapter 5).

In the present work, mean maximum bite force was very low in elderly complete denture wearers in comparison to previously published figures for age-matched dentate subjects, although it may be suspected that force recording in edentulous subjects tends to slightly underestimate the true MVC because the subjects are closing on denture bases and not on their natural teeth. It is likely that decreased voluntary bite force in the edentulous subjects could be due to reduced jaw muscle bulk and changes in contractile and functional muscle properties (Larsson, 1978; Newton and Yemm, 1986; 1990; Newton et al, 1993), or to limitations in the ability to fully activate the jaw-closing muscles due to altered afferent inputs.

Edentulous patients with TMD have been found to have significantly lower maximum bite force values than healthy edentulous subjects (see Chapter 5). A likely explanation for the reduced bite force could be that subjects with jaw muscle pain and discomfort develop less muscle activity during maximum clenching. Another interpretation is that the patients with TMD may have innately weaker jaw-closing muscles. Additionally, it has been reported that these patients have a lower pain tolerance, which, in combination with their dysfunction in the masticatory muscles, may contribute to lower bite force.

6.1.2 AMG, EMG and bite force in the masseter muscle

In this investigation, bite force was measured between the anterior teeth with the mandible in a protrusive position using the anterior force transducer. For the detection of AMG, the microphone carrier was designed so that placement of the skin-contact piston of the microphone in between the EMG electrodes was achieved in each case. This made it more likely that AMG was being recorded from the same group of masseter muscle fibres as the EMG.

A positive and linear relationship was shown to exist between the integrated surface EMG of the masseter muscle and bite force in a non-fatigued state at sub-maximum clenching levels (Chapter 4). Although the relationship between force and EMG is known to vary in different muscles, the linearity seems well documented in jaw-closing muscles during isometric contractions. The range of linear correlation coefficients has varied between 0.53 and 0.99 in other studies (Garett et al, 1964; Kawazoe et al, 1979; Ahlgren et al, 1985; Bakke et al, 1989; Lindauer et al, 1991). There seem also to be at least two situations where non-linear EMG-force relations may exist: (1) when a muscle is activated at high contraction levels (Pruim et al, 1978; Hagberg et al, 1985); and (2) when synergistic muscles contract simultaneously (Bakke et al, 1989).

Many physiological factors have been reported to affect the EMG-force relation in a particular muscle, such as the distribution and quantity of slow twitch and fast twitch fibres within the muscle, action of antagonistic and synergistic muscles and recruitment patterns and firing rate properties (Lawrence and De Luca, 1983).

The masseter muscle is composed of about 62-72% type I, slow twitch fibres, but the distribution varies slightly in different parts of the muscle with type I most markedly predominate in the anterior part (Eriksson and Thornell, 1983). It has been hypothesised that the predominance of type I fibres in the anterior masseter permits this part of the muscle to have more precise control over dental forces, and the posterior part could be then contribute to the maintenance of the mandibular posture (Mao et al, 1992). This latter, because it contains relatively fewer type I fibres, is supposed to contribute to more forceful and faster functional acts. It has also been suggested that the EMG-force relation for human limb muscles of uniform fibre-type composition tends to be linear, whereas there is deviation from linearity in muscles of mixed fibre types (Woods and Bigland-Ritcie, 1983). However, it is known that neither masseter nor temporalis has a uniform fibre-type composition (Eriksson and Thornell, 1983; Hannam and McMillan, 1994).

The functional entity of the muscles is the motor unit, and the motor unit activity is controlled by the interaction of afferent inputs, causing them to be recruited or change their firing rate in relation to the required force. Two neural mechanisms control the force output: recruitment of motor units and changing the firing rate of active motor units. At the beginning of a contraction up to 30% MVC, recruitment of additional motor units is the main factor, progressively larger motor units being recruited as the force increases. For force levels above 30% MVC the main factor is the increase in firing rate of active motor units; above 75% little recruitment occurs (Milner-Brown, Stein and Yemm, 1973c; De Luca, 1979).

It has also been shown that the smaller, slower, motor units are activated first when isometric bite force increases. In addition to an increase of firing rate of smaller motor units, a number of larger, faster motor units are recruited (Yemm, 1977a; Goldberg and Derfler, 1977). This orderly recruitment of motor units is known as the 'Size principle' (Henneman, 1957).

Different relationships between AMG and force have been reported in different muscles (Oster and Jaffe, 1980; Rouse and Baxendale, 1990; Stokes and Dalton, 1991a; Smith and Stokes, 1993). These differences can be related to several factors e.g., different types of the contact transducers, muscles, contractions and range of forces that were used. However, in general there is agreement that AMG increases with increasing force, although a decrease at higher force may be expected. The reason for the initial increase is thought to be due to motor unit recruitment and to increased motor unit firing rate. There are a number of possible reasons why the IAMG may fall as force increases beyond 75%; increased intramuscular pressure and the consequent reduction in muscle compliance; the increase in firing rate may lead to increased muscle stiffness; the characteristics of the microphone may result in a reduced sensitivity to higher frequency sounds (Orizio et al, 1989b; Smith and Stokes, 1993).

It also seems that the reported variability in force/AMG relationships may be due to technical factors rather than physiological differences between muscles and individual subjects (Smith and Stokes, 1993). Therefore, guidelines for appropriate use of AMG need to be established before it can be used to assess muscle function. The principal factors which can affect the magnitude of AMG signals are (1) the position of the transducer over the muscle. When the sensor is placed over the tendon, instead of the belly of muscle, the magnitude of the AMG is reduced (Bolton et al, 1989; Stokes and Dalton, 1991a) and (2) the contact pressure between sensor and the belly of muscle (Bolton et al, 1989; Smith and Stokes, 1993).

In the present investigation, it was found that AMG increases with bite force during submaximum clenching levels like EMG, although IAMG amplitude had weaker linear correlation coefficient with force than IEMG amplitude. A likely explanation is that AMG source picked up from a wider jaw muscle territory.

6.1.3. Masseter muscle fatigue in edentulous subjects and those with TMD.

The endurance time has been previously studied during sustained maximal isometric contraction or during a sustained submaximum clench, as an indicator for the resistance to fatigue of the masticatory muscles, in dentate subjects (Christensen, 1981a; Dahlstrom et al, 1988; Lyons and Baxendale, 1990). In this study, similar endurance times were observed in the healthy edentulous group at 50% MVC compared to previously published figures for dentate subjects (see Chapter 1), and this may be related to the fibre composition of the masseter muscle in edentulous subjects. It has been suggested that endurance of a muscle during isometric contraction of constant force is related to the proportions of muscle fibre types of which is composed (Petrofsky and

Lind, 1979). Subjects with a high proportion of type I, fatigue resistant muscle fibres have a greater endurance than subjects with a low proportion of these fibres (Hulten et al, 1975; Viitasalo and Komi, 1978). Additionally, it is also known that the low bite forces involve type I, fatigue resistant fibres and are also associated with a long endurance time (Ringqvist, 1974b). Thus, the relatively long endurance time observed here for the healthy complete denture wearers together with the low biting forces suggests that there is a relatively large proportion of type I fibres in their jaw muscles. The shorter endurance time in the TMD group compared to the healthy edentulous group indicated either the impaired ability of the painful muscle to sustain the clench or lack of motivation of the patients as a consequence of pain.

Several studies have looked at the EMG frequency spectrum in the healthy jaw-closing muscles of dentate subjects during a fatiguing task (Naeije and Zorn, 1981; Lindstrom and Hellsing, 1983; Lyons et al, 1993). They have all demonstrated a decrease of the mean or median frequency during a sustained isometric contraction, which can be used to identify muscle fatigue. According to Basmajian & De Luca (1985a), Stylen (1985) found in the first dorsal interosseous and deltoid muscles that the greatest decrease in the median frequency occurred at 50% MVC. This shift in the median frequency of EMG during a fatiguing task can be explained by a reduction of the conduction velocity of the action potential along the muscle fibres and, therefore, to an increased duration of the motor unit action potential (MuAP). It has been shown that the duration of the MuAP affects the shape of the EMG power spectrum (Kadefors et al, 1973; Blinowska & Piotrkiewicz, 1978; Naeije and Zorn, 1982).

In this study the percentage change in median frequency as a result of the sustained clench was used as an indicator of the fatiguability of the masseter muscle in the edentulous subjects and those with TMD. In healthy edentulous subjects, there was no significant difference in the percentage changes in mean median frequency, as a result of the sustained clench, between the right and left masseter muscles. Naeije and Zorn (1981) also reported no significant difference in mean power frequency shifts between the right and left masseter muscles in healthy dentate subjects.

Since a lack of symmetry of jaw muscle function is often characteristic of myogenous TMD, a statistically significant difference in power frequency spectrum between the right and left masseter muscles may be indicative of pathology. In the present investigation, the painful muscles in TMD patients showed a higher decrease of median frequency during the fatiguing task than their non-painful muscles but the difference was not statistically significant. Other workers have found in dentate patients that the rates of change of mean power frequency was significantly greater for the painful muscle than for the healthy muscle, in both the masseter and temporalis muscle (Kroon and Naeije, 1992; Gay et al, 1994).

The magnitude of relaxation half-times was greater than expected, considering the results in limb muscles of Edwards et al (1972) and Bigland-Ritchie et al (1983). However, the jaw-closing system has a different arrangement of synergistic muscles acting bilaterally on a single mandible. It is possible that there was some residual activity from the anterior temporalis or the medial pterygoid muscles which may have resulting a small reduction in the state of fatigue in the masseter muscles.

Although, in the present investigation the bite strength of the masseter muscles was very low in the edentulous subjects in comparison to published figures for dentate subjects, these muscles were relatively resistant to fatigue, as may be seen from the endurance times and the small percentage changes in MF and relaxation half time following the sustained clench. This supports the contention that there is a relative increase in type I, fatigue resistant fibres which accompanies edentulism (Rinqvist, 1974b). Additionally, electrophysiological studies of jaw closing muscles have shown a prolongation of the contraction phase with age, indicating also the reduction of type II, fatigue susceptible fibres (Newton and Yemm, 1990).

6.2 CONCLUSIONS AND AREAS FOR FUTURE RESEARCH

The general conclusions to be drawn from this investigation may be summarised as follows:

1. The maximum voluntary bite force assessed bilaterally on posterior teeth with a strain-gauge transducer may be used as a simple indicator of jaw-closing muscle strength as a whole. However, aspects of design and position of the bite force transducer within the dental arch are important in the consistent measurement of bite force; maximum biting force was most reproducible when measured between the first molar and second premolar on one side only. Repeated assessment of voluntary maximum bite force ability is relatively consistent.

2. EMG and AMG amplitude may be used as an indirect measure of force in order to assess masseter muscle function. Positive and linear relationships have been found to exist between the integrated EMG and integrated AMG of the masseter muscle and bite force at different submaximum clenching levels. Further validation studies and some care are required before AMG can be used as an accurate monitor of jaw-closing muscle force production.

3. Mean maximum bite force was low in elderly complete denture wearers. This could have indicated a low level of muscle strength but bite force is likely to have been influenced by other factors, including mucosal discomfort produced by the dentures or pain from loading of the TMJ.

Bite force was further significantly reduced in edentulous TMD patients, and this could have arisen from discomfort from the dentures, adverse loading of the TMJ and inhibition from painful muscle.

4. The small percentage changes in the median frequency and the relaxation half time following the sustained contraction at 50% MVC measured in healthy edentulous subjects indicate a small amount of fatigue in their masseter muscles.

The endurance times were relatively long, but this is to be expected in view of the low recorded bite force values i.e the prolonged endurance times do not necessarily indicate an increased resistance to fatigue in these edentulous subjects.

Shorter endurance time, greater decrease in median frequency of EMG power spectrum together with a slower rate of relaxation before and after a sustained clenching in edentulous patients with TMD tend to indicate that their masseter muscles were more susceptible to fatigue than in healthy edentulous subjects. However the causes of the muscle fatigue are at the present unknown.

AREAS FOR FUTURE RESEARCH

The fabrication of miniaturised, more comfortable and smaller size bite force transducer might be useful for measuring maximum voluntary force output of the jaw-closing muscles, in studies with large groups of subjects.

The change in the AMG/EMG amplitudes ratio during fatiguing task may be examined in the jaw-closing muscles in order to differentiate between peripheral and central fatigue, without the jaw separation caused by a bite force transducer.

Further information is required on the physiology of the jaw-closing muscles in edentulous patients especially when considering the aetiology of TMD in this disadvantaged group.

APPENDIX

PUBLISHED ABSTRACTS

The variability of bite force measurement within and between sessions in different positions within the dental arch.

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The measurement of bite force is notoriously difficult, and the reliability of the measurement depends on a number of factors. The aim of this study was to investigate the effect of measuring bite force with three different transducers on different occasions and to estimate the variability of forces.

Maximum voluntary bite force was measured in eight fully dentate participants using three transducer positions, each with a different transducer: between the anterior teeth from canine to canine; between the second premolar and the first moral on one side; and between the second premolars and first molars on both sides. Visual feedback of force was provided. Five maximum clenches were recorded and then following a rest period, a five more were recorded. This sequence was repeated for each transducer and the experiment was repeated on three different days. Adequate rest periods were provided to avoid fatigue.

The highest forces were measured with the bilateral transducer (mean 579.3N) and the lowest on the anterior transducer (mean 286.7N). The standard deviations of the bite force mean values were used as an indicator of the variability and were subjected to a non-parametric ANOVA (Kruskal-Wallis). The forces recorded with each transducer position were significantly different between the transducers ($p<0.01$). There was little difference in bite force between the three different sessions using the same recording position ($p\geq0.05$), and the bite force was least variable when measured between the second premolar and first molar teeth and on one side only.

Acoustic Myography, Electromyography and force in the Masseter muscle.
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Journal of Dental Research, 1996, **75**, p1198

Acoustic myography (AMG) offers some advantages over electromyography (EMG) (Barry, Geiringer & Ball, *Muscle and Nerve*, 8: 189-194, 1985), but the use of AMG on the jaw-closing muscles has not fully tested. The study examined the relationship between AMG, EMG and force in the masseter muscle of nine healthy male subjects. AMG was recorded using a piezoelectric crystal microphone[1] and EMG was recorded using surface electrodes[2]. Force was recorded between the anterior teeth with a strain-gauge transducer.
A correlation analysis was carried out and Pearson's correlation coefficient was $r^2 \geq 0.884$ for force/AMG, and $r^2 \geq 0.924$ for force/EMG in all subjects, indicating a linear relationship between force, AMG and EMG at the four different submaximum force levels tested (25%-75% of maximum).
<u>It is apparent that AMG may be used as an accurate monitor of masseter muscle force production, although some care is required in the technique.</u>

[1] HP 21050-A, Hewlett Packard.
[2] Littman 2325 VP, 3M Company.

Bite force, endurance and fatigue changes in the masseter muscles of edentulous subjects.

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It is well known that bite force and EMG activity are considerably reduced in edentulous subjects (Helkimo, Carlsson and Helkimo, 1977; Miralles et al, 1989), but the susceptibility of their jaw-closing muscles to localised fatigue is less certain. The aim of this study was to determine maximum bite force, endurance, and changes in relaxation rate and power density spectrum of the EMG as measures of fatigue in the masseter muscles of edentulous subjects.

Maximum voluntary clenches (MVC) were recorded with a strain gauge transducer placed on the canine- first premolar region bilaterally. EMG was recorded from the masseter muscles bilaterally. Each subject was asked to clench to 50% MVC and immediately relax, the force output providing visual feedback. A constant force of 50% MVC was then maintained for as long as possible and endurance time was noted; when concluded, each subject quickly clenched to 50% MVC and then relaxed again.

The mean maximum bite force was 115N (SD 41N). Endurance time varied between 41s and 230s (mean 86s, SD±51). The reduction in mean median frequency as a result of the sustained contraction was 8.6% for the right masseter and 6% for the left masseter muscle. The mean relaxation half time was 41.75ms before the sustained clench and 53.8ms afterwards; the percentage change in the mean relaxation half time was 28.5%.

<u>In conclusion, the strength of the masseter muscles in edentulous subjects was very low in comparison to published figures for dentate subjects, but despite this, these muscles were relatively resistant to fatigue.</u>

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Bite force, endurance and masseter muscle fatigue in healthy edentulous subjects and those with TMD.

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It is well known that bite force is considerably reduced in edentulous patients, but the susceptibility of their jaw-closing muscles to localised fatigue is less certain. This information is even less clear for edentulous subjects who have TMD. Eleven healthy edentulous subjects and 10 edentulous subjects with TMD participated in this study. Maximum bite force was measured first, with the transducer placed on the canine-first premolar region bilaterally, and then two rapid relaxations were made from a voluntary clench of 50% of maximum. A sustained voluntary clench of 50% of maximum was then maintained and endurance time was noted. EMG was recorded from both masseter muscles and the median frequency of the power spectrum of the EMG from 2 s at the beginning of the sustained clench and 2 s at the end was subsequently calculated. Two more rapid relaxations from brief clenches were performed immediately after the sustained clench.

The mean maximum bite force in the healthy group was 115 N (SD 41) and in the TMD group was 75 N (SD 22), this difference being significant (P = 0.013). The mean endurance time in the healthy group was 86 s (SD 51) and in the TMD group was 63 s (SD 20). The percentage change in the mean median frequency in the healthy group as a result of the sustained contraction was 6% (left) and 8.6% (right) and in the TMD group was 13.9% (left) and 12.8% (right). The percentage change in the mean half-relaxation time for the healthy group was 28.5% and for the TMD group was 72%, a significant difference (P = 0.046).

<u>It was apparent that (1) the maximum bite force was low in edentulous subjects and was further reduced in edentulous TMD subjects (2) endurance time was reduced in TMD subjects (3) fatigue resistance of the masseter muscles was reduced in TMD subjects.</u>

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