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The Association Among Occlusal Contacts, Clenching Effort, and Bite Force Distribution in Man

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Abstract. *The contact area during habitual biting can vary* according to the activity of the jaw musculature. Forceful masticatory muscle activity may also induce deformations of the dento-alveolar tissues and the supporting skeleton, yielding various tooth loads despite an apparently even distribution of tooth contacts. To investigate this variability, we measured bite forces simultaneously at multiple dental sites during maximum-effort clenching tasks. In each of four healthy adults with complete natural dentitions, four strain-gauge transducers in the right side of an acrylic maxillary appliance occluded with the lower canine, second premolar, and first and second molars. These, and matching contralateral contacts, were balanced by means of articulating paper and a force monitor (type F appliance). Bite forces were recorded when the subjects, without visual feedback, clenched maximally on the appliance. Similar recordings were made when contralateral molar and all contralateral contacts were removed (type R and type U appliances, respectively). Although the relation between individual forces often changed during the initial increase in force, it was generally constant around the maximum. The maximum forces at the four dental locations varied in distribution between subjects, but were characterized by posteriorly increasing forces. Forces in the anterior region (especially at the canine) significantly increased (up to 10 times) when clenching took place on unilateral contacts only (type U) as compared with fully balanced ones (type F). Bite force distribution thus changed with biting strength and the location of occlusal contacts. Increased force in the canine region during unilateral clenching seems related to the pattern of jaw muscle co-activation and the physical properties of the craniomandibular and dental supporting tissues which induce complex deformations of the lower jaw.

Key words: Bite force distribution, human jaw biomechanics, occlusion, mandibular elasticity, jaw muscle activity.

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Introduction

Definitions of an "ideal" occlusion of the teeth in clinical dentistry usually specify even, simultaneous, and bilateral tooth contacts in the intercuspal position. These are assumed to provide a balanced distribution of occlusal force (Okeson, 1993). The contacts are generally revealed by articulating paper or ribbon placed between the teeth, or by interocclusal impressions in wax or some similar material (Korioth, 1990). The number of contacts during habitual biting can vary according to the biting pressure (Riise and Ericsson, 1983). However evenly distributed they appear to be, simultaneous tooth contacts made during habitual clenching or tooth-tapping do not necessarily mean that forces on the teeth are also distributed evenly. The dento-alveolar tissues and supporting skeleton do not form a rigid system when acted upon by the jaw muscles, and differential tooth loads are possible despite apparently "balanced" tooth contacts.

The maximum forces developed between the molar teeth are larger than those between incisors (Helkimo *et al.*, 1977; Hagberg, 1987). When recorded between isolated pairs of antagonistic teeth, these forces increase progressively in a non-linear but monotonic manner as the bite point moves posteriorly (Kraft, 1962; Rugh and Solberg, 1972; Mansour and Reynik, 1975). This distribution can be explained biomechanically, since the mandible functions as a class III lever, and the tension vectors produced by isometric contraction of the jaw-closing muscles lie between the mandibular condyles and the dental arch (Gysi, 1921; Gosen, 1974; Hylander, 1975; Picq *et al.*, 1987). To maintain static equilibrium, reaction forces produced at isolated bite points must increase progressively the closer the bite point is to the active muscle group. This effect is evident in theoretical rigid models used to predict biomechanical behavior in the human masticatory system (Barbenel, 1972, 1974; Pruim *et al.*, 1980; Throckmorton and Throckmorton, 1985; Osborn and Baragar,1985; Smith *et al.*, 1986; Van Eijden *et al.*, 1988a; Koolstra *et al.*, 1988; Koolstra and Van Eijden, 1992).

In addition to the lever effect, however, forces on the teeth are influenced by the strength and pattern of muscle contraction. Like dental lever arms, muscle tensions change with the bite point. Different muscles, each with a level of contraction specified by the central nervous system, are associated with tooth clenching at a particular site. Thus, the use of anterior bite points requires less contraction in fewer muscles than does biting on more posterior teeth (MacDonald and Hannam, 1984a,b). The physical need to maintain static equilibrium during clenching and the physiological constraints provided by periodontal sensory feedback (which regulates the bite force a given tooth can tolerate) together shape muscle contraction patterns. These are presumably responsible for the non-linear characteristic of the force-magnitude curve that is found anteroposteriorly in the dental arch.

In contrast, the distribution of bite force when several teeth make simultaneous contact is not so well-understood. The lack of experimental data available may be linked to technical difficulties in measuring forces at multiple bite points. Most devices are designed to record uni-axial forces between single, or small groups of, opposing teeth (Linderholm and Wennström, 1970; Ringqvist, 1973; Fløystrand *et al.*, 1982; Proffit *et al.*, 1983; Hagberg *et al.*, 1985; Sasaki *et al.*, 1989; Lindauer *et al.*, 1993; Waltimo and Könönen, 1993). Similar devices have been used to estimate dental forces bilaterally (Pruim *et al.*, 1980; Devlin and Wastell, 1985). Some investigators have incorporated strain gauges within restored teeth (Anderson, 1956; Graf *et al.*, 1974; De Boever *et al.*, 1978) or placed them below individual teeth in complete dentures (Atkinson and Shepherd, 1967). More sophisticated transducers have been used to sense bite force magnitude and direction in either two or three dimensions (Graf *et al.*, 1974; Hylander, 1978; Van Eijden *et al.*, 1988b; Mericske-Stern *et al.*, 1992; Osborn and Mao, 1993). The small sizes required in all load-sensing devices—regardless of their designs, the large forces involved, and the need for transducers to be customized to function in a hostile biological environment—seem to have discouraged the simultaneous use of multiple sensors.

Multiple-site force sensing has been attempted with sheets of polymer, photoplastic material (Dawson and Arcan, 1981; Amsterdam *et al.*, 1987), and electrically conductive film (Maness *et al.*, 1987; Chapman, 1989; Waltz, 1991). However, these studies have been of limited value, because complex tooth morphology affects quantitative assessments made with sheets and films that may compress or stretch during biting. Additionally, the methods do not reveal dynamically changing forces.

Lundgren and Laurell (1984) successfully managed to record bite force distribution at multiple locations by fixing conventional transducers within prosthetically restored dentitions. Later, they reported local maximum forces in fixed bridges to be about twice as large in the posterior as in the anterior region (Lundgren and Laurell, 1986a). In unilateral, posterior, two-unit, cantilever prostheses, however, the distal cantilever forces were smaller than those in the anterior regions (Lundgren and Laurell, 1986b). An 80-µm-thick occlusal interference on the distal cantilever unit resulted in a significant increase in local force (Laurell and Lundgren, 1987). These authors attributed their findings to the elasticity of the skeletal system and possible deflections of the bridge cantilevers.

Recently, differential tooth loading has been studied with a three-dimensional finite element (FE) model which quantified elastic deformation of the lower jaw (Korioth and Hannam, 1994a). During unilateral tooth-clenching on multiple contacts, this model predicted higher bite forces at the most posterior tooth locations, consistent with the lever theory. However, it also revealed distinct force peaks in the canine region. This non-uniform, non-monotonic grading of bite force along the dental arch—with the highest values on the molar teeth, the next-to-highest in the canine region, followed by the premolar and incisor regions—reflected complex bending of the mandible, and was attributed both to its form and its elastic properties (Korioth and Hannam, 1994b).

It therefore seems that previous assumptions based on data obtained at singular loading points may not hold for more common and unilateral multiple-point loading, which, theoretically at least, results in reversed force trends in the canine-premolar region. If true, such patterns could have implications for the design and function of occlusal prostheses, especially when these involve multiple load points and osseointegration.

The aims of the following study were: (a) to measure the amount and distribution of bite force at multiple occlusal sites simultaneously during clenching with increasing effort, and (b) to compare these forces for three different occlusal contact patterns. To accomplish this, we used four miniature sensors, placed unilaterally in a maxillary acrylic occlusal appliance, to record multiple bite forces when systematic changes were made to the number and distribution of occlusal contacts on the appliance's contralateral side.

Materials and methods

Prior to the clinical experiment, we tested the prospective design of the transducer system numerically to determine the possible range of forces. This information was used for calibration purposes and to predict whether the theoretical antero-posterior load gradient remained non-monotonic when clenching was simulated on stainless steel areas embedded within acrylic material.

The methods used to construct a FE model of the human mandible have been described in detail elsewhere (Korioth and Hannam, 1994a) and are summarized briefly here. Sections from an imaged jaw were reconstructed into a wireframe model which was in turn used to create a three-dimensional mesh of solid, linear elements (I-DEAS 6.0, Structural Dynamics Research Corp. Milford, OH). A total of 5926 elements made up the cortical and cancellous bone, the dental enamel dentin, the periodontal ligaments and lamina durae, and the articular fibro-cartilagenous tissues. Each element was assigned elastic properties consistent with its tissue type. The model was loaded with multiple force vectors representing muscle tensions scaled according to task. Its output consisted of physical events such as structural displacements, stresses, and strains affecting the jaw, and the reaction forces required to maintain the mandibular system in equilibrium.

A full-arch, maxillary acrylic-resin appliance was also modeled. Its dimensions were measured with calipers from the original and transferred to the model as required. Recesses were created on the left side of the appliance for the simulated load sensors. Material properties for acrylic resin and stainless steel

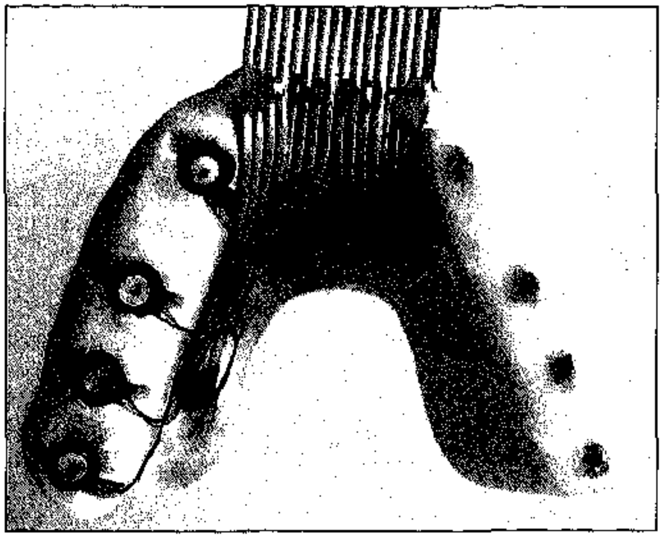

Figure 1. Occlusal view of a full maxillary appliance with four strain-gauge transducers mounted in the right side of the occlusal rim. Points occluding with selected lower teeth are marked bilaterally (Ca, canine; Pm, second premolar; M1, first molar; M2, second molar).

were assigned to the appliance and the transducer elements, respectively. A clenching task with unilateral contacts on the left side was simulated; each contact corresponded to the prospective transducer locations. The contacts between the teeth and the appliance were modeled without freedom of movement in any direction, thus neglecting any sliding action of the teeth on the acrylic and the transducers. Appropriate and different muscle activation levels were assigned to simulate maximum-effort clenching on the appliance. These values were drawn from electromyographic data reported for clenching on full-arch, acrylic-resin multiple occlusal supports (see MacDonald and Hannam, 1984a,b). Reaction forces at the tips of each supporting cusp were then calculated.

Acrylic resin, full-arch, maxillary stabilization appliances were used during the clinical phase of the experimental set-up. Initially, each made simultaneous occlusal contact against the mesio-buccal cusps of the first and the second molars, the buccal cusps of the first premolar, and the incisal edges of the canines on both sides of the dental arch. Tooth separation at the second molar region ranged between 5.0 and 7.5 mm, depending upon the subject. The occlusion of the splint was checked with articulating paper, and by asking the subject to tap his teeth together in unforced "centric relation". Thereafter, four cylindrical holes, 6 mm in diameter and 3 mm deep, were drilled on the right side of each splint so that each of the centers corresponded to the antagonistic canine, second premolar, and first and second molar teeth, respectively, with the base of each hole parallel to the occlusal plane.

Bite forces were registered with four miniature strain-gauge transducers (PS-70KAM260, Kyowa Electronic Instruments Co., Tokyo, Japan) 6 mm in diameter, and 2 mm high. A brass circular plate, 3.0 mm in diameter and 1.0 mm thick, was attached to the upper surface of each transducer. The plate made it possible for us to adjust the occlusion later by adding small amounts of light-cured composite resin to the surface of the brass plate, then grinding it as needed, prior to the experiment, and it permitted the development of simultaneous, even tooth contacts without changing the sensitivity of the transducers. Each transducer (with its attached plate) was calibrated separately with an Instron Tensile Testing Instrument (Model 4301, Canton, MA). Known forces were applied to the center of the brass plate through a small metal hemisphere resembling a tooth cusp. Amplified outputs from the transducers showed high linearities up to 200 N, which was the maximum force applied. The correlation coefficient between the applied load and the registered output was 0.99 in all four transducers. After calibration, the transducers were secured in the splint with small amounts of autopolymerizing resin, and their connecting wires were secured for protection (see Fig. 1).

The experiments were carried out on four adult males between the ages of 23 and 35 years, each with an apparently healthy masticatory apparatus. The subjects provided informed consent that was reviewed by the ethics committee of the University of British Columbia. At the beginning of the experimental session, each subject was asked to tap very lightly on the customized appliance and transducers. A bilateral, even, and simultaneous occlusion was carefully developed by monitoring the force signals on a digital oscilloscope. The acrylic surfaces on the transducers and appliance were adjusted with the aid of thin articulating paper and a slow handpiece until all four transducers yielded simultaneous force signals at high amplifier gain (Fig. 2). When balancing was complete, this was considered to be a type F (full-arch) appliance. The subject was asked to clench on

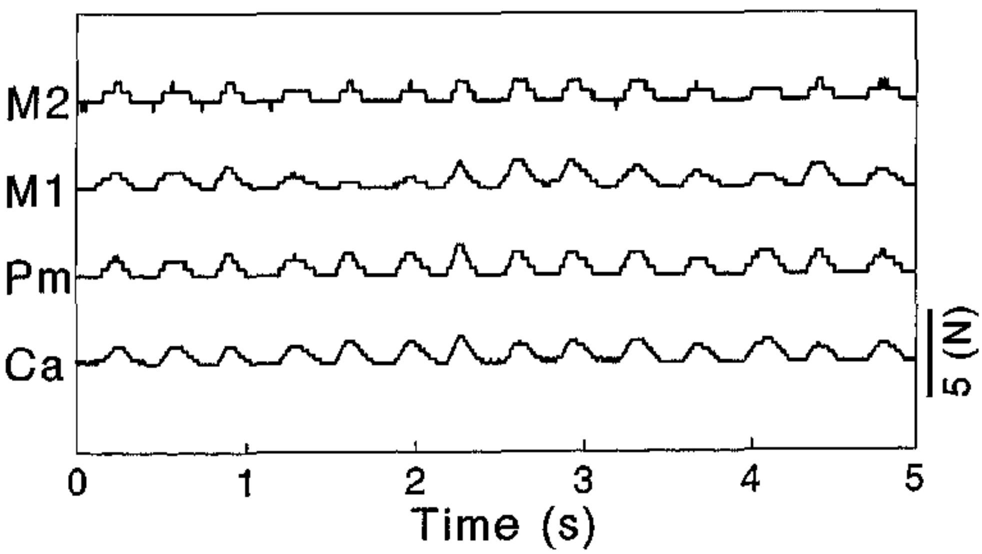

Figure 2. Simultaneous bite forces recorded at high gain during occlusal balancing of a type F (full-arch) appliance. The amplitudes are less than 5 N during extremely light tapping movements made on the appliance prior to maximum effort trials. Abbreviations as for Fig. 1.

the type F appliance for 5 s, and to increase effort until the maximum voluntary clenching level was reached. The trial was repeated three times.

The occlusal scheme was then modified twice while contact on the transducers was retained. In the second design, contacts on the contralateral (left) first and second molars were removed to produce a type R (removed contacts) appliance. In the third, all contralateral tooth contact was eliminated to create a type U (unbalanced) unilateral appliance. In each case, the three-trial clenching protocol was repeated (see Fig. 3).

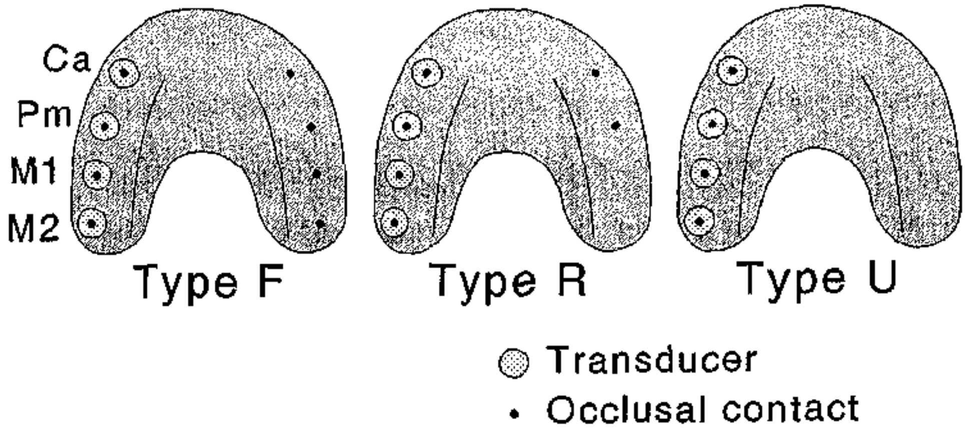

Figure 3. Occlusal contact schemes on the three types of appliances.

Amplified force signals for each of the nine trial clenches comprising a series for one subject were sampled at 200 Hz for 5 s by an analogue-to-digital converter, and stored in a computer (HP9000, Series 380, Hewlett Packard, Palo Alto, CA). These data were converted to forces by means of the individual calibration curves for each transducer.

Thirty-one data points (totaling 155 ms), equally distributed around the peak value of the highest and most stable region of each individual force curve, were extracted from each trial. These were pooled for each subject to make up a total of 93 values selected from the three repeated maximum clenching efforts on each appliance. The pooled data were used to calculate mean bite forces and their standard deviations at each transducer location for the three different appliances. Also, the raw force data were normalized in each instance to the total force for all four sensors. This revealed any changes in the relative contribution at each site, expressed as a percentage.

In addition to the standard deviations, the technical error of measurement (TEM) was computed as the square root of the squared differences between corresponding measurements for each appliance and subject (Knapp, 1992). TEM values between two normalized forces for each tooth were randomly selected from three trials with the same appliance.

Differences in bite force at each of the four locations for the three types of appliance were compared by means of one-way analysis of variance (ANOVA) and the Bonferroni multiple-comparison tests at the 5% level of significance.

Results

During simulated clenching on unilateral occlusal contacts, tooth loads were distributed similarly to those reported previously for the FE simulation using natural teeth, *i.e.*, a peak was evident in the canine region (see Fig. 4). The simulation experiment thus confirmed that, on theoretical grounds at least, the use of an acrylic appliance containing load sensors would make it possible to demonstrate force reversals in the canine regions should they occur in living subjects. We assumed that any elasticity inherent in the appliance/transducer complex would be unlikely to interfere significantly with the elastic properties of the mandible as a whole when experimental tooth-clenching was carried out.

Bite forces for three trials carried out by a subject clenching on a type F (fully balanced) appliance are shown in Fig. 5. Differences in the way the subject approached the timing of clenching in the trials are evident. Differences in timing when the task was commenced were common during the experiment, despite attempts to regulate them. Nevertheless, consistent relationships between the forces generated at each dental site are observed. For instance, the amplitudes of forces at the first and second molars cross at

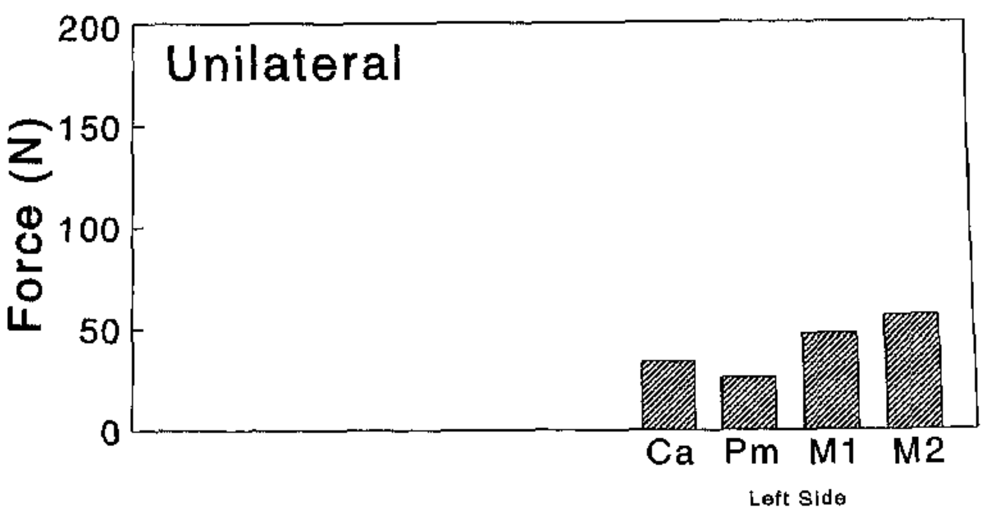

Figure 4. Bite force distribution predicted by the FE model during maximum unilateral clenching. The tooth locations at which forces were calculated are shown on the horizontal axis (Ca, canine; Pm, second premolar; M1, first molar; M2, second molar).

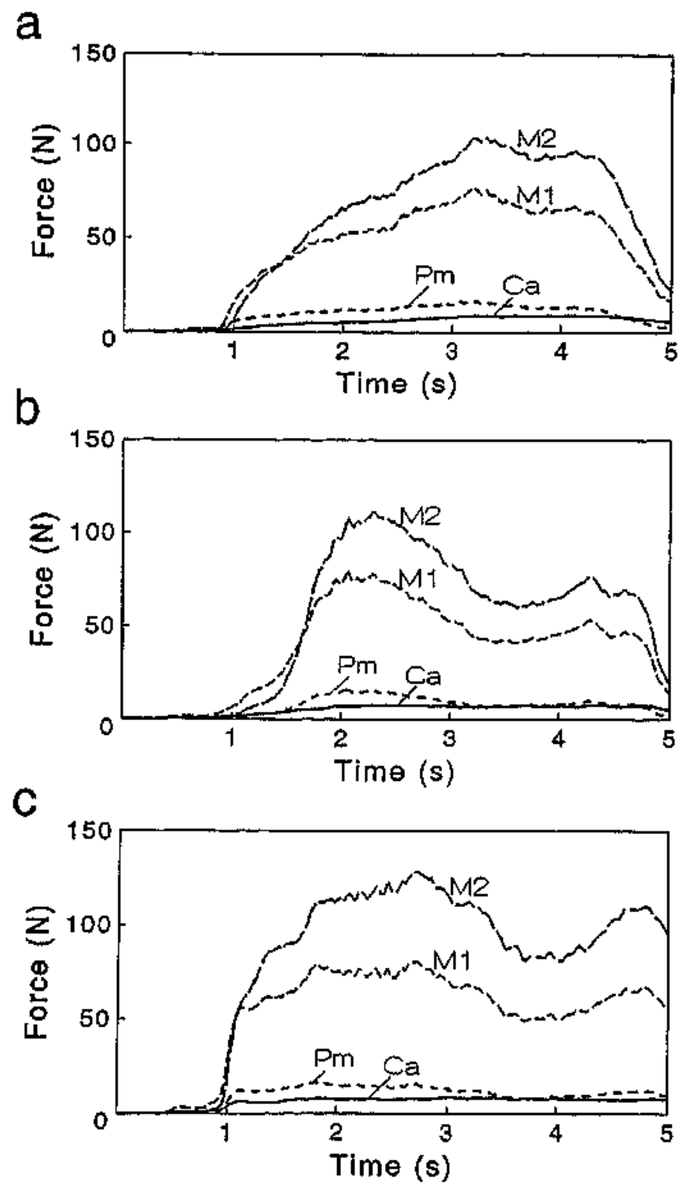

Figure 5. Bite force magnitudes produced by subject 1 during maximum voluntary clenching on a type F appliance. Forces are plotted against time for three trials (a, b, and c). Thirty-one data points distributed around each time-line in each trial were pooled to yield 93 values which were used to calculate the mean bite force for each transducer for all three trials (see text and Fig. 7). Abbreviations as for Fig. 1.

approximately 50 N, but later show synchronous changes at higher levels of biting in all three trials. This contrasts with forces in the canine and premolar regions, which remain low and relatively constant throughout all trials. This trend was apparent in all subjects; quantitative relationships in amplitude between bite forces at each location differed little throughout the trials, despite changes in temporal shaping.

Fig. 6 represents forces and ratios for the type F appliance for selected trials of all four subjects. In each case, relative forces are expressed as percentages of the simultaneous total forces (100%). The graphs in this Fig. are representative of the consistent tendencies found in the experiment. As biting effort began to increase, the relative contribution of bite force at the canine suddenly decreased, while that at the second molar increased. These were evident as early "cross-overs" in the graphs depicting force ratios. Overall, the bite force ratio for each tooth was generally constant at high bite levels.

Mean bite forces and standard deviations for all subjects and types of appliances are shown in Fig. 7. Bite force profiles differed between subjects, *e.g.*, subject 3 produced the lowest mean forces for the group, while subjects 1 and 4 produced the highest. However, the latter differed in the regions over which their largest forces were produced. Subjects showed an acceptable level of repeatability, since no TEM values exceeded 10% (see Table).

When using the fully balanced (type F) appliance, two subjects (1, 2) revealed bite forces which increased monotonically as the location of the transducer moved posteriorly. Subject 3, however, showed force peaks at the premolar and second molar, while Subject 4 produced his highest forces in the premolar and first molar region. These individual patterns generally remained constant when the balancing-side molar contacts were removed (type R appliance), but they altered in Subjects 1 and 2 when the unilateral (type U) appliance was created. Here, respective forces in the canine and premolar regions increased disproportionately.

When bite forces at individual locations on the fully balanced appliance (type F) were compared statistically with their counterparts on the unilateral (type U) appliance, there was a significant increase at the canines in contrast to the second molars, which showed little change. Thus, the highest forces in the canine regions were always reached when the appliance with the unilateral contacts was used. Although only one subject showed a marked peak in the canine region which seemed to confirm the FE predictions, peaks in the approximating premolar region were evident in all other subjects when this appliance was used. This confirmed the non-monotonic distribution of anteroposterior tooth forces in all subjects and limited the generalized extrapolation of the predictive results.

Fig. 8 demonstrates the distribution of estimated total force for all three types of appliances. The total force was defined as the sum of the four mean forces measured by each transducer. It ranged between 128 N and 330 N, depending upon subject and appliance, and increased significantly in only two of the four subjects.

Discussion

Despite the generally high repeatability of maximum bite force distribution in this study, variations occurred when tasks were repeated by some of our subjects. In these instances, it is possible that equal balance between occlusal contacts was not completely maintained. Although strenuous efforts were made to ensure balance before each experiment (by light tooth-tapping, with the requirement for simultaneous forces to be visible at high gain), small shifts in jaw position during repeated runs, or changed load-displacement profiles between sequences for individual

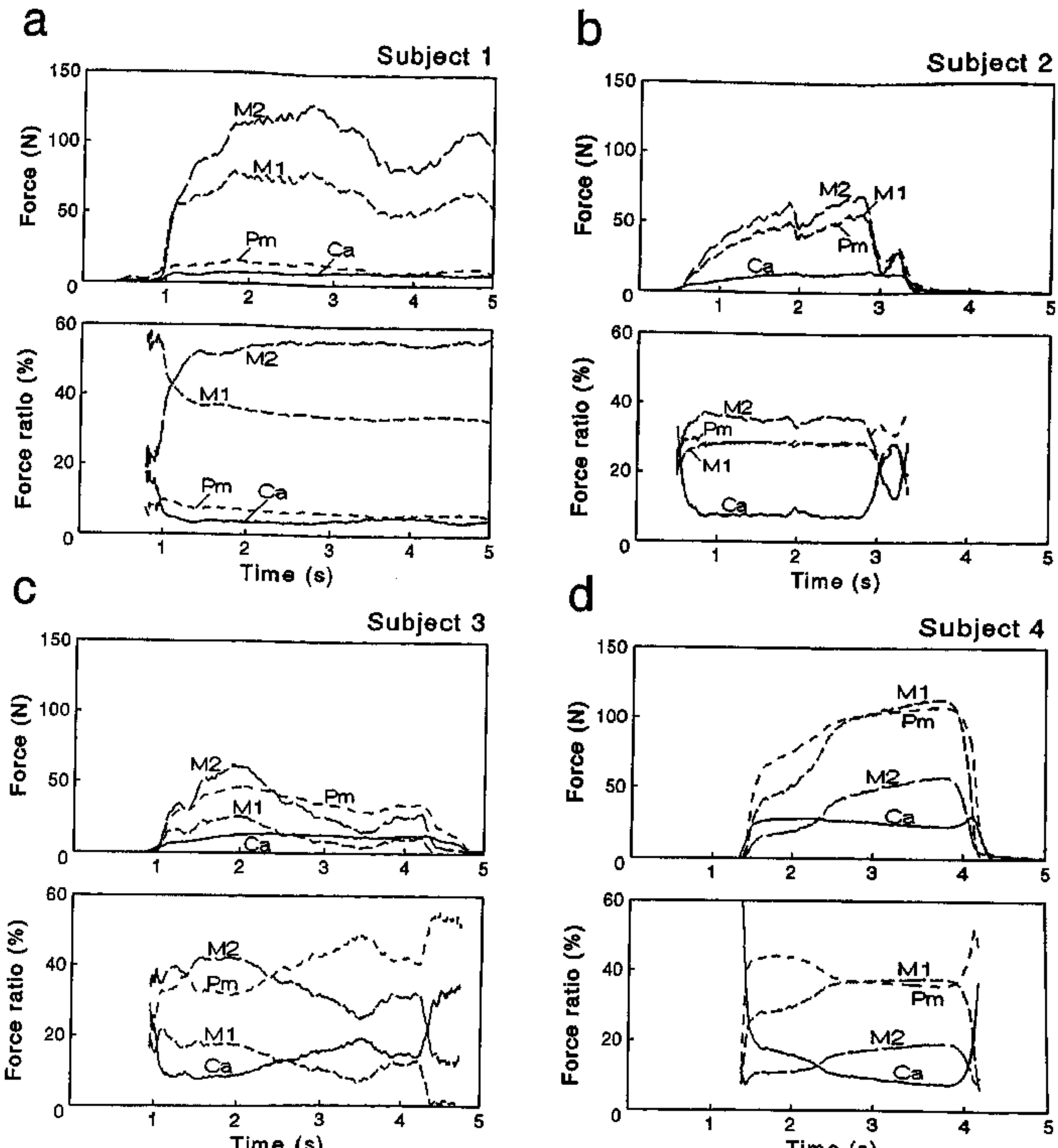

Figure 6. Forces and corresponding force ratios plotted against time when representative maximum voluntary clenching trials were performed by all subjects with a Type F appliance (a, b, c, and d). The force ratios were expressed as percentages of the total forces (100%) measured with the four transducers. Abbreviations as for Fig. 1.

teeth, could both have altered the relative balance of loads recorded from clench to clench. It is known that force distribution in dentitions with fixed-cantilever prostheses can be sensitive to occlusal increases as small as 80 µm (Laurell and Lundgren, 1987, 1992).

Clenching on the bilaterally supported occlusal appliance produced a distribution of maximum bite forces similar to that reported in implant-supported dentitions (Lundgren *et al.*, 1987; Falk *et al.*, 1989; Falk, 1990). In both cases, the forces increased posteriorly. It should be noted, however, that

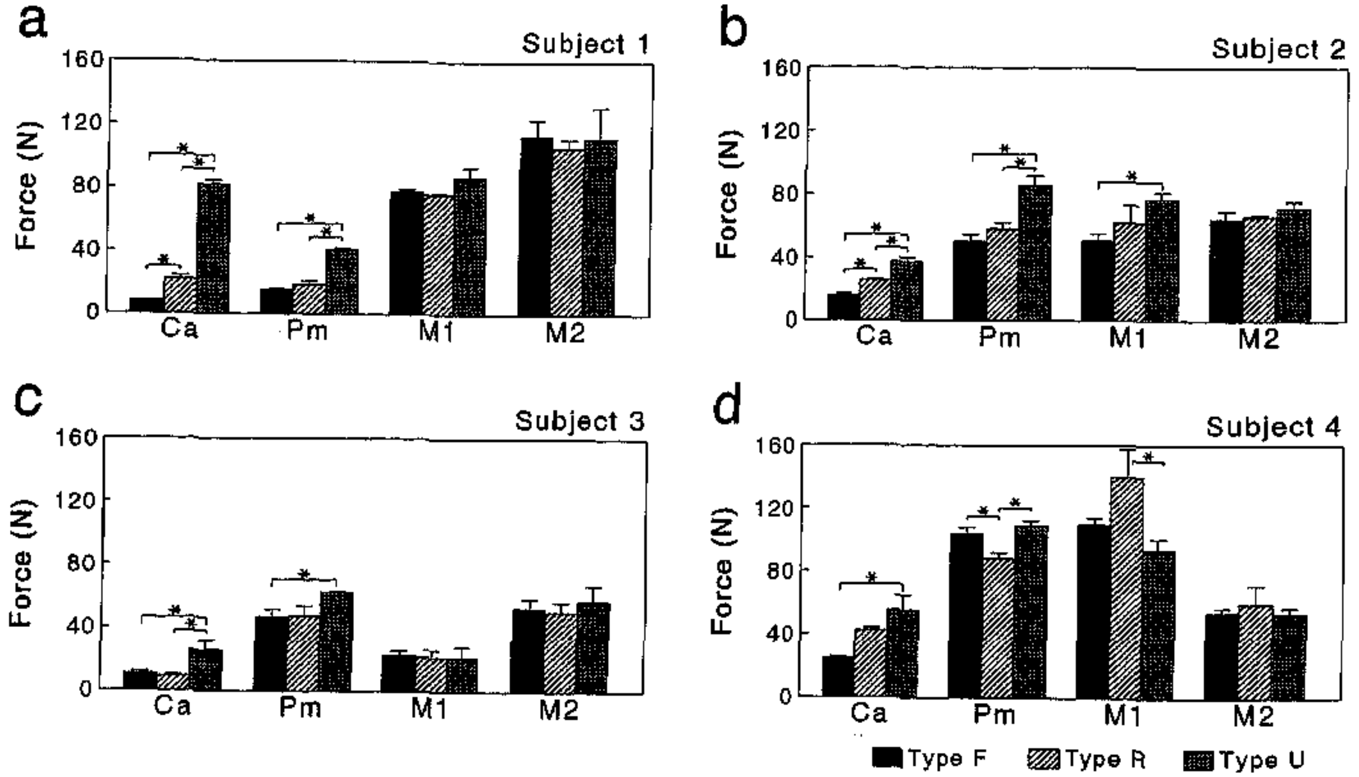

Figure 7. Mean maximum bite forces (plus standard deviations) obtained simultaneously at different locations on the right side of the dental arch for each subject using the three different types of appliances. Significant differences ($p < 0.05$) between forces on the three different appliances are indicated with asterisks. Abbreviations as for Fig. 1.

forces were measured at two (Lundgren *et al.*, 1987) and four (Falk, 1990) contact points on each side of the dental arch in the implant studies, and that these locations were more anterior than those used in the present study. Experiments in dentitions restored with fixed bridges with bilateral extensions have also revealed bite forces that are larger posteriorly than anteriorly (Lundgren and Laurell, 1986a).

Differences in the antero-posterior bite force gradient observed during maximum effort for bilateral dental support could be due to intersubject differences in masseter muscle angulation and/or mandibular ramus height, as predicted by an earlier jaw model (Ward and Molnar, 1980). In this model, higher anterior occlusal loads were evident for more anteriorly placed simulated masseter muscle forces, and were also seen when the vertical distance between the condyle and the occlusal plane approximated 2 cm.

The extent to which bite forces measured on appliances such as those used in our study represent forces normally generated between natural teeth is uncertain. In natural dentitions, each tooth has a contact pattern, usually consisting of more than one point or area when opposing another tooth or teeth. This situation differs from the present study, where individual contacts were represented as points on one supporting cusp for each mandibular tooth. Thus, relatively higher bite forces could be generated at the molars in natural dentitions, because natural teeth have more potential contact points per tooth, offering the opportunity for angles of tooth force other than those offered by the tip of one supporting cusp. This is made more likely by relationships that have been established between the activities of jaw muscles and the distribution of occlusal contacts (MacDonald and Hannam, 1984a,b), and by predictions from static models which correlate changes in the direction of bite force with variations in muscle activity (Van Eijden, 1990; Van Eijden *et al.*, 1990). In our study, we measured only forces aligned perpendicularly to the occlusal plane. These forces, however, were assumed to be considerably larger than any horizontally directed components that might have been present in the natural dentitions. Jaw stability was an important factor in the experiment, and the presence of any significant horizontal components of force would have increased the likelihood of undesirable motion of the mandible.

Maximum forces consistently increased at the more anterior locations when the occlusal scheme on the appliance was modified from bilateral dental support to unilateral contacts on the side of the transducers. This change was most significant at the canine, and in marked contrast to the second molar, where bite force generally remained constant, independent of the type of appliance used. These increases in anterior bite force could be attributed to several factors. If indeed the bilateral muscle activation remains unchanged when all contralateral occlusal support is removed during clenching on an acrylic appliance (Wood and Tobias, 1984), then the absence of contacts may have allowed the contralateral corpus to undergo forceful twisting and

Table. Mean and standard deviations (SD) of a total of 93 selected bite forces (31 forces by 3 trials each) *per* appliance type and subject

		Subject 1		Subject 2		Subject 3		Subject 4	
Appliance	Location	Mean	SD	Mean	SD	Mean	SD	Mean	SD
	Ca	8.0	0.6	15.9	1.4	10.7	1.6	24.7	1.3
Type F	Pm	15.6	0.5	50.7	4.4	46.5	4.9	104.1	3.9
	M1	77.3	1.9	50.8	4.6	22.9	2.6	109.6	4.3
	M2	112.8	10.5	64.6	5.1	52.4	6.4	53.9	2.7
	Ca	21.9	2.1	25.9	0.6	9.4	0.9	42.3	1.5
Type R	Pm	18.9	2.1	58.1	4.5	47.1	6.5	88.0	3.9
	M1	75.7	1.0	62.6	10.8	21.5	4.6	140.2	17.4
	M2	105.5	5.3	66.1	1.5	50.2	6.5	59.2	11.9
	Ca	81.2	2.9	37.5	2.5	25.5	6.1	55.4	9.1
Type U	Pm	41.0	0.8	85.3	6.0	62.4	0.7	108.7	3.2
	M1	86.7	6.5	76.7	4.4	21.2	7.0	93.6	6.8
	M2	111.5	19.8	71.4	3.9	57.2	10.0	53.0	3.7

ɔending (Hylander, 1979; Korioth and Hannam, 1994a). While unrestrained parasagittal bending on the contra-ateral side would mainly ıccount for vertical forces, the ıuperimposed twisting of the :orpus would transmit these orces across the fused ymphysis, to be countered at he ipsilateral tooth contacts. Since the canine is the most ınterior tooth, it would resist he complex bending forces irst. This effect, summed vith the residual resistance orce generated by the psilateral musculature, could ccount for the relatively ıigher force peaks predicted ıy deformable jaw models nd observed in the present tudy. However, the specific amounts of these loading ffects, which may explain the relative increase in bite force ı the canine region during unilateral clenching on multiple eeth, remain to be shown.

The biomechanical properties of alveolar bone and the eriodontal ligament significantly influence bite force and ıe stress-bearing capabilities of the jaw (Daegling *et al.*, 992). Since the modulus of elasticity of the periodontium ıpproximately 2 to 3 MPa; Ralph, 1982; Mandel *et al.*,1986) s much less than that of mandibular cortical bone ıpproximately 10 to 20 GPa; Dechow *et al.*, 1993), the ıaximum compressibility of the periodontal ligament will each its limit prior to that of the mandibular cortical bone. During clenching, the molar teeth can resist more ompression than the anterior eeth, due to their larger eriodontal areas. In addition, ıe molar region of the ıandible experiences more ompressive load than the nterior region, due to the roximity of the masseter and ıe medial pterygoid muscles. ollectively, these two factors ıay account for the significant ıanges in bite force ratios bserved in the present study. orce ratios at the second molar nd the canine changed ynamically in a consistent, but pposite, manner during the ıitial stage of biting. Whereas ıe force ratio at the second ıolar increased, the force ratio : the canine decreased nmediately after the onset of iting in all subjects. These ıanges were not evident at either the first molar or the premolar. A possible explanation for these shifts may be the combined effect of the elastic characteristics of the mandibular tissues, in particular those of the cortical bone and the periodontal ligament. If these tissues were assumed to consist of rigid materials, the ratio of force distribution should be relatively constant and independent of changes in biting strength. Even so, some changes in bite force ratio could occur due to the influence of muscle co-activation and changes in the direction of muscular force.

The total maximum bite force generated on the appliance with unilateral support was larger than that on the appliance with bilateral support. However, the true total force for the appliance with bilateral support should be

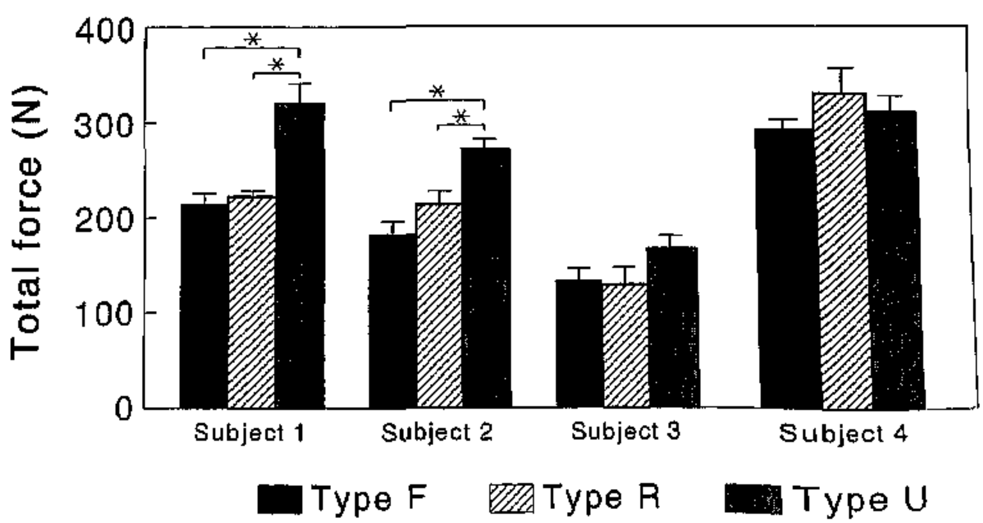

Figure 8. Total mean maximum bite forces (plus standard deviations) plotted against the types of appliances for each subject. Those forces are defined as the sum of four forces simultaneously measured with the transducers. Significant differences ($p < 0.05$) between forces on the three different appliances are indicated with asterisks.

estimated at twice the measured value, since bite force was measured unilaterally with this appliance. The estimated true total maximum forces during bilateral clenching therefore ranged from 265 to 585 N. These values seem to be generally lower than those previously reported by Pruim *et al.* (1980) for seven subjects biting bilaterally at the first premolars (633 ± 210 N), first molars (965 ± 276 N), and second molars (756 ± 289 N), or by Gibbs *et al.* (1981), who used a sound transmission system to predict total forces around 740 N. However, our results agreed well with the total maximum bite forces reported by Lundgren and Laurell (1986a) and Falk *et al.* (1989), whose subjects' dentitions had been restored with fixed bridges (320 ± 117 N) or with implant-supported prostheses occluding with complete dentures (336 ± 97 N).

There are many experimental factors which determine maximum bite forces, including methodological, functional, physical, and psychological influences (Carlsson, 1974; Hagberg, 1987). Among the craniofacial variables that determine the mechanical performance of the masticatory apparatus and thus could affect the bite force gradient are jaw muscle size and the direction of muscle action lines (Weijs and van Spronsen, 1992), and, most importantly, spatial skeletal relationships such as zygomatic arch width, ramus height, and gonial angle.

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