Control of Jaw-Clenching Forces in Dentate Subjects

Marco Testa, PT
Senior Lecturer
Master in Musculoskeletal Disorders Rehabilitation
School of Medicine
University of Genoa–Campus of Savona
Savona, Italy

Mara Rolando, MEng
Research Assistant
Department of Neuroscience
Physiology Division
University of Torino Medical School
Torino, Italy

Silvestro Roatta, PhD
Researcher
Department of Neuroscience
Physiology Division
University of Torino Medical School
Torino, Italy

Correspondence to:
Dr Silvestro Roatta
Dip. Neuroscienze, Sez Fisiologia
Università di Torino
c.so Raffaello 30
10125 Torino, Italy
Fax +39 011 6708174
Email: silvestro.roatta@unito.it

Aims: *To characterize the control of jaw-clenching forces by means of a simple force-matching exercise.* ***Methods:*** *Seventeen healthy subjects, provided with visual feedback of the exerted force, carried out a unilateral force-matching exercise requiring developing and maintaining for 7 seconds a jaw-clenching force at 10%, 30%, 50%, and 70% of the maximum voluntary contraction. The task was repeated three times in each of two sessions. Motor performance was assessed, for both left and right sides, by different indices quantifying mean distance (MD), offset error (OE), and standard deviation (SD). Their dependence on force intensity, side, and time was assessed by ANOVA.* ***Results:*** *All error indices increased with the intensity of contraction in absolute terms. After normalization with respect to force level, the average performance in the second session was characterized by MD of 8.1% ± 2.6, OE 4.8% ± 2.9, and SD 12.7% ± 6.7 (mean ± standard deviation). Assessment of performance exhibited good reliability for all indices (intraclass correlation coefficient ranging from 74% to 88%). The motor performance improved with repetition (*P *< .01), varied considerably between subjects, was not correlated with gender or age (*P *> .05) but was highly correlated between left and right side (*P *< .01).* ***Conclusion:*** *The adopted approach is adequate to provide for an objective assessment of individual force control, although the presence of a learning phase must be taken into account.* J OROFAC PAIN 2011;25:250–260

Key words: accuracy, force transducer, masseter muscle, motor control, precision

Mastication is a complex function strongly based on sensory-motor integration that is impaired by several clinical conditions affecting the orofacial area; among them, temporomandibular disorders (TMD) are the most common chronic pain disorder.[1] In fact, pain is known to affect motor control in terms of modulation of inhibitory and excitatory motor reflexes of the jaw,[1–4] proprioceptive and sensory functions as assessed by interdental discrimination of thickness and repositioning tasks,[5] as well as masticatory performance and bite force.[6,7]

One important aspect of motor control concerns the accuracy and precision (steadiness) of muscle force in voluntary contractions.[8–10] These variables are conveniently investigated in isometric contractions by instructing the subjects to grade muscle force according to constant or dynamic targets superimposed to the visual feedback of the developed force. Most of these studies have been conducted on limb muscles while only a few have focused on jaw muscles.[11–13] While accuracy, ie, how close one gets to the target, essentially depends on the central motor command, force steadiness is potentially influenced by both peripheral and central mechanisms.[10] Accordingly, force steadiness was shown to be affected in both musculoskeletal[14–17] and central nervous system disorders.[18,19] More specifically, reduction in force steadiness was reported in patients affected by subacromial impingement syndrome[16] and by knee osteoarthritis[15] and was also observed to occur after bed rest[14] and prolonged muscle unweighting,[17] as well as during experimental pain.[20,21] Force steadiness was also suggested as a possible clinical indicator for cerebral palsy[22] and stroke-associated disorders.[19] Importantly, both accuracy and steadiness are known to decrease with an increasing level of contraction and to worsen with age.[8–10,23] Recently, Marmon et al[24] described a significant relation between measures of steadiness and of hand function, suggesting that the physiological mechanisms behind force steadiness also contribute to the performance of fine hand motor tasks.

Besides the large number of studies on limb muscles and the potential clinical application of the objective and quantitative characterization of the control of muscle force, only a few studies appear to have investigated the control of jaw-elevator muscles by this approach, ie, by assessing the ability to adjust the clenching force according to visually-presented target levels.[11–13] The results of these studies indicate that, also in jaw muscles, the matching error increases with the force level[11,12] and with the age of the subject,[11] but only small force ranges were explored (0.3 to 4 kg) and were not adjusted to the individual maximum voluntary contraction (MVC). Van Steenberghe et al[13] elegantly compared the performance of jaw muscles with forearm and calf muscles in a similar force-tracking task (force level: 15% to 20% MVC) in the same subjects. They provided evidence of a larger matching error in jaw muscles, supporting the notion that visual control of force is poor in these muscles, possibly because the mandible is not usually controlled by a visual feedback. However, they did not provide any quantitative measure of the accuracy or precision in the execution of the tasks.[13]

The aim of the present study was to characterize the control of jaw-clenching forces by means of a simple force-matching exercise. To address this aim, different performance indices, including accuracy and force steadiness, were correlated and tested for reliability. A novel force sensor, based on a piezoresistive film transducer, was employed for the measurement of unilateral clenching force.

Materials and Methods

Seventeen healthy subjects (9 females; 8 males), aged 24 to 40 years (28.4 ± 6.67, mean ± standard deviation), volunteered in this study, approved by the University of Torino Ethical Committee. The subjects, recruited among the attendants of a postdegree master course, were informed about the experimental procedure and gave their written consent.

Inclusion Criteria

Inclusion criteria were an absence of TMD signs and symptoms assessed by a physiotherapist with experience in TMD (MT), according to the Research Diagnostic Criteria for TMD,[25] and occlusion in class I according to Angle's Classification.

Force Transducer

Force measurement was based on a piezoresistive force transducer Flexiforce A201 (Tekscan), featuring a load range of 100 lb, equivalent to 440 N, and a sensitivity of 0.05 V/lb). A special housing (Fig 1) was developed to allow the sensor to measure voluntary clenching force without undergoing permanent damage and to reduce discomfort for the subject during clenching. The external thick rubber layer provided the possibility of small yielding of the surface under the teeth, thereby generating a wider contact surface and lower local pressure. It also provided improved comfort during the clenching, as compared to a hard surface. The thin hard plastic foil gave a flexible support and a graduated handle for the housed sensor. These two layers were stuck together by a thin bi-adhesive foam. The force transducer was inserted in between an inner hard plastic layer and a metal disk (10 mm in diameter) located exactly below the sensory area of the transducer, which granted that all the force lines between upper and lower teeth were conveyed through that area.

The housed sensor was then inserted into a disposable latex glove for protection from saliva. The final thickness was 9 mm and decreased approximately to 5 to 6 mm after some pressure was exerted by the teeth, slightly accommodating in the rubber layer and compressing the bi-adhesive foam layer. A graduated handle provided a measure of the anteroposterior location of the sensor in the mouth, thus facilitating accurate repositioning in different sessions (Figs 1b and 2). A subsequent laboratory testing verified that the original transducer properties were not impaired by the housing. The linearity error of both the bare and housed sensor was below 5%, while the difference between the output of the bare and housed sensor was always below 6%.

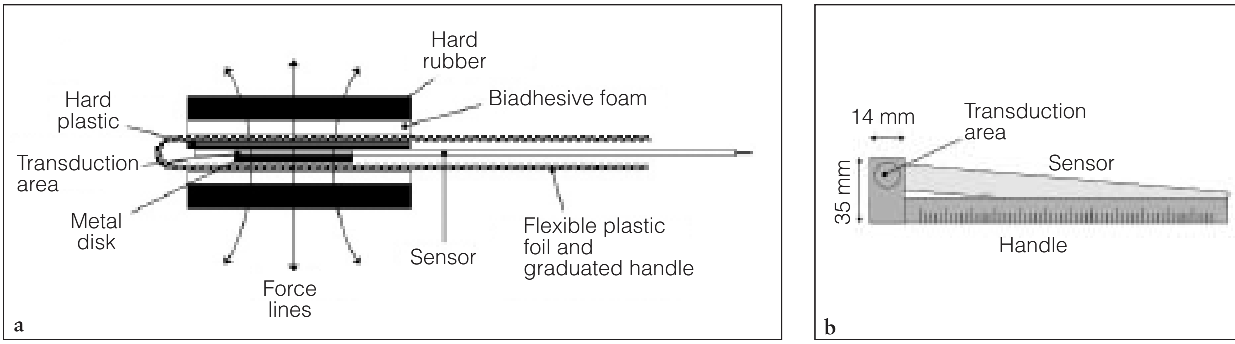


Fig 1 Scheme of the sensor housing. *(a)* Cross section; *(b)* top view.

Fig 2 Photograph of the sensor in situ. Note how the anteroposterior placement of the sensor can be controlled by the position of the incisors on the graduated handle.

Experimental Setup

The subject sat on a comfortable chair without head support, with the trunk in an erect posture and natural head position. One housed load sensor was located unilaterally in between the molar teeth; the distance between the sensor and the front teeth was noted in order to maintain the same position during all recordings. The force signal was digitally acquired on PC (USB-6211 DAQ module, National Instruments) and displayed on a screen to provide the subject with a visual feedback of exerted biting force. The acquisition software was developed under LabVIEW (graphical programming environment, National Instruments).

Experimental Protocol

First, the force produced during a MVC lasting 3 to 4 seconds was recorded three times separated by 2-minute resting periods, the subject being provided with a visual feedback of the exerted force. The force signal was low-pass filtered with a 0.5-second moving average and the maximum value observed over the three contractions was taken as MVC and used as reference for the exercise in that session; the procedure was performed for both sides in random order at the beginning of each session. The subject was familiarized with the visual feedback and was allowed to rest for 2 minutes before starting the exercise.

The subject was then visually presented with a sequence of upgrading and downgrading different target force levels (10%, 20%, 30%, 50%, and 70% of MVC) (grey line in Fig 3). On the same display, the current force level exerted by the subject was also displayed (black trace in Fig 3).The subject was instructed to regulate the clenching force so as to reach and maintain as precisely as possible the level indicated by the target. Each force level had to be maintained for 5 seconds. A resting interval of 7 seconds was inserted in-between the different target levels (Fig 3).The whole sequence was preceded by an additional target at 10% MVC (not shown in Fig 3) that was excluded from the analysis and was meant to accustom the subject to the force scale of the display.

This 2-minute long exercise was repeated three times alternatively by both sides, with resting periods of 4 minutes during which the sensor was re-

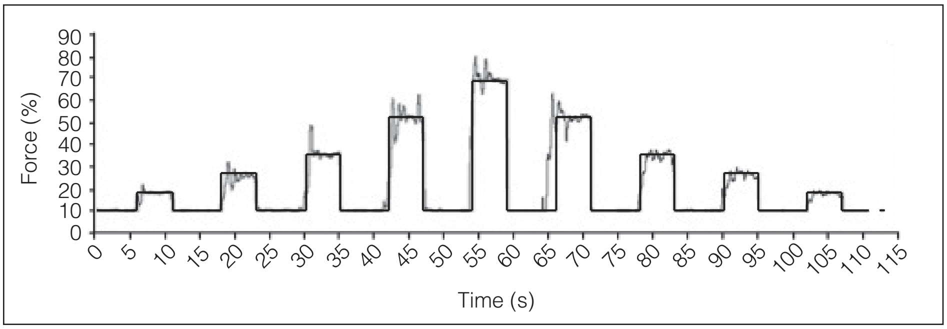


Fig 3 Display of the exercise performed by one subject. The black solid line indicates the force levels (10% to 70% MVC) to be reached and maintained for 5 seconds. The gray jagged line is the force exerted by the subject, updated in real time.

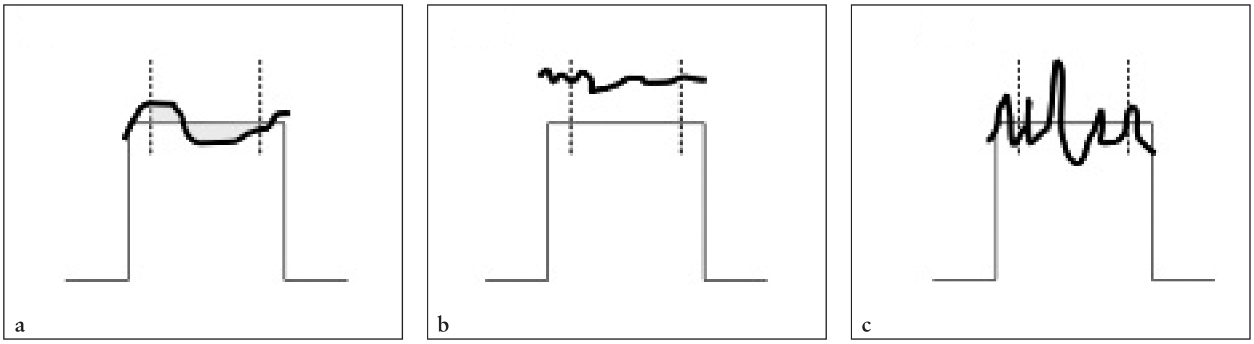


Fig 4 Schematic illustration of the meaning of the different indices. *(a)* MD is proportional to the area enclosed between the force signal and the target; *(b)* example resulting in high OE (indicating displacement of the average force level from the target) and low SD (indicating the variability of the force signal); *(c)* example resulting in low OE and high SD. Thick line = force signal. Thin line = target force to be matched. The dashed lines delimit the interval for analysis.

moved from the mouth and the subject was allowed to move the mandible and talk. In each exercise, the sensor was accurately repositioned in the same location with the help of the graduated handle (Fig 1b). The subjects started to exercise with the left or right side on a randomized basis. The whole sequence of bilateral MVC assessment and six repetitions of the exercise (three repetitions per side) was repeated in a second session on the following day. The exerted force was continuously acquired during each exercise and saved for offline processing.

Analysis

The individual performance was assessed by means of three indices: mean distance (MD), offset error (OE), and standard deviation (SD). All the indices were computed over the central 3-second interval of target presentation (see Fig 4). MD is the average of the absolute difference between the target level and the recorded force and is a global indicator of the goodness of match. The OE is computed as the difference between the average force and the target and reflects the accuracy of the matching: it indicates whether the force signal is, on average, centered or displaced above (positive OE) or below (negative OE) the target level, irrespective of force steadiness. The SD component is the standard deviation of the force signal and reflects force variability, irrespective of its vicinity to the target. To avoid reciprocal elision between positive and negative terms when averaging OE over different subjects, the absolute error was also computed and indicated as |OE|.

All indices were also evaluated in relative terms, ie, normalized to the target force level, expressed in (%) and averaged over all target levels in the task, thus providing an average individual score of motor performance.

Statistical Analyses

Data are presented as mean ± standard deviation in the text and mean ± standard error in the bar graphs.

A first analysis was performed by a two-way ANOVA for repeated measures to investigate the dependence of the three variables (MD, |OE|, and SD) on the intensity of contraction (n = 5 target levels) and on side (n = 2).

On the normalized variables, averaged over target level and over the two sides, a multivariate two-way ANOVA for repeated measures, followed by the Tukey Honestly Significant Difference (HSD) post-

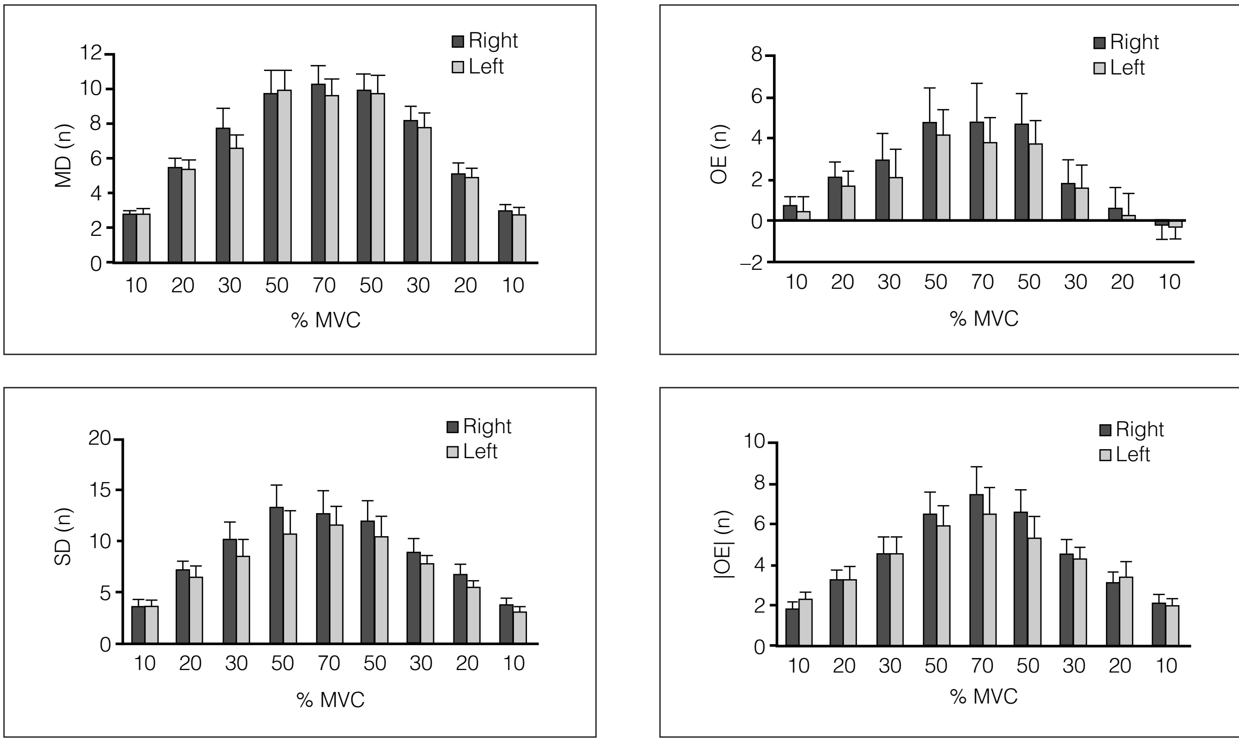


Fig 5 Performance of the 17 subjects when reaching for the nine force levels in the task, as indicated on the X axis. Each graph is obtained by averaging over all six trials and all subjects.

hoc test, was used to test the dependence of the three variables on Session (two levels) and Trial (three levels). Dependency of MD% on gender was assessed by the Student *t* test and its correlation with age by linear regression analysis.

Test-retest reliability was evaluated by means of the intraclass correlation coefficient (ICC), computed using a random-effect one-way ANOVA. Normality of distribution and equal variances of the groups examined were verified in advance as required by ANOVA.

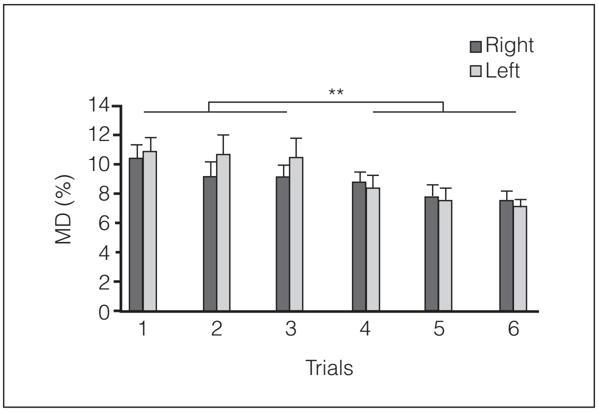
Results

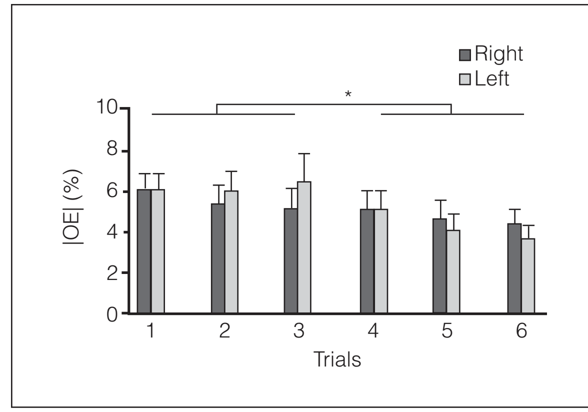
Experimental Series on Healthy Subjects

The MVC was not significantly different between sides ($P > .05$). MVC on the first day was 308.1 ± 102.1 N and increased to 355.8 ± 55.1 N on the second day ($P < .01$; average of the two sides).

The error indices MD, OE, and SD were found to be dependent on the intensity of the contraction ($P < .01$, for all variables), although this dependency appeared to lessen at high force levels flattening at about 50% MVC, particularly for MD and OE. This effect is illustrated in Fig 5, showing the average performance in the task (average of all tasks in all subjects). None of the variables was dependent on side ($P > .05$). Notably, OE was above 0 in all but the latest contraction level, indicating that the subjects tended to overshoot the target level (Fig 5b). To avoid elision of positive and negative terms, the |OE| was computed (Fig 5d) and used in the analysis.

To minimize the dependence of the errors on the level of contraction, the error indices were normalized to the target force level. Although normalization did not completely eliminate the dependence of the normalized variables on the force level, which remained statistically significant (MD: $P < .01$; OE $P < .05$; SD: $P < .01$), it reduced the relative range of change of the variables, MD and OE in particular. In addition, all normalized variables now decreased with increasing contraction level. The coefficient of variation (CoV) of MD values over the different contraction levels reduced from 42% to 14% after normalization. CoV reduced from 42% to 13% for OE and increased 40% to 44% for SD after normalization. In spite of the residual dependence with target level, an overall indicator of the individual performance over the whole trial was obtained by averaging the normalized variables over all target levels in





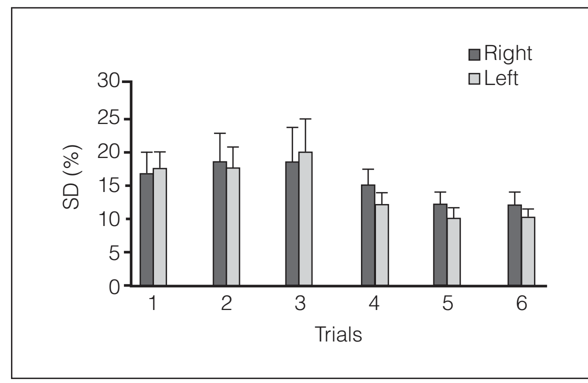


Fig 6 Average performance in the six trials in the first (trials 1, 2, 3) and the second session (trials 4, 5, 6) as assessed by the three error indices MD, |OE|, and SD. The errors are expressed in relative terms as percent of the target force level. MD and |OE| were significantly decreased in the second session with respect to the first (**$P < .01$; *$P < .05$) (n = 17 subjects).

the task. Thus, for each variable, this average value expressed the error in terms of percentage of the target force level. Figure 6 shows the average performance of the subjects in the six trials over the two sessions. Multivariate ANOVA revealed a dependence on session ($P < .05$) but not on trial ($P > .05$). Therefore, by pooling together data from the two sides and the three trials in each session, MD was found to decrease from the first to the second session from 10.1 ± 3.3 % to 8.1 ± 2.6 % ($P < .01$), OE from 6.2% ± 3.2 to 4.8% ± 2.9 ($P < .05$), and SD from 18.1% ± 15.0 to 12.7% ± 6.7 ($P > .05$).

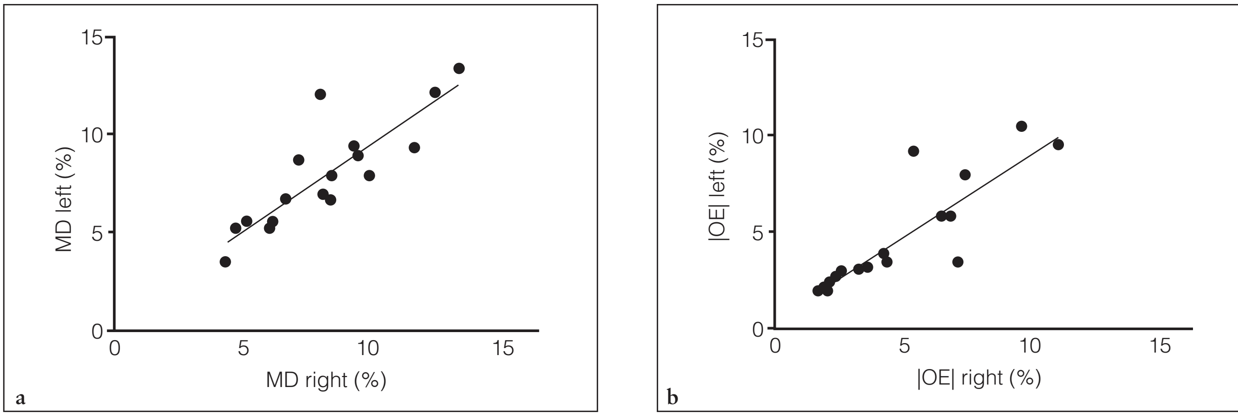
The study also investigated whether the nonsignificant difference between right and left sides was the result of a balance between those performing better on the left and those performing better on the right or instead resulted from a real and consistent symmetry in performance in the examined subjects. The correlation between the performance of the two sides was rather poor in the first session (r^2 values for MD: 0.49, $P < .01$; OE: 0,28, $P < .05$; SD: 0.12, $P > .05$) but markedly increased in the second session (r^2 values for MD: 0.72, $P < .01$; OE: 0.74, $P < .01$; SD: $r^2 = 0.56$, $P < .01$), as shown in the scatter plots of Fig 7. The spreading of the dots along the midline, particularly for MD and |OE|, indicates indeed a general symmetry in the performance of the two sides.

Scatter plots were used to investigate whether there was a correlation between the indications of performance provided by the three indices (Fig 8). In all cases a significant correlation was observed, |OE| versus MD: $r^2 = 0.85$, $P < .01$; SD versus MD: $r^2 = 0.76$, $P < .01$; SD versus |OE|: $r^2 = 0.83$, $P < .01$ (second session, all trials, both sides).

The method exhibited good reliability for all indices as indicated by the ICC evaluated both in the first session (MD: 74%; OE: 76% SD: 88%) and between the first and second sessions (MD: 60%; OE: 80%; SD: 44%). No correlation was found between the motor performance and individual characteristics of the subject, such as gender ($P > .05$) and age ($r^2 = 0.13$) (Fig 9).

Discussion

A simple task was devised to assess the individual capacity of controlling unilateral biting force, based on a newly developed force sensor and on a computerized procedure. In a force-matching task, the subjects had to sustain short, nonfatiguing contrac-



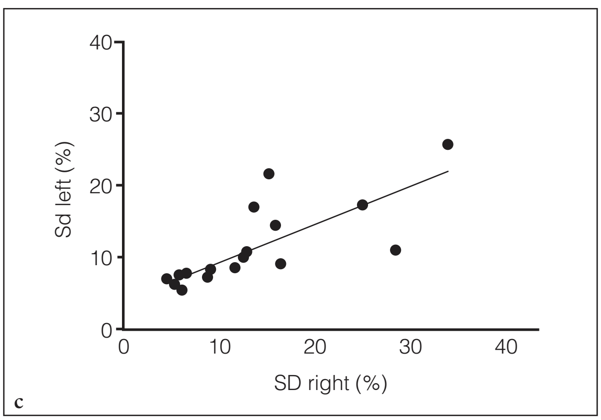
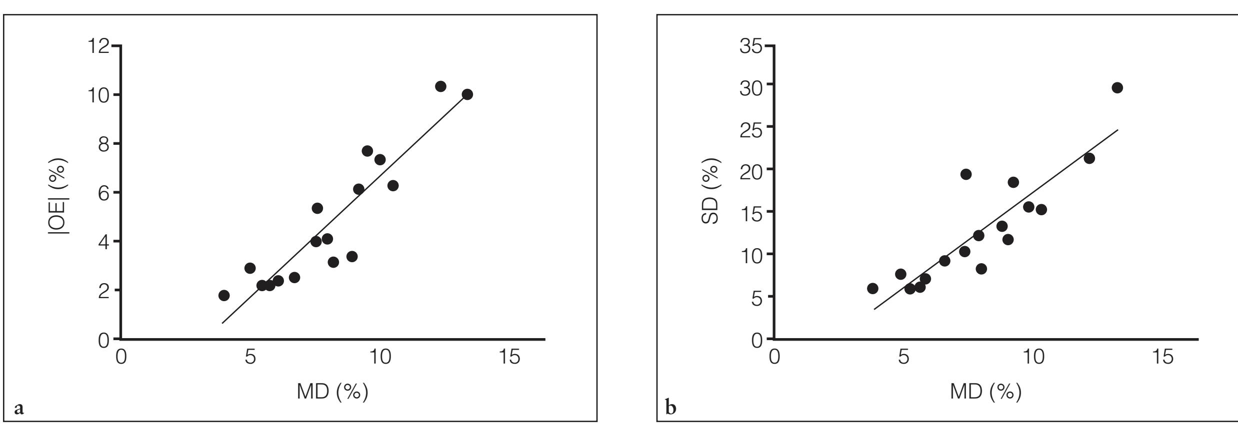


Fig 7 Correlation between the performances of the two sides in the second session. Each symbol is the average performance of one subject in the three trials of the second session. The different indices are expressed as percent of target level.



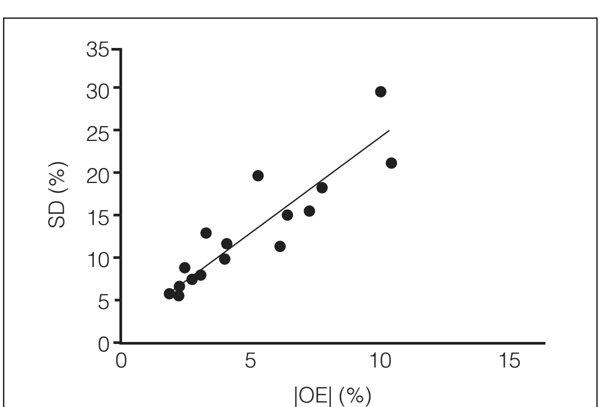
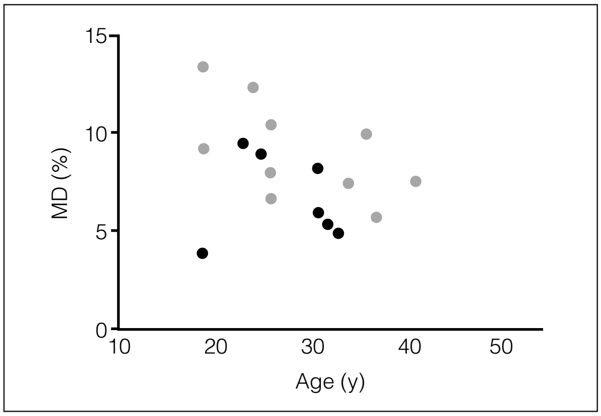


Fig 8 Correlation between the three indices scored by the different subjects, each dot representing the performance of one subject (average of both sides over the three trials of the second session).

Fig 9 Scatter plot of the MD score (second session) versus age of the subjects. Males and females are represented by black and grey dots, respectively.



tions while attempting to match target force level at different percentage of their MVC.[9] The motor performance was characterized by three error indices: the MD, the OE, and the SD, accounting for different aspects of the motor performance. These parameters exhibited good reliability and large intersubject variability that was not correlated with sex and age. In addition, a good symmetry was observed in the performance of the left and right sides.

Several studies investigating motor control in limb muscles during constant-force isometric tasks focused on the variability of the force as indicated by SD and SD% (also known as the coefficient of variation, CoV).[9,26,27] The unsteadiness of the exerted force may originate from recruitment and synchronization of motor units, from the variability in their discharge rate[26,28] as well as on the possible synergic activation of agonist muscles.[10,27] Earlier studies on jaw muscles provided evidence of a linear relationship between SD and contraction intensity,[11,12] but the investigated force levels were limited to 4 kg. Increased force variability at higher contraction intensity was also reported for jaw muscles during force ramp tasks and attributed to the progression of motor unit recruitment.[28] However, several studies on limb muscles have reported that the initially linear relationship lessens its steepness when the contraction level exceeds about 60% to 70% MVC, depending on the muscle group,[9,29,30] ie, the increase in variability lessens at higher contraction levels. This force level, at which the force variability versus force curve starts decreasing its slope, was considered to reflect the completion of motor unit recruitment in the muscle.[9] The present results strongly support a qualitative similarity between jaw and limb muscles. Interestingly, there was a tendency for SD to flatten at about 50% MVC (Fig 5c), which fits with the notion that motor unit recruitment in the masseter muscle is completed at a relatively early stage.[31,32]

Normalization of force variability to the average force level, which yielded SD% (the CoV) is a recommended procedure in order to compare different subject populations and muscle groups.[9,10] This operation often has shown that the larger variability in relative terms occurs at low force levels,[9,10] which was confirmed also in the present study. As for quantitative aspects, SD% stabilized around 13% in the second session and revealed a considerable unsteadiness compared to 1% to 2% in knee extension,[9] 1% to 3% in ankle dorsi- and plantar flexion,[33] 2% to 3% in pinch grip,[34] and 3% to 5% in index finger abduction.[10] These data support the early observation that visual control of force is worse in jaw muscles as compared to limb muscles.[13]

While SD characterizes the precision (steadiness) of muscle force, OE indicates the accuracy of the force matching, ie, how close the target is to the average exerted force. OE was expected to yield positive and negative values in individual scores, which would result in the reciprocal elision of opposite-sign values when averaging over all subjects. Instead, all subjects consistently overshot the target levels, thus producing positive offset errors. In spite of the given instructions, simply requiring a match to the target as precisely as possible, the subjects probably intended to increase their "safety margin" by adding some extra force, as was also observed in other studies.[35] In the literature, the matching error is often assessed by means of a mean absolute error (as for MD in the present study) or a root-mean-square (RMS) error. It should be noted that both these measures are also dependent on the variability of the force signal (SD) while OE is, in principle, independent of SD. However, MD and |OE| exhibited similar qualitative results in terms of dependence on contraction level, on session, and on side.

The dependence of accuracy on contraction level is qualitatively similar to what has been reported for limb muscles in pinch grip[8] and knee extension.[15]

Van Steenberghe et al[13] observed larger RMS matching error for jaw than for limb muscles but did not provide the actual values. MD%, averaged over the 10% to 70% MVC force range, was shown to be around 8% to 10% in the present study, while others have reported 3% to 4% error in knee extension[15,36] and a 9% to 16% in pinch grip.[8] The many differences in experimental conditions, ranging from the force range investigated, the preliminary motor training, the subject age, etc, may account for the large variability in matching error reported in the different studies. Thus, on the basis of the present data, it is not possible to conclude whether accuracy of the force control is also worse in jaw as compared to limb muscles.

In the second session, MD% spanned a large range, 4% to 14%, within the subject group. This provided evidence of a large intersubject variability and emphasizes the potential of this index to discriminate between bad and good performers.

As shown by the scatter plots, the three indices appeared to be correlated to each other, and all exhibited values that were lower in the second session than in the first one, which is suggestive of a learning effect. However, based on the present data, it cannot be excluded that an additional improvement in performance would have been achieved with further training.

One important outcome of this study was the good reliability of the three indices in assessing the individual motor ability. Two reasons may have contributed to this achievement: *(1)* adoption of relative (related to individual MVC) rather than absolute target levels; and *(2)* adoption of a comfortable sensor housing allowing accurate repositioning. In fact, discomfort caused by awkward transducers, particularly when recording from molar teeth, and imprecise sensor repositioning are acknowledged as potential sources of variability in the assessment of clenching forces.[37]

The voluntary control of muscle force is a basic outcome of sensory-motor integration and can be affected by disorders acting at peripheral or central neural levels. Several studies have investigated the precision of the force output in limb and hand muscles, with important implications in diagnostic and rehabilitation processes.[38–41] Of particular interest also is the analysis of the left/right symmetry in performance. The possibility of repeating the task on the two sides allowed the present study to detect that there was a good symmetry in motor performance. At the same time, the fact that comparable values were scored by the two sides in individual subjects supports the consistency of the technique in assessing motor performance. Since symptoms and functional impairments in TMD are often unilateral, the possibility to detect asymmetries in motor control of the mandible may be a useful approach for the characterization of the clinical condition and may assist diagnosis and follow-up of craniomandibular dysfunctions.

Advantages and Limitations

Film-based force sensors provide a potentially interesting mean to measure biting force. Rubber shielding of the sensor has been shown to improve comfort and stability of the force signal[42,43]; due to this rubber shielding and its limited thickness (5 to 6 mm), the housed sensor was well tolerated by all subjects. They easily learned to control the biting force to match the target levels. In addition, the subjects' confidence with the device appeared to increase further in the second session, as compared to the first one, as suggested by the increased MVC force.

Stress and anxiety levels were not measured, so it cannot be excluded that a decrease in anxiety occurred along with increased confidence with the setup in the second session. In principle, this may have contributed to the improvement in performance from the first to the second session, since stress and sympathetic activation may affect motor output and steadiness at different levels.[44,45] On the other hand, the occurrence of learning processes when practicing novel motor tasks and their overnight consolidation has been well documented[46,47] and remains the most likely explanation for the observed improvement in performance.

Finally, this study was not aimed at investigating the dependence of performance on gender and on age. Although no significant differences in performance could be observed between male and females and no correlation was observed between MD and age, a larger sample size will be necessary before more firm conclusions can be derived.

Conclusions

Individual force control of the jaw-elevator muscles can be reliably assessed in visually-guided isometric contractions. The motor performance was not correlated with age and gender, was highly correlated between left and right side, and was significantly improved in the second of the two sessions.

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References

1. Lobbezoo F, van Selms MK, Naeije M. Masticatory muscle pain and disordered jaw motor behaviour: Literature review over the past decade. Arch Oral Biol 2006;51:713–720.
2. Capra NF, Ro JY. Experimental muscle pain produces central modulation of proprioceptive signals arising from jaw muscle spindles. Pain 2000;86:151–162.
3. Wang K, Arendt-Nielsen L, Jensen T, Svensson P. Reduction of clinical temporomandibular joint pain is associated with a reduction of the jaw-stretch reflex. J Orofac Pain 2004;18:33–40.
4. Cadden SW. Modulation of human jaw reflexes: Heterotopic stimuli and stress. Arch Oral Biol 2007;52:370–373.
5. Clark GT. A critical evaluation of orthopedic interocclusal appliance therapy: Design, theory, and overall effectiveness. J Am Dent Assoc 1984;108:359–364.
6. Duarte Gaviao MB, Durval Lemos A, Diaz Serra M, Riqueto Gambareli F, Nobre Dos Santos M. Masticatory performance and bite force in relation to signs and symptoms of temporomandibular disorders in children. Minerva Stomatol 2006;55:529–539.
7. Hansdottir R, Bakke M. Joint tenderness, jaw opening, chewing velocity, and bite force in patients with temporomandibular joint pain and matched healthy control subjects. J Orofac Pain 2004;18:108–113.
8. Lindberg PG, Feydy A, Maier MA. White matter organization in cervical spinal cord relates differently to age and control of grip force in healthy subjects. J Neurosci 2010;30:4102–4109.
9. Christou EA, Grossman M, Carlton LG. Modeling variability of force during isometric contractions of the quadriceps femoris. J Mot Behav 2002;34:67–81.
10. Enoka RM, Christou EA, Hunter SK, et al. Mechanisms that contribute to differences in motor performance between young and old adults. J Electromyogr Kinesiol 2003;13:1–12.
11. Lewis GR. Aging and the accuracy of jaw muscle control. Gerodontology 1988;7:139–144.
12. Lewis GR, Yemm R. Measurement of the ability to control jaw muscle forces in man. J Dent Res 1983;62:821–824.
13. van Steenberghe D, Bonte B, Schols H, Jacobs R, Schotte A. The precision of motor control in human jaw and limb muscles during isometric contraction in the presence of visual feedback. Arch Oral Biol 1991;36:545–547.
14. Shinohara M, Yoshitake Y, Kouzaki M, Fukuoka H, Fukunaga T. Strength training counteracts motor performance losses during bed rest. J Appl Physiol 2003;95:1485–1492.
15. Hortobagyi T, Garry J, Holbert D, Devita P. Aberrations in the control of quadriceps muscle force in patients with knee osteoarthritis. Arthritis Rheum 2004;51:562–569.
16. Bandholm T, Rasmussen L, Aagaard P, Jensen BR, Diederichsen L. Force steadiness, muscle activity, and maximal muscle strength in subjects with subacromial impingement syndrome. Muscle Nerve 2006;34:631–639.
17. Clark BC, Pierce JR, Manini TM, Ploutz-Snyder LL. Effect of prolonged unweighting of human skeletal muscle on neuromotor force control. Eur J Appl Physiol 2007;100:53–62.
18. Smits-Engelsman BC, Rameckers EA, Duysens J. Muscle force generation and force control of finger movements in children with spastic hemiplegia during isometric tasks. Dev Med Child Neurol 2005;47:337–342.
19. Lodha N, Naik SK, Coombes SA, Cauraugh JH. Force control and degree of motor impairments in chronic stroke. Clin Neurophysiol 2010;121:1952–1961.
20. Del Santo F, Gelli F, Spidalieri R, Rossi A. Corticospinal drive during painful voluntary contractions at constant force output. Brain Res 2007;1128:91–98.
21. Bandholm T, Rasmussen L, Aagaard P, Diederichsen L, Jensen BR. Effects of experimental muscle pain on shoulder-abduction force steadiness and muscle activity in healthy subjects. Eur J Appl Physiol 2008;102:643–650.
22. Bandholm T, Rose MH, Sonne-Holm S, Jensen BR. Assessment of torque-steadiness reliability at the ankle level in healthy young subjects: Implications for cerebral palsy. Eur J Appl Physiol 2008;104:609–615.
23. Sosnoff JJ, Newell KM. Are age-related increases in force variability due to decrements in strength? Exp Brain Res 2006;174:86–94.
24. Marmon AR, Pascoe MA, Schwartz RS, Enoka RM. Associations among strength, steadiness, and hand function across the adult life span. Med Sci Sports Exerc 2011;43:560–567.
25. Dworkin SF, LeResche L. Research diagnostic criteria for temporomandibular disorders: Review, criteria, examinations and specifications, critique. J Craniomandib Disord 1992;6:301–355.
26. Halliday DM, Conway BA, Farmer SF, Rosenberg JR. Load-independent contributions from motor-unit synchronization to human physiological tremor. J Neurophysiol 1999;82:664–675.
27. Shapkova EY, Shapkova AL, Goodman SR, Zatsiorsky VM, Latash ML. Do synergies decrease force variability? A study of single-finger and multi-finger force production. Exp Brain Res 2008;188:411–425.
28. Sowman PF, Brinkworth RS, Turker KS. Mandibular physiological tremor is reduced by increasing-force ramp contractions and periodontal anaesthesia. Exp Brain Res 2008;184:71–82.
29. Loscher WN, Gallasch E. Myo-electric signals from two extrinsic hand muscles and force tremor during isometric handgrip. Eur J Appl Physiol Occup Physiol 1993;67:99–105.
30. Sherwood DE, Schmidt RA. The relationship between force and force variability in minimal and near-maximal static and dynamic contractions. J Mot Behav 1980;12:75–89.
31. Goldberg LJ, Derfler B. Relationship among recruitment order, spike amplitude, and twitch tension of single motor units in human masseter muscle. J Neurophysiol 1977;40:879–890.
32. Scutter SD, Turker KS. Recruitment stability in masseter motor units during isometric voluntary contractions. Muscle Nerve 1998;21:1290–1298.
33. Tracy BL. Visuomotor contribution to force variability in the plantarflexor and dorsiflexor muscles. Hum Mov Sci 2007;26:796–807.
34. Christou EA. Visual feedback attenuates force fluctuations induced by a stressor. Med Sci Sports Exerc 2005;37:2126–2133.
35. Cole KJ. Grasp force control in older adults. J Mot Behav 1991;23:251–258.
36. Hortobagyi T, Tunnel D, Moody J, Beam S, DeVita P. Low- or high-intensity strength training partially restores impaired quadriceps force accuracy and steadiness in aged adults. J Gerontol A Biol Sci Med Sci 2001;56:B38–B47.

37. Tortopidis D, Lyons MF, Baxendale RH, Gilmour WH. The variability of bite force measurement between sessions, in different positions within the dental arch. J Oral Rehabil 1998;25:681–686.
38. Jones RD, Donaldson IM. Measurement of integrated sensory-motor function following brain damage by a computerized preview tracking task. Int Rehabil Med 1981;3:71–83.
39. Kriz G, Hermsdorfer J, Marquardt C, Mai N. Feedback-based training of grip force control in patients with brain damage. Arch Phys Med Rehabil 1995;76:653–659.
40. Seynnes O, Hue OA, Garrandes F, et al. Force steadiness in the lower extremities as an independent predictor of functional performance in older women. J Aging Phys Act 2005;13:395–408.
41. Tracy BL, Enoka RM. Steadiness training with light loads in the knee extensors of elderly adults. Med Sci Sports Exerc 2006;38:735–745.
42. Fernandes CP, Glantz PO, Svensson SA, Bergmark A. A novel sensor for bite force determinations. Dent Mater 2003;19:118–126.
43. Waltimo A, Kononen M. A novel bite force recorder and maximal isometric bite force values for healthy young adults. Scand J Dent Res 1993;101:171–175.
44. Christou EA, Jakobi JM, Critchlow A, Fleshner M, Enoka RM. The 1- to 2-Hz oscillations in muscle force are exacerbated by stress, especially in older adults. J Appl Physiol 2004;97:225–235.
45. Passatore M, Roatta S. Influence of sympathetic nervous system on sensorimotor function: Whiplash associated disorders (WAD) as a model. Eur J Appl Physiol 2006;98:423–449.
46. Born J, Rasch B, Gais S. Sleep to remember. Neuroscientist 2006;12:410–424.
47. Landsness EC, Crupi D, Hulse BK, et al. Sleep-dependent improvement in visuomotor learning: A causal role for slow waves. Sleep 2009;32:1273–1284.