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**ASSESSMENT OF FORCE AND FATIGUE IN ISOMETRIC
CONTRACTIONS OF THE UPPER TRAPEZIUS MUSCLE BY SURFACE
EMG SIGNAL AND PERCEIVED EXERTION SCALE**

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LIST OF ACRONYMS USED IN THE MANUSCRIPT

Acronyms	Definition
BRC	Borg rate of change
CV	muscle fibre conduction velocity
<i>Corr</i>	correlation coefficient between maps
EMG	electromyogram
ET	endurance time
FD	fractal dimension
IZ	innervation zone
MNF	mean frequency of the power spectrum
MU	motor unit
MVC	maximal voluntary contraction
R	Pearson's correlation coefficient
RMS	root mean square value
UT	upper trapezius muscle

ABSTRACT

Quantifying muscle force and fatigue is important in designing ergonomic work stations, in planning appropriate work-rest patterns, and in preventing/assessing the progress of disorders. In fourteen subjects (7 males, 7 females), muscle force and fatigue were estimated by subjective perception (based on Borg scale CR10) and objective indexes extracted from surface electromyogram (EMG). The experimental protocol consisted of an isometric task selective for the upper trapezius muscle at different force levels (10%-80% of maximal voluntary contraction – MVC, in steps of 10%MVC) and one fatiguing contraction (constant force level at 50%MVC until exhaustion). Surface EMG signals were detected by a two-dimensional array of electrodes placed half way between C7 and the acromion. The following variables were calculated from EMG signals: muscle fibre conduction velocity (CV), root mean square value (RMS), mean frequency of the power spectrum (MNF), fractal dimension (FD), and entropy. All detected signals were also used to build topographical maps of RMS. Both subjective and objective indications of force and fatigue can provide information on exerted force and endurance time. In particular, Borg ratings, RMS, and entropy were significantly related to force, and the rate of change of CV, MNF, FD, and Borg ratings were predictive of the endurance time. Moreover, significant differences were found in Borg ratings between males and females. The correlation coefficient of pairs of topographical maps of RMS was high (of the order of 0.8). This reflects a characteristic spatial-temporal recruitment of upper trapezius motor units that is not affected by force levels or fatigue.

1. INTRODUCTION

Work related musculoskeletal disorders of the shoulders and neck have been reported for different occupations [1,2]. Shoulder and neck conditions (e.g., rotator cuff tendonitis, myalgia, thoracic outlet syndrome, and radiculopathy [2]) can be caused by repetitive or sustained work, short work cycles, and localised muscular loadings.

Perceived exertion ratings (e.g. VAS, Likert, OMNI, and Borg) have been used to study the subjective feeling during an exercise. Different scales were compared in the literature [3]. The Borg scale CR10 was considered in this study. This is a subjective scale with values from 1 to 10 indicating the perceived level of effort. A good correlation between quantitative measure of physiological response (e.g., metabolic acidosis, ventilation, oxygen intake, heart rate, and respiration frequency) and perceived exertion has been shown [4]. The Borg scale was applied to both resistance [5] and endurance [6] exercises, and to both isometric [7] and dynamic [8] contractions. Thus, perceived exertion ratings can be considered an acceptable approach to work related studies [7,9] measuring the perceived response at different loads [10] or testing the effect of rest periods on recovering perceived efficiency [11].

An objective and non invasive assessment of muscle activity can be provided by surface electromyography (EMG). The use of surface EMG techniques in ergonomic studies has been documented in different tasks, including repetitive work at a car assembly line [12], work at the visual display terminal [13], and other studies [1]. In occupational health, upper trapezius muscle (UT) is usually investigated by surface EMG, as it is a superficial muscle and its activity is influenced by neck or shoulder pain [14]. The structure of UT is complicated. Contradictory descriptions of the direction and disposition of its motor units (MUs) are available in the literature [15].

The relation between EMG and force strongly depends on MUs control by the central nervous system. This can change depending on muscle pain [16] or fatigue [17]. Due to the high inter-

subject and inter-muscle variability, the estimation of the EMG-force relation requires a calibration model [18].

Muscle fatigue consists of myoelectric and mechanical phenomena, the former ones preceding the latter. Myoelectric manifestation of fatigue includes both “peripheral” and “central” adaptations of the muscles [19]. Interesting indications have been obtained from EMG studies concerning fibre type distribution of the muscle [20], prediction of endurance time (ET) [21], and pathological conditions [22].

To increase the reliability of the information extracted from surface EMG, high density, two-dimensional (2D) detection systems have recently been applied. Information pertaining to a large spatial area of a muscle was obtained, supporting the investigation of the spatial-temporal MUs recruitment [23,24].

This paper focused on the application of subjective (based on Borg scale CR10) and objective (based on surface EMG signals) methods to investigate UT. Surface EMG signals were recorded by a 2D detection system. The relation between subjective/objective estimations and muscle force, expressed as percentage of maximal voluntary contraction (MVC), and muscle fatigue, expressed as a function of ET, was assessed. Differences related to gender were investigated. Furthermore, the analysis of topographical maps of RMS supported the analysis of the activation pattern of UT during the isometric task performed by the subjects.

2. METHODS

2.1 Subjects

Fourteen healthy subjects, 7 males and 7 females (mean±standard deviation; age: 25±3 years, height: 172±10 cm, weight: 63±13 kg) participated in the measurements. The study was approved by the Local Ethics Committee of the Health Department of Piemonte Region, Italy, and a written informed consent was obtained from all participants. All subjects had right side dominance, assessed by the Edinburgh Handedness Inventory.

2.2 EMG and force measurement

Selective and isometric contractions (shoulder elevation) were performed using the experimental station showed in Figure 1a. The subject sat on a chair, with the arms straight alongside the body to reduce the involvement of biceps and deltoid muscles, the back leaned against the back of the chair to avoid the activity of the erector spinae muscles, and the feet rose to avoid contributions from the lower limb muscles. Bilateral pull was performed to avoid trunk bending.

Figure 1

Surface EMG signals were detected in single differential configuration from the dominant UT using a 2D, disposable and flexible, adhesive array of 64 circular electrodes (3 mm diameter), with 8 mm inter-electrode distance (LISiN-SpesMedica, Battipaglia, Italy). Surface EMG signals were amplified (EMG-USB amplifier, LISiN-OTBioelettronica, Torino, Italy; CMRR>96 dB, gain variable from 100 to 10000, bandwidth 10-750 Hz, and equivalent input noise less than $0.8 \mu\text{V}_{\text{RMS}}$), and stored on a PC (sampling frequency 2048 Hz, 12 bit A/D converter).

Force was measured by two independent load cells with a linear operation range 0-2000 N (UU-K200, DACELL, Korea). Each load cell was connected to a handle, which could be pulled by the subject (Figure 1a). The force signals obtained from each load cell were amplified (MISO-II amplifier, LISiN, Torino, Italy; bandwidth 0-10 Hz) and stored on a PC (sampling frequency 50 Hz, 12 bit A/D converter). The force signal obtained from the load cell corresponding to the dominant UT was used to provide biofeedback to the subject, by indicating in the middle of a screen the force target to reach. The corresponding force level was not indicated to avoid any bias.

2.3 Experimental protocol

The skin above UT was gently abraded and cleaned with water. The 2D adhesive array was placed over the skin, centred midway between C7 and the acromion (Figure 1b). The cavities of the electrode were filled with 20 μ l of conductive gel. The reference electrode was applied to the dominant wrist.

The subject performed three trials of shoulder elevation of 4 s each and was asked to reach the MVC within 2-3 s, with 4 minutes of rest in between. Verbal encouragement was provided to help the subject reach a higher level in each trial. The highest level achieved was selected as reference MVC.

After 5 minutes of recovery, the subjects performed two sequences of 8 contractions, from 10%MVC to 80%MVC in steps of 10%MVC, with a 2 minute rest after each contraction. Subjects were asked to reach each contraction level within 2-3 s. Acquisition then started and this lasted for 4 additional seconds. Contractions were randomised in each sequence in order to avoid cumulative and learning effects. At the end of each contraction, subjects estimated their subjective indication of effort according to the Borg scale CR10.

Finally, after 10 minutes of recovery, the subjects performed an isometric contraction at 50%MVC sustained until exhaustion. UT was contracted as long as possible, until the force value decreased to below 90% of the target (ET). Subjective indication of effort was provided every 5 s from the beginning to the end of the contraction.

2.4 Signal processing

Surface EMG signals were digitally filtered (bandwidth 20-400 Hz) in order to reduce slow transients and instrumentation noise, and were divided into epochs of 250 ms, where the surface EMG signals were supposed to be stationary. For each column of the 2D array, the main innervation zone (IZ) was identified by visual analysis of the signals. Channels closest to the acromion showing propagating signals were selected. Each epoch of each signal was used for the estimation of the following EMG variables: root mean square value (RMS), mean frequency of the power spectrum

(MNF), fractal dimension (FD), estimated by the box counting method [19,25], muscle fibre conduction velocity (CV, estimated by a multichannel algorithm [26], values accepted if the average correlation coefficient between adjacent channels was higher than 0.75), and entropy [23].

Estimates of variables obtained from single channels (all variables, except CV) were averaged over the N channels selected by visual analysis (e.g. channels which showed a clear signal propagation and absence of artefacts). The average of CV values obtained over each column was considered as the estimate of CV.

The RMS values of individual single differential channels were interpolated (using the bicubic interpolation) to provide a topographical map of RMS for each force level, averaged over the 4 s of the acquisition. Thus, for each subject, eight maps were obtained corresponding to force levels from 10%MVC to 80%MVC in steps of 10%MVC. For the sustained contraction, RMS was evaluated in non overlapping epochs 2 s long. The epochs were located at 10% to 80% of ET, with a 10% step. Thus, for each subject, the eight maps obtained represented the RMS spatial distribution from 10% to 80% of ET. Maps were compared pair wise by estimating the cross correlation coefficient (*Corr*).

2.5 Statistical analysis

Having preliminarily verified by Kolmogorov-Smirnov testing that the estimated EMG variables at each force level were not normally distributed across subjects, the non-parametric Kruskal-Wallis test was performed on EMG variables, computed for each contraction at different force levels. Considered factors were trial, gender, and contraction level. Significance was set to $p < 0.01$. Results were reported as median, min-max values, and 25%-75% percentile intervals. When the Kruskal-Wallis test indicated significant variations, pair-wise comparisons were performed with post-hoc Mann-Whitney test. Moreover, any correlation between Borg values and RMS estimates at different force levels was tested using Pearson's correlation coefficient (*R*).

Linear regression over time was performed on the EMG variables and on the Borg values estimated during endurance contractions. *R* was computed to assess a relation between ET and normalized

rate of changes of the regression line (normalized with respect to the initial value) of different variables. Non linear regression was performed between Borg rate of change and ET.

3. RESULTS

3.1 Representative signals and results

Portions of signals at 20%, 40%, 60%, and 80%MVC are shown in Figure 2a. Channels chosen by visual analysis for the subsequent global analysis are indicated by arrows. Estimates of CV, MNF, RMS, FD, entropy, and Borg rate at different force levels are shown in Figure 2b. Two examples of topographical maps obtained for 20%MVC (upper map) and 80%MVC (lower map) are shown in Figure 2c.

Figure 2

Representative signals recorded during the endurance contraction are shown in Figure 3a. Global parameters estimated from the EMG signal are shown in Figure 3b. Force signal (normalized with respect to 50%MVC) is also provided. Two examples of topographical maps obtained at the beginning (10%ET, upper map) and close to the end of the contraction (80%ET, lower map) are shown in Figure 3c.

Figure 3

3.2 Subjective and objective estimation of force

MVC obtained from the load cell corresponding to the dominant UT was higher for males (mean±standard deviation, 613.5±108.0 N) than females (246.0±51.4N).

Kruskal-Wallis test did not show statistical dependence of any subjective or objective variable on trial. Furthermore, EMG variables were not statistically dependent on gender. Significant

differences in RMS, Borg ratings, and entropy considered at different force levels were found. Post-hoc Mann-Whitney test disclosed pair-wise differences between Borg, RMS, and entropy values obtained at low force levels (in the range 10%MVC-40%MVC) and at high force levels (50%MVC-80%MVC). The variations of Borg, RMS, and entropy at different force levels are shown in Figure 4a, 4b, and 4c, respectively.

The observed increasing trends of MNF, CV, and FD versus force level were not statistically significant.

The averaged (across subjects) RMS estimates in relation to the averaged Borg values at different force levels are shown in Figure 4d. The correlation coefficient between these parameters was very high ($R=0.99$). Correlation coefficient between RMS and Borg values for each subject ranged between 0.88 and 0.99.

Figure 4

Statistically significant differences were found in Borg ratings between the two genders (mean \pm standard deviation of Borg rating, females 3.2 ± 2.1 and males 4.5 ± 2.5).

3.3 Subjective and objective estimation of fatigue

Normalized MNF, CV, FD rate of changes (indicated by nMNF, nCV, nFD rate of changes, respectively), and Borg rate of changes are shown in Figure 5 in relation to ET. Force level was quite constant during the endurance contraction for each subject. Maximum amplitude of variation was about $\pm 5\%$ of the required level set at 50%MVC. The normalized rate of change of MNF, CV, and FD showed a linear relation with ET. The correlation between nCV and ET ($R=0.58$) was lower than that obtained for nMNF ($R=0.71$) or nFD ($R=0.72$). Since the rate of change of Borg ratings as a function of the ET did not show a linear relation, the non linear relation (1) was used to fit these variables:

$$BRC = \frac{A}{ET} \quad (1)$$

where *BRC* indicates the Borg rate of changes and *A* is chosen to provide optimal fit of the data in the least mean square sense (optimal $A=5.6$). The average Borg rating for 50%MVC was 4.57 ± 1.14 (mean \pm standard deviation) so that the variation of Borg ratings during the endurance task was about 5.4 (as the Borg rating at ET was 10). Hence relation (1) is approximately equal to the ratio between the variation of Borg ratings and ET.

Figure 5

3.4 Assessment of spatial activation patterns by RMS maps

Pair wise correlation coefficients between RMS maps at different force levels and time epochs of the sustained contraction were estimated. The correlation coefficient was obtained for all the possible combinations of two force levels (from 10%MVC to 80%MVC) and time epochs of the sustained contractions (from 10%ET to 80%ET). The correlation coefficient between maps at different force levels and during the endurance contraction were high in all cases, with $Corr > 0.7$.

4. DISCUSSION

4.1 Subjective and objective estimation of force

A high correlation ($R=0.99$) was found between the perceived effort (Borg ratings) and the objective measure of exerted force (RMS values) at different force levels (Figure 4d). This is in line with the good correlation between Borg scale and force levels documented in the literature for different tasks [27].

A statistically significant dependence was demonstrated between force level and EMG amplitude (measured by RMS). Averaging RMS values from a large number of channels was useful to reduce the estimation variance. Nevertheless, large epoch to epoch fluctuations of RMS were observed

during constant force levels, indicating that force estimation by surface EMG amplitude could be feasible only to distinguish between quite different force levels or by averaging over long epochs. The relation between RMS and force was subject-dependent, confirming that a calibration model is needed for force estimation [19].

Signal entropy was also related to force. In the field of information theory, entropy is introduced as a measure of complexity. For increasing force level, more and more MUs are recruited and their firing rate increases, so that interference EMG signals have a more and more stochastic behaviour. For the same reason, it was expected that FD would increase with increasing force level (as also shown in [25] and in [28]). Nevertheless, for the specific muscle and task considered, FD was not statistically related to force.

4.2 Subjective and objective estimation of fatigue

Perceived fatigue measured by the rate of change of Borg ratings was related to ET, which can be predicted assuming a constant rate of change and estimating the time needed to reach the maximum value of Borg scale.

Surface EMG amplitude increased during the contraction while MNF, FD, and CV decreased, providing the expected pattern of the fatigue plot. Signal amplitude showed high fluctuations in time, larger than those obtained in the simulation study [19], which accounted only for the stochastic behaviour of EMG signal, but not for possible load sharing between synergistic muscles. Thus, the high variation in amplitude could be due to momentary changes of load sharing with other synergistic muscles of UT, such as the sternocleidomastoid, scalene, splenius capitis, and deltoid muscle. The normalized rate of change of MNF, CV, and FD were also strongly correlated to ET, indicating that these parameters could provide indications of fatigue and could be predictive of ET. The fractal index considered in [28] did not show a relationship with fatigue (whereas it was related with the force level). A direct comparison between our results and [28] is difficult as the muscle

under study and the indexes of the fractal dimension were different (biceps brachii and a spectral indicator was used in [28]).

Entropy is not affected by fatigue (Figure 3b), as was also suggested by simulations [19]. Entropy is affected by the force level (Figure 2b). Hence, entropy could be useful to distinguish between myoelectric changes due to fatigue and force.

4.3 Spatial activation patterns by RMS maps

UT is controlled differently in different tasks, as indicated in [29] where subunits were observed to be involved differently according to the task.

In this study, correlation between maps at different force levels was in general high ($Corr > 0.7$). There were no evident variations of the activation pattern (either a shift of IZ or recruitment/release of different spatial portions, also called subunits [29]) of UT when the force level changed. This result is in contrast with [24], where a shift of the RMS distribution was observed in isometric contractions of UT with increasing force level. The discrepancy could be related to the different experimental station. In [24] subjects were standing, therefore the contribution of leg muscles was not compensated. Moreover, the shoulders were fixed by straps, possibly causing discomfort at high contraction levels.

During the 50%MVC endurance contraction, correlation coefficients between maps of RMS were higher than 0.9. Hence, there is no evidence of different spatial muscle activation in UT during a sustained contraction. This result is in contrast with [24], probably due to the different experimental station as stated above. It is also in contrast with [23] which showed change in spatial distribution of UT activity during abduction of both arms at 90° with elbows fully extended and forearms 90° pronated, without hand load. The difference with respect to [23] could be due to our selected task being more selective for UT. Moreover, the same 2D electrode array was used, but it was placed in a more proximal location and with the longer side transverse to the fibres. Hence, the upper part of the 2D electrode array was probably covering a portion of muscle strongly involved in the task.

However, the lower part was probably covering a portion of muscle with activity varying over time when compared to the upper portion, thereby causing the observed pattern changes.

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FIGURE CAPTIONS

Figure 1 System for measurement of force generated by selective and isometric contraction of the upper trapezius muscle. a) Position of the subject on the chair, lateral and front view. Subject's feet are not touching the floor, arms are kept straight along the body and the back against the chair back. This is a bilateral task to avoid trunk bending. b) Position of the two-dimensional electrode array: the array's columns are parallel to the line connecting C7 and acromion, the centre of the array is located half way. c) Two-dimensional array with inter-electrode distance of 8 mm.

Figure 2 Processing of a representative signal recorded during contractions at different force levels. a) Portions of the surface EMG signals at 20% MVC, 40% MVC, 60% MVC, and 80% MVC, and channels chosen for global analysis (see arrows). b) Estimates of CV, MNF, RMS, FD, entropy, and Borg rate at different force levels: EMG variables are averaged across the channels indicated in a) and over a 4 s interval. c) Example of maps of RMS at 20% MVC (upper map) and 80% MVC (lower map). The innervation zone is under rows 7-8 for both contraction levels. The spatial distribution is similar (except for an increase in amplitude at the highest contraction level). The correlation coefficient between the two maps is 0.91.

Figure 3 Processing of a representative signal recorded during the endurance contraction. a) Portion of the surface EMG signal and channels chosen for global analysis (see arrows). b) Fatigue plot, showing normalized force signal, CV, MNF, RMS, FD, and entropy estimated over 250 ms long epochs and averaged across all selected channels indicated with arrows in a), and with the Borg rate given by the subjects every 5s along the endurance contraction. c) Example of maps of RMS from 2 s long signal epoch at the beginning (10% of endurance time) and close to the end of the contraction (80% of endurance time). The innervation zone is under rows 7-8 during the entire endurance

contraction. The RMS spatial distribution is similar (except for an increase in amplitude during the contraction). The correlation coefficient between the two maps is 0.95.

Figure 4 Borg values, RMS, and entropy related to the contraction levels are shown in a), b), and c), respectively. Results are reported as median, quartile range and min/max values. Note that the Borg scale has only integer values and therefore the median value overlaps with the quartile edge or the quartile edge overlaps with the min value. The relation between Borg ratings and RMS estimates at different force levels (mean \pm standard deviation) is shown in d).

Figure 5 Normalized rate of changes of MNF (nMNF), nFD, nCV, and Borg rating related to endurance time are shown in a), b), c), and d), respectively. R is the correlation coefficient of the regression line. The regression curve in d) is $BRC = \frac{A}{ET}$.

Figure 1

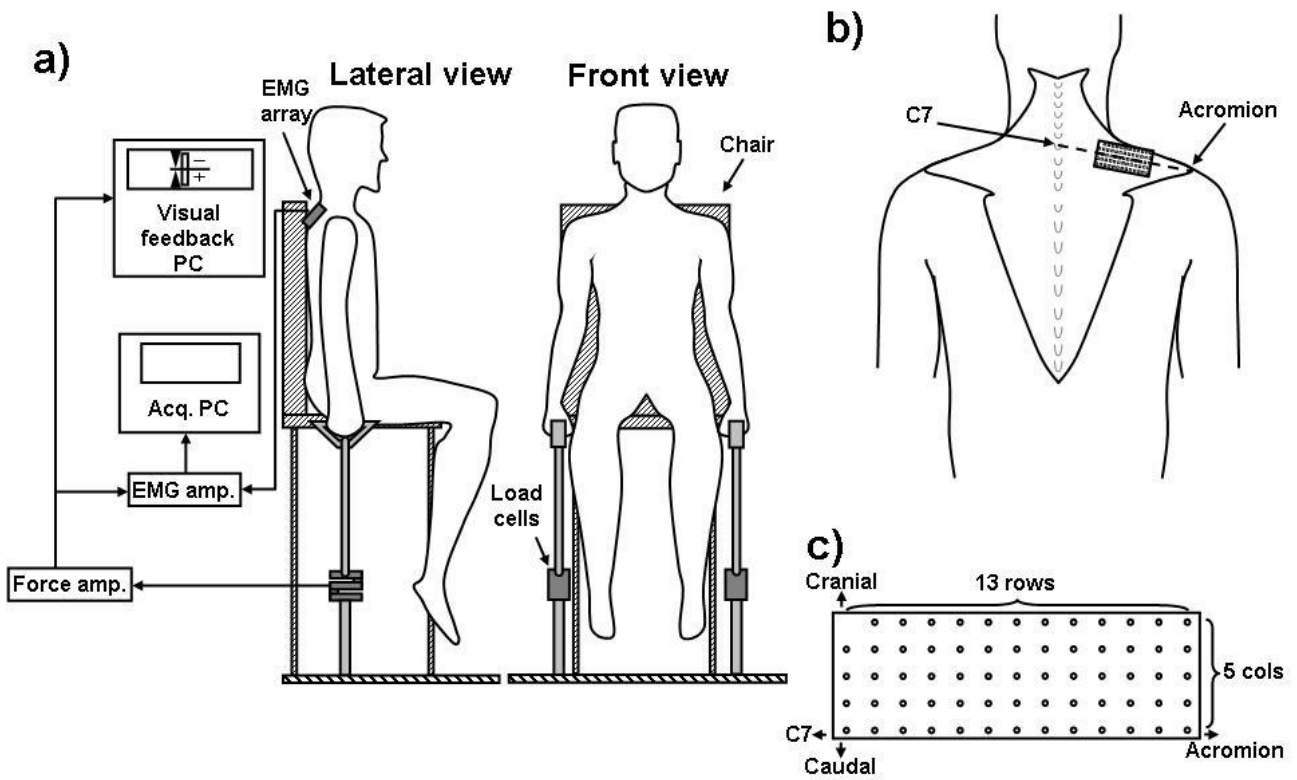


Figure 2

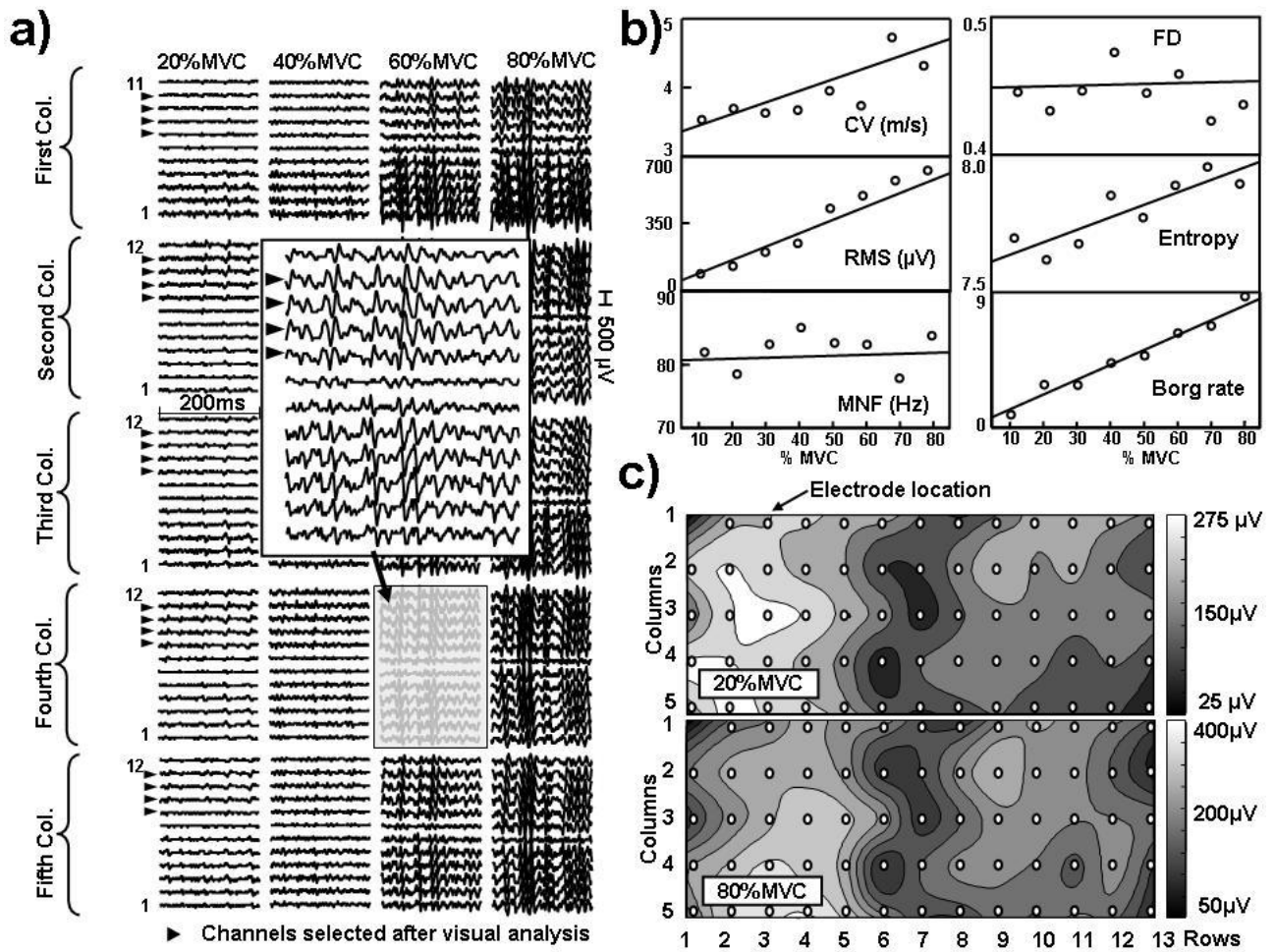


Figure 3

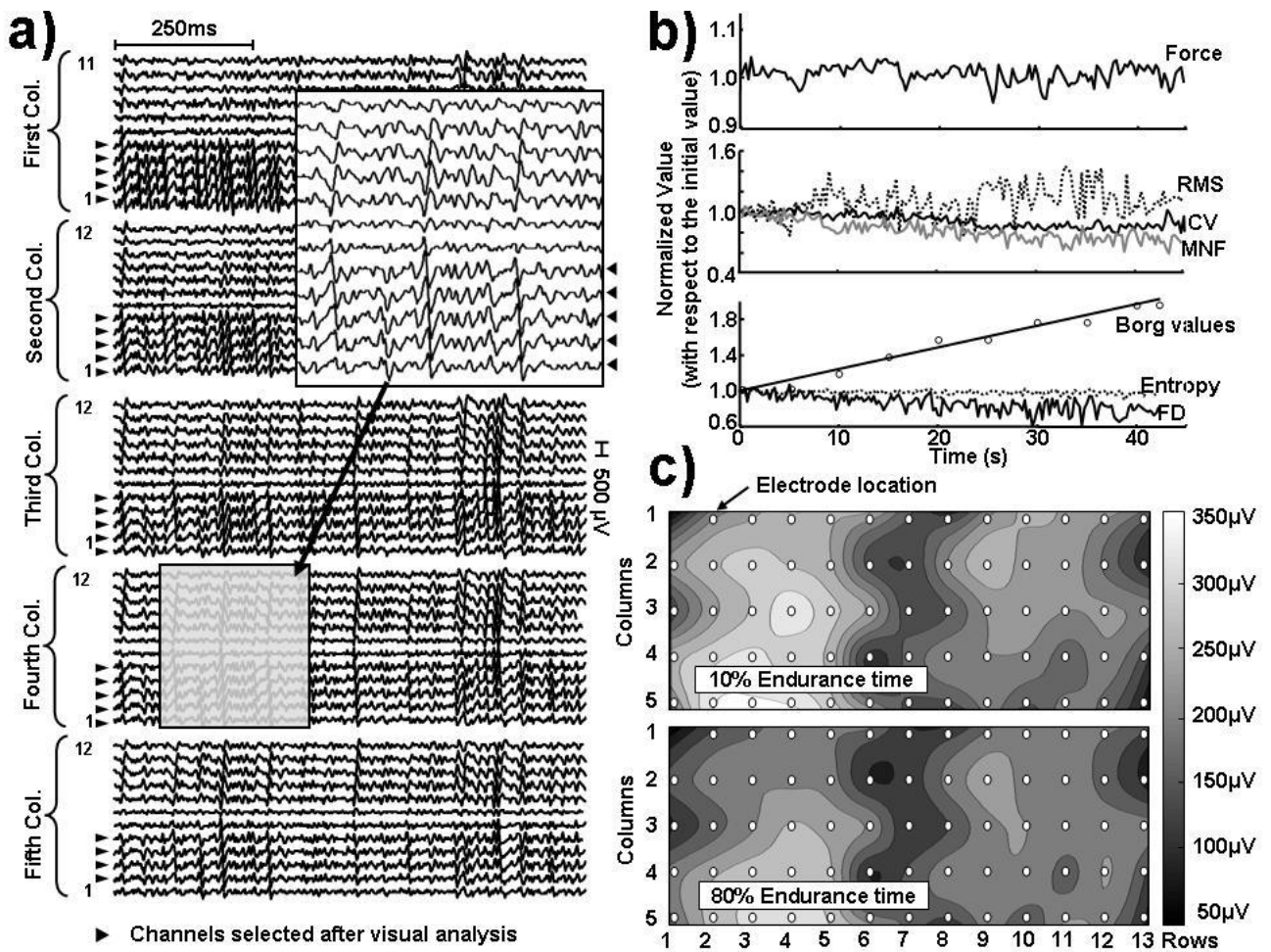


Figure 4

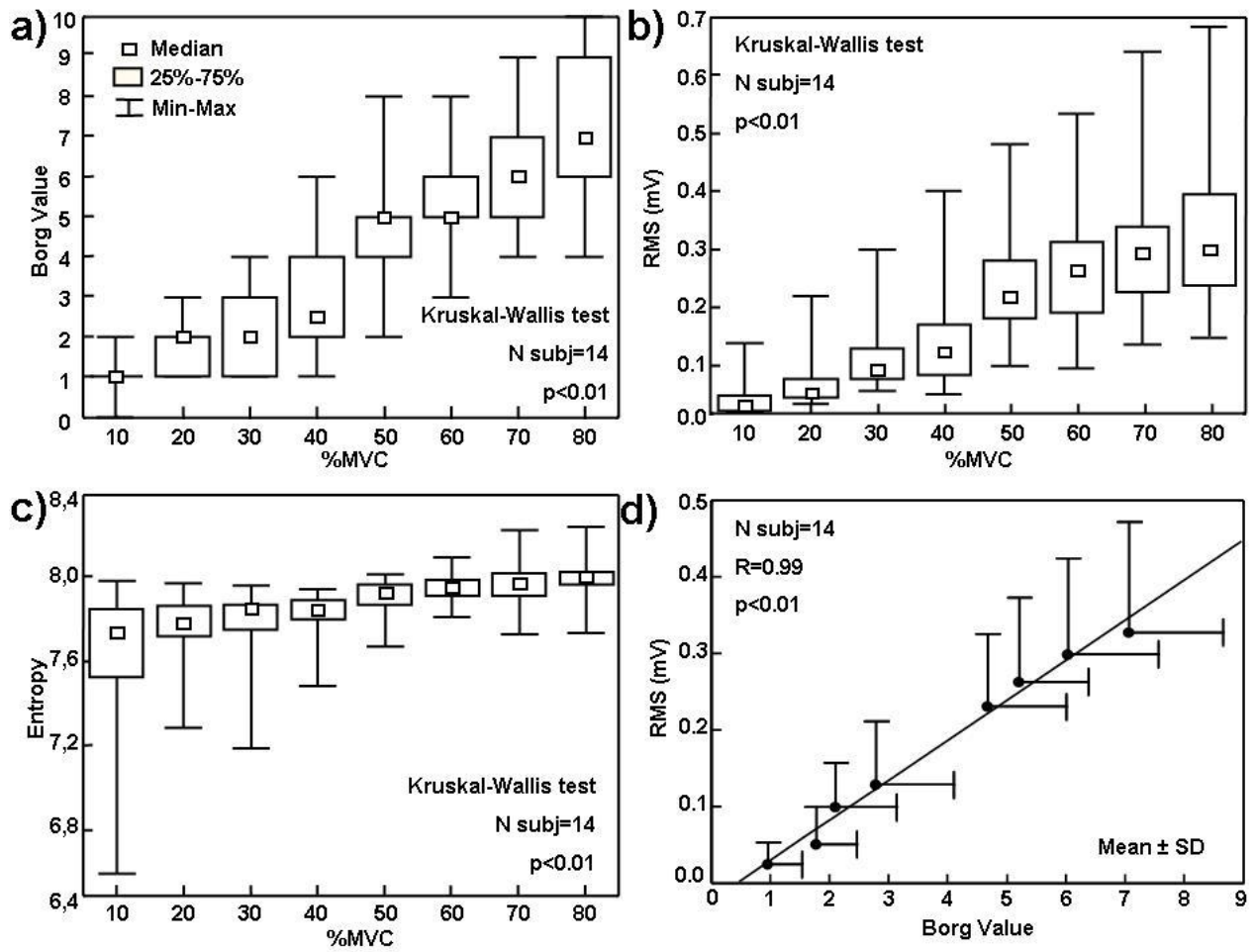


Figure 5

