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INDEX

SESSION 1: EPIDEMIOLOGY

LECTURES

ORAL PRESENTATIONS

POSTERS

SESSION 2: INFORMATION EXTRACTION FROM EMG AND MMG

LECTURES

ORAL PRESENTATIONS

POSTERS

SESSION 3: ADVANCED TECHNIQUES FOR NEUROMUSCULAR AND CLINICAL ASSESSMENT

LECTURES

ORAL PRESENTATIONS

POSTERS

Introduction

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Abstract *– A summary of the achievements of project NEW in both the technical and life-sciences fields, and an outline of the Symposium are provided.*

1. The project NEW: objectives and structure.

Science is the process of answering questions dealing with the understanding of nature. The question that the European RTD Shared Cost Project *"Neuromuscular Assessment in the Elderly Worker"* (NEW, March 1 2001 - February 28 2004) intended to answer was:

Is there a correlation between measurable physical variables and quantitative indicators from functional tests, on one side, and the condition or symptoms reported by the worker, his/her age, gender, living habits, job type and working condition, on the other side ?

This question can be broken down into many secondary questions, defining the specific objectives described in the proposal. Quite obviously there is a first dichotomy into two groups of questions:

1. what are the measurable variables of interest, how can they be measured, with what degree of error, reliability, repeatability? Can they be measured in the research lab as well as in the field at the working place, without excessive noise and undue discomfort of the subject?
2. how do we select cases and controls, report or describe symptoms, living habits, working conditions?

Both issues are very interdisciplinary. Point 1 was addressed by partners with a prevalent engineering and technological knowledge, whereas point 2 was mostly addressed by life scientists, ergonomists, physiologists and experts of occupational medicine. The pathologies of interest concern mostly female workers and the neck, shoulder and back muscles, so the research was restricted to this gender and these muscles. Secretaries, computer operators, nurses are the categories most affected by neuromuscular disorders. The study was restricted to these jobs.

The measurable variables of interest were grouped as biomechanical and electromyographic obtained from surface EMG signals detected on the skin above specific muscles. The main fields of work were described in nine workpackages concerning: 1) definition of jobs and subject selection criteria, definition of questionnaires and protocols, muscles to be investigated, 2) sensor development, optimisation, upgrade and testing, 3) instrumentation and software development for lab and field testing, optimisation and upgrade, 4) laboratory and field measurements, 5) data processing, 6) development of information extraction systems based on novel EMG signal processing techniques, 7) development of tools (models) for signal interpretation and teaching for ergonomists and occupational health specialists, 8) dissemination of results and reporting 9) integration with existing telemedicine or teleconsulting systems from the home or occupational setting. Nine partners and three subcontractors, from eight countries were involved in the project.

2. The project NEW: results

2.1 Sensors and instrumentation

Isometric dynamometers for measurement of shoulder elevation and limb joint torques have been developed. The EMG signal represents something easy to detect but the information contained in it is difficult to extract and interpret and can be obtained only with appropriate detection and processing techniques. For this reason a considerable fraction of the total effort was devoted to the development of signal processing tools suitable to extract information from this signal.

One dimensional and two dimensional EMG electrode arrays have been developed as well as mechanomyogram (MMG) sensors, a 16 ch data-logger and the related software for signal acquisition, display and processing.

2.2 Information extraction methods

The sources of information were the questionnaires, the results of the clinical and functional tests (coordination, muscle strength, stress, performance, work capacity), the biomechanical and EMG variables. Major research efforts were focused on the latter issue.

Significant results have been achieved in the field of decomposition of the surface EMG into the constituent motor unit action potential (MUAP) trains. Figure 1 shows an example of decomposition of multi-channel surface EMG signals from Abductor Digiti Minimi (ADM) muscle during a low level isometric voluntary contraction. The detected action potentials are classified as belonging to different MUs and shown superimposed to each other.

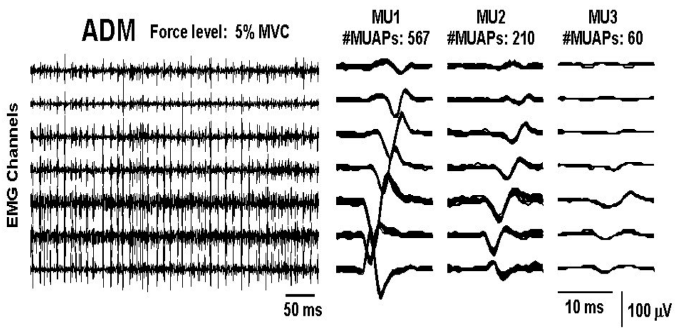


Fig. 1. Example of decomposition of multi-channel surface EMG signals from the ADM muscle. The action potentials belonging to the first three classes are shown superimposed to each other in the right panel.

The individual MUAPs can be used for spike triggered averaging of the MMG signal and extraction of the single motor unit MMG which is associated to the mechanical properties of the motor unit. An example of this procedure is shown in Figure 2 for a representative MU.

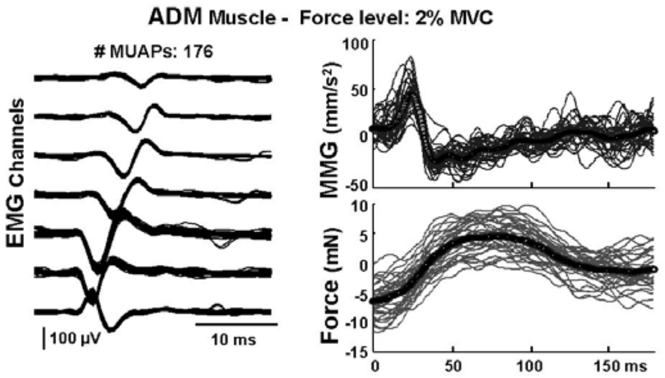


Fig. 2. Examples of single MU MMG and force signals for a representative contraction. The MMG and force response were obtained by averaging the signals following the occurrences of single MU action potentials used as triggers. All the detected occurrences in the 60 s long contractions were used for the averaging.

A very critical EMG variable is the muscle fiber conduction velocity (CV), an important physiological variable whose estimate is affected by many factors such as the geometry of the electrode array, the length, depth and inclination of the fibers and the properties of the volume conductor. Progress was achieved in this direction as well. CV can now be estimated with much greater accuracy than a few years ago but at the price of a very careful electrode placement. This time consuming requirement makes the procedure more suitable for lab applications than for field measurements.

Another long term problem addressed within NEW was crosstalk. Advanced techniques for “blind source separation” have been developed and are ready for off-line application.

2.3 Laboratory and field results

The main field protocols concerned the neck and the low back in computer workers and nurses. The neck study involved the upper trapezius of 88 cases and 164 controls. The low back study involved the erector spinae muscles of 59 cases and 225 controls. Collateral laboratory studies focused on specific muscles and issues. A NEW database was prepared to host the subject data and the results of the tests and measurements. The major finding on the large-scale epidemiological study is the significant difference in upper trapezius muscle strength in the neck protocols. This finding was supported to be of muscular origin by the lower EMG RMS value in the cases compared with the controls.

2.4 Dissemination and teaching

Dissemination of results among partners, to the scientific community and to potential users was carried out through three main channels :

a) publications on international peer reviewed journals and international congresses (59 published or accepted works, 28 presentations at international meetings)

b) thirty-one visits of researchers among the partners, with durations ranging from a few days to four months

c) seven two-day courses aimed at experts of occupational and physical medicine, ergonomists, physiotherapists

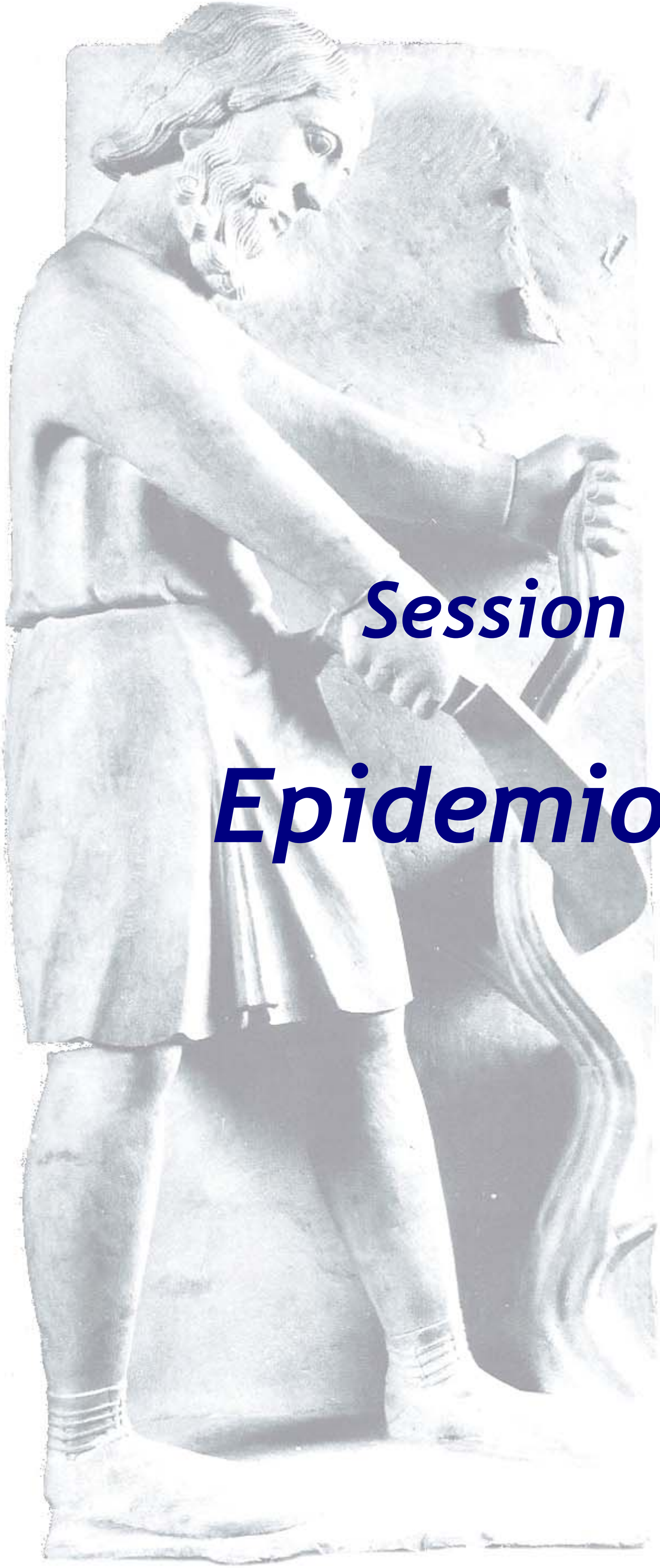
The dissemination activity received specific support from the Regional Administration of Piemonte (Italy) which provided support for courses and applications in specific fields (e.g. carpal tunnel syndrome).

3. Outline of the symposium.

The International Symposium on Neuro-muscular Assessment in the Elderly Worker is structured in lectures given by the NEW partners and contributions from other participants. The Friday morning (Feb. 20, 2004) lectures provide an overview of the problem and the methods of investigation. The afternoon lectures address the issue of information extraction from the EMG and MMG signals. The contributions from other speakers and the posters of the day concern movement and EMG analysis, age effects and issues related to ergonomics. The Saturday (Feb. 21 2004) morning lectures complete the issue of information extraction from the EMG signal and describe the results obtained from questionnaires while the afternoon lectures describe the integration of different findings, the biofeedback applications and the application of NEW results in sport, space, and rehabilitation medicine. The contributions from other speakers and the posters of the day concern muscle fatigue, spatial filtering, the correlation between biomechanical, bioelectrical and biochemical variables and other issue such mastication.

Acknowledgements

Project NEW received support from organisations besides the EU. Among these are: Compagnia di San Paolo, Fondazione CRT and Regione Piemonte, Italy.



Session 1:

Epidemiology

Evaluation of muscle tension of the shoulder and neck during microscope work using surface EMG.

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Abstract *– Musculoskeletal symptoms such as neck and low back pain are very often reported by professionals using microscopes on a daily basis. The goal of this study, to investigate which ergonomic adaptations to the microscope workstation of a chemical company, will result in a reduction of musculoskeletal stress. We used surface EMG recordings from the m. trapezius in 27 healthy volunteers while they were working both in a simulation of the real workstation and in a simulation with ergonomic changes. A discomfort rating (Category Ratio-10 Borg scale) was used to measure the subjective stress. We can conclude that by using the back of the chair and a maximum deviation of 5-10° from the glance there is a significant reduction of musculoskeletal stress.*

1. Introduction

Musculoskeletal disorders are the most important cause of sick-leave [1]. Because of the enormous sick-pay costs there is a growing interest from companies in ergonomic workstations to anticipate this problem.

Lee *et al.* [2] proved that there is a significant muscle tension caused by sitting a long time in a fixed position. Also a clear correlation between muscle tension in the *m. trapezius pars descendens* and the development of musculoskeletal discomfort in the upper part of the body on the other is reported [3,4]. However, there is lack of scientific studies that investigate work-related physical disorders caused by long-term microscope work [5,6]. Haines & McAtamney carried out a study with cytotechnologists that showed that about 60-80% complained about head and neck ache, while 75% suffered from shoulder-and backache. The use of an ergonomic microscope workstation would lead to a rise of production of 5-25% [5].

A chemical company asked to investigate their microscope workstation at the division quality control and to apply ergonomic changes that would result in a reduction of musculoskeletal tension for

2. Method

27 Healthy volunteers participated, 16 women and 11 men, aged between 19 and 31 years. They all had a normal sight but no microscope experience.

The tests were carried out in a simulation setting reflecting the real workstation in the chemical company. The subjects were examined in 2 situations: the simulation of the workstation ("non-ergonomic condition") and a simulation of the workstation with ergonomic changes ("ergonomic condition")

In the non-ergonomic condition we tried to simulate the real workstation as good as possible. The height of the worktable measures 85 cm and the height of the chair (Ergocomfort) was adapted to each subject in such a way that the angle between thigh and abdominal wall was 90°. The task they were asked to do is to count cells in a microscopic specimen. The forearms were completely supported by the worktable. Similar to the real work of the professional staff in the company, the back of the chair is not used and the neck is positioned in a high cervical extension.

In this condition the subject worked (doing the same task) in a workstation with some ergonomic changes. The height of the worktable, the angle of the hip and the support of the forearms remains the same. But the subjects had to use the back of the chair and the microscope was placed in such a way that the test-persons only had to deviate 5-10° from their horizontal glance.

A discomfort rating (Category Ratio-10 Borg scale) was used to measure subjective stress.

Surface electromyography (Biograph) with a sampling rate of 32/sec. was used as an objective measure for the quantitative assessment of physical stress. The electrodes were attached in paramedian way on the upper and lower neck at the level of the 3rd and 7th cervical vertebra (fig.1). Muscle selection was based on the rational concept of upper body muscle involvement during microscope operation.

First maximal voluntary contractions (MVC) were carried out for each point of measurement in order to calculate individual procentual rms-values. The experiment was carried out according to a standard protocol: EMG-measurements were performed for 20 minutes, data were selected in the 1^{st}, 5^{th},10^{th}, 15^{th} and 20^{th} minute. The collected data are represented by the mean for each minute in the two conditions.

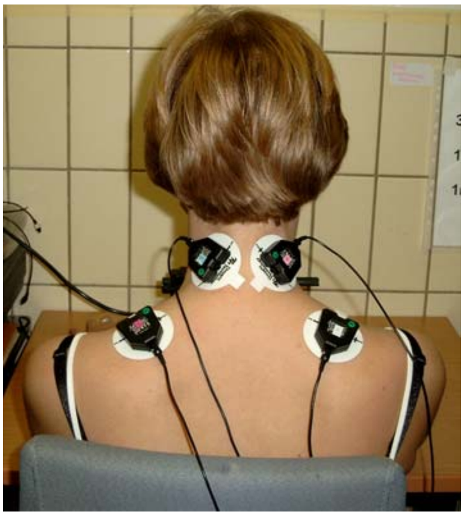

FIG. 1 : Paramedian attachment of the electrodes on C3 and C7

To eliminate the effect of learning, 14 subjects started with the ergonomic setting and 13 with the non-ergonomic condition

The Kolmogorov-Smirnov test was used to examine normal distribution. Significant differences were tested using the Wilcoxon test (p= 0.05).

3. Results

Working at the non-ergonomic workstation results in an elevated EMG activity in both the upper and lower region of the *m. trapezius.* Muscle tension in the upper part (C3) is significantly higher than in the lower part (C7) both in the non-ergonomic and ergonomic condition. The relative

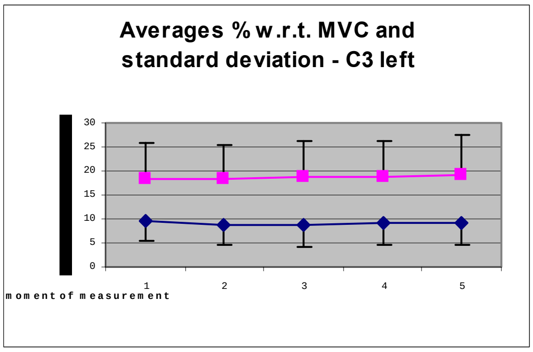

FIG. 2 : Averages % w.r.t. MVC and stand. dev. C3 left

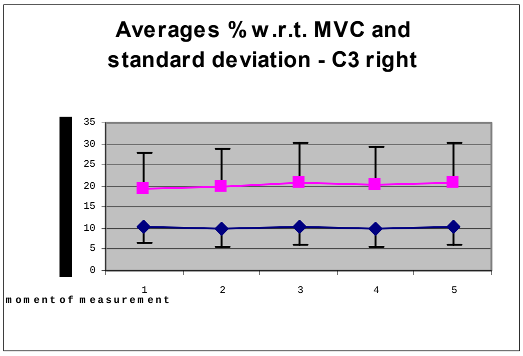

FIG. 3 : Averages % w.r.t. MVC and stand. dev. C3 right

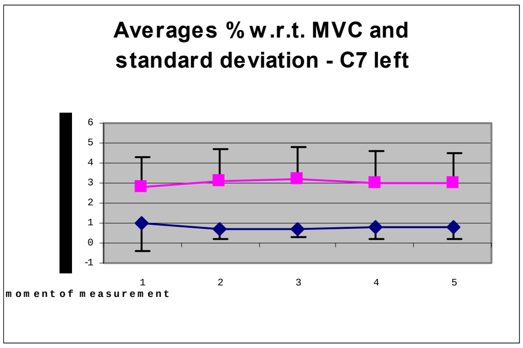

FIG. 4 : Averages % w.r.t. MVC and stand.dev.C7 left

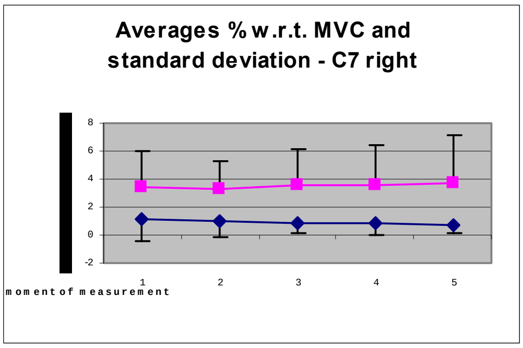

FIG. 5 : Averages % w.r.t. MVC and stand.dev.C7 right

reduction in muscle tension created by the ergonomic working condition is most pronounced in the lower muscle part (C7: factor 3-4, C3: factor 2). This major reduction in distress in the shoulder region was also recorded by the Borg scale. All volunteers experienced the ergonomic working position more relaxing.

With sustained work (20 min), EMG activity did not change significantly over time in both conditions.

There was just a bad correlation between left and right in the non-ergonomic condition both for C3 and C7 (0.508 and 0.180 resp.). The ergonomic conditions results in a good correlation between both sides (0.803 and 0.713 resp.) The right side of the body shows a slightly higher muscle tension. The sequence in which the tests were performed (first ergonomic or non-ergonomic) did not influence the results.

4. Discussion

A significant reduction in muscle tension during microscopic work can be achieved by using the back of the chair and reducing the deviation from the glance to less than 5-10°. The positive effect of a minimal high cervical extension created by this situation was already shown by Menozzi *et al.* [7]. A few adaptations, necessary to keep the normal physiological posture (e.g. height of the chair, microscope closer to the edge of the table, full support of the forearms) clearly effect distress and muscle tension during sedentary microscopic work [8,9,10,11]. Also Kumar and Scaife [12] showed that even very little changes to a workstation result in a significant decrease in muscle tension. Because there are no significant changes in the EMG-pattern during the 20 minutes, we cannot say that fatigue of the muscles occurs. Based on the high correlation between the different points of measurement within each condition, we conclude that EMG during 2 or 3 minutes is enough to become an objective measurement of the muscle tension during microscopic work. Often EMG-measurement is only performed on one side. The moderate correlation between left and right in the non-ergonomic condition of this experiment leads us to the conclusion that bilateral measurement is very important. The fact that there was no difference between the group who started with condition 1 and the others shows that there is no effect of learning. This study shows, as other studies [2,13], a high correlation between the changes in objective EMG-values and the subjective feeling of muscle tension. In many subjects not only an elevated tension of the *m. trapezius* was noted but also at the *m. rhomboideus*. There is also a very clear relationship between the static tension of the *m. trapezius* and the development of musculoskeletal discomfort [3,4].

In future research on strain in the neck region, it could also be interesting to place electrodes on the *m. rhomboideus* or add modifications resulting in a changed angle of the hip. It is clear that some minor (low-budget) changes will help to prevent the distress-syndromes for professionals using microscopes and clearly results in a reduction of muscular tension in the neck and shoulder region.

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Explaining disabilities in women with work-related neck-shoulder pain

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1. Introduction

The fear-avoidance model[1] is generally accepted to explain disabilities in patients with low back pain and because of suggested similarities between low back pain patients and patients with work-related neck-shoulder pain[2], this study explored the applicability of this model in the last group.

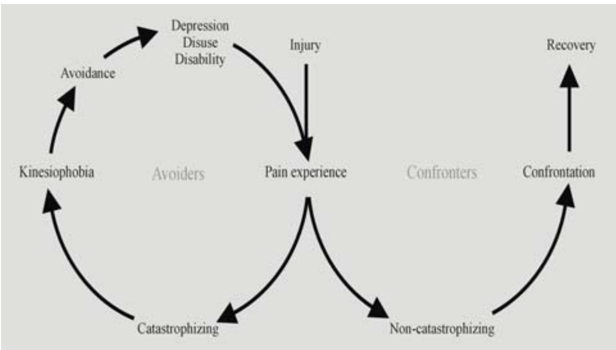

FIG. 1 : Fear-avoidance model of Vlaeyen *et al* (1995)

2. Methods

The sample consisted of 58 patients with work-related neck-shoulder pain and 45 healthy controls. In a cross-sectional study a number of questionnaires were completed; Borg scale for pain intensity, Coping Strategies Questionnaire subscale catastrophizing (CA), Neck Disability Index (NDI), Tampa Scale for Kinesiophobia (TSK) and the Fear-Avoidance Beliefs Questionnaire (FABQ). In addition, subjects performed a static maximal voluntary shoulder elevation contraction (MVC) of the trapezius muscle. Group-differences were tested using t-tests. Relationships between parameters were investigated using Pearson correlation coefficients. Three multiple regression analyses were performed for pain-related fear, MVC and NDI as dependent variable.

3. Results

Cases were significantly more disabled. All the scores on the questionnaires were very low compared to their maximal possible ranges. Catastrophizing was significantly correlated to only one of the three pain related fear measures namely FABQ-W (r=0.27) but catastrophizing had no additional value in explaining pain related fear over age and pain duration. FABQ-W and TSK explained a significant portion of performance. Nevertheless, total portions explained variances were extremely low (<9%). The strongest correlation existed for FABQ-W and NDI (r=0.46). Remarkable is also the significant strong correlation between pain intensity and NDI (r=0.44). The regression analysis with disabilities as dependent variable showed that catastrophizing, FABQ-PA and pain intensity were the most powerful predictors for disabilities.

4. Discussion

In line with the fear-avoidance model a specific form of pain-related fear appears to be present in our case group which strongly correlated with disabilities, namely fear-avoidance beliefs about work. Counter to the fear-avoidance model no elevated levels of kinesiophobia and catastrophizing were found in patients with computerwork-related neck-shoulder pain as compared to healthy controls. In addition, total proportions explained variances found in our study were lower when compared to others, especially in explaining behavioural performance[3]. Also counter the fear-avoidance model is the great role of biomedical variables (age, pain duration and pain intensity)[4]. As most results found in the present study are in contrast to the fear-avoidance model it is concluded that this model is not as applicable to patients with work-related neck-shoulder pain as it is to patients with low back pain. Counter to physical activity, patients are afraid of working-activities. This suggests that other models than the fear-avoidance model are of importance.

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Aging effects on development of musculo-skeletal disorders among workers: assessment by WAI (Work Ability Index) in chemical, pharmaceutical, public, sanitary, energetic and building fields

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1. Introduction

In the developed countries, the progressive people aging and the working age extension have caused social and economic consequences. The elderly people often suffer from chronic diseases that increase difficulties in worker's ability. In fact, by aging there are structural and functional body changes and workers are more sensitive to job risk factors. Insurance data show that work's accidents severity increases by aging (1).

During last years we observed a reduction of "typical" professional diseases, whereas "work-related diseases" increased: these are often chronic-degenerative or psycho-somatic diseases, like musculo-skeletal disorders or osteo-arthropathies due to repetitive and heavy tasks with awkward postures (2,3,4,5,6).

The human functional ability decreases after 45 years and produces a reduction of work ability. The muscular capacity lower by aging (7): the force, strength and muscular mass reduce themselves. Rachis and upper limbs are the most affected.

The authors present the results of a study on work ability in elderly workers and discuss data related to musculo-skeletal disorders.

2. Materials and Methods

The WAI questionnaire (8,9,10) evaluates the self-perceived work capacity. This index (illustrated in Appendix A) has been obtained by worker's answers to questions on physical and mental task demands; moreover healthy status and employer's opinion on their own work ability were investigated. We administrated the questionnaire (to 629 workers from chemical, pharmaceutical, public, sanitary, energetic and building fields, from 45 to 65 years old.

The average age was 51.2 years (± 4.9) and the obtained prevalence data, related to musculo-skeletal disorders, are illustrated in Tabs 1 and 2.

We used the χ-square test ($P \leq 0.05$) for the statistical analysis and then we evaluated the correlation between WAI score and aging.

TAB 1: Prevalence of musculo-skeletal disorders

	PREVALENCE OF DISEASES (%)		
Field (n. of workers)	*Cervical C.*	*Lumbar C.*	*Upper limbs*
Chemical (111)	23 (20.7)	39 (35.1)	15 (13,5)
Energetic (83)	25 (30.1)	32 (38.5)	21 (19.02)
Public (112)	39 (34.8)	27 (24.1)	29 (25,8)
Pharmaceutical (113)	9 (7.9)	31 (27.4)	20 (17,7)
Sanitary (110)	18 (16.3)	22 (20)	24 (21,8)
Building (100)	18 (18)	32 (32)	30 (30)
Total 629	132 (20.9)	183 (29.1)	139 (22.1)

TAB 2: Musculo-skeletal disorders by age group

	PREVALENCE OF DISEASES %		
Age group (n. of workers)	*Cervical C.*	*Lumbar C.*	*Upper limbs*
45-49 (260)	35 (13.4)	63 (24.2)	45 (17.3)
50-54 (215)	59 (27.4)	64 (29.7)	40 (18.6)
55-59 (106)	23 (21.7)	23 (21.7)	26 (24.5)
60-65 (48)	15 (31.2)	15 (31.2)	23 (47.9)
Total 629	132 (20.9)	165 (26.2)	134 (21.3)

3. Results

The final WAI score decreases by aging. We obtained a significant negative correlation between age and WAI score, especially since 50-54 age group and in working sectors with prevalent psycho-physical engagement (r = -0,45) (Fig. 1).

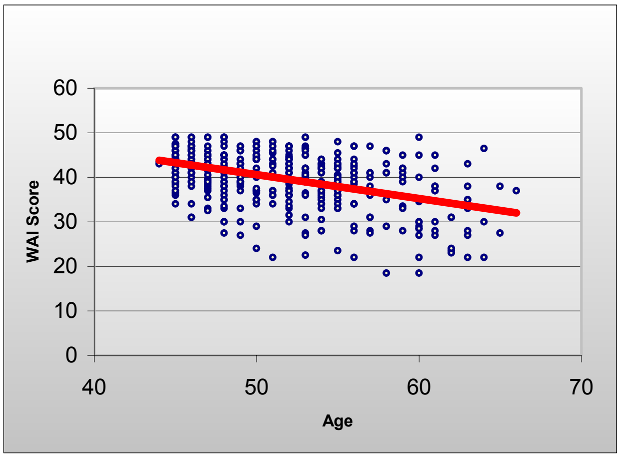

FIG. 1: Relationship between WAI and Age

We observed that cervical and lumbo-sacral column and upper limbs musculo-skeletal disorders were over 30% of all diseases. Particularly, cervical rachis disorders are prevalent in public sector, lumbo-sacral disorders prevail in chemical sector and upper limbs disorders are more frequent in public and sanitary fields.

4. Conclusions

Work ability decreases by aging, especially after 50 years and in the sectors with prevalent psycho-physical engagement. The use of WAI to evaluate the relation between work and aging is an useful index to predict the risk of work inability and it will allows to adjust the work environment to functional changes related to aging of worker.

The physical capacity decreases by aging, so the job physical load should be reduced, more than the preview reduction for an analogue case in younger worker (e.g. by greater reduction of heavy handling or repetitive tasks).

Some "micropauses" (1-3 minutes) should be immediately allowed after tiring works in order to aid a better physical recovery, and other pauses should be allowed when required by workers. Moreover, we believe it is necessary to think about ergonomic solutions preserving work ability (e.g. to reduce muscular effort) and to put more attention to awkward postures.

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Appendix A

WAI - WORK ABILITY INDEX (modified)

BACKGROUND	
Sex	
Female	1
Male	2
Age years	
Marital status	
Unmarried	1
Married	2
Unmarried but co-abitating	3
Separated	4
Divorced	5
Wedow/er	6
Basic education	
Elementary school	1
Comprehensive school	2
Intermediate school	3
Secondary school	4
Other education (specify)………………………………………..	5
Vocational/professional education	
Vocational course for the unemployed (at least 4 months)	1
Other vocational course (at least 4 months)	2
Vocational School	3
Vocational Institute/College	4
University	5
Other training (specify)…………………………………………..	6
Occupation (specify)	
Work task (specify)	
Industrial branch of employement (to be filled out by occupational health personnel)	
Workplace	
Department	
Are the demands of your work primarily?	
Mental	1
Physical	2
Both	3

WORK ABILITY INDEX
1. Current work ability compared with the lifetime best
Assume that your work ability at its best has a value of 10 points. How many points would you give your current ability? (0 means that you cannot currently work at all)
0 1 2 3 4 5 6 7 8 9 10
Completely unable to work … Work ability at its best
2. Work ability in relation to the demands of the job

How do you rate your current work ability with respect to the **physical** demands of your work?	
Very good	5
Rather good	4
Moderate	3
Rather poor	2
Poor	1
How do you rate your current ability to the **mental** demands of your work?	
Very good	5
Rather good	4
Moderate	3
Rather poor	2
Poor	1

3. Number of current diseases diagnosed by a physician

In the following list, mark your disease or injurie. Also indicate **whether a physician has diagnosed** or treated these disease. For each disease, therefore, there can be 2, 1 or no alternative.	Yes	
	Own opinion	Physician's diagnosis
Injury from accident		
01 back	2	1
02 arm/hand	2	1
03 leg/foot	2	1
04 other part of body, where and what kind of injury	2	1
Musculoskeletal disease		
05 disorder of the upper back or cervical spine, repeated instances of pain	2	1
06 disorder of the lower back, repeated instances of pain	2	1
07 (Sciatica) pain radiating from the back into the leg	2	1
08 musculoskeletal disorder affecting the limbs (hand, feet), repeated instances of pain	2	1
09 rheumatoid arthritis	2	1
10 Other musculoskeletal disorders, what? ……..………………………………………………..	2	1
Cardiovascular disease		
11 Hypertension (High blood pressure)	2	1
12 Coronary hearth disease, chest pain during exercise (angina pectoris)	2	1
13 Coronary thrombosis, myocardial infarction	2	1
14 Cardiac insufficiency	2	1
15 Other cardio-vascular disease, what? ………………………………………………………	2	1
Respiratory disease		
16 repeated infection of the respiratory tract (also tonsillitis, acute sinusitis, acute bronchitis)	2	1
17 chronic bronchitis	2	1
18 chronic sinusitis	2	1
19 bronchial asthma	2	1
20 emphysema	2	1
21 pulmonary tuberculosis	2	1
22 other respiratory disease, what? ………………………………………………………	2	1

Mental disorder		
23 mental disease or severe mental health problem (for example severe depression, mental disturbance)	2	1
24 slight mental disorder or problem (for example slight depression, tension, anxiety, insomnia)	2	1
Neurological and sensory diseases		
25 problem or injury to hearing	2	1
26 visual disease or injury (other than refractive error)	2	1
27 neurological disease (for example, stroke, neuralgia, migraine, epilepsy)	2	1
28 other neurological or sensory disease, what? ……………………………………………………….	2	1
Digestive disease		
29 gull stones or disease	2	1
30 liver or pancreatic disease	2	1
31 gastric or duodenal ulcer	2	1
32 gastritic or duodenal irritation	2	1
33 colonic irritation, colitis	2	1
34 other digestive diseases, what? ……………………………………………………….	2	1
Genitourinary disease		
35 urinary tract infection	2	1
36 kidney disease	2	1
37 genital disease (for example fallopian tube infection in women or prostatic infection in men)	2	1
38 other genitourinary disease, what? ……………………………………………………….	2	1
Skin disease		
39 allergic rash/eczema	2	1
40 other rash, what? ……………………………………………………….	2	1
41 other skin disease, what? ……………………………………………………….	2	1
Tumor		
42 benign tumor	2	1
43 malignant tumor (cancer), where? ……………………………………………………….	2	1
Endocrine and metabolic diseases		
44 obesity	2	1
45 diabetes	2	1
46 goiter or other thyroid disease	2	1
47 other endocrine or metabolic disease	2	1
Blood diseases		
48 anemia	2	1
49 other blood disorder, what? ……………………………………………………….	2	1
Birth defects		
50 birth defect, what? ……………………………………………………….	2	1
Other disorder or disease		
51 What? ………………………………………..	2	1

4. Estimated work impairment due to diseases

Is your illness or injury a hindrance to your current job? Circle more than one alternative if needed.	
There is non hindrance/I have no diseases	6
I'm able to do my job but it causes some symptoms	5
I must **sometimes** slow down my work pace or change my work methods	4
I must **often** slow down my work pace or change my work methods	3
Because of my disease, I feel I am able to do only part time work	2
In my opinion, I am entirely unable to work	1

5. Sick leave during the past year (12 months)

How many **whole days** have you been off work because of a health problem (disease or health care or for examination) during the past year (12 months)?	
None at all	5
At the most nine days	4
10-24 days	3
25-99 days	2
100-365 days	1

6. Own prognosis of work ability two years from now

Do you believe that, from the stand point of your health, you will be able to do your current job **two years from now**?	
unlikely	1
not certain	4
relatively certain	7

7. Mental resources

Have you recently been able to enjoy your regular daily activities?	
often	4
rather often	3
sometimes	2
rather seldom	1
never	0
Have you recently been active and alert?	
always	4
rather often	3
sometimes	2
rather seldom	1
never	0
Have you recently felt your self to be full of hope for the future?	
continuously	4
rather often	3
sometimes	2
rather seldom	1
never	0

The personnel data will be treated exclusively for survey or research in the respect of the privacy-law

Self-assessed and measured physical capacity of elderly female computer users – the NEW study

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1. Introduction

The most accurate estimation of physical capacity is obtained by measuring relevant physical parameters during functional tests. An alternative way is to let the subjects self-assess their physical capacity. This method is obviously less accurate but in big scale studies the accuracy may be sufficient to investigate the significance of high physical capacity as a modulating factor for the incidence of muskuloskeletal disorders. To avoid artificial categories and value-laden words visual analogue scales (VAS) can be used to score the self-assessed physical capacity (6).

The aim of this study was to compare low-back cases and controls regarding measured and self-assessed physical capacity.

2. Methods

2.1 Participants and case definitions

Elderly female computer users above 45 years were recruited. A total of 16 subjects fulfilled the criteria for low back cases (LB) by reporting trouble (ache, pain, discomfort) in the low back region for more than 30 days during the last year and by having trouble (more than 30 days) from no more than two other body regions. A total of 32 controls (Con) without trouble in the low-back region for more than seven days during the last year and a maximum of three other body regions with trouble (more than 30 days) were defined.

2.2 Measured physical capacity

Functional lifting capacity was evaluated by the lower and upper Progressive Isoinertial Lifting Evaluation test (PILE) by Mayer 1988 (5), back extension endurance was tested by the modified Biering Sörensen test (3,7) and back flexion endurance was tested by a similar procedure as Ito et al. (4). In addition maximal isometric back extension, back flexion and shoulder elevation strength were tested (2).

2.3 Self-assessed physical capacity

Five questions were asked related to five different aspects of physical capacity: aerobic fitness, strength, endurance, flexibility and balance/coordination. All questions were phrased like this: “How would you score your aerobic fitness compared to women on your own age?”. To answer, a vertical mark was set on a 100mm long VAS with a descriptive word and figure at each end of the line. Similarly, three more questions were asked about the strength, endurance and flexibility of the back. The answers were scored on a VAS with guiding words but without figures.

2.4 Statistics

Normal distributed data was tested by students t-test and not-normal distributed data was tested by Mann-Whitneys rank sum test. Significance level was set at $p<0.05$ and the results in the text are displayed as mean±sd.

3. Results

The LB and Con were comparable according to both age (LB:54.1±4.2yr Con:53.8±5.4yr) and body mass index (LB:26.2±4.2 Con:25.5±4.2). No differences were seen in the PILE tests (lower PILE: LB:14.9±5.3 kg Con:16.5±4.0 kg; upper PILE: LB:10.8±3.7 kg Con:11.6±2.6) kg or in the back flexion and extension strength (flexion: LB: 98±18 Nm Con: 104±29 Nm and extension: LB: 110±28 Nm Con: 109±28Nm). However, there was a tendency to a lower back extension endurance time in LB compared to Con ($p=0.064$).

In the five general aspects of the self-assessed physical capacity no differences were seen between the LB and Con. In contrast, all three scores concerning the physical capacity specifically of the back, were significantly lower in the LB compared to Con ($p<0.001$).

4. Discussion and conclusion

No significant differences were seen between the LB and Con in measured physical capacity. However, the back extension endurance time showed a tendency to be lower in LB compared to Con. A difference in this parameter could have been expected since previous studies found the back extension endurance being a better predictor of low back pain than the maximal low back extension strength (1).

The difference in the self-assessed capacity between LB and Con could not be confirmed by a significant difference in any of the measurements of functional capacity.

Further, the lower self-assessed capacity was only evident, when the region of trouble was the focus for the self-assessment.

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Movement problems of elderly drivers: women provide evidence of noticeable difficulties

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Abstract - *Within a research carried out with Fiat Auto on the elderly population we discovered that elderly women report more problems than men of equivalent age for almost all movement aspects that were examined: in turning the head during bends and manoeuvres, in entering and exiting from the car, in reaching a completely open door and in seat belt activation. This fact can probably depend partly on smaller female anthropometric dimensions and lower strength and partly on the higher incidence in women of muscular-skeletal pathologies, that come out from our data analysis on the pathologies declared by the subjects. Objective data obtained through observational tests carried out on a sub-sample confirm gender differences only for seat belt pulling. This fact can depend on the limited number of subjects used in the observational tests and on a certain inadequateness of the adopted indicators in evaluating women's particular difficulties encountered during those operations. In any case, as higher female longevity will considerably increase the number of women that will continue to drive, it is important to conceive new cars paying special attention to elderly women's needs through the introduction of features that can allow them to overcome the main problems that tend to limit their mobility.*

1. Introduction and methodology

Within an ergonomic oriented research carried out in co-operation with Fiat Auto intended to analyse the transformations with ageing of human characteristics involved in human-car interaction, we discovered that elderly women show more movement problems than men of equivalent age.

On a sample of 312 subjects, found among Fiat Pensioner Group members and, for the younger segment, among the company workers, subdivided by gender and age group (50-59, 60-69, >70 years), we collected a series of anthropometric and biomechanical data and also information on personal data, lifestyle and medical problems. The first part of the research also investigated the most problematic aspects for elderly drivers and passengers and the factors that tend to limit car use by older people.

Subsequently, on a sub-sample of 25 particularly significant subjects over 60, defined on the basis of our anthropometric data analysis, we observed, through video-camera recording, the body movements during some operations that, according to the assessment of the total sample, proved to be particularly critical for the more elderly subjects: getting into the car, getting out of the car, accessibility of the door, fastening and unfastening the seat belts.

Detailed aspects of the methodology and of the results have been described in other papers [1] [2]; here we will examine some particular outcomes related to the occurrence of significant gender differences both in the declared movement problems and in the observational results on body movements.

2. Results

2.1 Problems surveyed from questionnaires

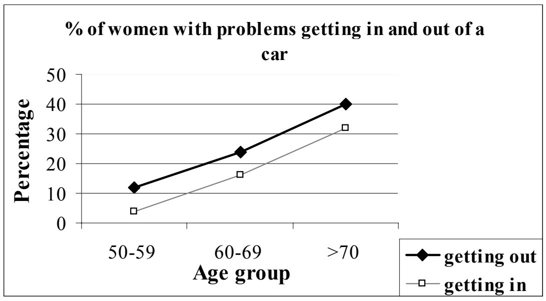
Statistical analysis of questionnaire data through ANOVA tests shows that in most cases women report more problems than men driving and using cars. Some problems are mainly psychological, as the fact that many of them stop driving when very young because of insecurity and fear. Other problems are related to critical environmental situations (they tend more to avoid driving in bad atmospheric conditions, at sunset or by night, in tunnels and on highways) and to dislike long distance driving. They declare to a higher extent to have vision problems in various circumstances but mainly in night visibility and while entering and exiting tunnels.

Besides visibility, the main difficulties reported by all the participants that can be directly correlated with car design are movement, reaching and activation problems (Table 1), and, in particular, entry and exit movements, reaching a completely open door and seat belt activation, in which women show percentages that are significantly higher than men.

Particularly relevant for women are movement problems for getting into a car and, especially, for getting out of a car. In Figure 1 it can be seen that for women there is a dramatic increase in these problems with age, while men show a considerably lower percentage of problems, and the increase is only visible for the 70-year-old group.

TAB. 1: Percentage of problems reported by men and women of different age classes

MAIN REPORTED PROBLEMS		% Men			% Women		
		50/59	60/69	>= 70	50/59	60/69	>= 70
Movement	Exit	10	7	26	12	24	40
	Entry	8	7	19	4	16	32
	Turning head	13	15	13	31	35	31
Reaching	Seat belt	32	30	27	30	39	26
	Compl.open door	27	16	27	46	46	43
	Dashboard	8	4	2	8	4	5
	Pushbuttons	4	7	4	4	7	16
Activation	Seat belt	10	4	22	18	23	23



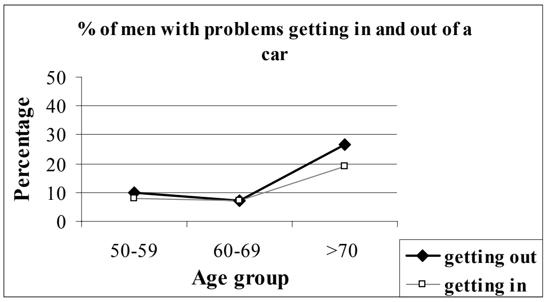


FIG. 1: Percentage of people with problems getting in and out of a car

The higher percentage of problems among women can be due not only to a greater prevalence of muscular-skeletal pathologies among the females, that come out from our data analysis on the pathologies declared by the subjects, but also to the smaller female anthropometric size and to wearing skirts.

The higher incidence of difficulties in reaching a completely open door and in seatbelt activation can also be related to their smaller dimensions, but the latter problem can probably also depend on their lower strength.

In effect we found that the females of our sample could develop handgrip forces very low in comparison with males (Figure 2).

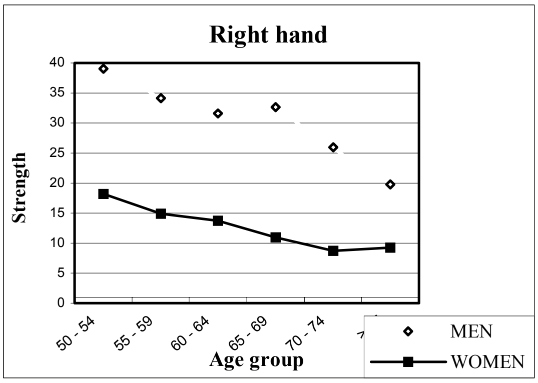


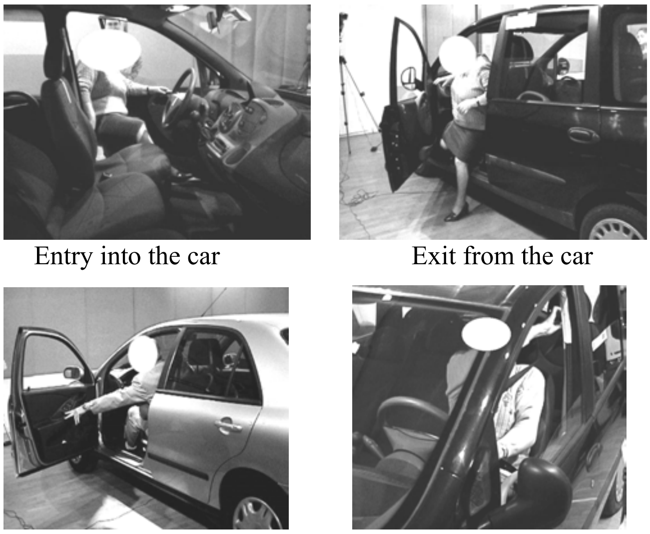
FIG. 2: Average handgrip strength (in Kg) by age

2.2 Problems evidenced during observation tests

On a sub-sample of 25 subjects (12 males and 13 females) over 60 we carried out specific observation tests with video recordings of body movements during operations that from the questionnaires proved to be more critical for older subjects (Figure 3):

- entry into the car
- exit from the car
- door accessibility
- seat belt fastening and unfastening.

All subjects performed such operations on 4 different car models chosen as very different in dimensions, like size of door, door weight, sill height, door step dimension, height of the seat from the ground, which can influence the analysed operations. The order in which the cars were evaluated was balanced by a Latin Square Design so as to minimise learning effects. Female participants who normally wore skirts were asked to wear them during the tests too, so as to investigate problems of accessibility related to wearing these garments.



Entry into the car | Exit from the car

Door accessibility | Seat belt fastening and unfastening

FIG. 3: Operations analysed during observation tests

For video analysis each operation was subdivided into a series of phases so as to standardise the evaluations that mainly regarded:

- the different strategies used by participants in the analysed operations
- the support zones utilised by people that kept hold of different car elements
- the impacts of subjects into different car parts
- the average number of seat belt pulls.

We found that the percentage of subjects who hold on to some part of the cars while entering or exiting was always rather high (in average 48% in entering and 89% in exiting) and also the percentage of subjects who bump into something while getting in and out of a cars (53% in entering and 29,5% in exiting) - and it is then very important to consider these aspects in the design of cars that can suit the elderly - but we could not find significant differences between men and women. We could not find significant gender differences either in movement strategies during entry and exit, as could be expected from the fact that women were asked to wear a skirt. In the same way we did not find significant gender differences in door accessibility, that was mainly affected by door angle and handle layout and design.

Vice versa for seatbelts, we found that women did a higher number of pulls, that is the number of times that the belt is pulled to overcome the obstruction of the subject's body and reach the fastening. In Table 2 it is possible to see the number of mean seatbelt pulls in the different cars for men and women: the average number is quite high, especially for women, and this fact can probably depend on their lower strength.

TAB. 2 : Average number of mean seat belt pulls in the different cars

Car	Men	Women
A	4,0	4,7
B	3,5	4,2
C	3,5	4,3
D	3,5	4,1

3. Discussion

Our observational data on entry and exit movements and on door accessibility do not confirm the gender differences about problems declared by subjects in the first part of the research. This fact can depend on two elements: the limited number of subjects used in the observational tests and a certain inadequateness of the adopted indicators in evaluating women's particular difficulties during those operations. It would be interesting to verify if EMG techniques applied to the elderly during the execution of the above-described operations could evidence specific body movement problems in women.

4. Conclusion

In any case, our results show that it is necessary to pay special attention to women's needs, who significantly perceive to have more problems than men during car use. Higher female longevity will considerably increase the number of women that would like to continue to drive, as mobility, in our society, is one of the main aspects of quality of life allowing more control and freedom in the choice of daily life activities and free time occupations. It is then important to conceive new cars that can allow them to overcome the main problems that tend to limit their mobility, taking into particular account their differences in movement capacity, in size and in force exertion. The final goal is to design cars with features that tend to facilitate their use to everyone, passenger or driver, regardless of their age or sex or the presence of degenerative disorders, and comfortable for the greatest possible number of people.

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Relation between cervical posture on lateral skull radiographs and EMG activity of masticatory muscles in Caucasian adult women. A cross-sectional study

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1. Introduction

The relationship between cervical and oro-facial areas was previously investigated. These studies mostly analyzed morphological interrelationships [1] and functional interrelationship [2].

The aim of this study was to evaluate correlation between sEMG recordings of head and neck muscles and cervical posture in homogeneous sample for ethnic origin, sex, and age (Caucasian adult females). If a significant correlation could be found between cephalometric variables and sEMG recordings of muscles, the complex relationship between head morphology and head posture could be better understood.

2. Materials and Methods

2.1 Study population

The sample comprised 54 females, aged 25-35 years, average 26.8, admitted to the Department of Orthodontics and Gnatology, University of Chieti, for orthodontic evaluation. The criteria for selection were sex, European ethnic origin, confirmed day of birth, adult age.

2.2 Cephalometric Tracings

Lateral skull radiographs were taken using Instrumentarium Imaging®, Orthopantomograph OP100. Exposure data were 70 KV and 20 mA/s. The distance between the head and radiological tube was 1,5 m. High-speed intensifying screens were used. The radiographs were exposed with the subjects standing in the [1], defined as the intention position from standing to walking, looking into a mirror [2]. In order to minimize external influence, no ear rods were used in the cephalostat, as shown in Figure 1. The radiologist was asked to register on lateral skull radiographs all the neck and the sixth cervical vertebra. Sixteen craniofacial morphological variables have been studied.. Lordosis of the cervical spine was measured on the lateral skull radiographs according to Hellsing.

2.3 SEMG Recordings

The study was performed using a Key-Win 2.0 surface electromyography (Biotronic s.r.l., S.Benedetto Tronto, Ascoli Piceno, Italy) All monitoring was performed with the patients in a standing position. The subjects were asked to make themeselves comfortable, to relax their arms by their sides, and to look straight ahead and make no head or body movements during the test. The sEMG activity of eight muscles was studied bilaterally [1] with the mandible at the rest position, [2] during maximal voluntary clenching

(MVC). The muscle were: the right masseter (RMM), left masseter (LMM), right anterior temporalis (RTA), left anterior temporalis (LTA), right digastric (RDA) and left digastric (LDA), as masticatory muscles; the right sternocleidomastoid (RSM), left sternocleidomastoid (LSM), right posterior cervicals (RPC), left posterior cervicals (LPC), right upper trapezius (RUTR), left upper trapezius (LUTR), right lower trapezius (RLTR) and left lower trapezius (LLTR) as postural muscles. The sEMG recording time for each analysis was at least 15 seconds, and the values were expressed in microvolts per second (μVolts/sec). For each subject, sEMG recordings were considered as the sum of the recordings of the left and the the right sides. [3]

2.4 Data treatment

The error variance was calculated using Dahlberg's formula: [4]

A Pearson's correlation coefficient was performed to evaluate the association between cephalometric and sEMG recordings. Due to the small sample and in order to increase the power of statistical tests, $p<0.01$ was considered as statistically significant.

3. Results

Intraobserver method error variance for all variables was found to be less than 5% of the biological variance for the whole sample, as shown in Table 1. Table 2 shows descriptive statistic (Mean, Standard Deviation – SD – Range, Minimum, Maximum and Variance) for the variables describing head posture, cranio-cervical angulation, cervical posture, cervical lordosis and sEMG activity. Table 3 describes the results of the correlation analysis for the variables studied in the whole sample. Due to the small sample and in order to increase the power of the statistical test, only correlation with $p<0.01$ was accepted as statistically significant.

4. Discussion

The results

The most important finding in this study was that a significant correlation was found between cranio-cervical angles and sEMG activity of masseter muscle at MVC, as the higher the sEMG activity of masseter muscle was, the smaller the cranio-cervical angulation was. On the other hand, subjects with a lower sEMG activity showed a greater cranio-cervical angulation.

This finding seems to be in accord with previous important observations. In fact, Solow and Tallgren [2] noticed that the position of the head in relation to cervical column (that is cranio-cervical angultion) showed more correlations with facial morphology than the conventional measures of head posture, that is the position of the head in relation to true vertical. In those studies, subjects with a large cranio-cervical angle showed reduced facial prognatism, large mandibular plane inclination, and large lower anterior facial height. In addition, these findings were also supported by Thompson and Opdebeek.

This study limited to a description of relations between cervical posture and sEMG activity of head and neck muscles in Caucasian adult women. Because of the transversal structure of the study, no conclusions were possible concerning the mechanism.

Future *transversal studies* of a different population (samples including men, skeletal III class, etc) and future *longitudinal studies* are required to analyze the detailed nature of the mechanism at work. They should be directed to investigate the extent of environmental and genotype influences on cervical vertebral growth. A clearer appreciation of these determinants will clarify the complex inter-relationship between form and function in craniofacial and cervical vertebrae morphogenesis.

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Table 1
Descriptive statistic of postural variables on Lateral skull radiographs

Postural variables on Lateral skull radiographs	Mean	SD
OPT/SN	90.09	20.23
CVT/SN	100.64	9.60
CVT/SN	100.64	9.60
EVT/SN	112.92	9.42
EVT/SN	112.92	9.42
OPT/FH	79.61	14.98
OPT/FH	79.61	14.98
CVT/FH	89.41	8.62
CVT/FH	89.41	8.62
EVT/FH	103.07	9.30
EVT/FH	103.07	9.30
OPT/pns-ans	85.3	9.46
OPT/pns-ans	85.3	9.46
CVT/pns-ans	91.9	9.31
EVT/pns-ans	105.21	10.01
OPT/GoGn	60.77	9.09
OPT/GoGn	60.77	9.09
CVT/GoGn	67.49	8.86
CVT/GoGn	67.49	8.86
EVT/GoGn	80.91	9.87
EVT/GoGn	80.91	9.87
OPT/GoRasc	7.72	6.51
OPT/GoRasc	7.72	6.51
CVT/GoRasc	11.92	8.29
CVT/GoRasc	11.92	8.29
EVT/GoRasc	24.62	8.50
EVT/GoRasc	24.62	8.50
CVT/EVT	13.87	7.92
CVT/EVT	13.87	7.92

Table 2
sEMG Activities (μV/sa) of the Different Monitored Muscles at the Mandibular Rest Position (Scan 9) and in the Maximal Voluntary Clenching (Scan 11)

Muscles	Mean	SD
Maximal Voluntary Clenching		
Anterior Temporalis	183,4960	75,1893
Masseter	193,9555	130,7081
Masseter	193,9555	130,7081
Digastric	9,8310	6,1189
Digastric	9,8310	6,1189
Sternocleidomastoid	17,2030	6,1708
Sternocleidomastoid	17,2030	6,1708
Upper Trapezius	30,0818	12,4216
Upper Trapezius	30,0818	12,4216
Lower Trapezius	23,1218	12,5459
Lower Trapezius	23,1218	12,5459
Cervicals	14,3515	6,6171
Cervicals	14,3515	6,6171
Mandibular Rest Position		
Mandibular Rest Position		
Anterior Temporalis	14,5555	4,1284
Masseter	14,5788	10,9367
Masseter	14,5788	10,9367
Digastric	8,1785	3,8895
Digastric	8,1785	3,8895
Sternocleidomastoid	11,9855	3,7753
Sternocleidomastoid	11,9855	3,7753
Upper Trapezius	17,3327	4,5497
Upper Trapezius	17,3327	4,5497
Lower Trapezius	18,8258	8,4505
Lower Trapezius	18,8258	8,4505
Cervicals	15,6400	7,4791
Cervicals	15,6400	7,4791

Table 3

Variable 1	Variable 2	Pearson Correlation	Sig. (2-tailed)
Maximal Voluntary Clenching			
Masseter	CVT/SN	-0.416	P = 0.008
Masseter	CVT/FH	-0.564	P = 0.000

Masseter	OPT/pns-ans	-0.518	P = 0.001
Masseter	CVT/pns-ans	-0.480	P = 0.002
Masseter	OPT/GoGn	-0.529	P = 0.000
Masseter	CVT/GoGn	-0.409	P = 0.009
Masseter	CVT/GoRasc	-0.576	P = 0.000
Masseter	CVT/EVT	0.474	P = 0.002
Digastric	OPT/GoRasc	0.451	P = 0.003
Lower Trapezius	OPT/FH	-0.576	P = 0.000
Mandibular Rest Position			
Anterior Temporalis	OPT/SN	0.409	P = 0.009
Lower Trapezius	CVT/EVT	0.414	P = 0.008

The NEW-study. Perceived work demands, felt stress, and musculoskeletal symptoms.

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Abstract *The aim of this cross-sectional study was to test a structural model of the relationship between perceived work demands, felt stress, and musculoskeletal neck/shoulder symptoms. Felt stress was hypothesized to be a mediating variable in this relationship. The study was based on a questionnaire survey among Danish, Dutch, Swedish, and Swiss female computer users aged 45 or older (n = 148). The proposed structural model was tested using structural equation modelling. The results indicate that the relationship between perceived work demands and musculoskeletal neck/ shoulder symptoms is fully mediated by felt stress.*

1. Aim

The aim of the present study was to test a structural model of the relationship between an aspect of the psychosocial work environment – perceived work demands – and musculoskeletal neck/shoulder symptoms, with felt stress as a proposed mediating variable. The model was tested using structural equation modelling using Mplus version 2.0 1.

2. Methods

The present cross-sectional study was part of the NEW-study, a European case-control study.

2.1. Participants

The study sample consisted of 148 female European (Danish, Dutch, Swedish and Swiss) computer users aged 45 or older, with at least 5 years seniority, and working at least 20 hours/week. A case was defined as having had symptoms in the neck and/or shoulder for more than 30 days during the past 12 months and a control was defined as having had no symptoms at all in the neck and shoulder, or symptoms for up to 7 days during the past 12 months.

2.2. Measures

The perceived work demands were assessed using the short version of the Copenhagen Psychosocial Questionnaire (COPSOQ) 2, felt stress was assessed using the two-dimensional mood adjective checklist 3, and musculoskeletal symptoms were assessed using the general Nordic Musculoskeletal Questionnaire (NMQ) 4.

3. Results

The proposed model showed good fit to the data. There was an indirect positive effect of work demands on musculoskeletal neck/ shoulder symptoms through the intermediary of felt stress. As the direct effect of work demands on symptoms when the mediating variable was included in the equation was non-significant the results indicate complete mediation, which means that all of the effect of perceived work demands on musculoskeletal symptoms could be attributed to the stress mechanism.

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Neuromuscular assessment of fatigue of low back muscles in elderly nurses

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Abstract *– The first aim of this study was to compare surface EMG variables in a group of elderly female nurses suffering from low back pain and in a control group. The second aim was to assess the repeatability of surface EMG on low back muscles of elderly nurses. Eleven cases suffering of low back pain and 9 healthy controls participated to the experiment. EMG signals were detected bilaterally from the longissimus dorsi and multifidus muscles with adhesive electrode arrays. The EMG measures were repeatable as indicated by the absence of differences between measures obtained in different trials. The two subject groups showed similar MNF slope values, despite a shorter endurance time for the cases.. The amplitude patterns suggested different activation strategies in the two groups, probably due to the muscle pain.*

1. Introduction

Surface EMG techniques have been extensively applied to the analysis of low back muscles, both in healthy subjects and low back pain patients [1,3,5-7]. Surface EMG analysis has shown promising for objective fatigue assessment and has been applied in many rehabilitation fields (e.g., for classifying healthy subjects and low back pain patients) [1,7]. Moreover, myoelectric manifestations of muscle fatigue were shown to be predictor of the trunk extensor endurance time, thus reflecting mechanical fatigue [4]. However, despite the numerous works which showed potential usefulness of the surface EMG approach in clinical routine, this technique still presents limitations related to repeatability, sensitivity to electrode location, and type of contraction performed. The aim of this study was to compare surface EMG variables in a group of elderly female nurses suffering from low back pain and in a control group. Moreover this study focused on the repeatability of the EMG signals detected with linear electrode arrays on low back muscles.

2. Methods

2.1 Subjects

Cases (11 subjects, age, mean ± std. dev., 50.1 ± 5.5 years, height, 157.1 ± 7.8 cm, weight, 64.2 ± 17.6 kg) were nurses, who, at the time of the study, had worked at least 20 hours/week on the same job for at least 5 years and reported trouble (ache, pain, discomfort) in low-back for more than 30 days during the last year. Nine control subjects were selected to match the cases for age and body mass index (age, mean ± std. dev., 50.1 ± 3.8 years, height, 160.3 ± 6.7 cm, weight, 64.7 ± 11.7 kg).

2.2 EMG detection

Multichannel surface EMG signals were detected bilaterally from Longissimus Dorsi (LD) and Multifidus (MF) muscles, at the level of the lumbar vertebrae L3 and L5 respectively, using four 4-electrode, 10 mm interelectrode distance adhesive arrays aligned with muscle fiber direction (FIG. 1) [2].

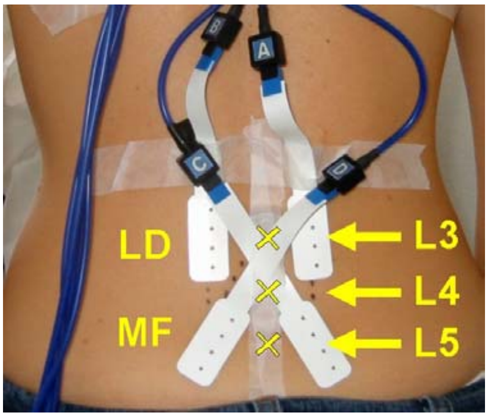

FIG. 1: Detail of the experimantal set-up for surface EMG signal detection. Four adhesive electrode arrays (four electrodes, 10 mm inter-electrode distance) were located on the Longissimus Dorsi (LD) and Multifidus (MF) muscles, at the level of the lumbar vertebrae L3 and L5 respectively, aligned with the fiber direction.

The single differential detection technique was used. The subject was held in horizontal position, with trunk and arm leaning off a bed (FIG. 2), and asked to perform a sustained contraction of the low back (Sörensen test [1]) while keeping the trunk horizontal until exhaustion (endurance time reached).

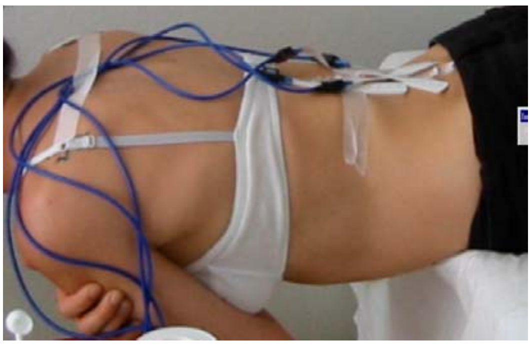

FIG. 2: Experimental set-up for the acquisition of EMG signals from the Longissimus Dorsi and Multifidus muscles during a Sörensen Test.

At the end of the test, the subject was asked to provide an indication of fatigue according to the Borg scale, and an indication of pain in the low back region on a VAS scale. The test was repeated on three non-consecutive days to assess repeatability. For each array, the three single differential signals were summed to obtain a bipolar recording acquired with 30 mm inter-electrode distance. Average rectified value (ARV) and mean power spectral frequency (MNF) were computed on 1 s long signal epochs (FIG. 3). Linear regression analysis was performed to compute ARV and MNF initial values and slopes.

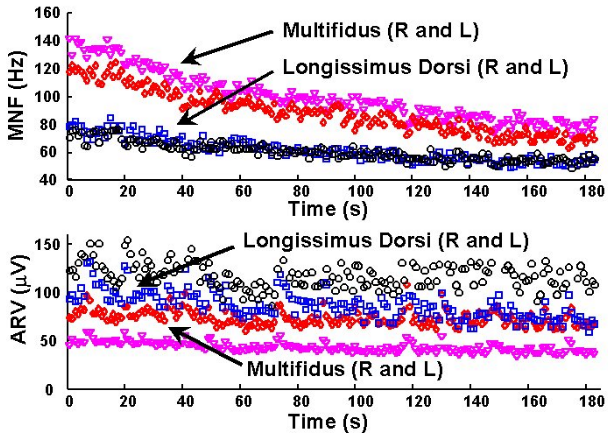

FIG. 3 : Example of EMG analysis on the 30 mm inter-electrode distance single differential signals. The four muscles are represented with different symbols (LD right, left: o, □; MF right, left: ◊, Δ). Note the larger values of MNF for the Multifidus muscle with respect to the other muscles.

2.3 Statistical analysis

Data were analysed using two-way repeated measures analysis of variance (ANOVA), followed by post-hoc Student-Newman-Keuls (SNK) pair-wise comparisons, when required. Statistical significance was set to $P < 0.05$. Statistical significance was set to $P < 0.05$. Data are presented as mean ± standard error (SE).

3. Results

Endurance time was significantly shorter and the pain score higher for cases with respect to the controls (Fig. 4), while no differences were found for the Borg scale.

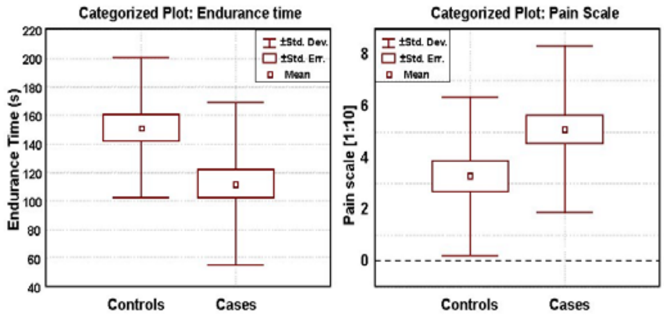

FIG. 4: Descriptive graphs for the Endurance time (left) and the Pain scale (right) for the cases and the controls. Note the lower endurance time and the higher Pain scores for the cases.

ARV initial value and slope were significantly different in the two groups, with lower amplitude for the cases and a larger ARV decrease for the control group (Fig 5). No significant difference was observed in the comparison of signal variables in the three subsequent trials.

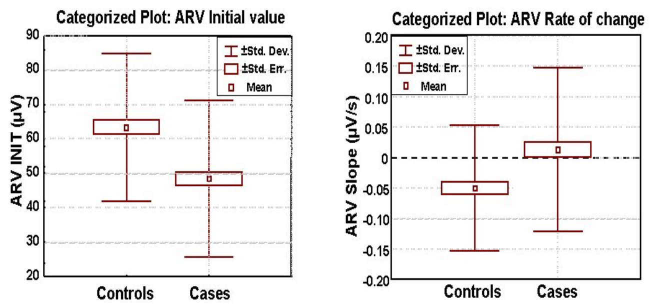

FIG. 5: Descriptive graphs for the ARV initial value (left) and the ARV rate of change (right) for cases and controls. Note the lower initial values and the negative rate of change for the cases.

4. Discussion

The EMG measures were repeatable as indicated by the absence of differences between measures obtained in different trials. The two subject groups showed similar MNF slope values, despite a shorter endurance time for the cases. This can be explained by the subjective pain scores, which suggest that the cases stopped the contraction because of low back pain rather than for fatigue. The amplitude patterns suggest different activation strategies in the two groups, probably due to the discomfort caused by muscle pain.

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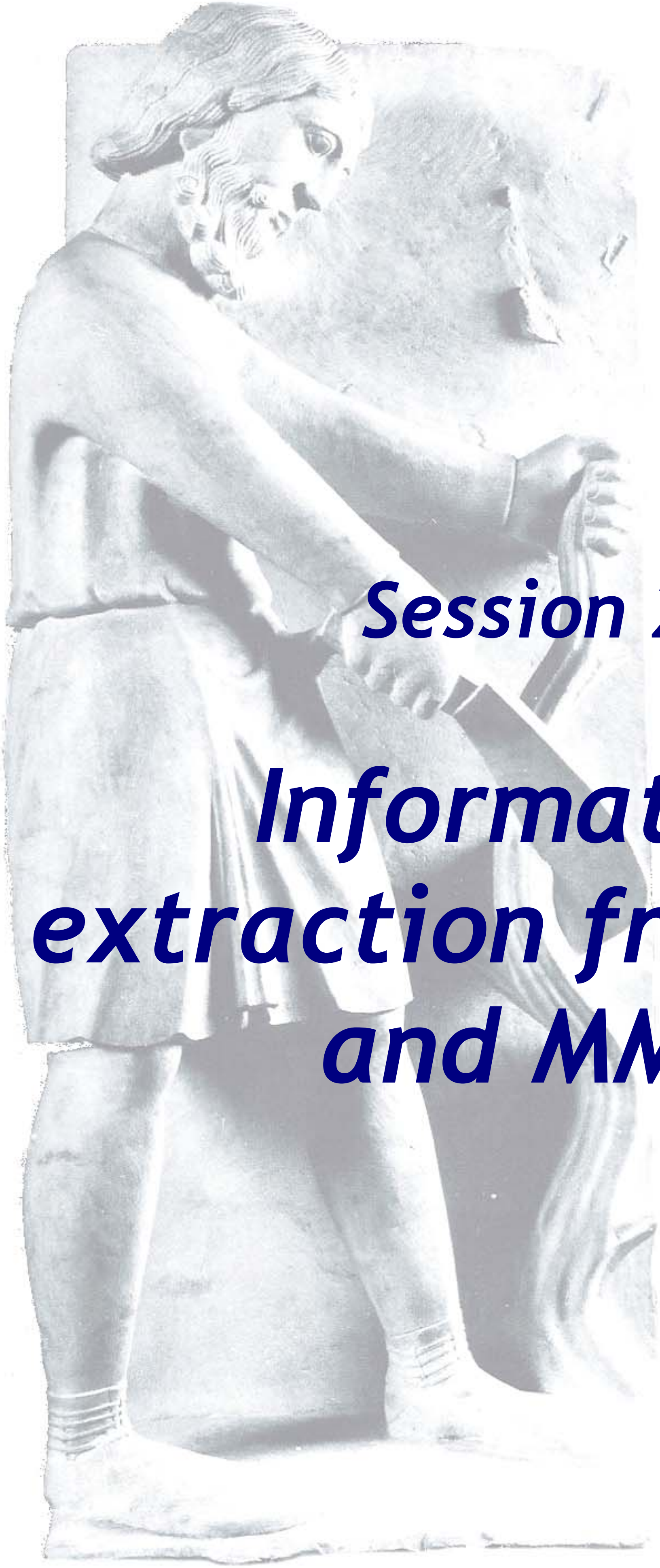

Session 2:

Information extraction from EMG and MMG

What information can be extracted from the surface EMG

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Abstract *– This lecture presents an overview of the type of information that can be extracted from the surface EMG. The factors affecting surface EMG signal features include both physiological and non-physiological mechanisms. The effect of the non-physiological factors should be properly recognized and minimized as a basis for surface EMG interpretation. Techniques based on the global surface EMG variables or on the extraction of variables related to single motor units are listed and briefly discussed.*

1. Introduction

The surface EMG comprises the sum of the electrical contributions of the active motor units (MUs) as detected by electrodes placed on the skin overlying the muscle. Thus it carries information on both the peripheral and the control properties of the neuromuscular system. The extraction of this information is, however, complex due to the many factors that affect the signal features. The aim of this work is to present the factors most affecting the surface EMG features, the type of information that can be extracted from this signal, and the way it is extracted. Part of this work is taken from [1].

2. Factors that influence the surface EMG

The surface EMG features depend on both physiological and "non-physiological" factors. Table I reports the main of these factors. In some conditions, non-physiological factors may have a more relevant influence on the signal features than the physiological mechanisms under study. The situation is significantly worsened by the fact that the influence of some of the factors of the generation surface EMG system is not intuitive. The use of models of signal generation is in this case fundamental for proper interpretation of the EMG characteristics.

An example of counterintuitive phenomena in surface EMG interpretation is the effect of crosstalk among nearby muscles. The simulation of crosstalk signals revealed that past approaches for its identification or reduction may be ineffective or misleading [1][2][3].

3. Information that can be extracted from surface EMG

The extraction of information from the surface EMG is based on the minimization of the influence of the non-physiological factors listed in Table 1, while selectively understanding the physiological mechanisms (possibly one at a time). Due to the large number of parameters which influence the surface EMG, often it is not possible to selectively associate a single physiological mechanism with each variable but rather the variables extracted from the signal are affected by many mechanisms.

In some cases, the influence of the non-physiological factors may be so large to completely mask any information related to the physiology under study. As an example, variations of EMG amplitude in a dynamic task may be due to a relative shift of the muscle fibers with respect to the detecting electrodes rather than to a variation in the intensity of muscle activation [4].

To reduce the number of parameters, often the surface recording is constrained by specific experimental requirements, that is the case of isometric contractions.

TAB. 1 : Factors that influence the surface EMG. (Reproduced from [1]).

Factors that influence the surface EMG		
"Non-physiological"	**Anatomical**	Shape of the volume conductor
		Thickness of the subcutaneous tissue layers
		Tissue in-homogeneities
		Distribution of the MU territories in the muscle
		Size of the motor unit territories
		Distribution and number of fibers in the motor unit territories
		Length of the fibers
		Spread of the end-plates and tendon junctions within the motor units
		Spread of the innervation zones and tendon regions among motor units
		Presence of more than one pinnation angle
	Detection system	Skin-electrode contact (impedance, noise)
		Spatial filter for signal detection
		Inter-electrode distance
		Electrode size and shape
		Inclination of the detection system relative to muscle fiber orientation
		Location of the electrodes over the muscle
	Geometrical	Muscle fiber shortening
		Shift of the muscle relative to the detection system
	Physical	Conductivities of the tissues
		Amount of crosstalk from nearby muscles
Physiological	**Fiber membrane properties**	Average muscle fiber conduction velocity
		Distribution of motor unit conduction velocities
		Distribution of conduction velocities of the fibers within the motor units
		Shape of the intracellular action potentials
	Motor unit properties	Number of recruited motor units
		Distribution of motor unit discharge rates
		Statistics and coefficient of variation for discharge rate
		Motor unit synchronization

Surface EMG can provide information by the analysis of its "global" characteristics (such as amplitude or spectral content) or by the identification, from the interference EMG signal, of the contributions of single MUs active during the contraction. Table 2 reports the main information that can be obtained by the global and single MU analysis of the surface EMG.

3.1 Global EMG variables

Global EMG variables describe macro features of the signal, such as its amplitude or frequency content. They represent the electrical outcome of the MUs within the detection volume and may be related to the underlying physiological processes.

A number of variables computed from the interference surface EMG signal have been proposed in the literature. Amplitude of the signal can be estimated by a standard demodulation, smoothing, and relinearization scheme which may be preceded by signal whitening. Demodulation rectifies the EMG and then raises the result to a power (e.g., 1 for average rectified value, ARV, or 2 for root mean square value, RMS). Smoothing filters the signal whereas relinearization inverts the power law applied during the demodulation stage, returning the signal to units of EMG amplitude.

Spectral analysis of surface EMG signals has been extensively applied for studying myoelectric manifestations of muscle fatigue or for inferring changes in MU recruitment. Characteristic spectral frequencies can be computed by classic periodogram and autoregressive based approaches or by more advanced methods based, e.g., on Cohen's class time-frequency distributions and wavelets. The latter techniques have been used in recent years for dynamic or isometric variable force contractions. From the signal theory point of view, these methods may be better than classic ones in case of highly non-stationary signals. However, the most important limitations of spectral analysis for muscle assessment are intrinsic in the spectral properties of the surface EMG signals. Thus, they depend on the signal generation system rather than on the method applied for signal analysis.

Other global variables extracted from surface EMG signals are the zero crossing rate, the spike properties, variables extracted from non-linear surface EMG analysis, the cross-correlation between surface EMG signals. In addition, average CV may be estimated from the interference surface EMG signals if there are at least two detection points along the direction of propagation of the muscle fibers.

The global surface EMG variables may easily be subject to erroneous interpretation since they are based on indirect relations between signal features and the underlying physiological mechanisms. As an example, methods based on the analysis of the characteristic spectral frequency trends for assessing the end point of motor unit recruitment are based on assumptions that are not valid when the active motor unit population changes over time.

3.2 Single MU analysis from the surface EMG

In recent years, techniques for the analysis of single MU properties from surface EMG signals, at least in specific conditions, have been developed. The common strategy was to increase the amount of available information from surface EMG and to improve the spatial selectivity of the recording (see also D. Farina, E. Schulte, "NEW advances in surface EMG modelling and processing" in this Proceedings). These approaches are thus based on the increase of the number of electrodes placed over the muscle with respect to the classic bipolar recordings.

Multi-channel surface EMG allows the extraction of motor unit anatomical and physiological properties, such as the location of the innervation zones, the fiber length, the muscle fiber conduction velocity, and the time of occurrences of action potentials. Since single MUs are analyzed, most of the critical issues related to a global surface EMG analysis are not of concern. A direct observation of the MU control strategies is indeed performed which avoids the limitations due to an inverse modelling approach, typical of the global surface EMG analysis.

4. Conclusions

Interpretation of surface EMG features for extracting information on the underlying physiology is a complex task, with many issues still to be solved. However, when standardized techniques for signal detection and processing are applied, valuable information, not gained with other techniques, on the neuromuscular system properties can be obtained. The European project "Neuromuscular Assessment in the Elderly Worker" (NEW) contributed to the identification of problems related to EMG interpretation for a better comparison of results among different studies [1].

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TAB. 2 : Information that can be extracted from the surface EMG.

Type of information	Most common methods	Requested detection system (minimum)	Notes
Intervals of muscle activity	Threshold/double threshold/time frequency or wavelet based detectors	Single channel bipolar	Crosstalk may significantly affect this information
Intensity of muscle activation	Envelope/average rectified value/root mean square value	Single channel bipolar	It may be critical in case of non-standardized electrode placements. In dynamic conditions it may be largely affected by the muscle shift under the electrodes
Average muscle fiber conduction velocity of the active motor units	Various time and frequency domain estimators of delay between travelling signals	Two channels	If more EMG signals are available, this measure shows lower variance and higher repeatability with respect to the use of only two channels. The estimate may be affected by geometrical factors.
Myoelectric manifestations of muscle fatigue	Spectral and amplitude analysis	Single channel bipolar	The interpretation of fatigue indexes may be very complex since they depend on both central and peripheral properties of the neuromuscular system
Degree of synchronization of motor units within the same muscle	Recurrence plot analysis	Single channel bipolar	The recurrence analysis parameters are affected by other physiological factors in addition to the degree of synchronization
Synchronization among motor units of two muscles	Crosscorrelation between signals generated by two muscles	Single channel bipolar for each muscle	This measure may be significantly affected by anatomical properties (such as the thickness of the subcutaneous layers). There are no simulation studies showing the limits beyond which this method should not be applied
Single motor unit conduction velocity	Delay estimation from signals detected along the muscle fiber direction	Two channels	Conduction velocity is a size principle parameter, being related to the twitch force of single motor units. The estimate may be affected by geometrical factors
Distribution of motor unit conduction velocity	Deconvolution/extraction of information at the single motor unit level by surface EMG decomposition	Two channels	The methods available should still be significantly improved for clinical applications
Motor unit firing rate	Detection of low frequency peaks in the power spectrum/signal decomposition	Single channel bipolar/multi-channel	Extracted only in very specific conditions
Types of recruited motor units	Spectral analysis/average conduction velocity	Single channel bipolar	The use of absolute characteristic frequency values is critical both from the physiological and the methodological point of view
Muscle force	Estimation of EMG amplitude	Single channel bipolar	The relation between force and amplitude is not a general property of specific muscles, thus it should be calibrated on a subject by subject and muscle by muscle basis
End of recruitment point	Analysis of trends of characteristic frequencies with increasing force	Single channel bipolar	Unreliable technique sometimes used in the literature
Location of the innervation zone(s) of the active motor units	Analysis of the spatio-temporal distribution of surface potentials	Multi-channel	Located with good resolution
Length of the fibers	Analysis of multi-channel signals	Multi-channel	Sometimes it is difficult to exactly locate the point of action potential extinction
Location of the tendon endings	Analysis of multi-channel signals	Multi-channel	Sometimes it is difficult to exactly locate the point of action potential extinction

What information is contained in the mechanomyogram (MMG) and how can it be extracted?

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Abstract– *This paper deals with the description of the evidence of single motor unit contribution to the surface mechanomyogram (MMG). Moreover it will be demonstrated how the MMG time and frequency domain parameters may reflect the motor unit activation pattern during voluntary contraction.*

1. Introduction

The muscle surface displacement of each recruited motor unit (MU) can be retrieved in the surface mechanomyogram (MMG). The spike triggered averaging technique, based on motor unit action potential amplitude recognition, was used to extract from raw MMG the contribution of single MUs [1, 2]. MUs can be classified by means of the analysis of the single twitch force response. **1)** On this basis two partners of the NEW project (LISiN and UNIBS) used the single MU action potentials, identified by the decomposition of surface EMG, as triggering events on force and MMG signals. This non invasive method allowed to prove that MMG is an interferential signal encompassing the single MUs contributions opening new perspectives in the assessment of single MU electromechanical properties.

2) Given that in MMG the activity of the recruited MUs is reflected the NEW partner UNIBS studied the possibility to follow the motor units activation strategy in different physiological situations.

2. Methods

2.1. In ten male subjects (25-30 years old) the first dorsal interosseous (FDI) and abductor digiti minimi (ADM) muscles were investigated at 2% and 5% of the maximal voluntary contraction (MVC) as well as during the activation of selected single MU driven by surface MU action potential visual feed-back.

2.2. In ten male subjects, 25 - 40 years old, EMG and MMG were detected from the belly of the biceps brachii by an integrated probe developed for the NEW project. Test exercise: isometric ramp from 0 to 90% MVC in 7.5 s. Fatiguing exercise: intermittent 50% MVC (6 s on – 3 s off) until the target was not reached. At that moment a new ramp, in which the 90% MVC corresponded to the 45% MVC of the unfatigued muscle, was administered. On .5 s EMG and MMG time windows the RMS and mean frequency (MF) of the signals spectra (estimated by FFT) were calculated.

3. Results

3.1. <u>At 5 % MVC</u>, the mean (± standard error) of single MU MMG peak-to-peak value was 11.0 ± 1.8 mm/s^2 (N = 17) and 32.3 ± 6.5 mm/s^2 (N = 20) for FDI and ADM muscles, respectively. The peak of the twitch force was, at the same contraction level, 7.41 ± 1.34 mN and 14.42 ± 2.92 mN for the FDI and ADM muscles, respectively. MUs activated by visual feedback. The peak-to-peak value of MMG was significantly different for the same MUs at 2% or 5% MVC. e.g. in FDI muscle, the MMG peak-to-peak value of individual MUs was 11.8 ± 3.8 mm/s^2 at 2% MVC and 21.5 ± 7.8 mm/s^2 when recruited by visual feedback.

3.2. MMG-RMS vs %MVC: at fatigue the MMG-RMS did not present the well known increment as the effort level increased followed by a clear reduction at near-maximal contraction levels. MMG-MF vs %MVC: compared to fresh muscle the fatigued biceps brachii showed a MF trend significantly shifted toward lower values and the steeper MF increment from 65 to 85% MVC was not present.The EMG-RMS vs % MVC did not changed significantly while the EMG-MF was shifted towards lower values.

4. Discussion

4.1 The difference in the peak-to-peak value of single MU MMG during visual feed-back of the MU action potential or during sustained low level contractions indicates a non-linear summation of the single MU contributions as already hypothesised [3]. The method proposed is promising for the non-invasive assessment of single MU properties and for the modelling of the MMG signal generation mechanisms.

4.2. The alteration in the MMG and EMG parameters vs %MVC relationships at fatigue have been discussed taking into account the possible influence of the changes in MUs recruitment threshold, in their firing rate and in time correlation between MUs discharges after fatiguing exercise. As a conclusion it seems that in fatigued muscle the impossibility to recruit fast, but more fatigable MUs, and the lowering of the global MUs firing rate could be reflected in the MMG and EMG properties during a short term isometric force ramp [5].

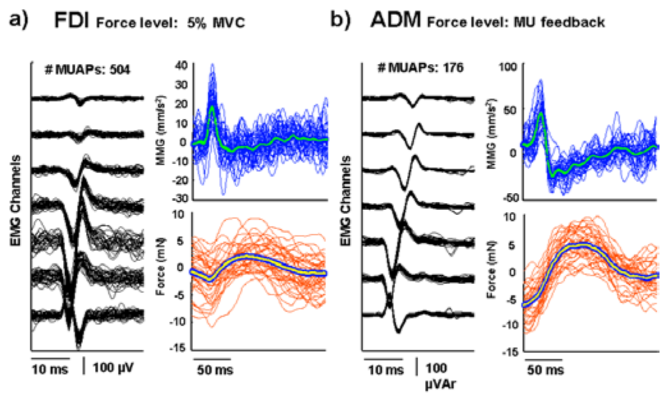

Figure 1. Single MU MMG and force signals. Spike triggered averaging on 60 s time window. Note the different MMG and force twitch response of the two muscles (redrawn from [4]).

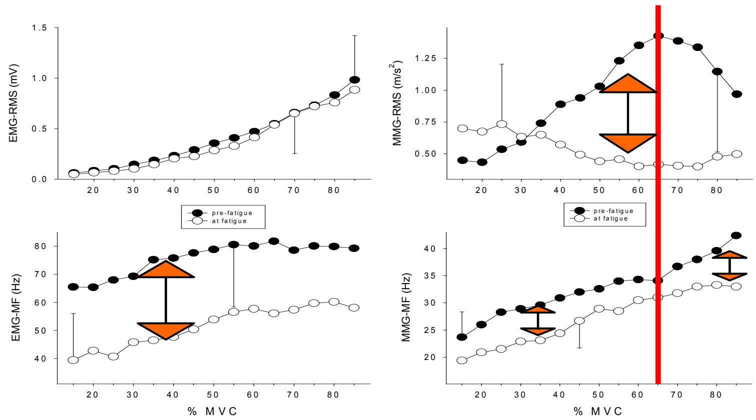

Figure 2. EMG and MMG RMS and MF vs %MVC in fresh (•) and fatigued (o) muscle. The difference in the parameters dynamics may be due to the impossibility to recruit fast twitch MUs.

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Influence of muscle size and force generating capacity on EMG response to low-level isometric contractions

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Abstract *– Surface electromyograms were recorded from muscles of different size (quadriceps femoris; biceps brachii; brachioradialis; abductor digiti minimi - manus) during knee and elbow flexion or finger abduction, respectively, from in total 34 subjects. In all cases a low-level isometric contraction at 15 % of the maximum voluntary contraction force of the respective muscle was performed for at least 2 min. The individual EMG response for the various muscles and persons differs widely. A comparison for the four muscles under test reveals different behaviour of the time characteristics for the EMG amplitude and frequency spectrum with respect to muscle size. For the larger muscles a change in the EMG amplitude was predominant, whereas for the smaller muscles a change was observed mainly for the Median Frequency. Furthermore an influence of the force generating capacity of the individual muscle on the EMG response was found. For the interpretation the distribution of fiber types within the muscles and the influence of recruitment, rate coding, slowing of action potential propagation and synchronisation during fatiguing contractions is discussed.*

1. Introduction

One of the most important applications of electromyography in occupational physiology deals with the determination of muscular fatigue. For this purpose commonly the temporal behaviour of electromyograms (EMG) is analysed and a time-related increase in the EMG amplitude and a shift in the EMG spectrum towards lower values are used as indicators of fatigue. During occupational work comprising high muscular forces usually a clear change in the above mentioned fatigue indicators is observed; for work with low-level forces, however, non-uniform results are often obtained for various muscles and different persons.

Since in occupational work - in particular at modern workplaces equipped with computer terminals - low muscle forces are predominant and it has to be assumed that, nevertheless, such activities contribute to the development of muscular complaints, the temporal behaviour of EMGs during low force production is worth to be investigated. Therefore in this study the temporal behaviour of the myoelectrical activity of various muscles was analysed during low-level isometric contractions.

2. Methods

In previous electromyographical pilot studies on various muscles (Luttmann et al. 1997, 1998) it was suggested that the EMG response to low-level contractions varies with the size and the force generating capacity of a muscle. In order to prove this hypothesis EMGs were recorded from muscles of different size (quadriceps femoris; biceps brachii; brachioradialis; abductor digiti minimi - manus) during knee extension, elbow flexion or finger abduction, respectively, from in total 34 subjects. In all cases a low-level isometric contraction at 15 % of the force at maximum voluntary contraction (max. F) was performed for at least 2 min. Temporal changes of the EMG amplitude and spectral distribution were determined by calculating regression functions for various amplitude and frequency parameters.

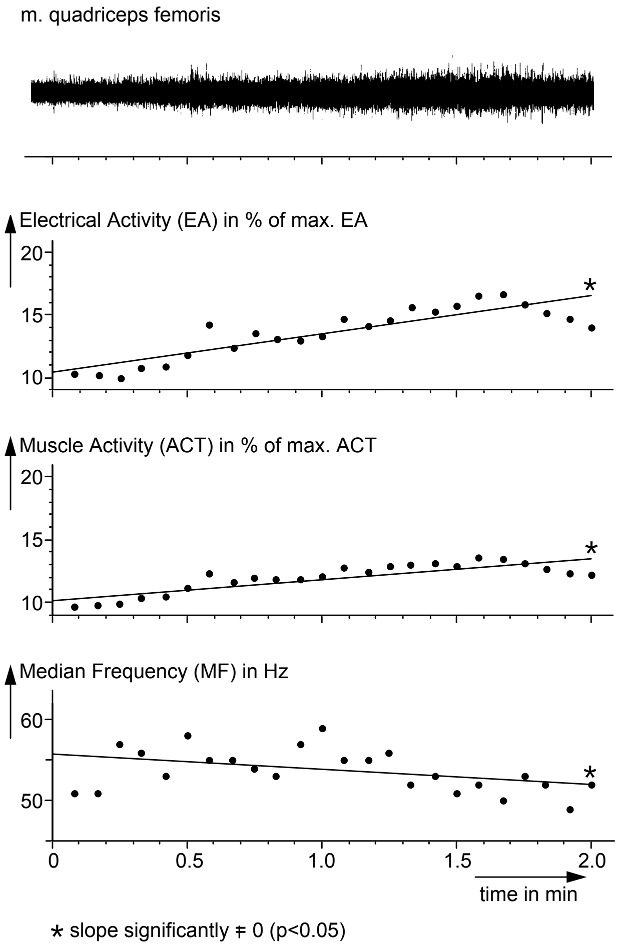

FIG. 1: Original recording of the raw EMG of the right quadriceps femoris during loading with 15% of the maximum voluntary force. In the lower time courses several amplitude and spectral indicators are provided. Each dot refers to the mean value in a 5-s period. The straight lines indicate regression lines. Max. EA and max. ACT refer to EA and ACT during maximum voluntary contraction.

EMG amplitude is quantified using the rectified and averaged EMG (Electrical Activity EA) and the "Muscular Activity" (ACT) defined by Spaepen et al. (1987). The spectral distribution is characterized using the Median Frequency (MF).

3. Results

In figure 1 an example for the EMG time course of one of the muscles under test is demonstrated. In the upper trace the original EMG of the thigh muscle is shown. In spite of the low force production of 15% of max. F in the example shown here a clear increase in the amplitude was observed. Accordingly for the amplitude indicators EA and ACT an increase over time was found. In the lowest diagram the time course of MF is shown indicating a decrease over time, on average.

Similar results were observed on average for all of the muscles under test. However, due to the low contraction force of 15 % of max. F the slopes of EA, ACT and MF for the different persons and muscles vary in a wide range. In figure 2 averaged time functions for the four muscles under test are shown for the amplitude and spectral indicators. All characteristics are normalized with respect to the corresponding initial value at t=0.

A comparison of the data for the four muscles under test reveals different behaviour of the time characteristics for the EMG amplitude and frequency spectrum. For the largest muscle under test - the thigh muscle – EA and ACT increase clearly over time, whereas the Median Frequency is changing to a small extent, only. By contrast, for the smallest muscle under test - the abductor of the

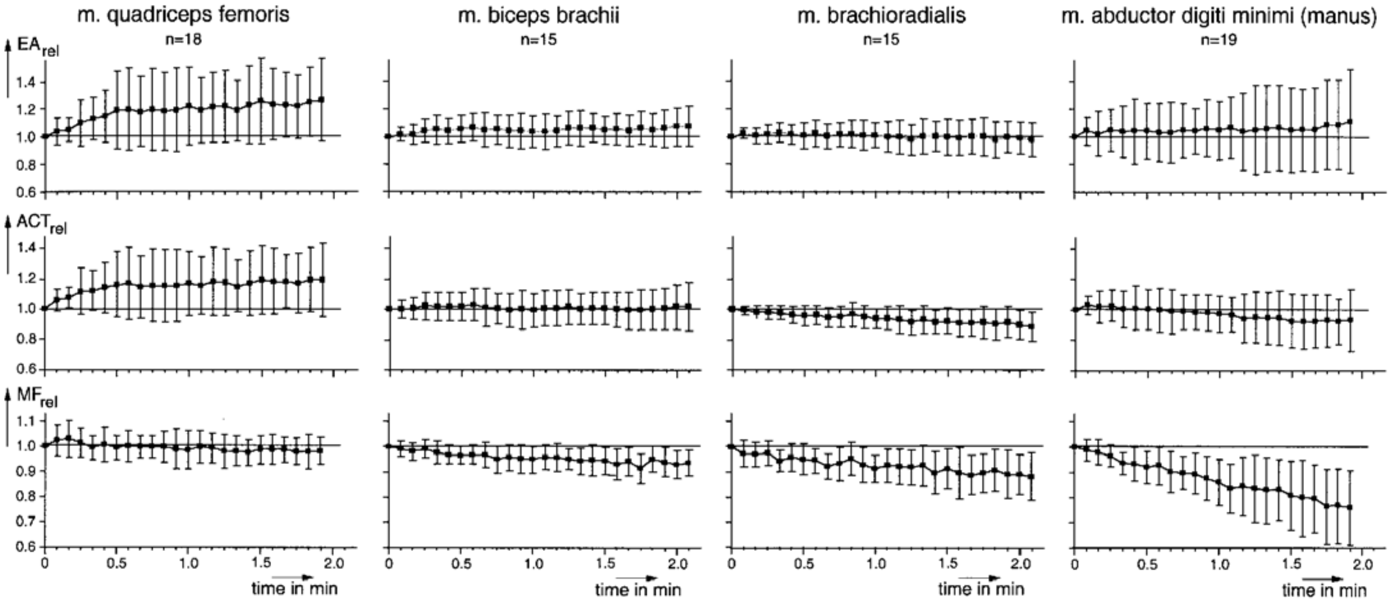

FIG. 2: Averaged time course and standard deviation for the relative change in the Electrical Activity (EA), Muscular Activity (ACT), and the Median Frequency (MF) for the right m. quadriceps femoris, right m. biceps brachii, right m. brachioradialis, and the m. abductor digiti minimi of the right hand during constant loading with 15 % of the maximum voluntary contraction force (relative values = actual value at a certain time t related to the initial value at t = 0)

fifth finger – a small time-related change is found for the averaged amplitude characteristics EA and ACT, whereas MF clearly decreases with increasing time. The time responses for the biceps and brachioradialis range between the two extreme situations found for the leg and finger muscles.

The high standard deviations shown in figure 2 confirm the aforementioned large variations in the individual time responses during the low-level contractions of 15 % of max. F.

In previous studies (Luttmann et al. 1997, 1998) it was concluded that besides the muscle size, the individual force generating capacity may influence the EMG response. To prove this suggestion for all muscles under test two subgroups were formed depending on max. F. For the subgroups different EMG responses were found: The increase in EA is higher for the group with lower max. F, in particular for the thigh muscle. For MF the highest temporal change was found in the finger muscle for the group with the lower max. F.

4. Discussion

The increase in the EMG amplitude during fatiguing contractions mainly indicates a rise in the number of action potentials per unit of time caused by an increasing firing rate and the recruitment of additional muscle fibers. This additional activation is needed to hold the force constant, because, as a result of fatigue, action potentials become less effective due to a diminished electromechanical coupling and a reduction in the force produced per action potential.

The shift in the EMG spectrum to lower frequencies is commonly interpreted as a central nervous synchronisation of motor unit action potentials and a slowing in the action potential propagation along the muscle fibers. Slowing is assumed to be caused by a change in the ional composition in the muscle tissue, results in a "broadening" of the biphasic action potentials measured with bipolar surface electrodes, and causes the spectral shift to lower values.

In this study two measures of the EMG amplitude EA and ACT are used. EA increases with recruitment and rate coding as well as with slowing and synchronisation, whereas ACT increases with recruitment and rate coding and decreases in particular with slowing. (Hermans 1996, Luttmann 2001). Therefore using both measures can help in the discrimination between the different types of changes in the electromyograms. It is concluded that recruitment and rate coding mainly occurs in the EMG of the larger muscles whereas slowing is more distinct in particular for the finger abductor.

The amplitude increase indicates a rise in the number of action potentials per unit of time and reflects the fatigue-induced decrease in the force produced per action potential. Insofar it is directly related to the mechanical fatigue phenomenon. Therefore, strictly speaking, the increase in the EMG amplitude is a more powerful indicator of fatigue than the spectral change, which represents an "accompanying phenomenon".

For the finger muscle the amplitude change is smallest and according to the aforementioned interpretation it can be concluded that the fatigability of this muscle should be relatively low. This suggestion correlates with the fiber composition. According to Johnson et al. (1973) the proportion of type I fibers amount to about 52% for the finger abductor in comparison to 40 to 44% for the other muscles under test.

In conclusion, in the larger muscles the amplitude increase caused by recruitment and increase in firing rate is more distinct than in the smaller ones. In the smaller muscles, spectral change resulting from slowing of action potential propagation of motor units is more dominant.

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Effects of bipolar electrode configurations on EMG based estimation of muscle force

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Abstract *– Limited experimental evidence exists on how electrode configurations affect the EMG signal. High-Density EMG array records many monopolar signals and thus allows constructing different bipolar electrode configurations. The aim of this experimental study is to investigate the effect of different bipolar electrode configurations on EMG based estimation of muscle force. We studied electrode size, inter electrode distance (IED), collection surface and surface density of the electrodes over the muscle belly. Eleven healthy subjects participated in the experiment and performed isometric right arm extensions at different contraction levels and different elbow angles. Surface EMG of the triceps brachii was collected with an active High-Density EMG array and synchronized with the measured extension force of the elbow. The quality of the EMG based estimation of muscle force was expressed as the root mean square difference (RMSD) between normalized force and EMG. The collection surface appeared to be the only important electrode configuration in the quality of muscle force estimation. Larger collection surfaces enhanced the quality of this estimation by about 25%. The results of this paper suggest that conventional bipolar electrodes allow the same quality of force estimation, when using multiple pairs of electrodes in uniformly distributed configuration over the muscle belly.*

1. Introduction

In EMG based estimation of muscle force various electrode configurations are applied. De Luca described the electrode configuration as an ‘extrinsic causative factor’ in the estimation of the EMG amplitude (De Luca, 1997). There is limited experimental evidence on how electrode configurations affect the EMG signal. High-Density monopolar EMG array collects many signals and thus allows constructing different bipolar electrode configurations over the muscle belly. The aim of this experimental study was to investigate the effect of different bipolar electrode configurations on EMG based force estimation. We studied electrode size, interelectrode distance (IED), collection surface and surface density of the electrodes over the muscle belly.

2. Methods

Eleven healthy subjects (age 28.0±4.1 years, weight 67.6±9.4 kg, and body length 1.8±0.1 m) participated in the experiment after signing an informed consent. The subjects performed isometric right arm extensions. The contraction was a block-shaped pattern and consisted of an isotonic contraction over a plateau of 5 seconds. These efforts were measured at 30, 50 and 80% of maximum voluntary contraction (MVC) and at three different elbow angles (60, 90 and 130º).

EMG and force output were measured simultaneously and synchronized. The force transducer (FUTEK L2353, advanced sensor technology, Irvine, USA) was attached orthogonal to the forearm at the level of the wrist and measured the extension force with a sample frequency of 1000 Hz. Surface EMG was collected with an active High-Density EMG array (BioSemi, biomedical instrumentation, Amsterdam, NL). This two dimensional EMG array consisted of 13x10 electrodes, covering a collection surface of 6.0 x 4.5 cm with gold plated electrode tips of 1.2 mm diameter and an interelectrode distance of 5 mm. EMG signals were collected monopolarly and sampled at 2048 Hz using a 16-bit A/D converter. After adequate skin preparation, the EMG array was attached to the middle of the upper arm with the longer array parallel to it, over the muscle belly of the triceps brachii muscle.

Data analysis consisted of the following steps: high pass filtering (10 Hz), compensation for the electromechanical delay (100 ms), construction of a specific EMG array configuration, full wave rectification, averaging over the EMG channels to get one time signal, low pass filtering (10 Hz) and normalization of the EMG and force signal. The quality of the EMG based estimation of muscle force was expressed as the root mean square difference (RMSD) between normalized force and normalized EMG.

3. Results and Discussion

3.1 Electrode size

The electrode size showed a significant effect on RMSD ($p<0.01$), but this effect was of a marginal magnitude (Fig. 1). Mean RMSD were 21.3±3.1% for the small electrode (12 mm^2) and 21.4±3.1% for the big electrode (113 mm^2). Additionally, the mean RMSD showed a significant ($p=0.03$) decreasing trend for increasing elbow angles.

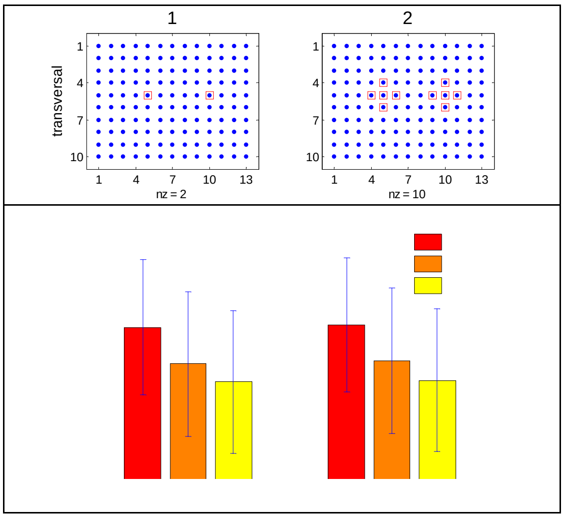

FIG. 1: The upper panel shows two different bipolar electrode size configurations for an interelectrode distance of 2.5cm. The label 'nz' indicates the amount of electrodes (symbolized with dots) used in the configuration. The lower panel shows the RMSD for the allocated electrode size at three different elbow angles. Bars represent the mean values over the different subjects and contraction levels. The error bars represent the standard deviation. Note the Y-axis scaling for better illustration of the effect between conditions.

A model of motor unit action potentials developed by Fuglevand et al. predicted little effect of electrodes size on recording depth (Fuglevand et al., 1992). This is in line with our findings.

3.2 Collection surface

The collection surface had a strong effect ($p<0.01$) on force estimation (Fig. 2). With a gradual enlargement of the collection surface, the

RMSD decreased linearly by about 25%. Elbow angle also affected the RMSD ($p<0.01$), in that a more extended arm position improved the force estimation quality by about 20%.

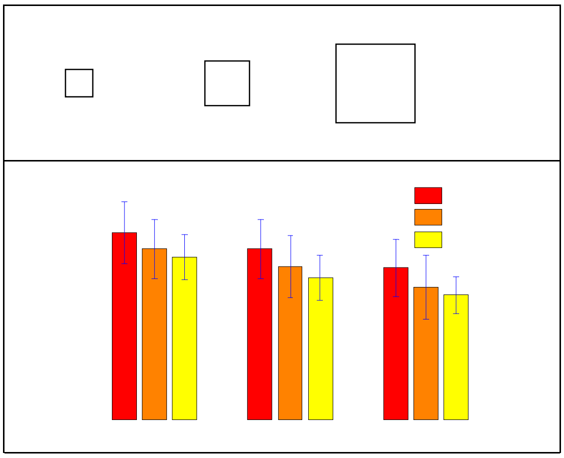

FIG. 2: The upper panel shows three different collection surface configurations (2, 6 and 20cm^2, respectively) constructed with 5-pairs of bipolar electrode channels (symbolized with triangles) having an IED of 5mm. The lower panel shows the RMSD for the allocated electrode size at three different elbow angles. Bars represent the mean values over the different subjects and contraction levels. The error bars represent the standard deviation

Considering that triceps brachii is a pennate muscle one might expect that at shorter muscle length more of the muscle fibres will be converted and thus collected by the electrode array.

3.3 Collection density

The collection density did not significantly affect ($p=0.29$) the force estimation quality (Fig. 3). Approximately 8 uniformly distributed bipolar electrode pairs predicted force as good as 60 pairs of electrodes over the same surface.

Other experimental studies, done with conventional bipolar electrodes could show an improvement of the EMG amplitude estimation by using a spatial combination of multiple recording sites (Clancy and Hogan, 1995).

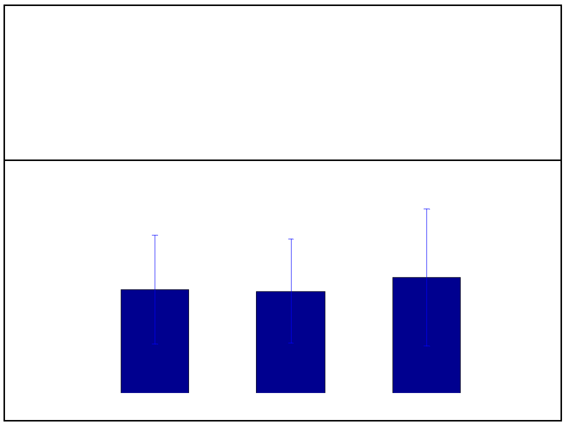

FIG. 3: The upper panel shows different collection density configurations over the same surface. The label 'nz' indicates the amount of bipolar electrode channels (symbolized with triangles) having an IED of 5mm. The lower panel shows the RMSD for the allocated electrode size at three different elbow angles. Bars represent the mean values over the different subjects and contraction levels. The error bars represent the standard deviation. Note the Y-axis scaling for better illustration of the effect between conditions.

3.4 Interelectrode distance

There was no significant effect ($p=0.14$) of IED on force estimation quality (Fig. 4). The two-way interaction, %MVC and IED, had a significant effect ($p=0.01$) on force estimation quality, showing a decreasing effect of %MVC for increasing IED. Between the one-pair electrode and the five-pair electrode configuration a substantial difference ($p<0.01$) could be seen. Mean RMSD was 18±2.9% for the one-pair electrodes and 15±2.5% for the five-pair electrodes, thus a 16% improvement of force estimation quality for multiple pairs of bipolar electrodes. Furthermore, the two-way interaction %MVC-configuration ($p=0.18$) and the three-way interaction, configuration-IED-%MVC, ($p=0.99$) was not significant, indicating a systematic effect of different collection surfaces across contraction levels.

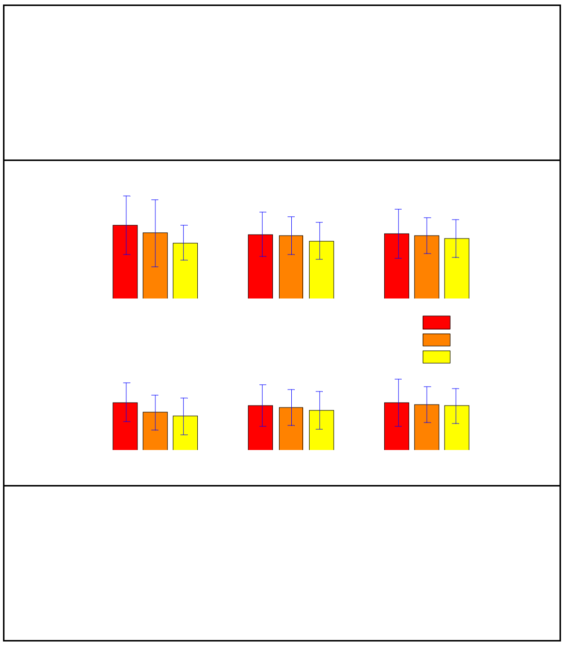

FIG. 4: Three different IEDs (0.5, 2 and 3cm) are presented in this Figure for two electrode configurations. The first consist of one-pair electrode in the middle of the array and is illustrated in the upper panel. The second consist of a transversal arrangement of five-pairs of electrodes, which is illustrated in the lowest panel. The middle panel shows different RMSDs. Bars represent the mean values over the different subjects and elbow angles and the error bars represent their standard deviation. Note the Y-axis scaling for better illustration of the effect between conditions.

Simulation studies showed an increase of collection depth for larger IEDs (Roeleveld et al., 1997, Fuglevand et al. 1992). Increasing IED could have been expected to improve the force estimation quality. This appeared not to be the case, suggesting that the muscle activation is fairly homogeneous in the direction perpendicular to the electrode array.

4. Conclusion

We studied the effect of bipolar electrode configurations on EMG based estimation of muscle force. The collection surface appeared to be the only important aspect of electrode configuration in muscle force estimation. Larger collection surfaces enhance the quality of this estimation by about 25% (Fig 2). In line with these results, more extended arm positions further improved the force estimation. These results suggest force prediction to be optimized by recordings from as many as possible of the active MUs independently. The other three configuration factors, electrode size, IED and density showed no effect on force estimation. However, a minimal density of about 8 pairs of bipolar electrode, evenly distributed over the collection surface, appeared to be necessary to obtain a force estimation of a quality that is comparable to using the whole array (Fig. 3). This suggests that conventional bipolar electrodes allow the same quality of force estimation when using multiple pairs of electrodes in uniformly distributed configuration over the muscle belly.

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The use of inhomogeneous muscle activation to investigate motor unit recruitment in different tasks

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Abstract *– Spatial inhomogeneous muscle activation can be caused by a non-uniform distribution of motor units in m. biceps brachii. To study the presence of spatial inhomogeneous activation, and examine a method to quantify the extent of inhomogeneous activation, 15 subjects performed isometric contractions with varying contraction levels with multi-channel surface EMG electrodes placed over m. biceps brachii. Consistent spatial changes in the distribution of EMG amplitude (RMS), over the three trials with different contraction levels, imply that this method is a valid and reliable tool to study the spatial homogeneity of muscle activation. The consistent changes in distribution illustrate that the inhomogeneous activation of m. biceps brachii occurs with minor variation in different trials. The changes in RMS distribution indicate that recruitment of motor units not are randomly scattered, but evolve in distinct regions in m. biceps brachii. A larger change in the spatial distribution of RMS at low contraction levels, compared to higher contraction levels, fits with the notion that recruitment is the main mechanism for force generation at low contraction levels. Dissimilar changes in spatial RMS distribution with increasing and decreasing contraction levels support the presence of dissimilar force gradation mechanisms in ascending and descending force generation.*

1. Introduction

Studies applying multi-channel surface electromyography (SEMG) have revealed spatial inhomogeneous activation of muscles in isometric contractions with different force levels (Kleine et al., 2000). Scholle et al., 1992 have observed that the unipolar SEMG amplitude distribution over the m. biceps brachii shifts its maxima from distal-lateral to proximal-medial during isometric contractions with increasing force levels. This is most likely caused by recruitment of different motor units (MUs), since MUs with a more proximal innervation zone become dominant at higher contraction levels (Masuda & Sadoyama, 1989). These findings can be due to a spatial non-uniform distribution of MUs in a muscle. Muscles containing functionally independent regions, with different histochemical muscle fiber composition (Chanaud et al., 1991), can cause the inhomogeneous muscle activation. Recruitment of MUs according to their size principle would, in a muscle with these characteristics, cause a spatial inhomogeneous activation in a single task with changing force levels. Studies applying multi-channel SEMG to examine inhomogeneous muscle activation have investigated the change in the root mean square (RMS) distribution over the muscle by topographical visualization (mapping) (Scholle et al., 1992). The main aim of the present study is to examine the validity and reliability of a method to quantify the inhomogeneous activation with multi-electrode SEMG, in addition to confirm the presence of spatial inhomogeneous muscle activation in a single isometric task with varying contraction levels.

2. Method

2.1 Subjects and equipment

Fifteen subjects voluntarily participated in the experiment. An isokinetic dynamometer (Kin-Com 500H, Chattanooga Group, Inc., Hixson, TN, USA) was used in the experimental setup. The force recording was carried out with 1000 Hz. A multi-channel system (modified ActiveOne, Biosemi Biomedical Instrumentation, Netherlands) with a 130-channel grid were used to acquire SEMG data at a sampling rate of 2048 Hz/channel. The multi-electrode array consisted of 13x10 gold covered pin electrodes placed at a holder of 7x5 cm. The inter-electrode distance was 5 mm. This SEMG grid was placed over m. biceps brachii. For the subjects to produce a predestined force level, a visual feedback device was utilized. The feedback device consisted of three arrays of diodes, two arrays showed the target force and the third (middle) illustrated the force produced by the subject. A trigger signal (in the beginning of each test) was used to synchronize the SEMG and force recordings.

2.2 Procedure

All muscle contractions in this experiment were isometric with 130 degrees of the elbow joint. To determine maximal voluntary contraction (MVC), the subjects performed two maximal contractions, lasting two seconds each. The force data recorded during the MVC of each subject was filtered using a 0.2 second moving average window. The MVC was determined as the highest recorded value after averaging.

Subsequently, the subjects were instructed to perform three contractions with sinusoidal force modulations between 0 to 80 percent of MVC. The subjects followed the force tracking displayed from the feedback system. Each sinusoid started at its lowest point (target force of 0, phase shift of -0.5π) and lasted 20 seconds, with 20 seconds (target force 0) rest between each sinusoid. The total test period was thus 100 seconds (three 20 sec sinusoids with two 20 sec brakes). Three contractions with both increasing and decreasing force modulations were performed to be able to investigate the consistency and reliability in this method to study inhomogeneous muscle activation. Contraction levels from 0 to 80% of MVC were applied to investigate the range where recruitment of MUs in m. biceps brachii occurs.

2.3 Data analyses

The force data was low pass filtered (15 Hz), and divided in periods of 500 ms. The mean force of each period was calculated.

The analyses of the SEMG data started with removal of the EMG-channels with bad connection by visual inspection. Then, the SEMG data was band-pass filtered (20-400 Hz), and bipolar configurations were made of the channels with good connection. Bipolar RMS was analyzed in periods of 500 ms (the same periods as the force data). Thereafter, in order to quantify RMS distribution changes, correlation coefficients were calculated between all RMS values at one time period, with the RMS values of the same electrodes at another time period. Correlation coefficients were obtained for all possible combinations within the recording period (100 seconds), resulting in a matrix of 200x200 correlations. In this way, a change in correlation in time compared to any other time instant was obtained. The correlations obtained with respect to two specific time instants were investigated in detail; 1) at maximal RMS and 2) at 25% of maximal RMS of the first inclining phase. The RMS correlations of all subjects were aligned in time, and the 95% confidence interval of the change in correlations of all subjects was calculated.

3. Results

The subjects managed to follow the sinusoidal target force (Fig.1). The EMG amplitude expressed as the median RMS value of all EMG channels, roughly followed the force curve in all subjects (Fig.1). The RMS correlation of all subjects varied with altering contraction levels in all three trials (Fig.2; $p<0.001$, t-test). The RMS correlation changed in a consistent manner in the three contractions of all subjects. In all subjects, the RMS correlation changed faster at low contraction levels, with minor variation at higher contraction levels (Fig.2). The correlations with respect to the time instant with low contraction level (Fig.3) were divergent in the ascending and descending force levels.

4. Discussion

The modification in RMS correlations with alternating force production of all subjects implies that recruitment of MUs are not randomly scattered, but evolve in distinct regions in m. biceps brachii. This finding is in accordance with results from other studies applying multi-channel SEMG with isometric contractions with changing contraction levels (Scholle et al., 1992).

The high correlation between the same force levels during the three trials indicates a stable and low variation in MU recruitment during one specific task. This illustrates that the inhomogeneous activation of m. biceps brachii in this constrained task is consistent in different trials. The consistent change in RMS correlation with changing contraction levels implies that this is a reliable and valid method to study spatial inhomogeneous muscle activation in humans.

The faster change in RMS correlations at low contraction levels, compared to higher contraction levels, fits with the notion that mainly recruitment of additional MUs causes force generation at low contraction levels, while primarily rate coding is responsible for force enhancement at higher contraction levels (Binder et al., 1996).

Romaiguere et al., 1993 have demonstrated that derecruitment of MUs during a slow decrease in isometric force may involve a different strategy than recruitment during the preceding slow force increase. The asymmetrical RMS correlations between the entire time period and a time instant of low force in the ascending and descending contraction levels support the presence of dissimilar force gradation mechanisms in ascending and descending force generation.

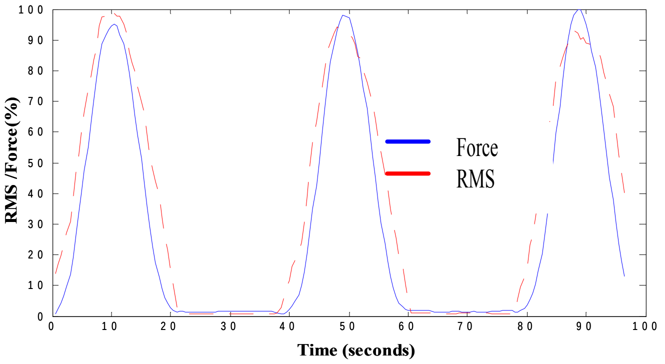

Fig.1 : Typical example of force and RMS changes in time of one subject.

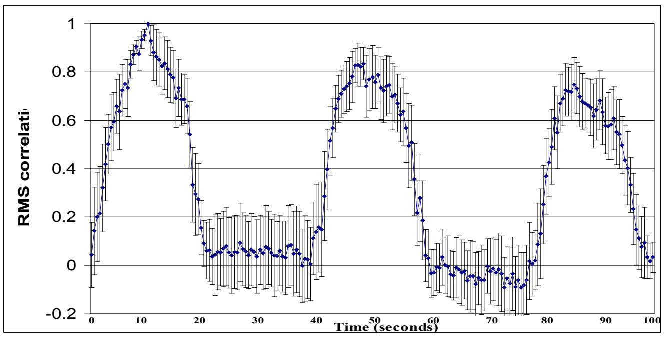

Fig.2: The mean RMS correlation with 95% confidence intervals of all 15 subjects calculated from the time instant of 100% of RMS with all other time instants.

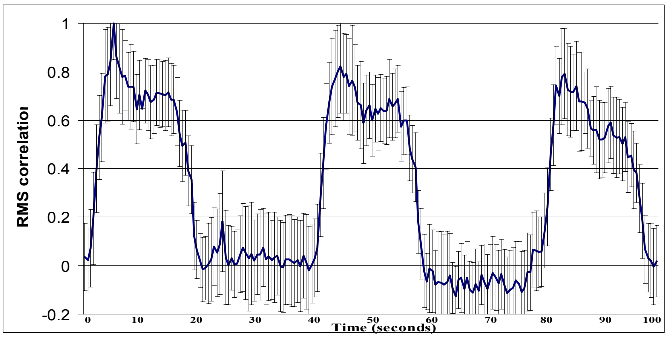

Fig.3: The mean RMS correlation with 95% confidence intervals of all 15 subjects calculated from the time instant of 25% of maximal RMS with all other time instants.

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Motor control during ramp and steady-state muscular efforts investigated by means of sEMG

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Abstract *-The possibility of making inferences on motor unit (MU) recruitment and MU firing rate modulation from the analysis of the myoelectric signal recorded non-invasively from the skin surface has been variously addressed in the literature, sometimes with controversial results. As a matter of fact, it has been widely confirmed that some time- or frequency-domain parameters extracted from the surface ElectroMyoGram (sEMG) are strongly related to these fundamental mechanisms of motor control. The Root Mean Square (RMS) or the MeDian power spectral Frequency (MDF) are just two of them. In the last decade, improvement of recording techniques allowed a better and reliable estimate of the average muscle fibre Conduction Velocity (CV) from sEMG. Moreover, the refinement of non-linear analysis technique provided a further investigation instrument (commonly referred as the percentage of determinism, %DET) able to detect the presence of repetitive hidden patterns in sEMG which, in turn, senses the level of MU synchronization within the muscle. In the present work, these four linear and non-linear parameters (RMS, MDF, CV and %DET) have been investigated in 9 subjects, while accomplishing a ramp-hold isometric contraction of the dominant arm biceps brachii. The ramp-up is from 0 to 100 .% of the Maximal Voluntary Contraction (MVC). Three different ramp-slopes have been used (5, 10 and 20 %MVC.sec^{-1}) and are followed by a 10 sec constant effort at the maximal force. The results, either in each subject or considered as the average for all subjects in all homologous tasks, confirm that: a)-MDF increases until reaching a maximum during the ramp execution, presumably corresponding to the Maximal level of MU Recruitment (MMUR);b)-CV increases to a maximum suitably positioned along the ramp phase, and then decreases till the end of the whole muscular task;c)-%DET, by sensing both CV and MU-synchronization variations, flattens on the middle of the ramp due to the opposite effects of these physiological variables on it; d)-RMS increases till the end of the ramp-hold trial, with two different rates suddenly varying during the ramp phase;e)-during the hold phase at the maximal effort, CV and MDF decrease as well as %DET increase manifest clearly the signs of muscular fatigue.*

1. Introduction

The accomplishment of a motor task is the result of complex interactions between several motor control mechanisms. Among them, motor unit (MU) recruitment and MU firing rate modulation predominantly operate at the muscle level [1,2]. In general, it's well accepted that amplitude and frequency content of the surface electromyogram (sEMG) reflect biophysical and biochemical modifications at the peripheral level of the neuromuscular system.

Typical parameters extracted in the time-domain, such as the root mean square (RMS) or the average rectified value (AVR), which measure the energy content of the sEMG during the observation interval, represent an estimate of the global level of muscle activation [3]. Moreover, simulation works [4, 5, 6] as well as experiments conducted on humans [2, 6, 7] demonstrated the relationship between motor unit activation and power spectra parameters, such as median (MDF) or mean (MNF) frequency.

Besides this classical approach, a new tool for non-linear time-series analysis, the recurrence quantification analysis (RQA), has shown to be valuable for the detection of state changes in drifting dynamic systems without any a-priori hypothesis on data stationarity and statistical distribution. Between the parameters introduced by RQA, the percentage of determinism (%DET) has proven its ability in revealing the presence of subtle rhythms hidden in sEMG [8, 9]. Furthermore, thanks to this capacity to reveal embedded determinism in an apparently stochastic signal, %DET has shown its sensitivity to short-term MU synchronization [10].

Finally, the use of electrode array recording techniques together with improvement of the algorithms for average fibre conduction velocity (CV) estimation provided further insight in motor control strategy.

The purpose of the present study is to gain greater insight into the mechanisms of MU activation by means of a non-invasive methodology. This approach is particularly attractive for our research group because it is the only reasonably available when dealing with athletes in sport medicine and/or patients in neurological clinics [11].

At this aim, the results obtained during ramp experiments from 0% to 100% MVC with three different rate of increase (ramp phase) and during the following 10 seconds of maximal muscular effort (hold phase) were compared. The main parameters extracted from sEMG recordings were:

- Mean conduction velocity (CV)
- Median frequency (MDF)
- Percentage of determinism (%DET)
- Root Mean Square (RMS)

and have been suitably analysed one versus the others or with the force profile.

2. Material and Methods

2.1 Subjects

Forty five experiments were performed on a group of 9 subjects. Subjects were all right handed and the arm chosen for the measurements was the dominant.

2.2 Experiment protocol

After a preliminary session for familiarizing with the experimental set-up and for measuring the maximal voluntary contraction (MVC), the subjects performed one attempt at any ramp slope, at 5, 10 and 20 %MVC·sec-1, followed by a 10 sec hold phase at the maximal effort.

2.3 Force measurement

The elbow flexion torque produced voluntarily during the biceps brachii (BB) isometric contraction was measured by means of a mechanic arm incorporating a strain-gauge bridge suitably amplified and sampled. A visual feedback of the effort exerted was provided on a monitor screen to the subject which was asked to follow a trajectory previously traced on the screen.

2.4 sEMG acquisition and processing

Surface EMG signals were picked up with a linear array of 4 silver electrodes placed over the BB muscle bellies. Signals from the electrodes were fed into an operational amplifier configuration, which gave the possibility to record both single differential (SD) and double differential (DD) sEMG signals. Myoelectric signals were A/D converted at fs=2048 samples/sec with 12 bit resolution. SD served for the extraction of amplitude (RMS), spectral (MDF) and non-linear (%DET) information whereas, according to the suggestions of Merletti et al [12], the DD data were used for the evaluation of CV.

3. Results

The average results obtained in all trials for all subjects during the ramp experiments at 5, 10 and 20 %MVC·sec^{-1} followed by the 10 sec. hold phase are respectively given in Figg.1, 2 and 3. The symbols used for force and parameters extracted are the same in all figures.

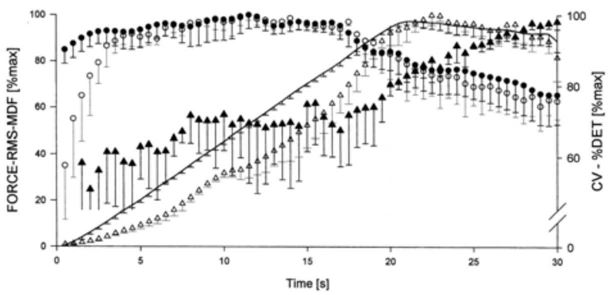

FIG.1: Average results for 5%MVC.sec^{-1} experiments

In order to render the results in each trial and in each subject comparable with those of the same or of other subjects, all data were normalized to the maximum value in each trial (which was assumed equal to 100).

As it can be seen by comparing the three figures, though the abscissas of peculaiar points shift according to the force gradients, the relative behaviours are substantially very similar. In fact, during the ramp phases we may observe that:

- Force tasks (continuous line) are better accomplished in case of slower gradients, as it is indicated by lower variances;
- CV (closed circles)) initially increases up to a maximum where, according to the Hennemann principle, the faster MUs are last recruited. Therefore CV decreases till the end of experiment;
- MDF (open circles *) monotonically increases up to a maximum. In corrispondence to that maximum value, it can be presumed that we are in presence of a maximum MU recruitment;
- %DET (closed triangles 4) initially increases, then flattens about at the center of the ramp and, finally, grows until the end of both ramp and steady state phases;
- RMS (open triangles 6) increases till the end of the ramp, with two different rates. At the beginning, the RMS increase is lower; at the center of the ramp, the exponential like growing behaviour assumes a faster time constant.

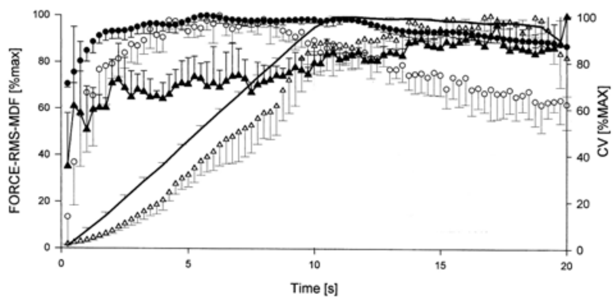

FIG.2: Average results for 10%MVC.sec^{-1} experiments

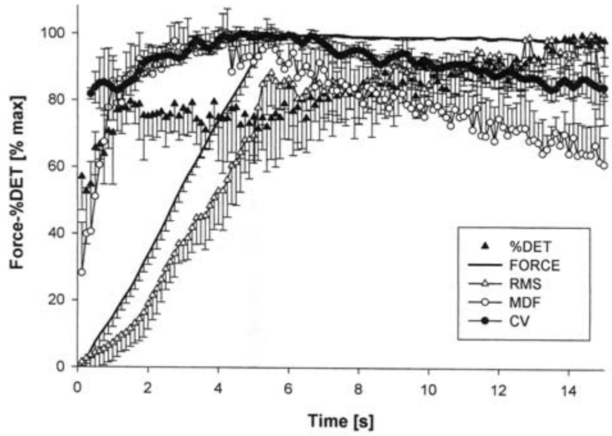

FIG.3: Average results for 20%MVC.sec^{-1} experiments

Concerning the 10 sec hold phase following the ramp phase, the main observations are:

- The greater errors, in terms of force performance, happens after the most fatiguing ramp, i.e. the ramp at 5%MVC.sec-1. In effects, at the end of this ramp, all subjects were not able to maintain the MVC for the whole hold phase;
- The prevalence of fatigue phenomena, which mainly provoke a pronounced broadening of the MU action potentials (MUAP) as well as an appreciable reduction of the CV [13, 14], in turn determines a MDF decrease;
- %DET increases for the concomitant effects of CV decrease and MU synchronization increase [10];
- Along the whole hold phase, RMS almost monotonically increases.

4. Discussion

In the present work, an experimental set-up comprising a ramp-phase up to the MVC followed by an hold phase of 10 sec at the maximal effort was employed to investigate the mutual relationship between MU recruitment, rate coding and MU Synchronization..

In order to deepen their relative interplay in comparably different experimental situations and to explore differences and/or similarities among relative neuromuscular tasks, we dealth with three different force gradients. Moreover, the use of both linear and non-linear tools for the analysis of myoelectric signals recorded non-invasively on the BB opened the possibility to infer the motor control strategy also by taking into consideration the MU synchronization [9, 10].

Referring to the whole set of data, our results suggest that, in a linearly varying effort, the main mechanisms for motor control interact in similar ways, as it is confirmed by the analogies among the general trends of the parameters investigated. At the contrary, concerning the remarkable points where these trends vary along the experiment (e.g., from increasing to decreasing, from increasing to flattening, and so on), they strongly depend on the force slope during the ramp phase.

The role and limits of intervention of MU synchronization in motor control strategy, though far for being exactly defined, find in the non-linear analysis a new developing and deepening tool. The already underlined behaviour of %DET can be interpreted as the result of two opposite actions on this parameter. In a recent study [10], we have shown that %DET is particularly sensitive to MU synchronization (i.e., increase of %DET with MU synchronization increase, and opposite behaviour in presence of MU synchronization decrease) more than to CV variations (i.e., decrease of %DET with CV increase, and viceversa). Within that interval of %DET flattening, we may speculate on the fact that CV increase is totally compensated by the MU synchronization increase. At the contrary, the subsequent %DET increase during the hold phase presumably is influenced by the contemporaneous actions of CV decrease and MU synchronization increase.

Concerning the sEMG amplitude estimation along the ramp-phase, the rate of RMS increase, which is mainly affected by the rate coding of active MUs, is well aligned with that of force increase. The different rates of RMS increase observed during the ramp phase may depend on the recruitment of smaller and lower threshold MUs and then on the activation of bigger and higher threshold MUs which improve their contribute to sEMG amplitude with their increasing firing rates.

Finally, an other consideration on RMS comes from its observed invariance irrespective of force slope during the ramp phase. This suggests an almost immutable strategy of sEMG amplitude adaptation to force increase rate. Likewise, the fact that the RMS maximum was always attained during the old phase is presumably joined to the presence of a functional reserve in the central control of the rate coding mechanism. This behaviour is not in contraddiction with the RMS decline along the hold phase succeding the most demanding ramp tasks (i.e., the slowest ramps at 5 %MVC.sec^{-1}) because this is a clear symptom of either myoelectric and mechanical muscle fatigue, also reported by all subjects.

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Force and MMG comparison during electrically induced muscle hysteretical behaviour

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Abstract*–This paper analyses the hysteresis of the force signal in two human muscles, biceps brachii and tibialis anterior, during stimulation with stimuli rate varying in a triangular fashion. The lowest frequency was 2 Hz and highest frequency rate was within the physiological range (fusion frequency + 20%). Simultaneously the MMG signal was recorded as an index of the degree of twitching for each stimulus. Compared analysis of the force and MMG hysteresis was aimed to clarify the role of the fusion process on the changes of the force frequency relationship in the on-going and down-going phases of the frequency triangle stimulation. The results support the hypothesis that the force hysteresis loop may be due to changes in the Ca^{++} dynamics maintaining muscle fibres shortened for longer time between one motor command and the following.*

1. Introduction

The force frequency relationship (FFR) can be defined as the input/output relationship of the neuro-muscular system. Changes in the motor units firing rate are used by the central nervous system to accomplish muscle force output modulation. On this basis FFR properties may be of interest to predict force from motor units activation strategy. Force control has to be exerted not only from low to high but also from high to low levels of effort. Moreover the two motor programs can follow one each other. Unfortunately it seems that FFR is not unique depending heavily on previous activity of the motor units [1]. No exhaustive data comes from studies on human muscle when stimulation rate are within physiological ranges [2]. The presence of an hysteretical behaviour of FFR during muscle stimulation by an on-going/down going trains of linearly varying frequency (triangular dynamics) has been clearly reported [1, 2].

Surface mechanomyogram (MMG) is generated by the muscle fibres dimensional changes during contraction [3]. When the interpulse time between one motor command and the following is too short to allow relaxation the muscle fibres can be considered in a fusion-like situation in which the dimensional changes are minimal. As a consequence the MMG amplitude during muscle electrical stimulation is inversely related to the frequency of the stimuli [4, 5].

On this basis this work is aimed to analyse the hysteresis of the FFR during a triangular pattern of stimulation in comparison with the simultaneously recorded MMG as an index of the degree of twitching of the evoked mechanical events. The final goal is to clarify the role of the fusion process in the FFR hysteresis.

2. Methods

Nine subjects (25-35 years old) with no neuromuscular diseases were recruited. The dominant arm or leg (in separate experimental sessions) were positioned in an anatomical device in order to keep the angle of the elbow joint at 115° and the ankle in the neutral position. An adhesive electrode (connected to the cathode of the stimulator) was placed to the most proximal motor point. The foot or the arm were connected by inelastic straps to a load cell for force measurements. EMG and MMG were detected by means of an integrated probe (developed within the NEW EU project) placed 1 cm distally from the motor point). The two experimental set-up are shown in Figure 1. The amplitude of the stimuli was set to the one eliciting the EMG M-wave. Muscle potentiation was obtained by administering a stimulation protocol similar to the one indicated by Lee and Binder-Macleod [6]. After potentiation, stimulation frequency was linearly varied from 2 Hz to fusion frequency (FF) + 20% and back to 2 Hz in 12 s. This pattern is named frequency triangle. FF was the frequency providing a force ripple lower than 5% of the single twitch amplitude. It was defined by a stimulation with 2 to 50 Hz increasing frequency. *Signal Analysis*. Single twitches (ST). The force and MMG responses at 2 and 3 Hz were averaged sample by sample to calculate the following parameters: the peak twitch force (P_t, measured in N), the normalized to P_t maximum contraction rate (MCR, %Pt/s) and the normalized to P_t maximum relaxation rate (MRR, %Pt/s). The same was done on the MMG responses to obtain the MMG peak to peak value. Frequency triangle. From the force signal during the on-going (OG) and the down-going (DG) 6 s phases the FFR was obtained as follows. For each stimulation frequency the average force was calculated as the total area below the force signal divided by the interpulse period. For stimulation frequencies with inter-pulse periods longer than ST duration the average force was calculated as the total area below the signal divided by the ST duration. Each average force value was used to define the FFR from 2 to FF+20% and vice versa. After this, the area beneath the FFR was calculated for both the OG and the DG phases. The ratio between the DG and the OG area was an estimation of the hysteresis process. The MMG signal was processed in a similar way. Instead of the average value the root mean square was estimated.

Figure 1. Experimental setup: biceps brachii (top) and tibialis anterior (bottom).

3. Results

In Figure 2 the force and MMG signals are reported for a representative subject.

In the left column it is evident that the force increases with the stimulation frequency while the MMG decreases. The right panels report the force and the MMG vs frequency relationships. A counterclock-wise and clock-wise hysteresis is present for force and MMG, respectively. On the average force hysteresis was 138.8±13.72 (biceps brachii) and 127.8±14.12 (tibialis anterior). The correspondent values for MMG were 56.0±11.55 and 71.9±19.48.

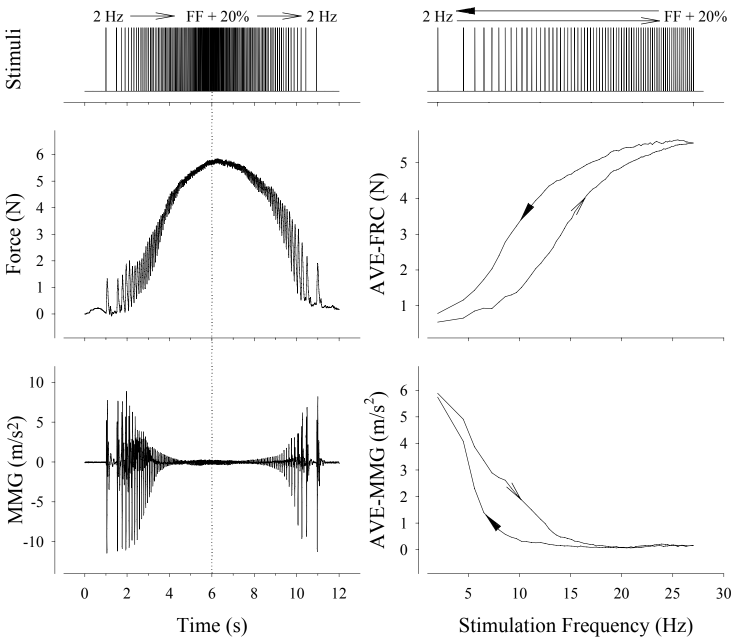

Figure 2. Biceps brachii. Left column. Stimulation pattern in time with the force and MMG responses in a representative subject. Right column. The force (average)/frequency relationship and MMG (RMS)/frequency relationship calculated from the signals on the left.

In table 1 the parameters of the force and MMG during single twitches and the areas below the force and MMG frequency relationships during the OG and DG phases are reported. Statistically significance ($p<0.05$) between OG and DG parameters has been searched by Anova for repeated measures and indicated by an asterisk.

	BICEPS BRACHII		TIBIALIS ANTERIOR	
	ON-Going	DOWN-Going	ON-Going	DOWN-Going
Amplitude (N)	2.646 ± 1.166	3.073 ± 1.166	2.549 ± 0.9511	2.887 ± 1.046
MCR (%Pt/s)	28.63 ± 7.47	25.61 ± 5.651	20.03 ± 3.05	21.16 ± 4.565
MRR (%Pt/s)	13.54 ± 5.078	11.67 ± 4.14	7.701 ± 3.543	8.45 ± 2.32
MMG (m/s^2)	5.061 ± 2.446	4.267 ± 1.937	1.444 ± 0.8897	2.363 ± 2.521
Area F/FR (N*Hz)	161.01 ± 71.54 *	212.86 ± 99.69 *	117.01 ± 60.56 *	154.47 ± 98.91 *
Area MMG/FR (m/s^2*Hz)	42.24 ± 26.98 *	22.44 ± 13 *	8.78 ± 7.91 *	6.23 ± 5.44 *

Table 1. Force and MMG parameters during single twitches and the areas from their relationships with stimulation frequencies.

4. Discussion

Contrary to the already described FFR during frequency triangular stimulation obtained in the range 4-100 Hz [2], our results attain to a frequency range (2 Hz/fusion frequency + 20%), suggested by Binder-Macleod & Clamann [1]. As a consequence they may provide useful data about the motor units firing behavior during voluntary isometric contractions encompassing a sequence of increasing/decreasing output force. Both biceps brachii and tibialis anterior presented a clear force hysteresis. This phenomenon has been attributed to different factors [1]:

1. Post-tetanic potentiation. We can exclude the force potentiation, as an effect of the ongoing part of the triangle, because the actual linearly varying frequency stimulation was administered after having fully potentiated the muscle.

2. Prolongation of the twitch. A decrease of the muscle relaxation rate may provide a better fusion of the mechanical events with more force produced at each stimulation rate. This seems not to be the

case given that MRR was not significantly different at the beginning and at the end of the triangular stimulation.

3. At each instantaneous frequency during the ascending limb of the hysteresis loop less force is generated, with respect to the descending limb, because the average tension for a given rate takes time to be developed and the preceding rate is lower than the actual one. On the contrary during descending limb the preceding higher frequencies help to reach a better tension at the actual rate. The factors determining this delayed tension development have been indicated by Binder-Macleod and Clamann [1] as the skeletal muscle catch-like property and the different rate or cross-bridge cycling during contraction and relaxation (probably due to a different p-Ca^{++}/force relationship during the Ca^{++} release or re-uptake from the sarcoplasmic reticulum).

MMG, at each frequency, during the descending limb of the force hysteresis loop is greatly decreased suggesting that the muscle fibre dimensional changes, due to muscle contraction (shortening) and relaxation (re-elongation), are reduced. The muscle contractile elements are, on the average, in a shortened situation with more tension placed on the connective tissue mechanically coupling them to the bone lever. On this basis we conclude that our data support the hypothesis that the force hysteresis loop may be due to changes in the Ca^{++} dynamics maintaining muscle fibres shortened for longer time between one motor command and the following, being this phenomenon more evident at intermediate frequencies.

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MUAP Rate: a new measure to assess motor control

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1. Introduction

Motor control is commonly investigated using bipolar surface Electromyography (EMG), resulting in a global view of muscle activation patterns. To study motor control in more detail, it is desirable to assess both the number of active motor units (MUs) and their firing frequency, since these are the parameters used by the Central Nervous System for motor control. As such, we propose to assess the number of MU action potentials (MUAPs) per second (MUAP Rate, MR) as a measure for motor control.

MR can be assessed with multi-channel electrode arrays and advanced signal processing techniques (1,2). The objective of this study was to explore the behaviour of MR in relation to force.

2. Method

EMG of the dominant upper trapezius of nine healthy subjects was recorded during a step-wise ramp contraction with a 2D 16-channel array (3).

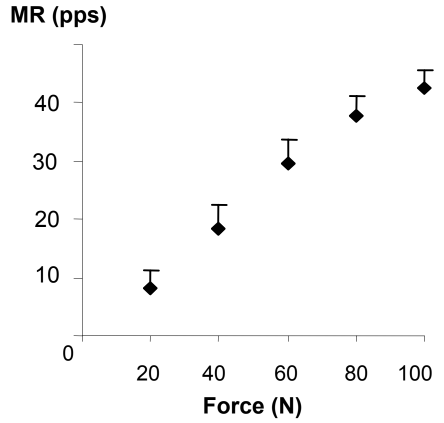

Fig. 1 Relation between MR and force (diamonds: mean , bars: standard error of mean)

The ramp consisted of five force levels of 20 to 100N in steps of 20N. Force feedback to the subjects was provided. The duration of each level was 10 seconds. MR was calculated per second from single differential signals (inter-electrode distance 10 mm) showing propagating MUAPs.

3. Results

In Figure 1, the relation between MR and force is shown. It can clearly be seen that MR increases with force. A linear regression analysis revealed a significant relation between MR and force, with 58% explained variance. The slope of the regression line was 0.44 (significantly different from zero, $p<0.000$).

4. Discussion

It can be concluded that MR increases almost linearly up to forces of 80N. The flattening for higher values is likely due to the occurrence of superpositions. The relatively high force value for the starting point of the flattening is probably related to the limited number of MUs that contribute to the EMG signal, due to the electrode configuration. MR seems to be a valuable tool for assessing motor control.

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The effect of intramuscular pressure on mechanomyographic activity

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1. Introduction

The mechanomyogram, MMG, has been suggested to reflect changes in motor unit activation and muscle fatigue development during low force contractions more potently than the electromyogram, EMG (Søgaard et al 2003). However, it is unknown if the changes in MMG are due to changes in the contractile mechanisms or in intramuscular pressure, IMP, that may also increase with fatigue due to muscle water accumulation.

2. Objective

To test the hypothesis that IMP *per se* affects the MMG.

3. Methods

Three subjects performed static elbow flexions at increasing force levels - ramp as well as step contractions - (0-60 %MVC) in combination with increasing external pressures (0-100mmHg) applied by a cuff around the upper arm. Simultaneous measurements were performed of force, bipolar surface EMG, MMG using an accelerometer placed above the belly of the muscle, and IMP using an intramuscular Millar tipped pressure transducer.

4. Results

IMP, MMG, and EMG all increased with increased ramp contraction force in a near-linear manner; the mean values at 60 %MVC being: 109 mmHg, 0.237 m/s^2, and 0.804 mV, respectively. When external compression of 50 mmHg was applied to the muscle, IMP increased to 269 mmHg but neither MMG nor EMG were affected. Correspondingly, for stepwise increasing contractions of 10, 20, and 40 %MVC, a rise in cuff pressure (20-100 mmHg) increased IMP: 16-141, 43-160, and 76-235 mmHg, while mean values for MMG remained at 0.042, 0.067, and 0.242 m/s^2 as well as EMG at 0.086, 0.138, and 0.224 mV, respectively, for the three contraction levels.

5. Conclusion

Muscle contraction - but not IMP *per se* - increases MMG activity in correspondence with EMG activity.

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Co-activation Of M. biceps brachii And M. triceps brachii In Children With Obstetric Plexus Lesion: Efficiency Of Botulinumtoxin In Therapy

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Abstract *– For the evaluation of the treatment of children with obstretic plexus lesion with botolinumtoxin, a new measurement procedure has been developed which permits a quantitative evaluation of the range and quality of motion. The procedure combines the assessment of muscular co-contraction by surface EMG with a three-dimensional motion analysis. The EMG data showed that the application of Botolinumtoxin together with physiotherapy reduced functional impairments due to co-contraction during elbow flexion for the majority of patients. Furthermore, an increased range of motion was found in the 3D data.*

Introduction

Patients with a lesion of the nervus plexus brachialis frequently suffer from functional impairments in the upper extremities. An example is the restricted range of motion due to coactivation, observed during flexion-extension of the elbow joint. The nerve poison Botulinumtoxin (Botox) paralyses a muscle for a certain period of time and has been used successfully in treatment of spastic gait patients with coactivation of leg muscles. The question thus arose, as to whether patients with plexus lesion could also be successfully treated through the use of Botox. Therefore a measurement procedure has been developed which permits a quantitative evaluation of the impact of Botox on both range and quality of motion.

Patients/Method and Materials

At the Institute for Biomedical Technologies, a procedure has been developed which synchronises bipolar surface electromyography and 3D motion capture. All measured patients were in the age range 2 to 6 years, suffered from a plexus lesion and showed clear coactivation of m. biceps brachii and m. triceps brachii. As part of the measurement procedure, the patients were asked to perform a flexion-extension movement of the elbow joint. EMG data of biceps and triceps were recorded according to the SENIAM recommendations, and subsequently rectified and enveloped. Crosstalk between the two muscles could be identified by comparing the healthy and affected sides. The 3D motion data were collected using the video-based Vicon 370 motion analysis system. By applying the motion data to a kinematic model of the arm, relevant joint angles were calculated. The patients were measured prior to the injection of Botox and afterwards, at intervals of 6 weeks, 3 and 6 months. Additionally they received physiotherapy.

Results/Discussion

The EMG data showed that the application of Botox together with physiotherapy reduced functional impairments during elbow flexion for the majority of patients. The analysis of the 3D motion data within a period of 6-12 months after treatment with Botox showed an increase in range of motion. Additionally, EMG measurements indicated that the paralysing effect of Botox had ceased. Consequently, a temporary paralysis of the antagonistic muscle seems to allow time for a permanent muscular improvement in the agonist. This results in the overall enhancement of muscle coordination and range of movement for patients with a plexus lesion.

Session 3:

Advanced techniques for neuromuscular and clinical assessment

NEW advances in surface EMG modelling and processing

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Abstract *– This lecture presents advances in surface EMG processing and interpretation made possible by the European project "Neuromuscular Assessment in the Elderly Worker" (NEW). The main topics addressed are 1) signal modelling, 2) estimation of muscle fiber conduction velocity, 3) spatial selectivity, and 4) surface EMG decomposition. Aim is to provide a summary of the NEW achievements in these fields.*

1. Introduction

The surface electromyographic (EMG) signal contains the information about the fiber membrane properties and the control strategies of motor units (MUs). Thus, it is an important window into the neuromuscular system. However, the EMG signal features are influenced by many factors not related to the physiology under investigation. The extraction of information from the surface EMG is thus a complex task which requires advanced processing methods.

In past and more recent years considerable work has been carried out on the extraction of information from surface EMG signals. In this abstract we will briefly summarize only the main advances in techniques for surface EMG processing and interpretation reached within the "Neuromuscular Assessment in the Elderly Worker" (NEW) project. The main topics addressed are 1) signal modelling, 2) estimation of muscle fiber conduction velocity, 3) spatial selectivity, and 4) surface EMG decomposition. These topics have been addressed in recent papers published within the framework of the NEW project. Some of these papers are quoted in this abstract and provide the detailed description of the methods presented in this work.

2. Surface EMG modelling

Surface EMG modelling finds important applications in the interpretation of signal features, the test of processing algorithms, and didactic purposes. The availability of advanced models was considered as one of the primary aims within the development of new methods for signal interpretation in the NEW project.

The main steps in the development of an EMG model are 1) the description of the source, i.e., the modeling of the generation, propagation, and extinction of the intracellular action potential, 2) the mathematical description of the local properties of the volume conductor (with a formulation based on partial differential equations), 3) the analysis of the geometry of the volume conductor and of the boundary conditions, and 4) the modeling of the detection system, i.e., of the spatial filter applied to the skin potential distribution (spatial arrangement, shape and size of the electrodes, inter-electrode distance). The proper modeling of the volume conductor is one of the most critical steps in the design of a model.

Models for surface EMG may be analytical or numerical. Analytical methods are usually computationally more efficient than numerical ones. They also allow an easier interpretation of the changes of signal features with system parameters than numerical approaches. Nonetheless, a numerical approach allows the description of complex muscle architectures, which is less tenable with analytical derivations.

Figure 1 shows a set of volume conductor geometries investigated within the NEW project. The solution of the mathematical problem of the computation of the potential distribution at the surface of the volume conductors shown in Figure 1 is provided in [1], [2], [3], and [4].

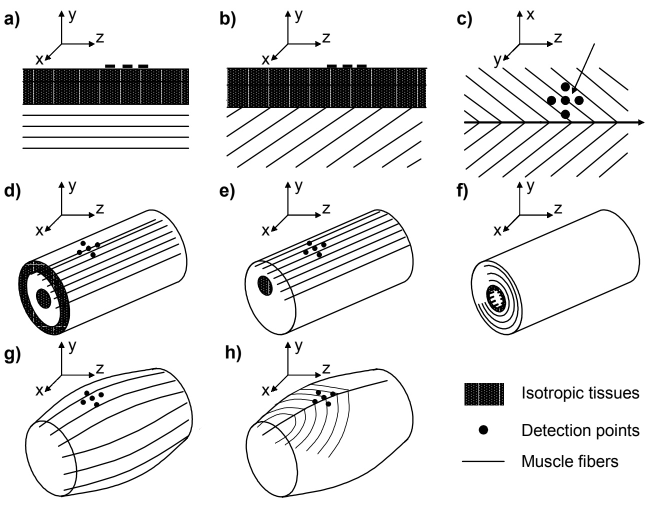

FIG. 1 : Examples of volume conductors. In some cases, both analytical and numerical solutions are available from the literature, while in others only numerical methods can be used. Analytical solutions exist for the cases reported in (a) [1], (b) [3], (c) [3], (d) [2], and (f) [2]. (From [4]).

2. Estimation of conduction velocity

Muscle fiber conduction velocity (CV) is an important parameter which can be extracted from surface EMG signals. It is indicative of muscle fiber membrane properties and fatigue. Although the issue of estimating CV from the delay of signals detected at two points over the skin is trivial in ideal conditions, it becomes complex in experimental situations. The shape of surface recorded signals detected along the direction of the muscle fibers may vary due to the end-plate and end-of-fiber components, electrode misalignment, local tissue inhomogeneities, etc. Thus, a method for delay estimation from surface EMG signals is implicitly a definition of delay between signals of unequal shape. An overview of the different methods for CV estimation described in literature can be found in [5]. The development of CV estimation methods within the NEW project was focused on the further improvement of the maximum-likelihood approach [6],[7], on the extension of the method to very short signal windows [8], and on the application to bi-dimensional recordings [9].

Method	PKI	PK	MLE2	MLEm
PKI		NS	**	NS
PK	NS		**	NS
MLE2	**	**		**
MLEm	NS	NS	**	

Filter	LSD	LDD	TSD	TDD	NDD
LSD		NS	**	**	*
LDD	NS		**	**	NS
TSD	**	**		NS	*
TDD	**	**	NS		*
NDD	*	NS	*	*	

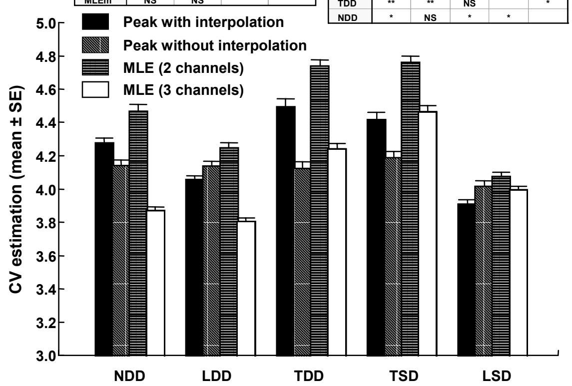

FIG. 2 : Influence of the choice of the spatial filter and estimation method on CV estimates. The methods compared were the estimation of CV from the peak delay with (PKI)

and without (PK) interpolation and the maximum-likelihood estimator for signal pairs (MLE2) and rows of signals (MLEm). Filters investigated were the longitudinal and transverse single differential (LSD,TSD), the longitudinal and transverse double differential (LDD, TDD), and the normal double differential filter (NDD). Experimental signals detected from the biceps brachii and upper trapezius muscle. (From [10]).

The comparison of different CV estimation methods based on experimental signals showed that the estimated values depend strongly on the adopted method, but also on the spatial filter used for signal detection [10] (Figure 2).

3. Spatial filter selectivity

The tissues interposed between the signal sources and the detection electrodes act as low-pass filters, causing a blurring effect on the surface detected potentials. As a consequence, over the skin the amplitude attenuation with increasing distance from the source (spatial selectivity) is small.

As in image processing, spatial selectivity can be enhanced by proper signal filtering. Since the volume conductor can be described as a spatial low-pass filter, a high-pass filtering in the spatial domain may be used to counteract the blurring effect of the tissues. In surface EMG, spatial filtering is performed by the weighted summation of signals detected by electrodes arranged in particular geometrical configurations.

Comparison of selectivity of spatial filters is a key issue. Theoretically highly selective spatial filters are usually comprised of more detection surfaces than less selective ones. For establishing if complex systems are worth to be used instead of classic ones, it is necessary to quantify the gain in terms of spatial selectivity obtained at the expenses of a more complex detection modality.

Comparison of spatial filter selectivity may be performed by modelling but it is always necessary to experimentally validate the model results since they can be largely dependent on the specific volume conductor considered [11]. Comparison of selectivity on the basis of experimental recordings revealed that the normal double differential filter (NDD) performs best among the most commonly used spatial filters [12][13]. An example of single motor unit action potentials detected by a number of spatial filters at different detection points over the skin is reported in Figure 3.

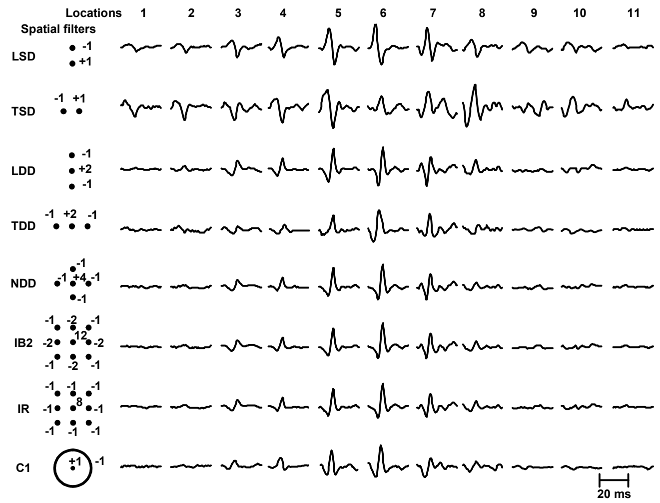

FIG. 3 : Single surface MU action potentials detected from the tibialis anterior muscle. The surface potentials were averaged using intramuscularly detected potentials as triggers. The results from 8 spatial filters are shown. The distance between each recording location is 5 mm, the inter-electrode distance of the point electrode spatial filters 5 mm and the radius of the ring system 5 mm. (From [12])

4. Surface EMG decomposition

In the past, the analysis of single MU properties was mainly conducted from EMG signals detected using intramuscular needles or wires because of the high selectivity of these detection systems. In recent years, techniques for the identification of single MU action potentials from surface recordings have been proposed. The common strategy was to increase the amount of information from surface EMG and to improve the spatial selectivity of the recording.

Within the NEW project, novel methods for the identification of single MU action potentials based on a segmentation/classification approach have been proposed [13][14]. The results of the application of one of these methods to surface recordings, filtered by commonly used spatial filters, are shown in Figure 4.

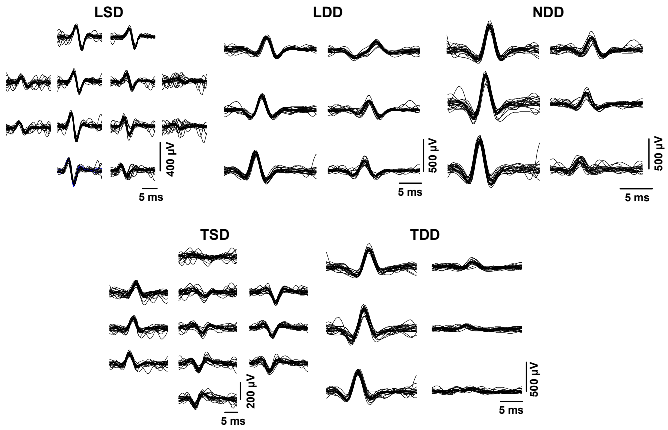

FIG. 4 : Surface MU action potentials extracted from interference EMG signals and classified as belonging to the same MU. Muscle biceps brachii. The MUAPs were classified from the NDD filtered signals and the detected times of occurrence were used to obtain the MUAPs as filtered by the other spatial filters. (From [13]).

The segmentation/classification methods suffer from a loss of performance when the superposition between potentials increases significantly. This may be limited to a certain extent by selective spatial filters (see previous section) but usually hinders the possibility of extracting the entire MU firing patterns.

Blind source separation [15][16][17] or higher order statistics [18] approaches have been proposed for the separation of different MUs and/or muscles (cross-talk) contributions from surface recordings. These approaches have not the strong constraint of requiring temporally separated MU action potentials, thus they appear promising for the complete surface EMG decomposition [15][18].

5. Conclusion

NEW tools for signal interpretation and processing allow a better insight into the neuromuscular system through the analysis of surface EMG signals.

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A Preliminary Report on the Use of Equalization Filters to Derive High Spatial Resolution Electrode Array Montages

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Abstract *– Recently, high resolution (HR) surface electrode arrays have been developed to monitor the activity of individual MUs. Most often, weighted combinations of monopolar electrode potentials are formed (in hardware) to produce a myoelectric signal that is spatially selective, e.g., the normal double differentiating (NDD) filter [1], [9], [10]. High-precision* **hardware** *differential amplifiers have been utilized to perform the channel weighting. These hardware systems exhibit a high common mode rejection ratio (CMRR) but are very inflexible, thus only a few HR channels are typically acquired per array, although numerous HR channels could be defined. In this paper, we describe* **preliminary** *work to develop a prototype system that attempts to achieve both high common mode rejection ratio (CMRR) and flexible electrode combination via the use of software channel equalization. This system would be less expensive to produce than a typical HR electrode array system and would permit the formation of any spatially selective filter.*

1. Background and Introduction

In recent years, surface electrode arrays that can monitor the activity of individual MUs have been investigated. These systems have an inter-electrode distance of 2–5 mm and an electrode diameter ≤ 2 mm. Electrodes are usually arranged in an equidistant rectangular grid. Studies have used these arrays, for example, to determine the locations of neuromuscular junctions [5], [6], measure conduction velocity [2], [3], [7], and non-invasively decompose the myoelectric signal [1], [4], [11]. Most often, weighted combinations of monopolar electrode potentials are formed (in hardware) to produce a myoelectric signal that is spatially selective. The simplest spatial filter is formed by a standard bipolar recording, which is a pair of electrodes with weighting coefficients (–1, +1). A simple longitudinal spatial filter can be formed from three successive electrodes (arranged equidistant along a line) with weighting coefficients (+1, –2, +1). To achieve two-dimensional spatial selectivity, two-dimensional electrode combinations are necessary, such as the normal double differentiating (NDD) filter that weights five electrodes, selected in the shape of a cross, with the weights (+1, +1, –4, +1, +1) — the central electrode uses the –4 weight.

In presently available EMG systems, high precision hardware differential amplifiers are required to perform channel weighting. A separate analog circuit is required for each spatial filter. Obviously, a flexible hardware system allowing many different spatial filter combinations is not readily possible, e.g., a 4x7 equal-spaced

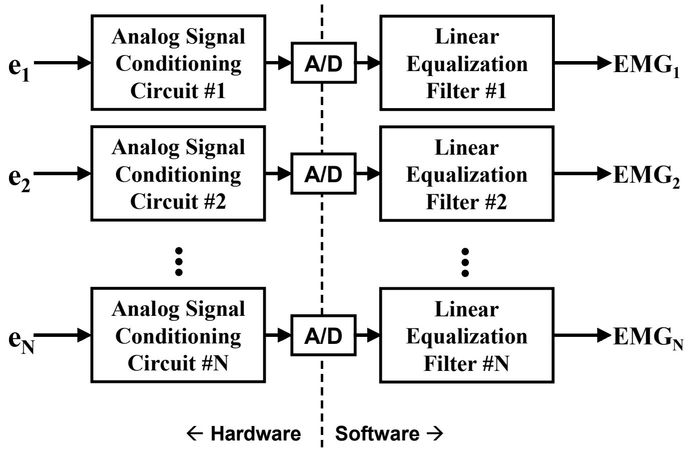

FIG. 1: Configuration of an EMG array system with channel equalization. Monopolar inputs ei are conditioned by analog circuits [typically, a high-pass filter, selectable gain, electrical isolation and low-pass (anti-aliasing) filter]. The conditioned signals are analog-to-digital (A/D) converted. Each channel is then individually equalized via a software filter to form outputs EMGi.

rectangular array has 28 monopolar electrode combinations, 84 standard bipolar electrode combinations, 32 standard LDD electrode combinations and 16 standard NDD electrode combinations—resulting in 160 possible standard myoelectric signal channels. Hence, existing systems either accept poorer weight matching by combining the monopolar channels in the post-acquisition software (which significantly degrades the common mode rejection ratio, CMRR, of the recording since it is not possible to match the hardware characteristics of the distinct channels with the required level of precision) or pre-wire (in hardware) a selected number of spatial filters. This conflict limits the technical performance of array systems.

We are developing a prototype system that will attempt to achieve both objectives (high CMRR and flexible electrode combination) — and be cheaper and simpler to construct — via the use of software channel equalization. Fig. 1 shows the general configuration of the system. Only monopolar electrode data are acquired by the hardware. In addition, the electrical characteristics (gain and phase response vs. frequency) of each analog signal conditioning circuit are carefully ***measured***. A distinct software equalization filter is cascaded with each monopolar channel and selected such that the overall frequency response of each channel is identical (or, as near so as possible). Thereafter, any weighted combination of the (equalized) monopolar channels (e.g., bipolar, LDD, NDD) of any set of electrodes can theoretically be derived with high common mode performance since the equalization filters match the hardware imperfections between the physical channels. In this manner, the myoelectric spatial filters desired need not be determined prior to data acquisition — any can be reconstructed during post-processing. A substantial advantage should result. Similar equalization filters have been used successfully in such applications as communication channels and phased-array radar.

2. Equalization Techniques

The equalization filters are "calibrated" by measuring the frequency-dependent gain and phase of each analog signal conditioning channel. The desired ("ideal") frequency response divided by the measured frequency response gives the frequency response of the equalization filter. Successful equalization relies on the assumption that the analog signal conditioning circuit frequency responses can be ***measured*** far more accurately than they can be manufactured in hardware. For measurement, we excite each channel input with a cosine chirp (sweeping in frequency from 0–2000 Hz) and compare the observed channel output with the output if the channel were "ideal". We initially attempted to perform equalization via time-domain finite impulse response (FIR) filters. However, these filters seem unsuited to the problem. The desired CMRR should exceed 80–100 dB, corresponding to errors of less than one part in 10,000–100,000, respectively. FIR filters that maintain this level of accuracy over a band of frequencies would require an excessive number of coefficients (c.f. [8]). Even if such filters could be properly designed, their excessive length would result in an unacceptably long filter start-up transient.

Hence, we are now pursuing frequency-domain equalization filters. In doing so, the dominant challenge has been the inevitable presence of broadband background noise in the calibration signal. In our prototype hardware, the noise standard deviation is typically 0.5–0.8 μV (referred to the input) — similar to EMG hardware systems reported in the literature. Although this noise level is quite small overall, both real data analysis (from a prototype hardware system) and simulation find this noise level unacceptably high when calibrating equalization filters. In particular, CMRRs of 18–44 dB at 60 Hz (bipolar montage) are typical when equalization filters are calibrated directly from the measured data.

As a consequence, our present focus is on signal processing methods for removing noise from the calibration recordings prior to determining the equalization filters. We are pursuing two de-noising methods — linear, time-variant (LTV), bandpass filtering and mixing. Both techniques recognize that the input excitation (the cosine chirp) is broadband when analyzed over the entire excitation period, but is localized in frequency at any particular time within the period. Thus, linear time-invariant (LTI) filters are not useful, but filters which modify their characteristics as the chirp frequency changes can be useful.

Perhaps the simplest de-noising filter is a LTV bandpass filter. Since the (time-varying) instant-by-instant frequency of the input excitation chirp is known, a LTV bandpass filter can be constructed with a passband centered at the excitation frequency. The passband location varies in time to follow the chirp excitation frequency. Ideally, much of the noise power can be eliminated. We have implemented an FIR LTV bandpass filter, designing the filter for each time index using the window method [8].

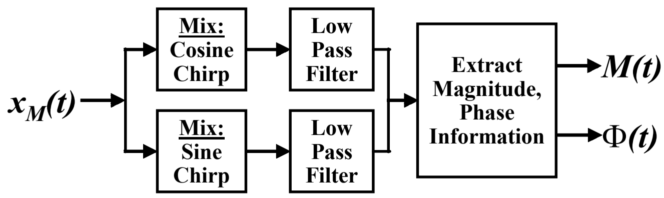

FIG. 2: Overview of mixing technique. Measured input $x_M(t)$ is the excitation chirp wave, modified by the frequency response of the analog signal conditioning circuit, plus measurement noise. The mixing technique produces an estimate of the measured chirp magnitude $M(t)$ and phase $\Phi(t)$ at each time index t.

We are also evaluating a mixing technique, depicted in Fig. 2. If the measured chirp waveform is multiplied (mixed) in software with another chirp utilizing the same sweep rate, the resultant output will consist of the sum of two chirps — one with double the sweep rate and one with a sweep rate of zero (i.e., modulated to DC). A low-pass filter is cascaded with the mixer to eliminate the double sweep rate term, leaving the DC modulated signal. In practice, two mixers — arranged 90º out of phase — are used to recover the complete calibration information. With this technique, the chirp input is not preserved, rather the magnitude and phase of the de-noised frequency response of an analog signal processing circuit is estimated.

3. Preliminary Results and Observations

For the LTV bandpass filter, only simulation results are available and only for testing the de-noising performance. We have evaluated filters with 100–400 coefficients, and reduced simulated noise by factors of 10–25. There exists a trade-off between the amount of noise reduction and the amount of distortion of the chirp signal. Smaller passbands (100–200 Hz) provide more noise reduction, but overly small bands cause distortion of the chirp signal. (Distortion is expected to degrade equalization performance.) Higher-order filters improve these issues (they reduce more noise *and* cause less signal distortion), but lead to increases in the filter startup transient.

We have more experience with the mixing technique, and have applied it to real data in our prototype hardware. With the real data, we are typically measuring CMRRs of 65–84 dB at 60 Hz (bipolar montage; after mixing and equalization). The most difficult technical issue is the design of the low pass filter stage, since this filter's cut-off frequency must be set to less than 1% of the Nyquist frequency. Presently, a large number of low pass filter coefficients (>4000) are required.

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Spatial intramuscular coordination in surface EMG signals

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Abstract *The origin and onset of muscles at the bone are mostly two-dimensional areas. This feature allows the change of force directions or lever lengths respectively if there is some intramuscular coordination. This phenomenon we could show experimentally by means of multichannel surface EMG for the static case and for dynamic movements [Scholle et al. 1992, Stegeman et al. 1998, Schumann et al. 2002, Scholle et al. 2001]. In the static case intramuscular coordination appears as a dependency of the spatial activity distribution on the force . In the dynamic case the activity distribution pattern within one muscle depends on the phase of the movement. These investigations were realised via the calculation of Surface EMG power (RMS) maps. RMS maps allow the discrimination of bio-mechanically relevant activity distributions (and neglects the details of the temporal activation pattern) by means of a two-dimensional interpolated representation of EMG data. It can be shown that the interpolation fulfils a spatial sampling condition due to time averaging. An improvement of the spatial resolution of the RMS map is possible via the calculation of cross covariance functions and their high pass filtering. This method also ignores the temporal activation pattern due to time averaging but it gives information about the spatial distribution of activity with higher resolution in cross fibre direction.*

1. Introduction

If we consider the spatial configuration of muscles, tendons and bones, there appear situations where the onset of the muscle has a spatial extend, perpendicular to the fibre direction, like the triceps muscle at the scapula in Fig. 1. In principle, such configurations could allow to control mechanical degrees of freedom via selective activation of muscles, because a muscle that performs a static contraction against a tension can be understood as a lever with two joints, see

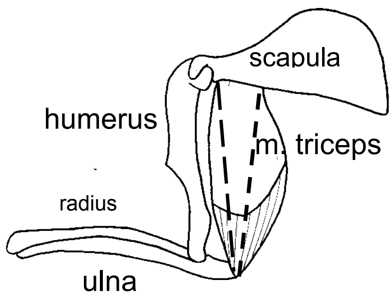

Figure 1: A sceme for the triceps muscle of a rat. The dashed lines indicate possible directions of contraction.

Fig. 2. Thus, there arises the question whether the motor control system uses such possibilities or not. Surface EMG is a valuable method for investigation of movement co-ordination. Because it does not irritate the muscle which is measured,

this method can be used e.g. for screening purposes, for sports and rehabilitation and furthermore such investigations can be easily performed with electrode grids, covering a whole muscle. So, multi-channel EMG is a valuable tool for the investigation of intra-muscular co-ordination.

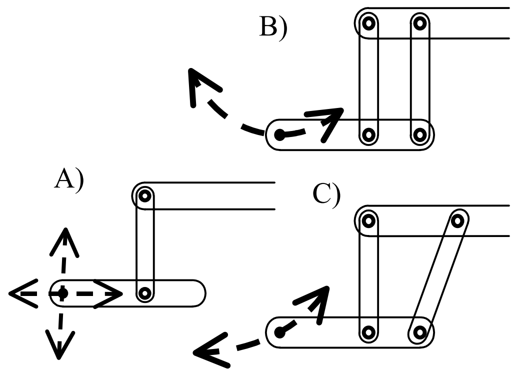

Figure 2: The control of mechanical degrees of freedom (DOF) via lever positions: A) two DOF B), C) one DOF with different trajectories.

2. Physiological results

For the static case intra-muscular co-ordination could be found in [1]. It proves to be an force dependency of the spatial position of the activation maximum. For dynamical movements the spatial location of the activation maximum within the muscle depends on the phase of the movement, see Figure 3 and [2,3]. Furthermore, surface EMG maps of the low back provide a clear distinction between healthy subjects and low back pain patients [4]. In this case the EMG power map reveals characteristic (inter and intra-muscular) spatial patterns although the grid of electrodes covers several muscles. For the further characterisation of such phenomena the investigation of single action potentials and their waveforms is not necessary. It is sufficient to get an estimation for spatial distribution of the force or the force density.

3. SEMG power (spectral) mapping

The force of a muscle is controlled via the recruitment of motor units and via the firing rate of each motor unit. Each activation potential causes fibre twitches and the mechanical system of tendons performs an average about such twitches.

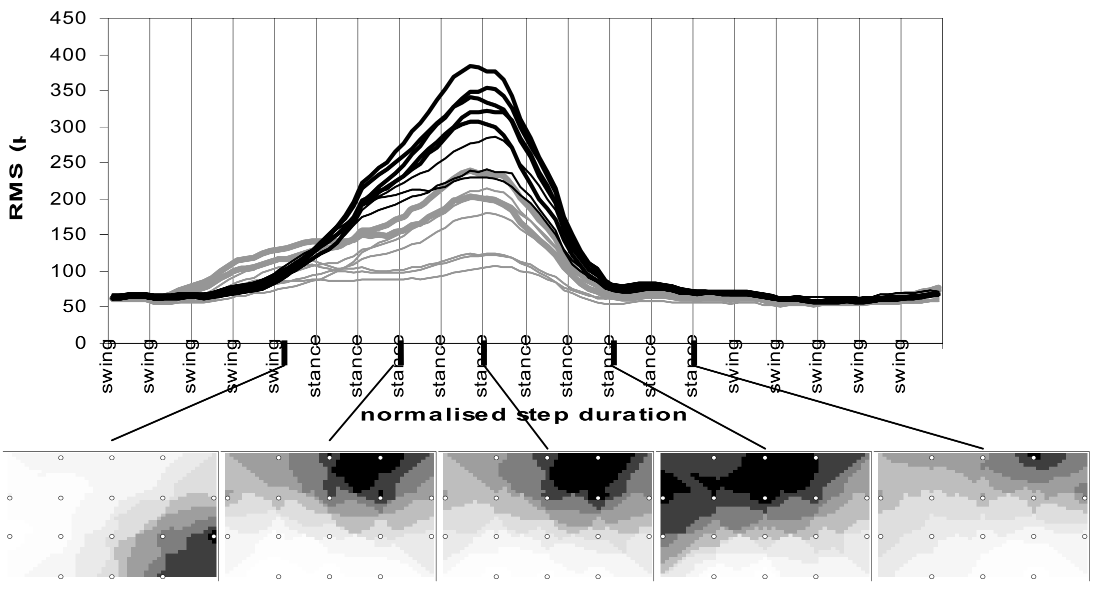

Figure 3: Time course of RMS from the triceps muscle of rats walking in a treadmill. The RMS map sequence clearly shows am inhomogenious activity distribution which depends on the phase of the movement. (These RMS values were calculated with a very short time window: 21ms., see [2])

Therefore, if we want to get a measure for the force, we have to perform an average about some action potentials which are contained in the EMG signal. This average is performed as the calculation of the root mean square with a time window of suitable length (at least the duration of a fibre twitch). If a muscle is covered with a grid of surface electrodes, it is possible to calculate an EMG activity distribution on the skin surface via spatial interpolation. In other words, an EMG power map is a projection of the EMG activity distribution onto the muscle surface, which is weighted (from source depth) and smoothed due to volume conduction. Nevertheless it gives a useful estimation of the force that is generated by each region of the muscle. A geometrical interpretation is possible immediately. The sampling condition which the electrode grid has to fulfil for power mapping can be weaker than for investigations of action potentials, see the simulations in figure 5.

4. Increased spatial resolution with high pass filtered cross covariance functions

Because the spatial activity distribution in the RMS map is filtered or smoothed respectively by volume conduction, some information from the (slowly varying) spatial distribution is lost in SEMG. This behaviour can be improved by temporal high pass filtering of the SEMG signal. But in this case there must be solved the problem how to judge the quality of the signal. One possibility that works up to MVC is the high pass filtering of cross covariance functions, see [6]. The time shift between the channels shows the AP

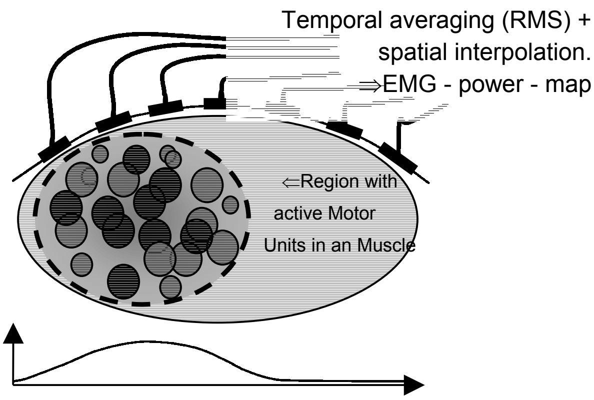

Figure 4: Due to the spatial and temporal averaging which the mechanical system performs about twitching fibres, the interesting force distribution (force density , intra-muscular coordiation pattern) is described by a slowly varying 3D-function.

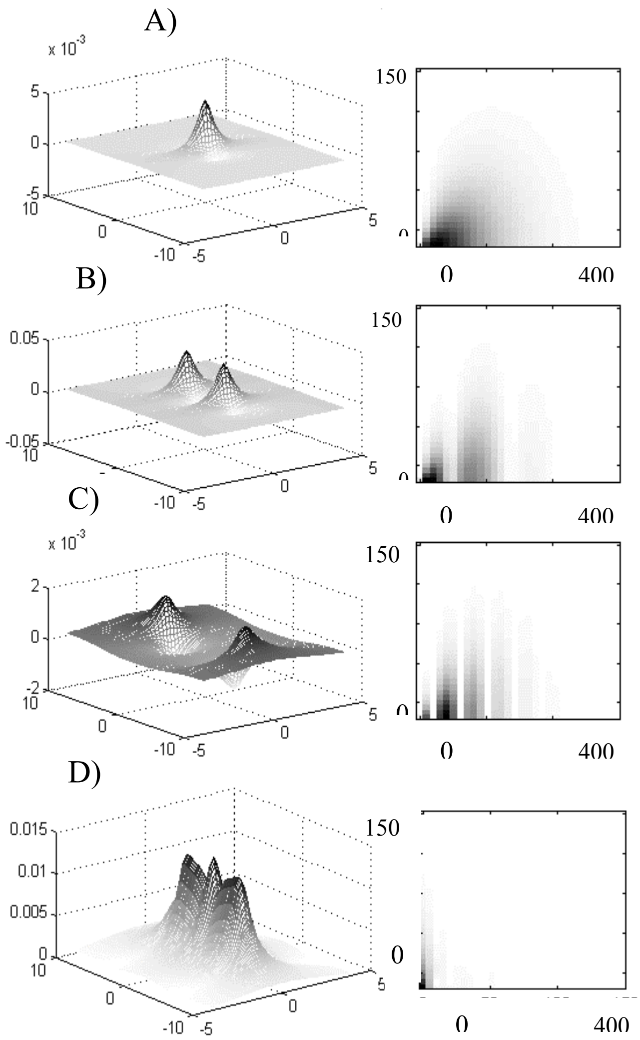

Figure 5: A) arising, B) travelling C) vanishing motor unit action potentials. and their spatial spectra. These spectra contain larger spatial frequencies than that of the RMS (D) of this action potential.

propagation, see Figure 6. A higher load forces the switch on of a further population of MUs with another position of endplates (intramuscular coordination), see Fig. 6B. For the upper column in

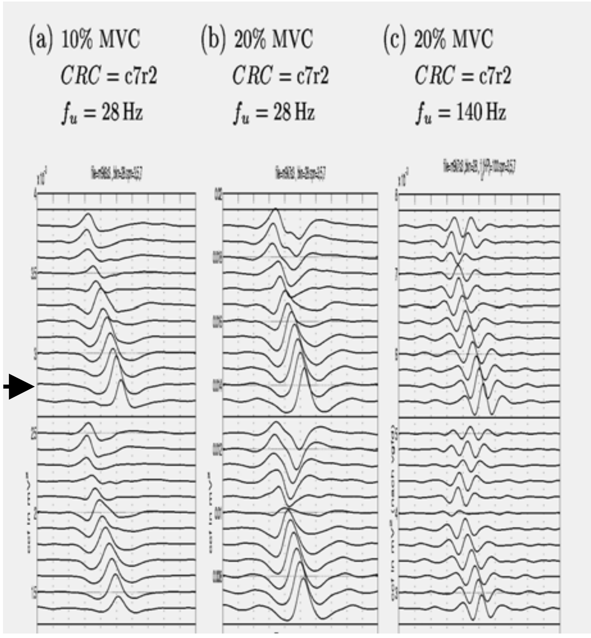

Figure 6: (a), (b) Cross covariance functions between bipolar SEMG channels which were measured at different load levels from the human biceps. The EMG channels were located in two columns parallel to the fibres. "⟶" is the covariance reference channel (CRC). (c) High pass filtered cross covariance functions.

the Diagram this contribution can be suppressed by high pass filtering (6C). But this method does not allow the investigation of single motor units.

Similar to RMS-Mapping the cross covariance method ignores the details of the firing pattern due to its integration time but it gives information about the spatial distribution of activity with higher resolution in cross fibre direction.

5. Conclusions

There is intramuscular co-ordination. Surface EMG (spectral) power mapping is an adequate tool for the investigation of such phenomena, especially for biomechanical problems. Due to the temporal averaging of power mapping, this method has a more generous sampling condition than the investigation (mapping) of action potentials (e.g., via triggered averaging). High pass filtering of cross covariance functions between bipolar SEMG channels increases the spatial resolution of power mapping [6]. It is also a valuable tool for biomechanical investigations.

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Clinical signs and physical function in computer users with and without neck and upper limb complaints - the NEW-study

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1. Introduction

When looking for morbidity contrasts the advantage of using clinical examination is larger than using questionnaires solely, as data from questionnaires is likely to result in a lower precision of the data. Previous studies have shown that workers with self-reported pain have a lower physical capacity than those not reporting pain (1). The aim was to test if workers reporting complaints had more clinical signs and a lower physical capacity than those not reporting complaints and to report the clinical pattern in the two groups.

2. Methods

Elderly female computer users (≥45 years) were recruited from 4 companies in Sweden and Denmark. Based on a questionnaire survey cases were defined as those reporting trouble in either the neck or shoulder >30 days and controls were not allowed to report more than a maximum of 7 days of trouble in both neck and shoulder within the last year, and both groups were not allowed to report more than a maximum of three other body regions with troubles >30 days. A total of 43 cases and 61 controls went through standardised clinical examinations (2) and physical measurements of maximal muscle strength in shoulder elevation, abduction, handgrip, endurance of shoulder elevation and a lifting test (cervical PILE) (3).

3. Results and conclusion

The most frequent diagnoses were trapezius myalgia: 16 cases (38%) and 4 controls (7%), tension neck syndrome: 7 cases (17%) and 1 control (2%) and cervicalgia: 7 cases (17%) and 0 controls. The cases had significant lower muscle strength in both right and left shoulder elevation compared to the controls (right: 59.3±15.5 vs. 70.1±22.9 Nm; left 52.6±15.0 vs. 67.8±25.9 Nm). The remaining physical capacity tests did not reveal any significant difference, though there were tendencies.

The results of the clinical examination showed a larger morbidity contrast than the questionnaire results, and the severity of the disorders could thereby be described. The difference between cases and controls seems to be detected by some of the used physical capacity tests.

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Searching precursors inside EEG and EOG time series in eye blinking

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Abstract *– In this work we try to recover information from the eye blinking activity. In this perspective, the transition from the awake to the asleep state on humans was experimentally investigated and mathematically interpreted by means of the spectral analyses of the time series resulting from processes underlying both the brain activity and the eye dynamics.*
Due to the inherently high non linearities involved, chaotic patterns were likely to occur during the experimental tests. As a consequence, the measured time series were analysed keeping in mind the chaos paradigm too.

1. Blinking activity

Blinking provides the important function of spreading the pre-corneal tear film and maintaining the moist condition of the anterior surface of the eye. Blinking may be defined as a rapid closure of the eyelids [1].

2. Materials and methods

Up to 70 hours of Electroencephalography (EEG) and Electroculography (EOG) activities from five healty subjects were recorded by means of the GALILEOTM acquisition system at a sampling frequency of 256 Hz. For each volunteer, the data collected 60 minutes before the transition phase up to 30 minutes after (Fig. 1) was analysed by means of the windowed fft technique, each window being 8 s long.
In this work we search for precursors that are particular parameters present before a particular event. We follow two procedures: the first one is the "traditional way" with spectral analyses, and the second one is a more "innovative way" with dynamical analyses. Now we report the results of these two different approaches.

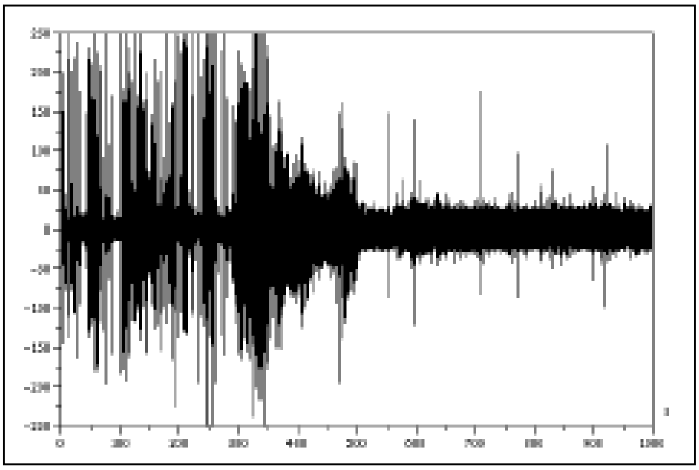

FIG. 1 : Subject 3 : awake – sleeping transition in the EOG.

3. Spectral analyses

First we analyzed the EOG time series with spectral analysis. The first question to be answered was "What type of frequency range are we dealing with?". With FFT (Fig. 2) we found that the highest frequency was approximately 30 Hz. Moreover, most of the energy is concentrated in the frequency range below 10 Hz. In order to look for the time evolution of the spectral content, we computed the so-called spectrogram, a tool commonly used in the analysis of dynamical systems in many fields like mechanics, geophysical phenomena [2] and neurophisiologic signals. As time evolves on the horizontal axis, each column represents the spectral energy distribution within the most interesting frequency range (0-10 Hz) by using different colours which indicate different percentage of energy in each frequency bin: in decreasing order red, yellow, green and blue. The vertical black line indicates the time of the transition from awake to sleeping state. From the spectrogram (Fig. 3) it can be clearly seen that before the transition there are two competitive phenomena. The first one is a general decrease of the frequency of the most energetic peak (indicated by the red colour). However, at the same time there is also a simultaneous spreading of the spectral energy over a much wider frequency range.

These two changes can be better summarized and highlighted by computing two scalar parameters derived from the spectrogram: the maximum frequency and the mean frequency. The maximum frequency is computed evaluating, for each single column representing 8 seconds of signal analyzed, the frequency bin for which the spectral amplitude is maximum. On the other hand the mean frequency is the weighted mean of the overall frequency distribution of each column of the spectrogram, i.e. a kind of COM of the spectrum. Looking again at the time evolution of these two parameters, we find that the maximum frequency (Fig. 4) and the mean frequency (Fig. 5) show opposite behaviour before the transition, highlighting both the decrease of the main spectral peak (Fig. 4) and the widening of the spectral content to higher frequencies (Fig. 5).

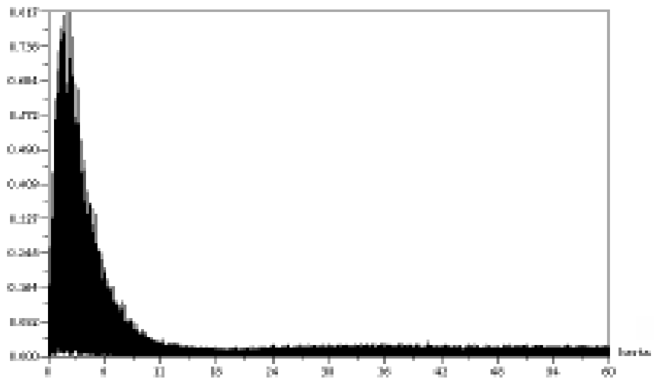

FIG. 2 : Subject 2 : spectral analyses with FFT.

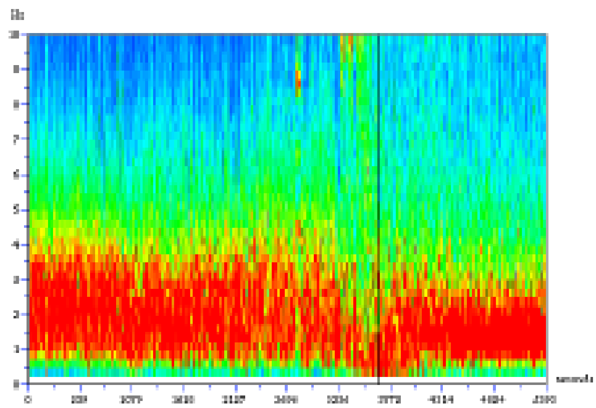

FIG. 3 : Subject 5 : spectrogram showing the time evolution of the spectral content.

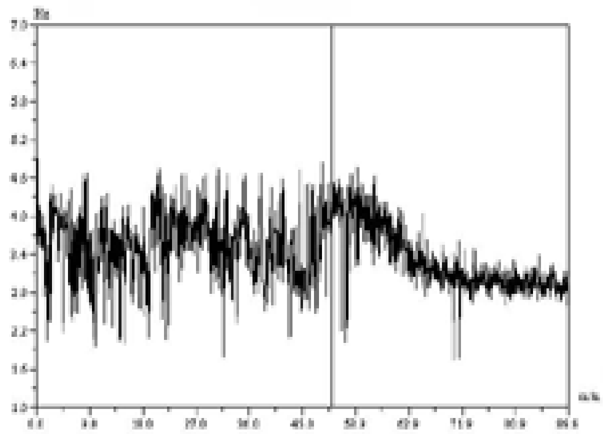

FIG. 4 : Subject 1 : time evolution of the maximum frequency.

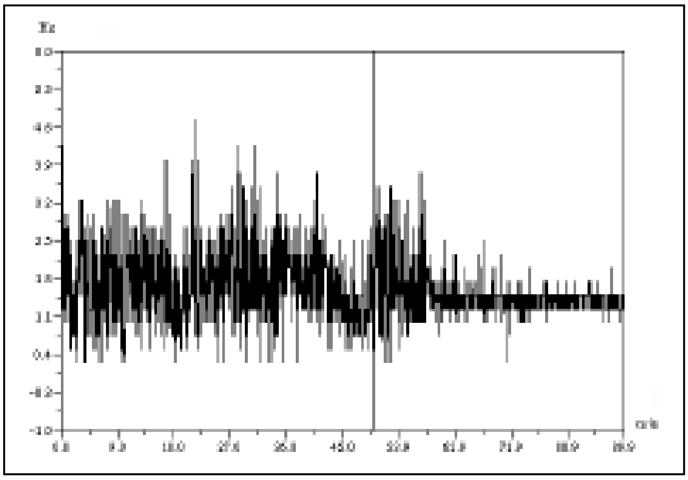

FIG. 5 : Subject 1 : time evolution of the mean frequency.

4. Dynamical analyses

The time series are difficult to study because the system that they represent is non linear. A dynamical system is one that evolves with time. Its state at any time is represented by a vector of state variables. We may denote an experimental time series with {xk} = {x(kτs)}, where N is the length of the series, k = 0, … ,N and τs is the sampling interval. The original dynamical system is continuous. However, when sampling, we are passing from the continuous dynamical system to a discrete one. The relation that links the time series to the original continuous dynamical system equation s'(t) = f(s(t)) is given by xk = h(s(kts)), where h is a measuring function, and s is the original state variables vector. The idea is that although we cannot obviously reconstruct the original attractor, we can try to reconstruct an embedding space where another attractor lies preserving the invariant characteristics of the original one. The space we reconstruct has a dimension m that is named embedding dimension; in general m is different from the unknown dimension [d] +1 if d is the dimension of the attractor.

A single time series (e.g. EOG data) is used to reconstruct an attractor with the delay coordinates method [3]. This method requires to know both the delay time τ, that is a multiple of the sampling interval τs, and the embedding dimension m (Fig. 6) [4]. The most common way to choose the delay time makes use of the autocorrelation function. This function is calculated by comparing the original time series and a delayed time series, where the second series is obtained by delaying the first series by τ. Repeating this computation for τ varying from 0 to a given τmax, the choice of the optimal delay time for the reconstruction goes to the first τ for which the autocorrelation is negative, i.e. the value of τ after the first zero crossing. This method induces an independence between the coordinates, which are guaranteed to be linearly uncorrelated (Fig. 7).

The other procedure is to detect the first minimum of the mutual information (redundancy) [5]. Infact the mutual information answers the question: “given x measured at time t, which is the amount of information that we have about x measured at time t+τ?”. In this case, the chosen τ guarantees a non-linear independence between the reconstructed coordinates.

The last step is to characterize the embedding dimension m with the method of False Nearest Neighbours [6]. The procedure identifies the so-called “false neighbours”, i.e. points that appear to be near only because the embedding space is too small. When, by increasing the embedding dimension, the percentage of FNN goes to zero for a certain value of m, the attractor is completely unfolded in that dimension, which is chosen as the optimal one for the reconstruction(Fig.8). As we did with the spectral parameters, also in this case the dynamical analyses are applied to 8-seconds long windows of the time series in order to highlight possible time variations of the computed parameters. Once again variations seen before the awake–asleep state transition can be regarded as precursors. An example is the evolution of the mutual information (Fig. 9). The time evolution of the other dynamical parameters is currently under investigation.

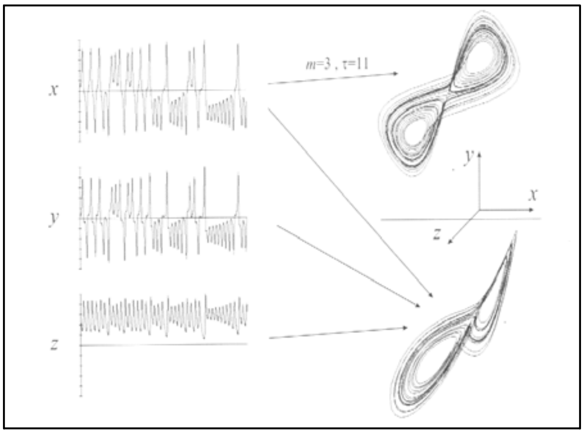

Fig. 6 : Reconstruction of an attractor using the delay coordinates method [4].

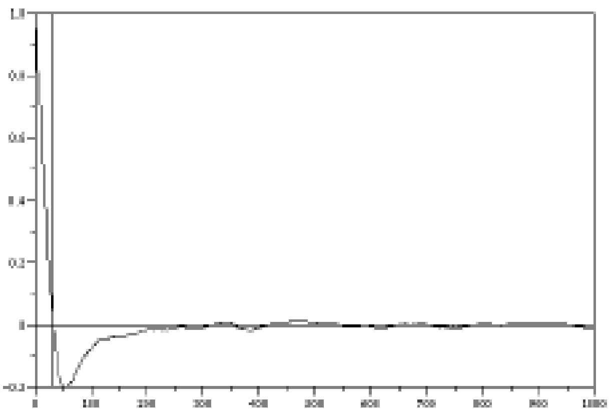

FIG. 7 : Subject 3 : values of the autocorrelation function for the first 1000 τs.

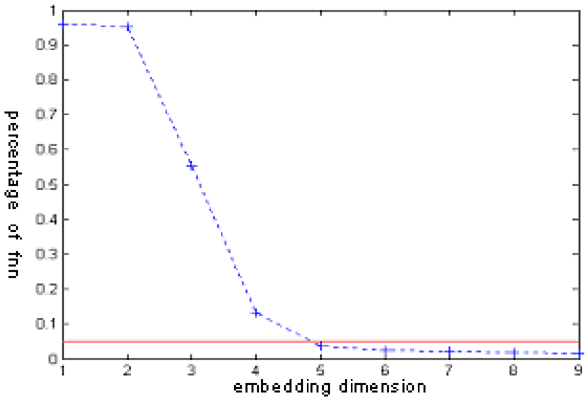

FIG. 8 : Subject 4 : percentage of FNN as a function of the embedding dimension.

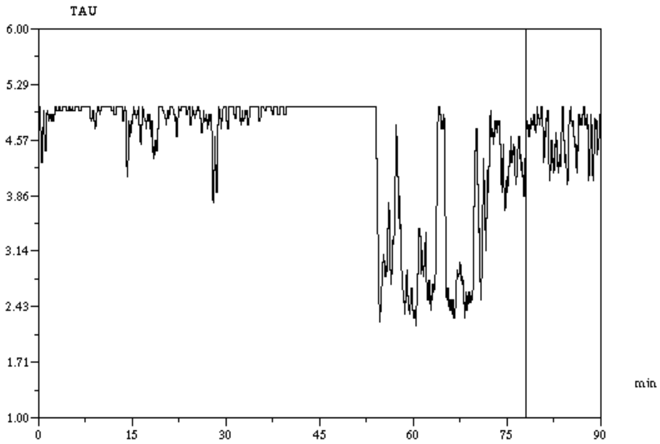

FIG. 9 : Subject 3 : time evolution of the first minimum of the mutual information.

5. Results

After studying the time series close to the awake–asleep state transition, we can indicate first results about the asleep state precursors. Blinking activity is less frequent and its amplitude decreases. There is also a redistribution of the spectral energy, with a decrease of the maximum frequency and an increase of the mean frequency with a simultaneous widening of the spectrum.

At the end of this study we have characterized the dynamical system using the chaos paradigm and we have found delay times near 30 τs with autocorrelation, delay time near 5 τs with mutual information and an embedding dimension of 5.

We can also apply spectral analyses and dinamical analyses in pathologies to found the evolution trend e.g. notion deficit and Parkinson desease.

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Tissue oxygenation and intramuscular pressure during dynamic and intermittent static muscle contractions in m. biceps brachii

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1. Introduction

Dynamic muscle contractions cause greater energy turnover and fatigue than static contractions performed at equal force level (Bridges *et al.* 1991, Vedsted *et al.* 2003). The greater fatigue may be due to greater reduction in tissue oxygenation.

1.1. Objective

To test if the reduction in tissue oxygenation is larger during dynamic than static low force contractions at identical time-tension development.

2. Methods

Four subjects performed elbow flexions for 1 min in 3 contraction modes resulting in identical time-tension products at two force levels (LOW/HIGH): 1) intermittent static (IST, 4 s contraction – 4 s rest) at 10 and 20 %MVC, 2) dynamic (DYN, 2 s conc – 2 s ecc – 4 s rest) at 10 and 20 %MVC (20° elbow movement), 3) continuous static (CST) at 5 and 10 %MVC. Force, bipolar surface electromyogram (EMG), intramuscular pressure (IMP) and reduction in tissue oxygenation from rest (TO_2) were measured simultaneously from m. biceps brachii.

2. Results

Mean TO_2 in IST were 4.9 and 8.2% and in DYN 5.5 and 6.1% in LOW and HIGH, respectively. Values in CST were 5.0 and 9.1%. Mean EMG_{RMS} of the contraction phase in DYN was generally higher than IST (0.08 vs. 0.05 and 0.14 vs. 0.09 mV in LOW and HIGH, respectively). Values in CST were 0.03 and 0.05 mV. Mean IMP in DYN was lower than IST (5.7 vs. 41.5 and 12.3 vs. 57.0 mmHg in LOW and HIGH, respectively). Values in CST were 24.0 and 33.3 mV.

3. Conclusion

Dynamic contractions did not result in larger TO_2 decreases in spite of larger EMG activity that may indicate larger energy turnover. Larger mean blood flow due to lower IMP during DYN than static contractions may play a role.

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Blind decomposition methods: application to surface EMG

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Abstract – *Three novel algorithms for the surface electromyograms (SEMG) decomposition are revealed. They were mainly tested on synthetic signals, but also a few experiments were done on real SEMGs. Two of them, utilizing the second- and third-order statistics of SEMGs, directly decompose motor-unit action potentials (MUAPs), which proved rather unreliable. The third approach is based on the inverse correlation of SEMGs, directly extracts the innervation pulse trains (IPTs), and gives good results also with real SEMGs.*

1. Introduction

In electromyography (EMG), one of the most challenging issues having been tackled since a long time certainly remains the EMG decomposition to its constituent components, i.e. to the motor-unit action potentials (MUAPs) and to the innervation pulse trains (IPTs). While the surface EMG (SEMG) measurements may be considered compound signals which are generated by statistically pretty independent signal sources, i.e. motor units (MUs), at least in low contraction force conditions, a variety of the so called blind decomposition methods seem applicable.

The blind source separation (BSS) [1] can be applied to the problems modelled as multiple-input multiple-output (MIMO) systems. These methods utilize statistical properties of the observed signals, i.e. SEMGs in our case. In mathematically exact and noise-free cases, the signal sources with a complete statistical independence (orthogonal sources) could be separated thoroughly and without error. In the SEMG case, this would mean exact reconstruction of IPTs and MUAPs. However, in reality we are forced to implement only the sample estimates, which introduces the first source of errors in our decomposition procedure. Three additional error sources appear when dealing with SEMG: the signals are noisy, the MUs can fire simultaneously – so the IPTs are no longer orthogonal, and the MUAPs can change their shape during a measurement session – they are nonstatoinary. Hence, although the available blind decomposition methods might allude to a quick SEMG decomposition solution in a close form, the reality necessitates much more sophisticated iterative optimisations.

2. Blind decomposition approaches

In the NEW project, we first tackled two statistical approaches, one based on convolutive BSS [2] and the other on third-order cumulants [3]. MIMO modelling was applied and the SEMG measurements were originally decomposed to the constituent MUAPs.

Our BBS-based method calculates first the shrunk pseudo Wigner-Ville distribution as follows:

$$\mathbf{PWV}_{y_{i1}y_{i2}}(t,f)=y_{i1}(t+\frac{1}{2})y_{i2}(t-\frac{1}{2})e^{-j2\pi f}\,, \quad (1)$$

The results of Eq. (1) are then scanned for most probable single auto-terms, i.e. the signal sources which happen to fire alone in a certain time instant [4]. These auto-terms are further "averaged" through the process of joint diagonalisation [5] giving a unitary matrix $\mathbf{U}$ from which the estimated MUAPs are obtained in $\hat{\mathbf{H}}$:

$$\hat{\mathbf{H}}=\mathbf{W}^{\#}\mathbf{U} \quad (2)$$

and $\mathbf{W}^{\#}$ stands for pseudoinverse of whitening matrix $\mathbf{W}$ [1].

Similarly, our HOC-based method utilizes the so called w-slices [6], where the weights $\mathbf{w}$ are calculated as follows:

$$\mathbf{w}=\mathbf{S}_a^{\#}\mathbf{D}=\mathbf{S}_a^{\#}\begin{bmatrix}\mathbf{E} & \cdots & \mathbf{0}\\ \vdots & \ddots & \vdots\\ \mathbf{0} & \cdots & \mathbf{E}\end{bmatrix}, \quad (3)$$

where $\mathbf{E}$=[1,0,…,0] and is of the length equal to the MUAPs duration, while $\mathbf{S}_a^{\#}$ stands for pseudo-inverse of matrix $\mathbf{S}_a$ which contains the anticausal 1D slices of third-order cumulants of the SEMG measurements. These weights are used in computation of the searched for MUAPs, which utilizes the causal parts of the cumulants [6]:

$$\hat{\mathbf{H}}=\mathbf{S}_c\mathbf{w}\,. \quad (4)$$

As the BSS-based approach tries to get rid of different decomposition errors by single auto-term search and joint diagonalisation, our HOC-based algorithm is from the same reason upgraded by an iterative gradient-ascent optimisation [3].

The methods which make a decomposition to MUAPs first are rather unreliable. Therefore, we developed a novel approach which relies upon the inverse of correlation (IC) matrix of measurements. The so called activity index

$$\begin{aligned}Ind(t)&=\mathbf{y}(t)^T\mathbf{R}_{\mathbf{y}}^{-1}\mathbf{y}(t)=\\ &=\mathbf{s}(t)^T\mathbf{H}^T(\mathbf{H}^T)^{-1}\mathbf{R}_{\mathbf{s}}^{-1}\mathbf{H}^{-1}\mathbf{H}\mathbf{s}(t)=\mathbf{s}(t)^T\mathbf{R}_{\mathbf{s}}^{-1}\mathbf{s}(t)\end{aligned} \quad (5)$$

clearly shows that the introduced computation cancels the contributions of the MUAPs. Designations $\mathbf{R}_{\mathbf{y}}$ and $\mathbf{R}_{\mathbf{s}}$ stand for the correlation matrices of measurements and signal sources, respectively, while the source signals themselves are indicated by $\mathbf{s}(t)$. The index from Eq. (5) depends solely on the signal sources, i.e. the IPTs. In the ideal circumstances, it would exactly indicate the time instants where individual MUs fire. Its shape would be stepwise and the height of the steps proportional to the sum of the inverse energy of superimposed sources. Of course, in the real SEMG recordings all the error sources highly degrade the ideal picture. Therefore, an additional advanced averaging of the obtained IPTs is necessary in order to eliminate the influence of errors to such a level that the innervation pulses become clearly discernable [7]. The IPT decomposition is based on the following relationship:

$$\mathbf{s}_{t_0}(t)=\mathbf{y}(t_0)^T\mathbf{R}_{\mathbf{y}}^{-1}\mathbf{y}(t) \quad (6)$$

where $\mathbf{s}_{t_0}(t)$ stands for the IPTs which fire at time instant t_0. The MUAPs are reconstructed by spike triggered sliding window averaging technique, using the detected MU firings as triggers.

3. Examples of the SEMG decomposition

Described decomposition methods were applied to both the synthetic [8] and experimental signals from the dominant biceps brachii of 5 healthy male subjects (age 27.8 ± 2.4 years, height 177.2 ± 4.5 cm and weight of 70.6 ± 4.9 kg). In both cases surface EMG signals were detected by an array of 13×5 electrodes (without the four corner electrodes, columns aligned with the muscle fibres) of size 1×1 mm and of inter-electrode distance 5 mm. Experimental EMGs were recorded in LISiN laboratory at Politecnico di Torino, Italy, at isometric voluntary contractions sustained at 5 % and 10 % MVC. Synthetic signals were used to evaluate the influence of the number of active MUs (3, 5, 10 and 20), and of signal-to-noise ratio (SNR). Decomposition results from synthetic signals are depicted in Table 1 and Figs. 1, 2a and 3. Figs. 3b and 4 illustrate the representative results on experimental signals.

4. Discussion and conclusion

The best results were obtained by the IC-based method. In the case of 5 and 10 active MUs almost perfect reconstruction of simulated pulse trains was achieved down to the SNR of 10 dB (Table 1, Fig. 3b). In the case of 20 MUs only a half of active sources were accurately reconstructed. Also the results from the experimental part are encouraging. On average 5±1.0 (mean ± standard deviation) and 7±1.6 MUs were identified during the contractions at 5% and 10% MVC, respectively (Fig. 4b).

TABLE 1: Normalized number (mean ± standard deviation) of accurately recognized pulses (T+), and of misplaced pulses (T-) in the sources reconstructed from the synthetic surface EMG signals (10 active MUs, SNR = 10 dB) by BBS-based method (BSS) and IC-based method (IC).

SNR [dB]		20	15	10	5	0
BSS	T+ [%]	100.0 ±0.0	99.02 ±0.89	96.12 ±2.21	90.04 ± 3.53	81.49 ±6.75
	T- [%]	0.0 ±0.0	0.34 ±0.26	1.07 ±0.96	4.71 ±3.31	9.97 ±7.32
IC	T+ [%]	100.0 ±0.0	100.0 ±0.0	100.0 ±0.0	98.35 ± 0.81	91.62 ±5.26
	T- [%]	0.0 ±0.0	0.0 ±0.0	0.0 ±0.0	0.57 ±0.63	2.83 ±3.64

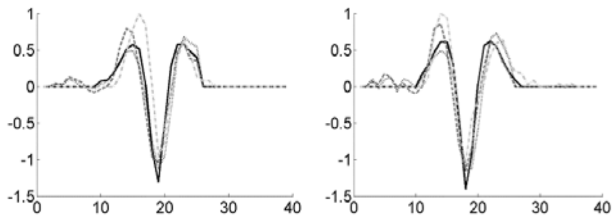

FIG. 1: The MUAPs decomposed from synthetic surface EMG signals (3 active MUs) by our HOC-based method. The original MUAPs are depicted solid, the decomposition results in the noise-free condition dashed, the results at SNR = 10 dB dotted and at SNR = 0 dB dash-dotted.

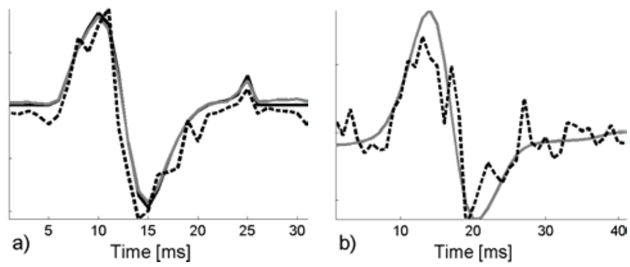

FIG. 2: The MUAPs identified by our BBS-based method (dashed) and the IC-based method (grey), from a) synthetic surface EMG signals (10 active MUs, SNR = 10 dB, original MAUP shape is depicted black-solid), and b) surface EMG recorded during an isometric 10 % MVC measurement of the right biceps brachii of healthy male subject (age 25 years, height 170 cm, weight 63 kg).

Although successfully decomposing the synthetic signals (Table 1, Figs. 1, 2a and 3a), the BSS- and HOC-based methods proved less reliable on the experimental signals (Figs. 2b and 3b). Finally, HOC-based method suffers from high computational complexity what makes it unsuitable for the decomposition of the surface EMG signals recorded at higher contraction levels.

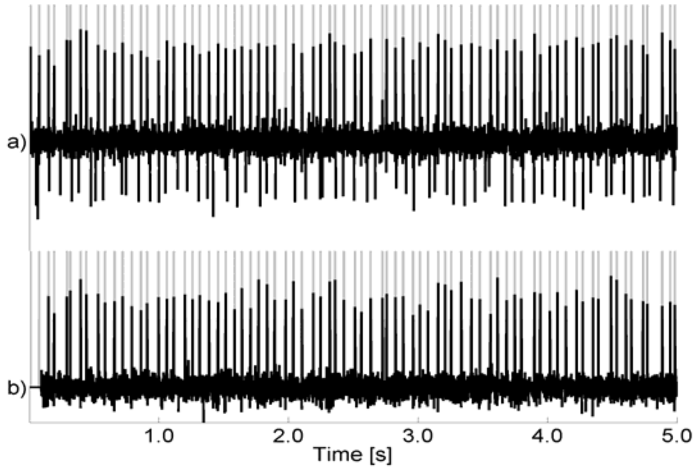

FIG. 3: IPT detected from synthetic surface EMG signals (10 active MUs, SNR = 10 dB) by a) BBS-based method (black), and b) IC-based method (black). Original IPT is depicted in grey.

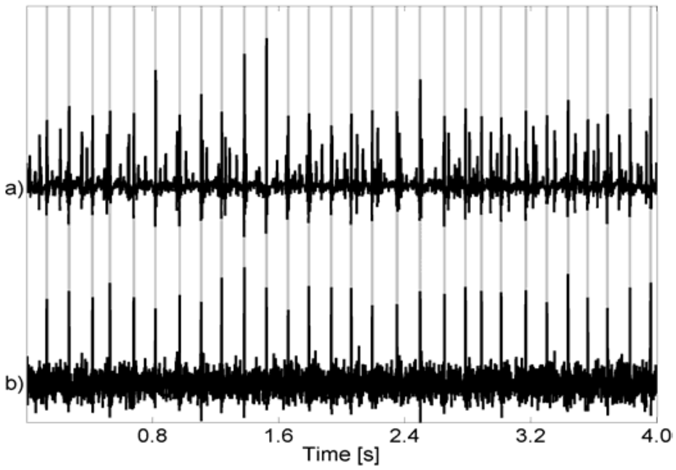

FIG. 4: IPT from real surface EMG recorded during an isometric 10 % MVC measurement of the right biceps brachii of healthy male subject (age 25 years, height 170 cm, weight 63 kg) as detected by a) BBS-based method, and b) IC-based method.

Acknowledgement

The authors are sincerely grateful to Prof. Roberto Merletti, Dr. Dario Farina and all the members of the LISiN laboratory at Politecnico di Torino, Italy, for the design and implementation of the experimental protocol. This work was supported by the European Shared Cost Project "Neuromuscular assessment in the Elderly Worker" (NEW) (Contract No. QLK6-2000-00139).

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Blind separation of convolutive mixtures of surface EMG signals

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Abstract *– This study deals with the application of a Blind Source Separation (BSS) method for separating surface EMG signals recorded from two closely located muscles. The detected signals are considered convolutive mixtures of the signals generated by the two muscles. We compare these results to those obtained considering linear instantaneous mixtures. The study is based on simulations of surface EMG.*

1. Introduction

Crosstalk is one of the major open issues in surface EMG detection. Crosstalk is the surface EMG signal generated by one muscle and detected over the skin surface of another muscle. It may seriously limit the applications in zones where several muscles are close to each other, such as in the forearm. In this case, a mixtures of several signals is recorded. To analyse the activity of each muscle separately, it is then necessary to separate each muscle's contribution (the "sources") from the recorded signals (the "observations"). It can be practically assumed to have more detection points than muscles (sources) since surface EMG amplifiers may currently detect hundreds of signals. As many unknown physiological parameters are involved in the activation of the different muscles, a possible solution to separate the signals is to use a blind approach.

Recent work in this direction [1] has led to promising results. In particular, Lebrun [1] applied a Blind Source Separation (BSS) algorithm (termed SOBI) to separate surface EMG signals recorded from two muscles, assuming a linear instantaneous combination of the sources. However, due to the filtering effect of the volume conductor, the assumption of linear instantaneous mixtures is not strictly valid, which limits the performance. In this work, we will extend the results reported in [1] by applying a BSS algorithm designed for convolutive mixtures, with the aim of improving the performance of the source separation.

2. Blind separation of convolutive mixtures

2.1 Model and method

BSS consists in recovering unknown signals, called sources, from the observation of their mixtures. The sources are not directly observed, and the structure of the mixtures is unknown.

The following model is considered:

$$x[t] = H(0)s[t]+\ldots+H(L)s[t-L]+n[t] \quad (1)$$

where

- $x[t] = [x_1[t], \ldots, x_m[t]]^T$ are the m observations, convolutive mixtures of the sources
- $s[t] = [s_1[t], \ldots, s_n[t]]^T$ are the n sources, assumed mutually independent, with m>n (over-determined mixtures)
- n[t] is an i.i.d noise vector, independent of the sources
- $H(k)=\{h_{ij}(k)\}$, k=0,…L are m×n matrices each describing the weight of vector s[t-k]

The key idea of the algorithm is to rewrite the model as a linear instantaneous mixture [2], considering the delayed sources and observations as new sources and observations. Let L' be an integer such that mL' ≥ n(L+L'), and

- $X[t] = [X_1[t]^T, \ldots, X_m[t]^T]^T$
- $S[t] = [S_1[t]^T, \ldots, S_n[t]^T]^T$
- $N[t] = [N_1[t]^T, \ldots, N_m[t]^T]^T$
- $A=\begin{bmatrix} A_{11}\cdots A_{1n} \\ \vdots \quad \vdots \\ A_{m1}\cdots A_{mn}\end{bmatrix}$

where $X_i[t] = [x_i[t],\ldots x_i[t-L'+1]]^T$, i=1,…,m

$S_i[t] = [s_i[t],\ldots s_i[t-(L+L')+1]]^T$, i=1,…,n

$N_i[t] = [n_i[t],\ldots n_i[t-L'+1]]^T$, i=1,…,m

$$A_{ij}=\begin{bmatrix} h_{ij}(0)\cdots h_{ij}(L) & 0 & \cdots & 0 \\ 0 & \ddots & \ddots & \ddots & \vdots \\ \vdots & \ddots & \ddots & & \ddots & 0 \\ 0 & \cdots & 0 & h_{ij}(0)\cdots h_{ij}(L)\end{bmatrix}$$

The problem can be stated as:

$$X[t] = AS[t] + N[t] \quad (2)$$

which is a linear instantaneous mixture problem, that can be solved more easily than the convolutive one. The algorithm is based on two main steps [3] :

1. Whitening of the observations
2. Rotation of the whitened observations

Since the sources are recovered up to unknown filters, a blind deconvolution step is then necessary to recover the original sources [4], [5].

2.2 Whitening of the observations

The whitening matrix W satisfies the relation $WAR_{SS}[0]A^HW^H = I_{n(L+L')}$, where $R_{SS}[0]$ is the autocorrelation matrix of the sources. To obtain W, the autocorrelation matrix of the observations should be diagonalized:

$$R_{XX}[0] = AR_{SS}[0]A^H+\sigma^2 I_{mL'}$$

Denoting $\lambda_1,\ldots, \lambda_{mL'}$ the eigenvalues sorted in decreasing order, and $h_1,\ldots,h_{mL'}$ the associated eigenvectors, an estimation of σ^2 (noted $\hat{\sigma}^2$), and of the whitening matrix (noted $\hat{W}$) ([3]), is

$$\hat{W} = \left[\left(\lambda_1 - \hat{\sigma}^2\right)^{-\frac{1}{2}} h_1,\ldots\left(\lambda_{n(L+L')} - \hat{\sigma}^2\right)^{-\frac{1}{2}} h_{n(L+L')}\right]$$

obtained as follows:

2.3 Rotation of the whitened observations

The aim of the rotation step is to find a unitary matrix U such that $U=WAR_{SS}[0]^{\frac{1}{2}}$. For this purpose, the covariance of the denoised whitened observations is considered:

$$R_{ZZ}[\tau]=U\left(R_{SS}[0]^{\frac{-1}{2}}R_{SS}[\tau]R_{SS}[0]^{\frac{-H}{2}}\right)U^H$$

As $R_{SS}[0]^{\frac{-1}{2}}$, $R_{SS}[\tau]$, and $R_{SS}[0]^{\frac{-H}{2}}$ are block-diagonal matrices, U block-diagonalizes $R_{ZZ}[\tau]$ for all τ. By block-diagonalizing a set of matrices $R_{ZZ}[\tau]$ corresponding to several τ (joint block-diagonalization, JBD), U can be obtained in a robust way. To assess the performance of the JBD, the index defined in [5] will be used.

The sources are recovered up to unknown filters [5] :

$$\hat{S}[t] = U^H W X[t] \quad \Rightarrow \quad \hat{S}[t] = F S[t]$$

where F is a block-diagonal matrix with (L+L') blocks of dimensions (L+L')×(L+L').

2.3 Blind deconvolution

The last step consists in a blind deconvolution, using the noise subspace method [4],[6], to recover each source $s_i[t]$, i=1,…,n.

3. Application to sEMG

Surface EMG signals generated by two muscles have been simulated by a structure-based EMG model. The algorithm described in the previous section as well as the SOBI method [3] have been applied to these simulated signals.

3.1 The model

The simulations focused on two muscles located in the forearm, the pronator teres and the flexor carpi radialis. These muscles are small with the motor units concentrated in a small volume. In this case, the filtering effect of the tissues is approximately the same for all the sources.

The characteristics of the two muscles are reported in Table 1.

TAB. 1 : Characteristics of the forearm and the muscles

Forearm	**Radius of the forearm**	50 mm
	Radius of the bone	20 mm
	Skin layer	1 mm
	Fat layer	2 mm
Muscles	**Shape of the muscles**	Elliptic
	First axix	8 mm
	Second axix	20 mm
Fibers	**Number of fibers per muscle**	16000
	Length of the fibers	90 mm
	Center of the inervation zone	22.5 mm

The characteristics of the simulated detection system are reported in Table 2.

TAB. 2 : Characteristics of the detection system

Detection system	**Type of electrodes**	rectangular
	Size	5×1 mm
	Configuration	Single Differential

The number of detection systems varied in the range 5-10.

3.2 Simulations

The simulated surface EMG signals were stationary.

Two preliminary simulation analyses [4] showed that the performance of the method is enhanced by having:

- at least 5 detection points; otherwise the JBD step is not efficient enough.
- the two sources not completely overlapped in the frequency domain (i.e., with exactly the same spectra).
- the two muscles "sufficiently" far from each other.

From these considerations, we investigated the configuration shown in Figure 1.

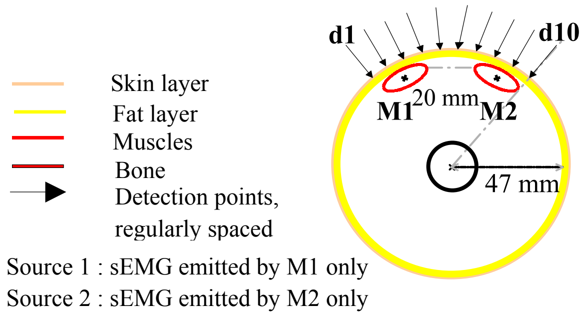

FIG. 1 : Configuration of the simulation.

3.3 Results

We applied both the SOBI algorithm and the convolutive one to the simulated mixtures, using L = 1 to 3 in the convolutive algorithm of Eq. (1), and considering 5 (d_1,d_3,d_5,d_8,d_{10}) or 10 observations. To assess the performance of the algorithms, we computed the correlation coefficients between the original sources and the recovered ones ($corr_1$ for M1, $corr_2$ for M2), and we computed the JBD criterion for the convolutive algorithm. Results are shown in Table 4.

The simulation parameters are shown in Table 3.

TAB. 3 : Parameters used for generation of the sEMG

MUAP library	**Number of fibers**	16000
	Number of MUs	70
Common features for the 2 muscles	**Recruitment of the last MU**	75% MVC
	Distribution of the MU territories	Uniform
	MU CV distribution	Random gaussian
	Signal duration	2 s
S1 (M1)	**Simulated Force**	80% MVC
	MUs selected	70 MUs
S2 (M2)	**Simulated Force**	55% MVC
	MUs selected	70 MUs
S3 (M1 or M2)	**Simulated Force**	55% MVC
	MUs selected	34 deapest MUs
S4 (M1 or M2)	**Simulated Force**	55% MVC
	MUs selected	36 superficial MUs

TAB. 4 : Results of the BSS for two sources

	m	5			10		
	L	1	2	3	1	2	3
Complete overlapping S1 and S2	JBD	0,86	0,81	0,78	0,82	0,92	0,90
	$Corr_1$	0,70	0,53	0,49	0,80	0,83	0,19
	$Corr_2$	0,70	0,39	0,38	0,74	0,62	0,31
	$Corr_1$	**0,97**			**0,94**		
	$Corr_2$	**0,86**			**0,95**		
Partial overlapping S1 and S4	JBD	0,86	0,80	0,77	0,84	0,94	0,91
	$Corr_1$	0,84	0,54	0,37	0,80	0,35	0,26
	$Corr_2$	0,44	0,75	0,76	0,35	0,31	0,56
	$Corr_1$	**0,97**			**0,94**		
	$Corr_2$	**0,95**			**0,94**		
Partial overlapping S1 and S3	JBD	0,89	0,79	0,78	0,87	0,88	0,82
	$Corr_1$	**0,84**	0,53	0,36	**0,79**	0,48	0,53
	$Corr_2$	**0,82**	0,28	0,62	**0,87**	0,22	0,33
	$Corr_1$	0,93			0,94		
	$Corr_2$	***0,03***			***0,06***		
Partial overlapping (S3 and S4	JBD	0,88	0,78	0,77	0,90	0,95	0,93
	$Corr_1$	**0,88**	0,26	0,54	**0,88**	0,33	0,71
	$Corr_2$	**0,90**	0,64	0,80	**0,58**	0,08	0,24
	$Corr_1$	***0,07***			***0,13***		
	$Corr_2$	0,99			0,99		

When there was a complete overlapping, or a partial overlapping including the superficial MUs, the multiplicative algorithm was more efficient than the convolutive one, although in the convolutive case, the JBD criterion was rather good. This may be due to an overestimation of L, which implies that the mixtures were not "convolutive enough".

When there was no overlapping, or a partial overlapping including the deepest MUs, the convolutive algorithm provided better results than the SOBI one.

4. Conclusion

The idea to use a convolutive mixture model seems promising. Indeed, in some conditions, the convolutive approach was better than the linear instantaneous one. However, the critical point is to understand the limit of the linear instantaneous approach, that is to understand when the mixtures are "sufficiently convolutive".

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A portable multichannel EMG acquisition system for long term recording

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Abstract *– The use of mono- and bi- dimensional EMG electrode arrays for the assessment of the neuromuscular system provides an insight on muscle physiology that could not be achieved with classical bipolar surface EMG. Among the advantages of multichannel EMG detection there are the possibility to estimate muscle conduction velocity even during highly non-stationary motor tasks. For these reasons, the development and use of multichannel surface EMG devices and techniques was chosen as the primary tool for the assessment of the neuromuscular condition in elderly workers within the European RTD Project "Neuromuscular Assessment on the Elderly Worker" ("NEW"). A wearable, battery-powered, multichannel EMG acquisition system (datalogger) has been purposely developed in the framework of the project. The system is comprised of adhesive electrode array, a wearable acquisition unit for multichannel EMG recording and storage on removable PCMCIA cards, and an optional USB interface for online signal display on a PC. The system has been extensively used within the Project NEW to acquire data from more than 300 subjects.*

1. Introduction

The use of mono- and bi- dimensional EMG electrode arrays for the assessment of the neuromuscular system provides an insight on muscle physiology that could not be achieved with classical bipolar surface EMG. Among the advantages of multichannel EMG detection there are the possibility to estimate muscle conduction velocity during both isometric or (by means of purposely developed techniques, [2]), during highly non-stationary motor tasks; to increase the number of detection points on a muscle to improve the performance of pattern-based EMG decomposition methods; and, for classical amplitude and spectral analysis, to allow the selection of the optimal bipolar EMG signal on which the analysis will be performed, using a criteria of minimum sensitivity to displacement.

For these reasons, the development and use of multichannel surface EMG devices and techniques was chosen as the primary tool for the assessment of the neuromuscular condition in elderly workers within the European RTD Project "Neuromuscular Assessment on the Elderly Worker" ("NEW"). The specific requirements of the project called for the availability of a user-friendly, small-sized EMG acquisition system for field use, suitable for multichannel EMG recording from one ore more muscles. A market survey pointed out that none of the commercially available EMG acquisition systems featured the desired specifications, nor could be easily adapted to this specific use.

To fill this gap, a wearable, battery-powered, multichannel EMG acquisition system (*datalogger*) has been developed in the framework of the Project NEW. The system is comprised of: a) adhesive electrode arrays for artefact-free EMG signal acquisitions during work or other activities, b) a wearable, user-friendly, battery powered acquisition unit for multichannel EMG recording and storage on removable PCMCIA cards, and c) an optional USB interface for connecting the acquisition to a PC for online display of the acquired signals.

The system here described has been extensively used within the Project NEW to acquire data, either on the workplace or in laboratory, from more than 300 subjects.

2. Methods

The system was developed by merging the knowledge of three different institutions. LISiN-, Politecnico di Torino (Coordinator of Project NEW) provided the expertise in the design of multichannel EMG detection, acquisition, and processing. SPES Medica s.r.l., a company in the field of electrode manufacturing, collaborated to the design, patenting and production of the adhesive arrays. Sirio Automazione s.r.l., technical partner of the project NEW, collaborated in the design, small series production and constant upgrade of the prototypes distributed to the partners.

The adhesive arrays have the structure depicted in FIG. 1A [2]. A flexible support carries a linear array of printed silver electrodes and the interconnecting tracks. The electrodes are coated with a layer of silver chloride. The support is covered with a thin plastic layer that isolates the tracks, leaving the electrode surfaces exposed. A small hole is provided laterally to each electrode for the insertion of conductive gel.

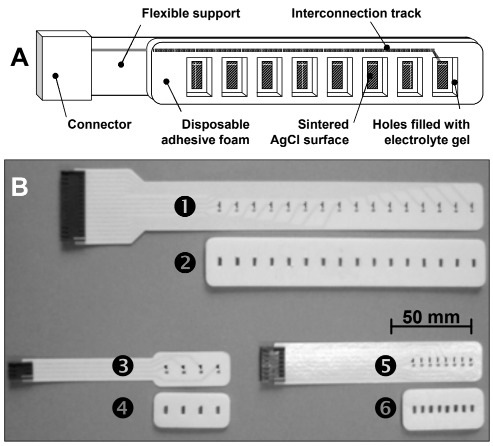

FIG. 1: A) schematic drawing of an adhesive electrode array, constituted by a semi-disposable flexible support and a double-sided adhesive foam. The array is placed on the muscle and the cavities between electrodes and skin are filled with electrolyte gel (see text). Figure modified from: Pozzo M, Farina D, Merletti R, *Electromyography: Detection, Processing and Applications*, in: Moore J, Zouridakis G. (ed), "Biomedical Technology and Devices Handbook", CRC Press, 2003. B) The three different types of adhesive arrays which have been developed: 16 electrodes, 10 mm i.e.d. (❶: semi-disposable flexible support, ❷: disposable adhesive foam); 4 electrodes, 10 mm i.e.d. (❸, ❹); 8 electrodes, 5 mm i.e.d. (❺, ❻).

For each application, a disposable, double-sided adhesive foam with pre-punched windows is placed on the support. The array is placed on the muscle and, by means of a precision dispenser, the cavities between each electrode and the skin are filled with a small quantity (20 ÷ 30 μl) of low density, ECG-type electrolyte gel. With this technology, the array is fixed on the skin with a stable and evenly distributed force; the electrode-skin impedance is lower than in "dry" metallic electrodes and is consistent between electrodes and over time [1]. Furthermore, the electrode-skin contact is stable even during highly-non stationary motor tasks, allowing the acquisition of multichannel EMG signals even during dynamic contractions. The arrays were designed in three types: 16 electrodes and 10 mm inter-electrode distance (i.e.d.), 8 electrodes and 5 mm i.e.d., 4 electrodes and 10 mm i.e.d (FIG. 1B). The 8-electrode, 5 mm i.e.d. array was selected as most suitable for the muscles (upper trapezius, longissimus dorsi and multifidus) investigated within the Project NEW.

The acquisition system is composed by (FIG. 2):

a) a wearable, battery powered, 64-channel acquisition unit (the datalogger itself), equipped with a graphical LCD display with touchscreen, which allows the control of all the functions by means of user-friendly, icon-based menus. Acquired data is stored on a removable, PCMCIA-type memory card, currently available in sizes exceeding 1 GB. Data can be acquired in three files formats (16-, 32- or 64-channels) at a selectable sampling rate of 1024 or 2048 S/S per channel.

b) a number of pre-amplified, 16-channel EMG probes; for each channel, each probe comprises a single differential (SD) input stage, a band-pass filter and fixed gain amplifier. Each probe also includes the 16-bit A/D conversion stage, an impedance monitor that continuously checks the electrode-skin contact, and a digital serial interface over which digital EMG data is transmitted to the acquisition system. Due to its small size, the probe can be easily fixed on the subject; cable splitters were usec to connect two 8-channel arrays to each probe (FIG. 3A and B).

c) an optional, opto-coupled USB interface, that allows the user to connect the datalogger to a PC for online signal display and quality check.

The datalogger incorporates several function, among which an auto-recognition feature that automatically identifies the number of plugged probes and configures the number of acquisition channels accordingly. In addition, a "scope" function allows the user to display the EMG signals, four channels at a time, directly on the LCD screen, allowing the datalogger to be operated as a fully stand-alone device.

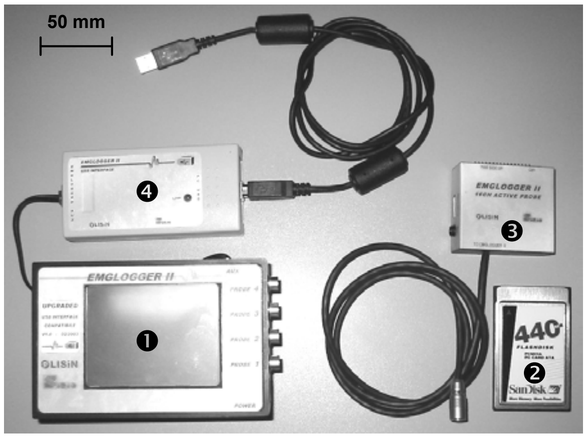

FIG. 2: The acquisition system, composed by: 1) the portable acquisition unit with graphical display interface and touchscreen; 2) a PCMCIA memory card; a 16-channel digital EMG probe; 4) an optional, optically isolated USB interface for online signal display on a PC.

A dedicated software (Figure 3C) has been developed for off-line signal review, filtering and processing. It allows the calculation of amplitude (RMS and ARV) and spectral (mean frequency, MNF and median frequency, MDF) parameters, and global conduction velocity (CV). In addition, a dedicated software package has been developed in Matlab™ for the batch analysis of all the data acquired by the four clinical partners of the Project NEW.

3. Discussion and Conclusion

A datalogger for the acquisition of raw surface EMG signals detected with linear electrode arrays has been designed and successfully tested. The system is battery powered and almost pocket size. A novel architecture has been implemented consisting of digital EMG probes, 16-channels each, connected via a digital serial bus to the acquisition unit. This solution allows data to be transmitted over a thin, 6-wire cable, with improved wearability with respect to classical analog solution, while featuring enhanced immunity against external electrical interferences.

The acquisition unit stores the data on a removable PCMCIA card, available in very large sizes, allowing long term acquisition of raw multichannel EMG data for off-line analysis and processing. The optional USB interface allows an easy online visual check of the signal quality on a PC screen, while preserving the optical isolation from the patient as required by international safety regulations. The system has been successfully used for field applications to study muscle activation at global and single motor unit level during standardized simulated work activities of computer terminal users and secretaries, during clinical tests of hospital nurses, and in a number of focused protocols. A commercial version of the system, with improved performance and additional functions, will be designed available by the end of 2004.

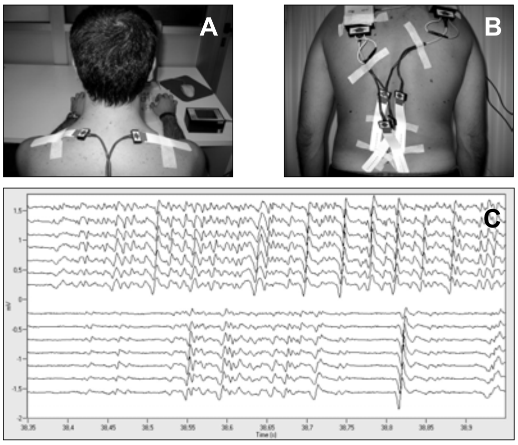

FIG. 3: Examples of experimental test protocols using the datalogger system: A) acquisition of multichannel surface EMG signals from left and right upper trapezius muscle during typing task. B) acquisition of multichannel surface EMG signals from low back muscles during 30° forward bending; C) Sample screenshot of the off-line display and processing software, showing signals acquired during typing task with the setup depicted in A).

4. Acknowledgements

This project was supported by the RTD European Project “NEW - Neuromuscular Assessment in the Elderly Worker” (QLRT 2000 00139).

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Session 4:

Applications in biofeedback, rehabilitation and sport medicine

Application of NEW results in sport, space and rehabilitation medicine.

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Abstract *– The results obtained within NEW have many applications in sport, space and rehabilitation medicine. A brief overview of these applications is provided. Focus is on an atlas of innervation zones of skeletal muscles, on sphincter EMG, on countermeasures to prevent muscle deterioration in microgravity conditions and on the potential for non invasive muscle fiber type assessment.*

1. Findings of general relevance.

The methods developed during the three year work carried out within NEW are certainly not applicable only to the neuromuscular assessment in the elderly worker. These methods have general applicability and the knowledge acquired within NEW is relevant in other fields. The issue of electrode location, partially addressed within projects SENIAM and PROCID, was completely investigated within NEW for the upper extremities (as indicated in Table 1, from Deliverable 1 and 4).

Table 1. Grouping of upper body muscles according to EMG quality.

1. Muscles in which the innervation zone and MUAP's propagation can be clearly detected and the conduction velocity measured : *Biceps brachii (long and short head), Deltoid (anterior and posterior), Trapezius (upper, middle and lower), Teres major, Abductor pollicis brevis, Abductor digiti minimi, Pronator teres, Palmaris longus*

2. Muscles in which MUAP's propagation is detectable but conduction velocity estimation is difficult: *Anconeus, Lateral and Medial head of Triceps Brachii, Infraspinatus, Opponent pollicis, Extensor digitorum, Brachioradialis*

3. Muscles from which surface EMG can be detected but propagation patterns cannot be found *Lateral deltoid, Flexor pollicis brevis, Flexor digiti minimi, Dorsal and First Palmar Interosseus, Lumbricals, Flexor carpi radialis and ulnaris, Extensor carpi radialis and ulnaris*

4. Muscles from which surface EMG signal quality is poor because of small muscle size, depth or cross-talk: *Abductor pollicis longus, Flexor digiti minimi, Opponens digiti minimi, Flexor digitorum profundus and superficialis, Flexor pollicis longus, Pronator quadratus, Extensor pollicis longus and brevis, Supinator, Second and Third Palmar Interosseus*

Other projects extended this classification to the lower extremities, leading to an atlas of innervation zones and suggested electrode locations. The atlas is currently in preparation [ATLAS]

The development of adhesive electrode arrays, the investigation of the effect of skin treatment on electrode-skin impedance and noise and the finding that rubbing with alcohol is not the best skin treatment are other observations of general relevance.

The combined use of EMG and MMG sensors first investigated within NEW (Deliverable 2) was further developed within the European Agency project *Microgravity effects on skeletal muscles investigated by surface EMG and MMG* (MESM) where the same techniques were applied to investigate electrically elicited EMG responses. The electrode array method was very useful for removing the stimulation artifact from electrically elicited EMG signals. The development of a 16 channels EMG datalogger, now evolving into the second generation version with 64 channels. (Deliverable 3), opens new research avenues not only in ergonomics but also in sport medicine and clinical rehabilitation. New perspectives are also opening thanks to the advanced signal processing techniques developed within NEW. In particular, the substantial progress in the field of crosstalk separation and conduction velocity estimation (Deliverable 5) are extremely valuable in sport medicine.

2. Applications in space medicine

Assessing the effectiveness of counter-measures (for prevention of muscle wasting) during long permanence in microgravity conditions is a major task in space medicine. This task involves processing electrically elicited EMG signals after removal or cancellation of stimulation artifacts. Multichannel signal detection and processing techniques developed within NEW were successfully applied for these purposes. In particular, the compound action potential obtained from a channel of the array distant from the stimulation electrode was successfully used to estimate artifact and response and adaptively eliminate the first from the second even in channels closer to the stimulation electrode where overlapping between artifact and response is substantial (Fig. 1) [MANDRILE]

a)

b)

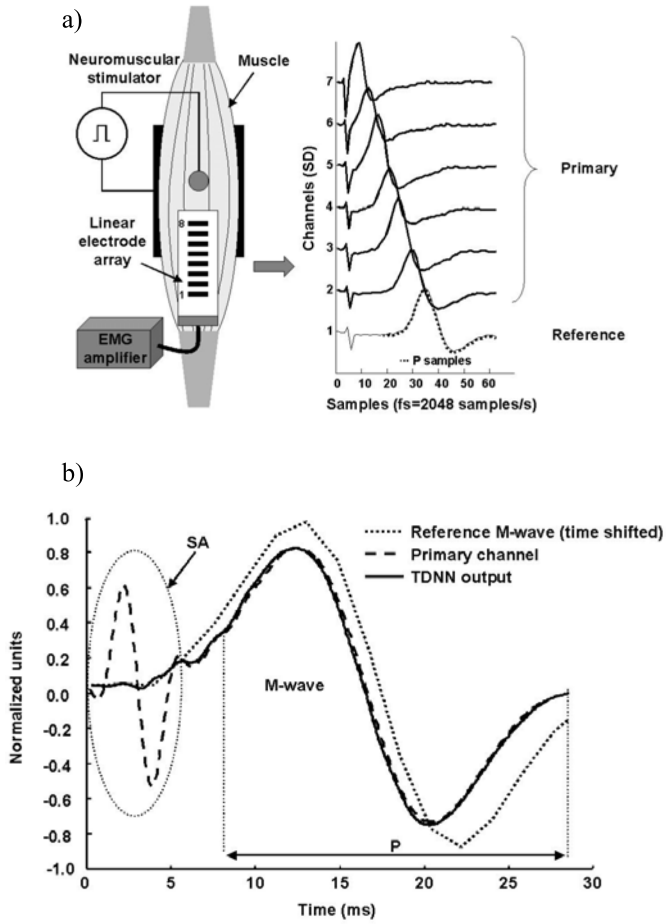

Fig. 1: Example of M-wave reconstruction using multichannel signal detection and processing. a) The compound action potential obtained from a channel (reference) of the array distant from the stimulation electrode was successfully used to estimate stimulus artifact (SA) and response and adaptively eliminate (using a Time Delay Neural Network) the first from the second even in channels (primary) closer to the stimulation electrode where overlapping between artifact and response is substantial.

3. Applications in pelvic floor muscle analysis

The signal acquisition hardware and the signal processing algorithms and models developed within NEW were adapted to the specific pelvic floor situation. Preliminary measurements led to interesting results and justified the preparation of the proposal "On Asymmetry in Sphincters" approved by the EU in 2002. The electrode array is no longer linear but circumferential and the motor units are placed in arcs, as indicated in Fig. 2. The innervation zone can be located on the basis of the single differential signal as is done for the skeletal muscles. The estimation of muscle fiber conduction velocity is more complex because the algorithm will estimate the angular velocity whereas the peripheral value is of interest and its estimation requires knowledge of the radial distance from the array.

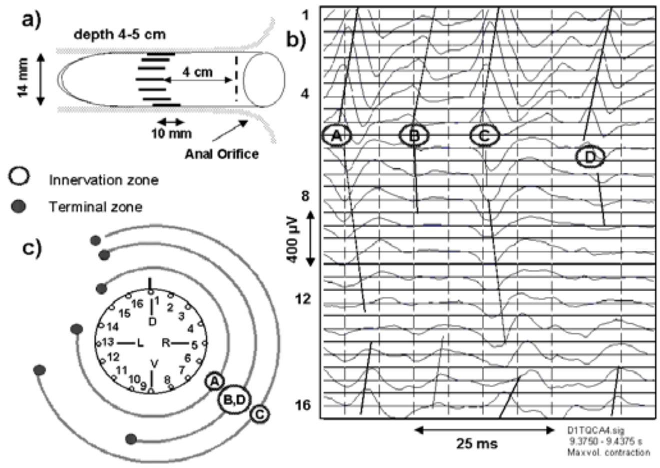

Fig.2 a) rectal probe carrying a circumferential electrode array with 16 contacts, b) examples of four single motor unit action potentials showing innervation under ch. 5-6, c) arrangement of the motor units as derived from b), L = left, R = right, D = dorsal, V = ventral.

Fig. 2 shows the importance of identifying the location of innervation zones in the external anal sphincter in order to avoid damage in the area during surgical interventions. [OASIS1] [OASIS2] [OASIS3]

4. Applications in rehabilitation medicine

The identification of innervation zones is important also for optimal use botulinum toxin. This expensive substance, inducing temporary denervation, has a maximal benefit/cost ratio with a minimal amount if it is injected in the innervation zone region of the muscle to be treated. Other important applications concern a) the EMG based gait analysis and the use of the atlas described in point 1 for the standardization of electrode locations,

5. Application in sports medicine

Muscular composition is usually assessed via bioptic and istochemical analysis. The high cost and the ethical problems associated to such procedures limit their use. In addition, the information obtained from small bioptic specimens is not necessarily representative of the whole muscle. These drawbacks motivate the development of alternative non-invasive methods for fiber type distribution assessment. Functional non-invasive assessment of skeletal muscles is usually obtained with the study of muscle fatigue. One widely used method, consists in the assessment of the maximal voluntary contraction (MVC) and of the lack of ability to repeat such MVC after a fatiguing exercise.

Integrated approaches that comprise the evaluation of the EMG signal features and the assessment of metabolic and inflammatory biochemical variables (i.e. acid lactic, the cytokine interleukin-6 and the cytokine myostatin) have been developed.

Preliminary findings seem in agreement with others already available in the literature in which the information about the fiber type and the athlete’s phenotype was obtained in invasive manner.

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Myoelectric Biofeedback and FES in Rehabilitation of Elderly Stroke Patients

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Abstract *– Strokesurvivers will in some cases be left with a compromised hand function which specifically can be attributed to lack of volitionally controlled hand opening. In this paper we present a case story where bio-feedback of the myoelectric signal from the paretic wrist extensors and following use of this signal to direct control electrical stimulation of the same muscles to help hand opening was provided. It is concluded that this method could be useful to advance rehabilitation when volitional hand opening is absent.*

1. Introduction

There are about 130.000 new cases of stroke each year in Italy, with a prevalence in the elderly part of the population. It often results in hemiparesis with loss of hand function. Normally there are few possibilities to accelerate the recovery of hand function and the disuse of the hand is often causing flexion contractions of the wrist and fingers leading to adverse effects such as pain and skin problems of the palm.

Electrical stimulation (ES) has shown to have a positive effect on residual wrist extension in hemiplegics [1] and though still debated, biofeedback of the volitional myoelectric signal may increase voluntary movement [2] and combining these two elements has shown even better results [3].

2. Methods

At the Centro di Bioingegneria-FDG a system (MeCFES) that can extract part of the volitional myoelectric signal from a stimulated muscle has been developed and tested [4]. Besides letting the myoelectric activity directly control stimulation level the system can be used to display the myoelectric activity on a computer screen in real time.

We tested the system with a 70 year old stroke patient who was clinically stable, 3 years after onset, with absolutely no movement of his left hand before treatment. He was receiving regular physiotherapy 2 times a week and had a severely closed hand with no sign of neither finger or wrist extension and resulting oedema in the palm. After initial assessment he was offered to add a biofeedback & FES session after his regular physiotherapy. At the assessment, a weak myoelectric signal from his wrist extensors was discovered, when the subject tried to open the hand. To condition the extensors, he was given an electrical stimulator to use 20 minutes daily at home for a month.

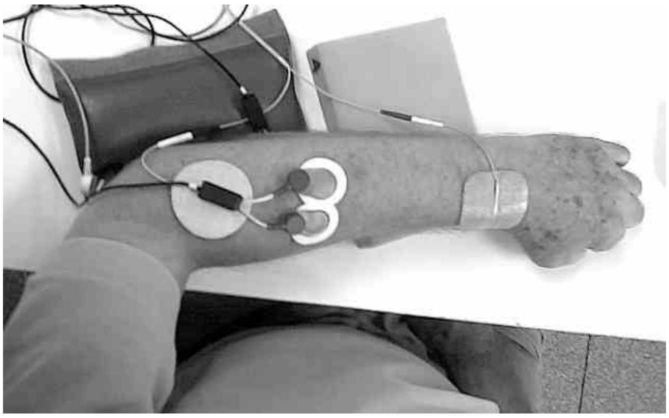

FIG. 1: Electrode placement with the two large stimulation electrodes, positioned by trial, to produce wrist and finger extension. The smaller EMG electrodes are placed over the long finger extensors in a configuration that minimises stimulation artefacts.

Stimulation sites were found by trial to produce extension of the fingers and wrist [5] and recording electrodes were placed perpendicular to the line of stimulation current and where the volitional myoelectric signal was strongest (see Figure 1)

FIG. 2: Biofeedback set-up. The position of the hand image is a linear function of the level of myoelectric signal. Every time the subject succeeds in catching the butterfly it changes position alternating between zero and a randomly chosen position.

After a month of conditioning the muscles at home biofeedback was given after each physiotherapy session. The level of myoelectric signal was shown as a vertical position of an image on a PC screen (Figure 2). The task of the subject was to place this image (which was a hand) on top of another image (a butterfly). Upon catching the butterfly the butterfly would move alternatively or to the baseline or to a new randomly selected vertical position. Different images were presented and points were given to motivate the subject. The subject would continue on will to 50 points (catches) or approximately ½ hour and this part took a month.

Finally after another month, when the myoelectric signal was sufficiently strong, the treatment was changed to let it control the stimulation and the subject observed how he could move the wrist and fingers with the aid of the MeCFES system and that proceeded for another month after which the planned physiotherapy period ended and the patient was dismissed.

3. Discussion

After conditioning the muscle for a month with neuromuscular stimulation at home the skin problems of the palm diminished and the subject became able to fold the hands.

Then after furthermore a month of biofeedback training the voluntary myoelectric activity became sufficient for the MeCFES system to control stimulation of the same muscles and the patient became able to extend the wrist by using the system. Finally after yet a month there was observed some carryover effect allowing slight wrist extension after the training.

Our conclusion is that myoelectric controlled functional electrical stimulation may be used as a tool in the physiotherapy to help re-educate movements where conventional therapy gives in and that there are grounds to pursue this method further.

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MUAP Rate in patients with chronic pain

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1. Introduction

Despite extensive research, little is known about deviations in motor control in people with chronic pain. Most studies have used bipolar surface Electromyograhpy (EMG) to investigate motor control, resulting in a global view of muscle activation patterns. Recently, we proposed a new measure, motor unit action potential (MUAP) Rate (MR), to study motor control in a more detailed way (1). The objective of this study is to assess differences in motor control between chronic pain patients and healthy controls.

2. Methods

EMG of the dominant upper trapezius during computer work tasks in a control group (n=13) and a patient group (n=10) was recorded. The protocol consisted of a unilateral gross motor task (dots task), a type task, an edit task, a mouse task and a stress task (STROOP test). EMG was measured with an 8-channel linear electrode array. MR was calculated by detecting and counting MUAPs per second. RMS was calculated per second as well.

3. Results

In Figure 1, at the left side, RMS is shown. A two-way ANOVA revealed a statistically significant dependency for task only ($F = 10.7$, $p < 0.000$). At the right side MR is shown. A two-way ANOVA revealed a statistically significant dependency for task ($F = 6.67$, $p < 0.02$) and group ($F = 3.92$, $p < 0.05$).

4. Discussion

The higher MR implies that the activity during an imposed task is higher in patients than in controls, although the biomechanical demands are the same. As can be seen from Figure 1, this difference is not as much reflected in RMS. Apparently, MR is more sensitive to differences in motor control.

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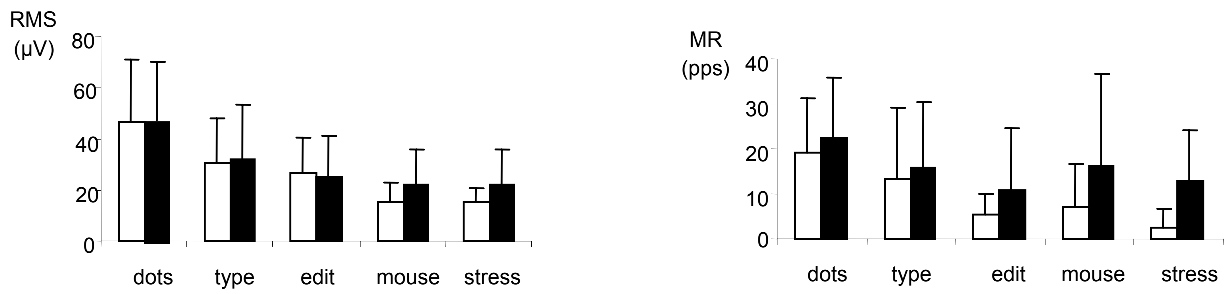

Figure 1 RMS (left) and MUAP Rate (right) for control (white) and patient group (black) for all tasks

The Influence of Different Intermittent Myofeedback Training Schedules on Learning Relaxation of the Trapezius Muscle while Performing a Gross-Motor Task

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1. Introduction

This study aimed at investigating the influence of three intermittent feedback training schedules as provided by a Cinderella-based myofeedback system[1], on learning relaxation and resistance-to-extinction of the trapezius muscle, in subjects performing a unilateral gross-motor task.

2. Methods

Eighteen healthy subjects (mean age 30.3±9.7) were randomly assigned to 3 groups. Three schedules were defined with feedback intervals of 5, 10, or 20 seconds. Measurements were performed 3 times; subjects were exposed to the schedules randomly. Bipolar surface EMG recordings were performed at the dominant upper trapezius muscle. Subjects performed a gross-motor task without (Baseline; B), with (Task 1-4; T1-4), and subsequently without feedback (Extinction 1-2; E1-2). Auditory feedback was provided when the pre-set level of muscle rest during 80% of the interval was not reached. Learning was defined as increased muscular rest (RRT) and decreased muscular activity (RMS).

3. Results

See figure 1. RRT was significantly increased during the tasks for the 5 (T1-4; p≤.049), 10 (T1-2; p≤.011) and 20 seconds schedules (T1; p≤.024) compared to baseline. RMS was only decreased under the 10 (T1; p≤.043) and 20 seconds schedules (T1-2; p≤.0.036). None of the schedules showed resistance to extinction.

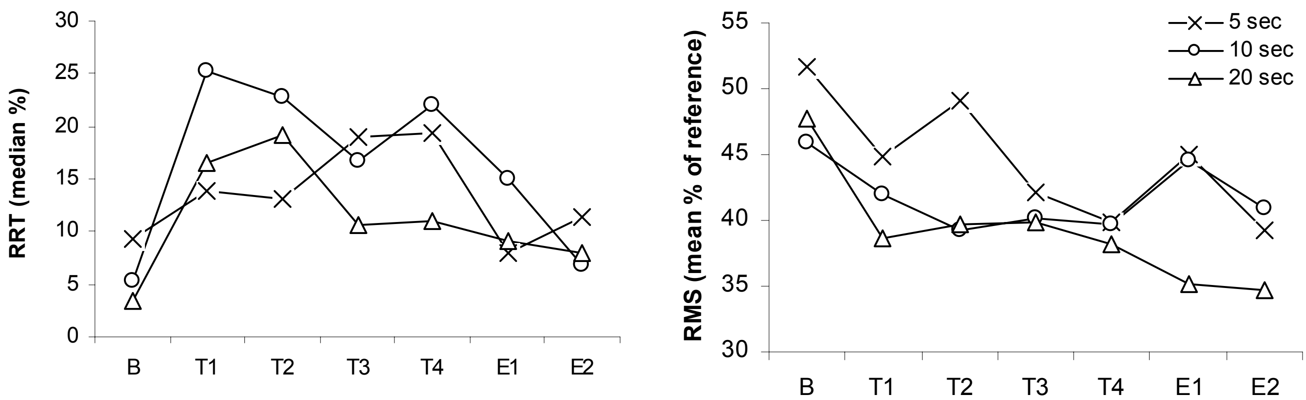

FIG. 1: RRT (left) and RMS (right) values for the three intervals during gross-motor task performance

4. Discussion

The 10 seconds schedule is preferred over the 5 and 20 seconds schedules in learning relaxation of the trapezius muscle in subjects performing a gross-motor task. Further studies are required to investigate the longer term effects of myofeedback training.

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Methodological considerations for muscle rest time calculation– a pilot study using Swedish NEW data from computer work

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Abstract *– In literature, different methods have been suggested for the computation of muscle relative rest time (RRT) from electromyography (EMG), and it is not clear which method is optimal for the large amount of data collected within the NEW project. In the present pilot study we utilized the Swedish part of the data to compare a commonly used individualised threshold, 0.5% of maximum voluntary electrical (MVE) activity, with another individualised, 3% of reference voluntary electrical (RVE) activity, and with a fixed threshold at 15 µV. Also the correspondingly computed RRT values were compared. The aim was to investigate if the 0.5 %MVE threshold could be replaced with any of the others without a significant change in the computed RRT, with respect to inter-subject comparability. The results show a considerable variation of both the RVE- and MVE-based RRT threshold values. They are however individually correlated. The 15 µV fixed threshold level was approximately four times as high as the average RVE- and MVE-based levels. Moreover, the estimated background noise correlated negatively with both the RVE- and MVE-based threshold values. An individual RRT threshold level, and compensation for background noise, should therefore be considered for the NEW data.*

1 Introduction

Work related musculoskeletal disorders (WMSDs) in the shoulder/neck area are a common and increasing problem among computer workers, especially women [1], in spite of the relatively low muscle activation levels that are required.

It has been postulated that long-term low-level workload with few periods of muscle rest can cause selectively over-usage of low-threshold muscle fibres, thus causing WMSDs [2]. This hypothesis was supported by a one-year prospective study of light manual workers [3], where subjects developing WMSDs showed significantly fewer short muscle rest periods (gaps) than subjects who stayed healthy. In another study, supermarket cashiers with self-reported shoulder/neck pain showed a significantly lower degree of muscle rest than controls [4].

Within the EU shared cost project NEW (Neuromuscular assessment in the Elderly Worker), an extensive amount of electromyography (EMG) data has been assessed from elderly (age 45-65 years) female computer workers in four countries. Thus, it is possible to investigate whether elderly computer workers with WMSDs show a lower degree of muscle rest than those without. In literature, different methods have been suggested for the computation of muscle relative rest time (RRT), and it is not clear which method is optimal for the NEW data.

The aim of the present pilot study was to utilize the Swedish part of the NEW data to investigate if the commonly used 0.5 %MVE threshold could be replaced by either 3 %RVE or a fixed 15-µV level without a significant change in the computed RRT, with respect to inter-subject comparability. Especially when studying subjects with WMSDs, a replacement by either RVE-based or a fixed threshold could be desirable since they both are less dependent on the subject's motivation.

2 Method

2.1 Subjects

In total, 27 subjects, all women, were included in this pilot study. Average age and BMI was 54.4 (SD: 5.0) years and 25.1 (SD: 4.3) kg/m2, respectively. The inclusion criteria were: (1) age range between 45 and 65 years, (2) average employment at least 20 hours/week, and (3) a minimum employment time of 5 years in present or similar work employments.

2.2 Protocol

The protocol started with a rest measurement, which was used for calculation and compensation for background noise of the EMG signals. For normalisation purposes, the subjects were also asked to perform reference voluntary contractions (RVCs) prior to the computer work tasks, and maximum voluntary contractions (MVCs) after the computer work. The RVCs and MVCs, as well as the calculations of the reference voluntary electrical (RVE) and maximum voluntary electrical (MVE) activities, were performed according to recommendations by Mathiassen et al. [5].

The subjects performed 4 computer work tasks, (1a, 1b, 2 and 3), 1-3 in a randomised order:

Text typing (1a; TY) on a computer keyboard, followed by an editing task (1b; ED) of the previously typed text. The first word with more than 5 letters should be changed to bold text, in the second such word the first letter should be capitalized, the third to bold, and so on. Each of the TY and ED tasks lasted up to 5 minutes, or until it was completed.

1. The STROOP colour word stress task (ST); a name of a new colour appeared approximately every second at a random place on the computer screen. The subjects were asked to respond, as accurate as possible, on which colour the word was written in by clicking on the correct icon alternative (4 alternatives) with the computer mouse. The task lasted for 5 minutes.
2. A precision task (PR), where the subjects were asked to connect lines between points according to a template by clicking on the points in the right order with the computer mouse. New points and a new template appeared on the computer screen when the previous figure was finished correctly. The subjects were asked to make as many figures as possible. The task lasted for 2 minutes.

2.3 Electromyography

Because of the interest to calculate muscle fibre conduction velocity, an array EMG electrode was developed within the NEW project. This 8-channel array surface EMG electrode with 5 mm inter-electrode distance (IED) [6] was placed on the left and right upper trapezius muscle, respectively. The electrodes were placed parallel to the line between the lumbar processus of the seventh cervical vertebra and the acromion, with the first channel 5 mm lateral, and the eighth channel 40 mm lateral, from the midpoint.

From each side, a 7-channel 5-mm IED bipolar EMG signal was recorded, 16-bit A/D converted and stored on a portable data logger [6] with a 2048 Hz sampling frequency for further analysis. The digitised signals were filtered with a 2^{nd} order Butterworth 10-400 Hz band pass filter and with an adaptive 50-Hz filter [7].

2.4 Signal quality assessment

Prior to further analysis, the signals were visually inspected in order to determine if some short artefact periods had to be removed or if any of the 5-mm IED signals had to be disregarded due to e.g. 50-Hz power line interferences or shortcircuited EMG electrode channels. Due to identified noise interferences below 20 Hz and above 300 Hz, an additional 6th order 20-300 Hz Butterworth band pass filter was applied.

2.5 RRT calculation

For each electrode array, 4 different 20-mm IED equivalent bipolar EMG signals were calculated as:

$$s20_i = \sum_{k=0} s5_{i+k} \quad , \quad i = 1,2,3,4$$

with $s20_i$ the ith 20-mm IED signal and $s5_{i+k}$ the (i+k)th 5-mm IED signal. From visual inspection, the most suitable signal considering signal quality, of these four was chosen. The complete file was disregarded in case of no usable combinations of 4 consecutive 5-mm IED signals.

Root mean square values (RMS) with a 50-ms window were then calculated. The RMS values were, in a power sense, compensated for background noise, which was defined as the minimum RMS value of a 1-second moving average window of the rest measurement.

RRT was defined as the relative cumulative time with RMS amplitudes below a certain threshold level for periods of 150 ms or longer. RRT based on three different threshold definitions were calculated:

1. individual, corresponding to 0.5 %MVE
2. individual, corresponding to 3 %RVE
3. fixed, 15 µV.

The 15 µV level was suggested within the NEW consortium, while the thresholds of 3 %RVE and 0.5 %MVE were in accordance with recommend-dations by Hansson et al. [8].

2.6 Statistics

The RRT threshold levels resulting from the definitions were compared by a Student's t-test for paired samples. The computed three different RRT sets were non-Gaussian distributions and therefore compared for the four different work tasks by a Wilcoxon sign rank test.

3 Results

Out of the 27 files for each work task, at average 21 (range: 20-23) were found to be analysable.

3.1 Average RMS levels

The average median RMS levels for the TY, ED, ST and PR tasks were for non-dominant side 3.8, 1.3, 0.6 and 0.6 %MVE, respectively, and for dominant side 4.0, 3.6, 1.3 and 1.2 %MVE, respectively. The corresponding absolute values were 25.0, 9.4, 4.0 and 3.5 µV for the non-dominant side and 32.0, 29.3, 9.9 and 8.9 µV for the dominant side. The average background noise level was 1.40 (SD: 0.75) and 1.39 (SD: 0.76) µV for non-dominant and dominant side, respectively.

3.2 RRT threshold levels

The subject-averaged RVE- and MVE-based threshold levels correlated significantly (Pearson test; r=0.50, P<0.01) with each other and were at average 3.9 (SD: 1.8) µV, and 3.3 (SD: 1.5) µV, respectively. No significant difference was found between the RVE- and MVE-based thresholds levels (P≥0.12), while both were significantly lower than the fixed 15-µV threshold (P<0.001). Histograms of the thresholds are plotted in figure 1. Moreover, there was a significant negative correlation between the subject-averaged noise levels vs. both the RVE and MVE levels (Pearson test; r=-0.45, P<0.01 and r=-0.54, P<0.01), and a tendency of negative correlation between BMI vs. the RVE and MVE levels (Spearman test; ρ=-0.36, P<0.08 and ρ=-0.33, P<0.09).

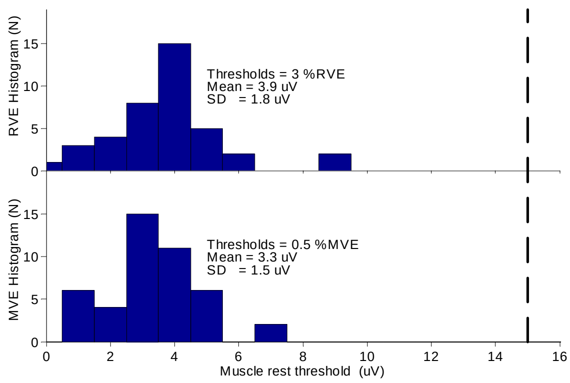

FIG. 1: RRT threshold histograms (27 subjects, both values for non-dominant and dominant side are included). Vertical line denotes the fixed 15-µV threshold level.

3.3 Relative rest time values

The median RRT values for the different work tasks are shown in table 1. For all work tasks and for both non-dominant and dominant side, the RRTs for RVE- and MVE-based thresholds were significantly lower than the RRTs for fixed 15-µV threshold ($P<0.001$). Eventual differences between RVE- and MVE-based RRT calculations could, except for the non-dominant side during the ED task, not be statistically verified (min. $P\geq0.1$).

TAB. 1 : Median RRT values for the four different work tasks. Asterisks denote significant differences per cell.

Work task	RRT [%] non-dom. side (fixed/rve/mve-based)	RRT [%] dominant side (fixed/rve/mve-based)
TY	22.3*/0.1/0	4.6*/0/0
ED	71.7*/1.1*/0.2	2.5*/0/0.2
ST	98.1*/21.6/11.5	69.1*/4.8/0.1
PR	98.6*/5.2/16.7	76.6*/1.1/0.1

4 Discussion and Conclusion

The results from the present study show a considerable individual variation of the RVE- and MVE-based RRT threshold values, which partly may be due to the relatively high inter-subject BMI variations. Moreover, the RVE- and MVE-based levels were, at average, approximately ¼ of the 15-µV fixed level. The background noise level plays an important role, and is also negatively correlated with the RVE and MVE values. Due to these findings, an individual RRT threshold level, and compensation for background noise, should be considered.

Considering the choice between the RVE- and the MVE-based approaches, a significant correlation, similar variances and no significant difference in average threshold level were shown. Nor could any significant differences be found for the computed RRT values, however with high variances between the subjects. The amount of data in the present study might therefore be too low for statement of choice between RVE- or MVE-based approaches.

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Surface EMG of the masseter and anterior temporalis in complete denture wearers during chewing

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Abstract *– This study investigates the EMG activity of jaw elevator muscles during mastication in a group of subjects with complete maxillary and mandibular denture. Subjects were analysed with the old denture, with the new one at the delivery, after one month, and after three months from the delivery of the new denture. Surface EMG signals were detected from the right and left masseter and temporalis muscles during chewing. The mandibular motion was measured with a kinesiograph with an accuracy of 0.1 mm. The main finding was that the EMG activity of the side of mastication significantly decreased at the delivery of the new denture while it returned to the values of the old denture after three months. The decreased EMG activity was accompanied by an increase of symmetry between the EMG activity of the two sides.*

1. Introduction

Mastication is a highly coordinated neuromuscular function involving fast effective movements of jaw and continuous modulation of force. The loss of teeth involves a number of changes, which affects bone, oral mucosa, and muscles. The alveolar bone tends to resorb, the overlying mucosa shows a decreased number of receptors and the muscular force reduces. These changes affect the mastication which must adapt to the complete denture: the width and the masticatory efficiency of the masticatory cycle are reduced if compared with the normal subjects [3,4,5,6,7].

Aim of this study was to compare and evaluate the electromyographic values during deliberate right and left chewing with the old denture, with the new one at the delivery, after one month and after three months.

2. Subjects and methods

Seven subjects (4 males, 3 females, mean age +/- standard deviation: 63.2 +/- 6.9) with complete maxillary and mandibular denture were selected for the study. They were asked to chew a soft bolus for 10 seconds: 3 times non-deliberately, 3 times deliberately on the right side and 3 times on the left side. Two identical sets of contractions were performed to obtain a total of 18 mastications.

Surface EMG signals were recorded with an electromiograph EM2 Myotronics Research

(Myotronics Research Inc., Tukwila, WA - USA) with eight channels. It is part of the K6-I WIN Diagnostic System, produced by Myotronics Research Inc., v Tukwila, WA-USA [1-2]. Electrodes were located over the muscle bellies identified by palpation.

The mandibular motion was measured with a kinesiograph (K6 -I, Myotonics Inc. Tukwila, WA – USA) (Fig. 1). The instrument measures jaw movements with an accuracy of 0.1 mm. Multiple sensors (Hall effect) in an extremely light weight (four ounce) array, track the motion of a tiny magnet attached to lower inter-incisor point. The configuration is relatively non-invasive. The kinesiograph was interfaced with a computer for the data storage and subsequent analysis.

3. Results

During chewing, the EMG activity was high from the beginning until the end of closure while in dentate, normal subjects, the maximum activity is reached very near the occlusion only.

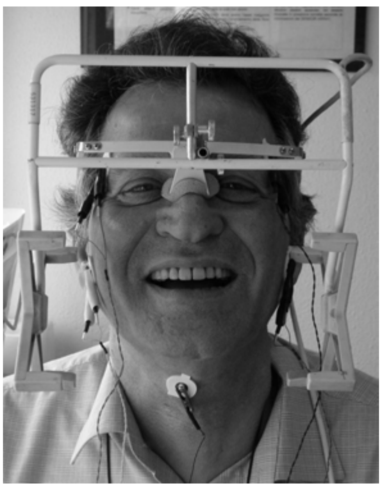

FIG. 1: Set-up of the Kinesiograph K6 and EM2 Myotronics on a full-denture wearer.

A four-way ANOVA was used to analysed the dependence of EMG amplitude on the following factors and the interaction between factors: trial, side of chewing, prosthesis (old prosthesis, new prosthesis when delivered, new prosthesis after 1 month, new prosthesis after 3 months), and recorded side. The EMG amplitude was not significantly affected by any of the factors investigated when considered independently from each other. There was, on the contrary, a significant interaction between the side of chewing and the recorded side ($F = 63.64$, $P << 0.001$). The EMG amplitude from the muscle of the side of chewing was significantly higher than that of the other side (SNK post-hoc test, $P < 0.01$). Moreover, there was a significant interaction between the side of chewing, the recorded side, and the prosthesis ($F = 3.55$, $P < 0.05$). In particular, with the old prosthesis the EMG amplitude was significantly larger for the chewing side than for the other (SNK post-hoc test, $P < 0.01$). This was not so for the right side when the new prosthesis was delivered (i.e., at the delivery of the prosthesis there was no difference between the EMG amplitude of the two sides when chewing from the right side). The difference between the two sides was recovered both after 1 and 3 months from the delivery (SNK post-hoc test, $P < 0.05$). Comparing the different prosthesis conditions, the EMG amplitude was significantly smaller at the delivery of the new prosthesis with respect to the case of the old prosthesis only for the side of chewing (and not for the other recorded side)

(SNK post-hoc test, $P < 0.05$). After 1 month, the EMG amplitude did not significantly increase with respect to the case immediately after the delivery and was still smaller for the side of chewing than the amplitude with the old prosthesis. After 3 months, the EMG amplitude significantly increased with respect to the case after 1 month (SNK post-hoc test, $P < 0.05$) and was not significantly different from the amplitude recorded with the old prosthesis.

There was a significant decrease in the EMG values of the masseter of the same side of the bolus and a loss of coordination between the two masseters; the coordination was recovered after one month and the return to the initial values (with the old prosthesis) after three months.

4. Conclusions

The main results of the study were:

1. The EMG activity in the masseter of the same side of the bolus is significantly higher than that of the masseter of the opposite side (as it happens in normal, toothed subjects).
2. The EMG activity of the masseter of the same side of the bolus significantly decreases at the delivery of the new prosthesis and the coordination of the right and left masseter is lost (the activity of the masseter of the same side of the bolus is significantly larger than that of the opposite one; it returns to the values obtained with the old denture within three months [8-9] but the coordination is recovered after 1 month).
3. The EMG activity decreases at the delivery only for the muscle of the side of mastication.ù

It is concluded that in the subject sample investigated the coordination of the masseter of the same side and of the opposite side of the bolus is the same as for toothed normal subjects. The masseter of the same side of the bolus is significantly higher than the masseter of the opposite side.

The stomathognatic system replies to an important change, such as a new complete maxillary and mandibular denture, with a decrease of the activity of the muscle of the side of mastication. This is probably due to the activation of peripheral receptors, especially nociceptors, which inhibit muscular contraction as a protective reflex [10].

Because the EMG amplitude significantly increased after three months from the delivery of the new complete denture and not after one month, we can conclude that three months can be considered the time needed for complete adaptation.

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Mechanical, electromyographical and biochemical variables after a fatiguing task in endurance and power-trained athletes

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Abstract *- This work aimed to identify, among mechanical, electromyographic and biochemical variables, those promising to be correlated with muscle characteristics of different athletes. Twenty subjects trained for power and endurance disciplines were studied before and after an isokinetic fatiguing exercise. Torque values, EMG variables estimated from signals recorded in both voluntary and electrically elicited isometric contractions and cortisol and lactate responses were compared to highlight possible correlations. Results allow to identify the rate of torque decrease during the isokinetic exercise and among different trials of isometric exercises, the lactate production and the recovery times of muscle fiber conduction velocity as the variables able to best distinguish between the two groups of athletes.*

1. Introduction

Several different efforts were spent in the past with the aim to correlate muscle fiber characteristics with variables obtained by non-invasive techniques alternative to the bioptic analysis. A number of authors have highlighted the correlations between mechanical output and fiber type distributions, showing that mechanical muscular performances are a function of the amount of fast fibers within the muscle [2, 3, 7, 14]. Other authors focused on electromyographic (EMG) signal modifications during fatiguing contractions. Muscle fiber conduction velocity (CV) and signal spectral characteristics (mean frequency, MNF or median frequency, MDF) were identified as suitable correlates with fiber type distribution [5, 8, 12]. A complementary approach could be provided by biochemical assays of blood samples in subjects acutely stressed by a demanding exercise.

In the present work, a comparison between power and endurance trained athletes was made. Isometric torque, EMG variables and serum cortisol and lactate were analysed before and after an isokinetic fatiguing exercise, with the following aims: to search for differences among the athletes and to assess the time course of all the studied variables.

2. Subjects and exercise protocol

Twenty male athletes (13 endurance-trained and 7 power-trained, age 27.7 ± 7.9 years) participated in hte protocol. Consistently with previous studies [1, 15, 16] we used the maximal voluntary knee extension torque value and its decrease rate during the repeated isokinetic exercise to characterize the functional phenotype of the athletes. The subjects fulfilled for each leg four sets of twenty maximal contractions of the knee flexor and extensor muscle groups at 180°/sec angular velocity throughout a constant range of motion (100°), starting from the non-dominant leg. There was a 30 seconds rest period between each set and a 3 minutes rest period between the two legs. Before, immediately after the exercise and in the subsequent 120 minutes of recovery, blood was sampled to determine cortisol and lactate concentrations in serum, Maximal

Voluntary Contractions (MVC) and electrically elicited contractions were performed by the dominant leg to assess mechanical and myo-electrical manifestations of fatigue.

3. Instrumentation and measurements

A Cybex 6000 device was used for isokinetic exercise and isometric MVC contractions. Surface EMG signals were detected from the vastus lateralis using a linear array [9, 10] of 8 electrodes (silver bars 10 mm apart, 5 mm long, 1 mm diameter) in single differential configuration (double differential signals were computed off-line for conduction velocity, CV, estimation). The EMG signals were amplified with a 10-500 Hz bandwidth. The signals were sampled at 2048 samples/s, digitized by a 12 bit A/D converter and stored on the disk of a personal computer. Since the subcutaneous tissue thickness under the electrode array strongly affect ARV and MNF estimates [4, 11], it was measured by using an ultrasound device at the end of each session.

Serum lactate concentrations were measured spectrophotometrically with an Aeroset analyser. Serum cortisol was measured by radioimmuno-assay.

4. Results

The studied mechanical variables were: 1) the peak torque produced by the nondominant leg during the isokinetic test (MVC_{ISK}), 2) the rate of change ($SLOPE_{ISK}$) of the maximal voluntary torque produced during the isokinetic test, and 3) the peak torque value recorded by the dominant leg during the MVC isometric trial before and after the test.

Surface EMG variables of interest were: mean frequency (MNF) of the power spectral density, the average rectified value (ARV) and the average muscle fiber conduction velocity (CV). Values of neuromuscular efficiency (NME) were calculated as the ratio between MVC and the corresponding initial ARV [17].

The MVC_{ISK} and those recorded during the pre-test isometric trial were compared checking for correlations. A linear correlation (r=0.72, p=0.013, n=18) was found between the two modalities of MVC recording.

In Figure 1 the values of $SLOPE_{ISK}$ are plotted in order from the highest to the lowest (if considered in absolute value) showing the functional phenotype classification of 18 subjects. On the basis of this output, we arbitrarily clusterized those with $SLOPE_{ISK}$ <-20 Nm/trial (N=6) as power and those with $SLOPE_{ISK}$ in the range –20/0 Nm/trial (N=12) as endurance subjects. The statistical difference between the two groups was shown to be significant (p<0.01 Mann Whitney U test).

With regard to the EMG variable (ARV, CV and MNF) estimates obtained during maximal voluntary contractions no differences were found between the two groups, while for the electrically elicited contractions statistically significant differences were found only for the MNF variable immediately after the test, with greater MNF decreases in the power than in the endurance group.

With regard to the NME variable no differences were found within groups throughout the sessions except for the power group at POST and +15 minutes.

The recovery times for each variable were assessed as the trial in which the estimates were no more statistically different (Wilcoxon paired test) with respect to the pre-test (PRE) trial (Table). Power athletes were shown to need a longer recovery time than endurance athletes in all the variables only for the voluntary contractions.

The isokinetic exercise test elicited significant serum cortisol and lactate responses. No differences were found between the cortisol responses, while the power athletes showed a higher lactate response in comparison to the endurance athletes (Figure 2).

Finally, no statistically significant difference was found in the subcutaneous tissue thickness between the two groups (E= 2.8±0.7 mm; P=3.1±0.9 mm).

5. Discussion

Our findings about mechanical variables (MVC and $SLOPE_{ISK}$) confirmed those already published [2, 3, 6, 7], providing in addition a correlation between isometric and isokinetic torque assessments not yet available in the literature.

No statistically significant differences were found in the EMG estimates between the two groups in both contraction modalities. This could be due to the partial overlap among the type of training activities in the recruited subjects.

On the other hand, when analyzing the recovery times it was possible to identify CV as the EMG variable most related with the athlete's functional phenotype. In fact, CV values of the power group recovered slower than those of the endurance group in both contraction modalities.

Data obtained for the NME variable are worth noting. The two groups showed a statistically significant difference in NME in PRE. However, since NME is proportional to 1/ARV, the role of the subcutaneous tissue thickness is of relevance. Ultrasound measurements did not evidence a significant difference between the two groups, although a trend to a thicker subcutaneous tissue in power athletes was observed. Since thickness increase generates a reduction in the ARV and MNF [4, 11] estimates, the findings about MNF and NME could be biased by this factor. However, in the power group a NME decrease was observed at POST and +15 min ($p<0.01$); moreover, this group recovered later than the endurance group. Since these findings were obtained with a within group analysis, they could be considered independent with respect to the subcutaneous tissue thickness confounding factor.

The recovery times of the mechanical and EMG variables showed two different behaviours: the EMG values returned comparable with those recorded in PRE faster than MVC values, especially for the athletes belonging to the power group. This particular dynamics could identify a further feature correlated with the skeletal muscle composition.

Our isokinetic exercise test elicited significant biochemical responses. The power athletes reached higher lactate levels than the endurance athletes in agreement with the literature [13], while no differences were found in the cortisol responses. One possible explanation refers to the small sample size and to the heterogeneity in group composition.

Taken togheter, these findings seem to provide promising bases to design an innovative multifactorial tool for noninvasive assessment of phasic/tonic characteristic of muscle fibers based on the following variables: $SLOPE_{ISK}$, MVC, lactate, CV and NME recovery times.

Future studies should focus on the response to exercise of others hormones or peptides, more specific than cortisol to the muscular impact of the physical effort, and more sensitive than lactate in differentiating individuals with different muscle composition.

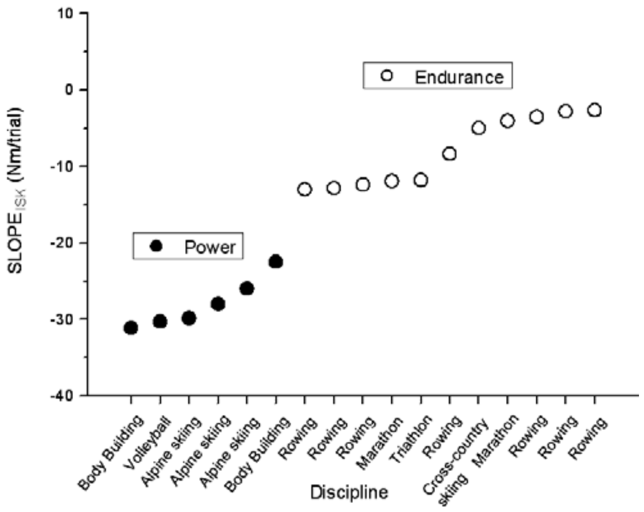

FIG.1: The values of SLOPE$_{ISK}$ (Nm/trial) are plotted in increasing order with respect to the discipline types

Table: Recovery times for the studied variables

	Subject classification	Voluntary contractions	Electrically elicited contractions
MVC	Power	+45	-
MVC	Endurance	+15	-
ARV	Power	+15	POST
ARV	Endurance	POST	POST
CV	Power	+15	+30
CV	Endurance	POST	+15
MNF	Power	+45	POST
MNF	Endurance	POST	+15
NME	Power	+30	-
NME	Endurance	POST	-

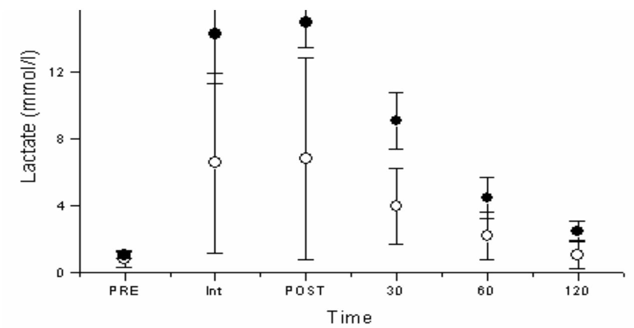

FIG.2: Lactate responses in the two groups

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Electromyographic activity of the Rectus Abdominis during an exercise performed with the Ab slider and crunch.

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1. Introduction

Abdominal training is important for improving athletes performance and may be useful in the prevention of back injury. Strength, endurance and coordination of the abdominal muscles has a role in most athletic movements either originate in or are coupled through the trunk [1].

Co-contraction of the trunk flexors and extensors has been found to play a fundamental role in providing lumbo-pelvic stability and impairment of this function seems to be related with acute and chronic LBP [2, 6,8].

From the evidence to date, it appears anyway that the main role in providing stability to the spine is related to a mechanism of feed-forward contraction of deep muscles of the trunk, especially transversus abdominis and multifidus [3]

Although the role of the rectus abdominis an the external obliques should not be ignored.

Acting primarily as spinal flexors these muscles also posses a stabilizer function during periods of high spinal loading and after the introduction of an unexpected trunk perturbation. During these activities, poor functioning of that muscles may increase the load on the deepest stabilizers, reducing overall spinal protection [4,7].

In terms of performance, it is the endurance, not the strength, of the superficial abdominal muscles that is considered to be most important. Previous research has demonstrated that a reduction in the endurance of the superficial abdominal muscles is more highly correlated with low back pain than a reduction in torque production [4,5].

Many methods and devices have been introduced to train abdominal muscles. One of the latest is the Ab Slider. It is a case with four rubber wheels connected to a spring. The exercise should be performed in quadrupedic position pushing out the case and the slowly return to starting position.

The Ab slider is spring loaded, and the further you extend, the more resistance it will create. Once you have stretched the spring will help you back to starting position. The aim of this study was to assess the muscles activity and the type of contraction of rectus abdominis while performing an exercise with the Ab slider and to compare it to a crunch exercise.

2. Materials and Methods

2.1 Study design

Repeated measures experimental design.

2.2 Subject

Eighteen students (mean age= 21,4 [SD=06], mean weight=65,3 Kg [SD=12,2], mean height=174,3 cm [SD=6,5]) from the School of Physiotherapy were included in this study.

All subject were in good health, with a low subcutaneous fat and reported no history of abdominal surgery, spinal deformities, acute and chronic low back injury or any other contraindication to the abdominal exercises.

The students had a previous training using the Ab slider and had no or minimal background knowledge of EMG testing.

All the subjects read and signed informed consent forms approved by the Vita-Salute University Office of Research prior to the experiment.

2.3 Data Collection

Electromyografic activity on rectus abdomins were recorded by a Telemg.

The skin was shaved and wiped with alcohol to lower cutaneous electrical impedance and a pairs of silver-silver chloride disk surface electrodes (diameter of 20 mm) were placed, parallel to the muscle fibers, three centimeters apart to the umbilicus [8].

Acquisition frequency of electromyographic signal was 1000hz. The signal was rectified, integrated with an integration period of 100 hz, filtered with a low pass filter of 500 hz and a high pass filter of 10 hz.

An opto-electronics motion analysis system, Elite (Bioengineering Technology System, BTS - Milan - Italy), was used to measure the movement of knee in the sagittal plane and to verify if any change in the distance between the pelvic girdle and the xiphoid process occurred during the Ab slider exercise.

Markers, before performing the Ab slider exercise, were placed on: lateral malleolus, lateral condyle of the femur, greater trochanter, superior rim of the iliac crest, anterior superior iliac spine, pubic symphysis, xiphoid process (Fig 1,2).

2.3 Exercise movement task

Each subject after recording an MVC of the rectus abdominis executed three repetitions of the crunch exercise and three repetitions of the Ab slider exercise, the exercise's order was randomized.

FIG. 1 : Surface electromyographic electrode arrangement, front view.

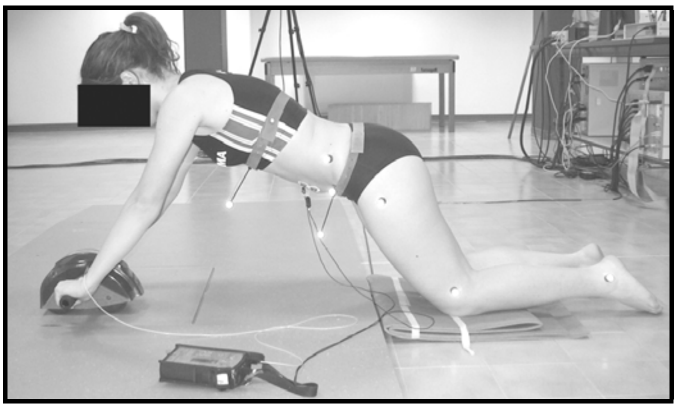

FIG. 2 : Surface electromyographic electrode arrangement, lateral view.

3. Results

It was recorded that the mean knee extension was 52,4 degrees (SD=10), the distance between the pelvic girdle and the xiphoid process always increased. During the crunch the mean muscle activity of the rectus abdominis was 76% of MVC (SD=24) and during the Ab slider the mean muscle activity of the rectus abdominis was 108% of MVC (SD=40). A Wilcoxon test for significance between the two exercise muscle activity showed p values of 0.004. (Fig. 3)

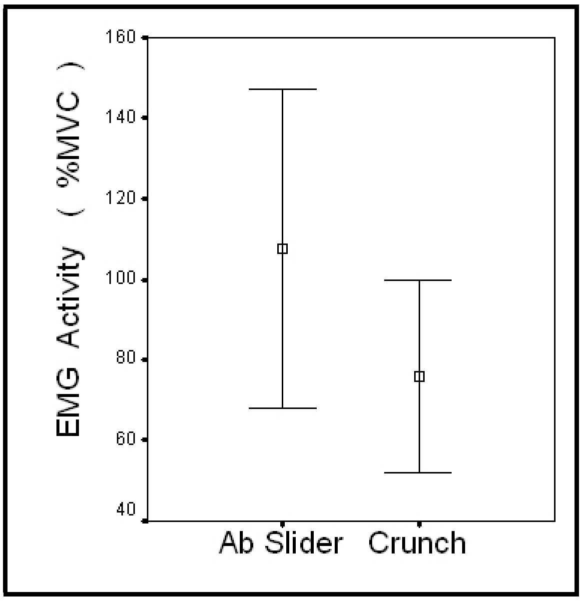

FIG. 3 : EMG activity detected on the rectus abdominis during Ab slider and Crunch exercises.

4. Discussion

The data showed that, during the exercise with the Ab slider, there is an important activation of the rectus abdominis if compared with the rectus activity during the crunch exercise. Cinematic analysis of the pelvic and xiphoid process movements indicates that the activity occurs in eccentric conditions, unfortunately it was not possible to quantify the variation of the distance between the starting position and the maximal extension because markers were placed on a bar strapped to the body.

5. Conclusion

Ab slider seems to be a good exercise to strengthen the abdominal muscles. In this exercise there is an important rectus activity obtained in eccentric condition.

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Coordination of right and left Masseter during mastication of a soft and a hard bolus in subjects with normal occlusion

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Abstract *– This study investigates the coordination between left and right jaw elevator muscles during mastication of a soft and a hard bolus. Surface EMG signals were detected from the right and left masseter and temporalis muscles during chewing. The mandibular motion was measured with a kinesiograph with an accuracy of 0.1 mm. The main finding was of an increase of the EMG activity in the controlateral masseter muscle significantly larger than the increase of activity in the same side of mastication when passing from a soft to a hard bolus. It was concluded that the larger are the load, the effort and the efficiency required, the more symmetric is the muscle activation.*

1. Introduction

The human posture is influenced by many factors: one of them is the equilibrium of the stomathognathic system, which includes teeth, muscles, bones, joint and central nervous system. Thus, the analysis of the functional charactheristics of the masticatory muscles assumes importance in many applied fields [1-3-5]. The aim of this work was to evaluate the coordination of the right and left masseter during mastication of a soft and a hard bolus in subjects with normal occlusion and function.

2. Subjects and methods

23 students (mean age +/- standard deviation: 25.6 +/- 2.6 years) with normal occlusion and function were selected for the study. Surface EMG signals were recorded with an electromiograph EM2 Myotronics Research [2-4] (Myotronics Research Inc., Tukwila, WA - USA) with eight channels. This system is part of the K6-I WIN Diagnostic System (Myotronics Research Inc., Tukwila, WA-USA). For the masseter, the belly was palpated during clenching and the EMG electrodes were fixed 2.5 cm above the mandibular angle; for the temporalis anterior, the belly was palpated during clenching and the electrodes were fixed along the anterior margin of the muscle 2 cm above the zygomathic arch. The mandibular motion was measured with a kinesiograph (K6 - I, Myotonics Inc. Tukwila, WA – USA) (Fig.1). The instrument measures jaw movements with an accuracy of 0.1 mm. Multiple sensors (Hall

effect) in an extremely light weight (four ounce) array, track the motion of a tiny magnet attached to lower inter-incisor point. The configuration is relatively non-invasive. The kinesiograph was interfaced with a computer for the data storage and subsequent analysis.

The patients were instructed to chew a soft, premasticated chewing gum and a hard bolus (winegum) non-deliberately and then deliberately on the right and on the left side. The duration of each test was 10 s and was repeated three times.

The recordings were performed in three consecutive days.

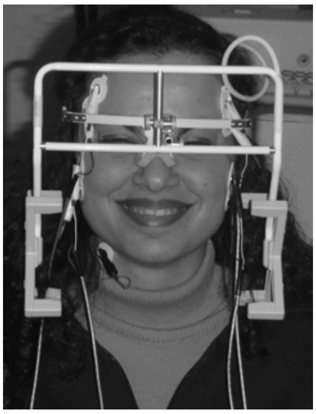

FIG. 1: Set-up of the Kinesiograph K6 and EM2 Myotronics in a young subject.

3. Results

3.1 Masseter

A four-way ANOVA was used to analyse the dependence of EMG amplitude on the following factors and the interaction between factors: day, side of chewing, bolus (soft or hard), and recorded side. The EMG amplitude was significantly affected by the bolus ($F = 13.52$, $P < 0.01$), with the soft bolus resulting in smaller EMG amplitude than the hard bolus (SNK post-hoc test, $P < 0.01$). Moreover, there was a significant interaction between the side of chewing and the recorded side ($F = 27.13$, $P \ll 0.001$). The EMG amplitude from the muscle of the side of chewing was significantly higher than that of the other side (SNK post-hoc test, $P < 0.01$) (Figure 2). Finally, there was a significant interaction between the side of chewing, the bolus and the recorded side ($F = 11.44$, $P < 0.01$). In particular, when chewing from the right side, there was no significant difference of EMG amplitude of the right masseter between the two boli (soft or hard) while the activity of the left masseter significantly increased in the same conditions when chewing the hard with respect to the soft bolus (SNK test, $P < 0.001$) (fig. 4). When chewing from the left side, for both the left and right side there was a significant difference in EMG amplitude between the soft and hard bolus (SNK test, $P < 0.05$ for the left side, $P < 0.001$ for the right side) (Fig. 4).

3.2 Temporalis anterior

For the temporalis muscle, there was a significant interaction between the side of chewing and the recorded side ($F = 31.31$, $P \ll 0.001$). The EMG amplitude from the muscle of the side of chewing was significantly higher than that of the other side (SNK post-hoc test, $P < 0.01$).

4. Conclusions

Coordination is the most important characteristic of the right and left masseter during mastication [7-8]. The results of this study showed that 1) there was no difference between the recordings in the different days, thus the measures were repeatable. 2) There was a significant difference between the chewing of a hard and a soft bolus resuting in higher EMG amplitude when chewing the hard bolus. 3) The masseter of the same side of the bolus was significantly higher than the masseter of the other side. 4) Comparing the mastication of a soft and a hard bolus we found a change in masseter coordination with a higher increase of activity of the masseter of the opposite side of the bolus. The masseter of the same side of the bolus increased its activity only slightly (Fig. 2-4).

FIG.2 EMG envelope of the right and left masseter of a normal subject during chewing a soft bolus (Chewing gum), deliberately on the left side. The left masseter (side of the bolus) shows a higher activity than the right masseter (opposite side of the bolus).

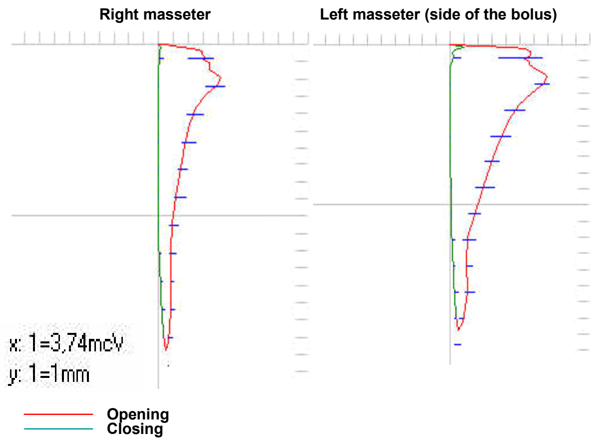

FIG.3 EMG envelope of the right and left masseter, of a normal subject during chewing of a hard bolus (winegum), deliberate on the left side. In comparison with the chewing of a soft bolus (Fig. 2), the EMG activity of the right masseter (opposite side of the bolus) increases more than the left (side of the bolus).

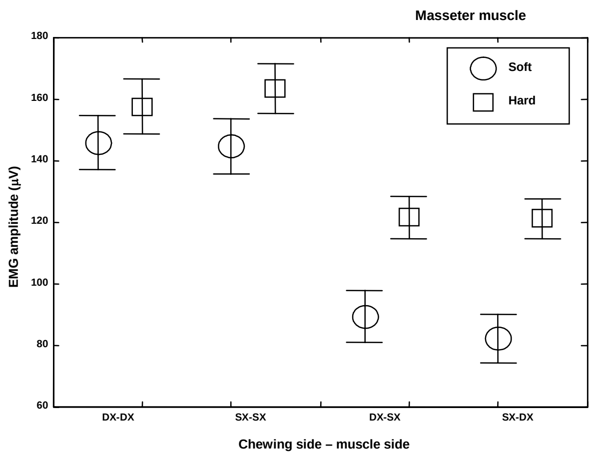

FIG. 4 EMG amplitude of the masseter of the same side of the bolus is significantly higher than the controlateral, but when chewing a hard bolus, in comparison with a soft one, the EMG activity of the masseter of the opposite side of the bolus increases significantly more than that of the masseter of the same side of the bolus.

It is concluded that the larger is the load, the effort and the efficiency required, the more the EMG activity becomes symmetric and the two masseters reach EMG amplitudes more similar than that showed with a soft bolus.

The increase of the masseter of the opposite side of the bolus, when chewing a hard bolus, from a clinical point of view, shows the stomatognathic system capability to apply load, to adapt to the effort of a hard bolus, and to increase the chewing efficiency, in a normal subject [6].

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Differentiating Deficits In The Range Of Motion Of The Arm Using 3-Dimensional Motion Analysis

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Abstract – *For the evaluation of patients with movement restrictions in the upper extremities it is essential to have information about their functional abilities during daily activities. Using three-dimensional motion analysis it is possible to objectively evaluate functional performance. 10 healthy subjects and 8 patients participated in the validation of the screening procedure developed. The movement tests were selected to closely approximate the activities of daily living. The data was analyzed and displayed using purpose-designed software. The results are presented in terms of a two-dimensional representation of the range of motion (ROM). This new screening procedure allows the objective and quantitative evaluation of motion in the upper extremities for a variety of tasks and greatly enhances documentation of the affected shoulder during therapy.*

1. Introduction

The shoulder or glenohumeral joint is the most complicated and most movable large joint of the body. The design of the joint is such as to allow a large range of motion (ROM). However, it is this very design which tends to make the joint unstable and easily subject to injury. Shoulder problems arise from disruptions of the soft tissues which stabilize the joint or from a degenerative process. For patients presenting with shoulder problems loss of mobility and a restricted ROM can often lead to varying levels of disability and a reduction in the quality of life. The doctor's typical decision-making process is dependent on the results of the clinical examination and his subjective assessment of the patient. However, such a screening procedure does not objectively evaluate changes in the performance of movements by the patients during tasks. This information is essential in order to comprehensively assess patients' status, improve the diagnosis of disability, determine the best course of treatment, better compare pathological with normal movements and to assess the effectiveness of therapy [1]. The objective evaluation of functional movement restriction is made possible by the procedure referred to as three-dimensional motion analysis. To date, clinical implementation of this procedure focuses primarily on the evaluation of movement in the lower extremities i.e. gait analysis. The application of this procedure for the analysis of motion in the upper extremities is rare since arm movements are complex, non-repeatable and non-cyclic. However, the expected scientific and clinical benefits of investigating unconstrained motion of the upper limbs are a motivating factor to also establish quantitative three-dimensional motion analysis as a screening procedure for the upper extremities.

2. Patients/Method and Materials

The degree of functional movement restriction was investigated for a control group (5 females, 5 males) and 8 male patients presenting with shoulder impingement syndrome. The subjects were screened using a three-dimensional motion analysis system. By means of infra-red cameras, the motion analysis system records, with high spatial and temporal resolution, the trajectories of passive reflecting markers attached to the limbs (c.f. Fig 1).

Fig 1: Marker triplets on body segments

Four movement tests were selected which, according to patient observations, posed the greatest difficulty to perform during their daily activities. These were:

- Abduction of the extended arm
- Elevation: lifting a package from the knee unto a shelf
- Reaching from the knee to the nape of the neck.
- Reaching from the knee to the lower back

Kinematic analysis was supported by a suitable biomechanical model that divides the body into rigid segments interconnected via ideal joints [2]. Each segment is defined by a triplet of passive infra-red light reflecting markers (c.f. Fig. 1). The biomechanical model permits the correlation of the marker paths with motion of the limb. Subsequent data analysis and depiction of the ROM were performed using software developed at the Helmholtz Institute. The patients were screened both before and after therapy.

3. Results

Functional ROM is analyzed by considering the changes in joint Cardan angles with respect to time. The range of shoulder motion is represented as a 2-dimensional depiction. The elbow path during each movement test was plotted onto a spherical coordinate system centred on the middle of the shoulder joint (c.f. Fig. 2)

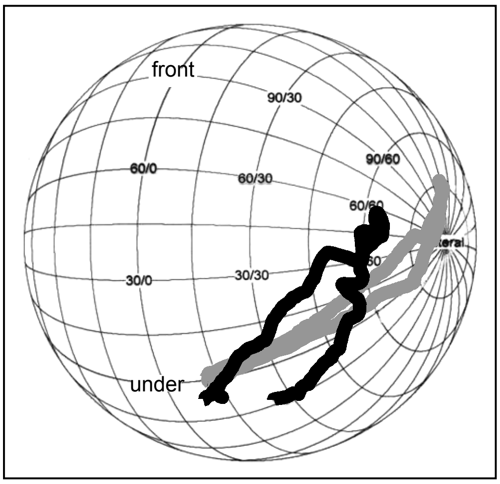

Fig 2: Graphical representation of the range of motion (ROM) of a patient before and after therapy using a spherical coordinate system.
Side: left shoulder, Movement test: Abduction of the extended arm
Legend: black curve: before therapy, grey curve: improvement after therapy

This depiction of the movement facilitates both the objective and quantitative assessment of ROM and greatly improves the comparison of the characteristic arm movement patterns of individuals or groups. This implies that the functional capability within the patient group before and after therapy and also with respect to the control group can be objectively assessed (c.f. Fig. 2).

4. Discussion

The visualization of functional movement restriction by means of a two-dimensional movement pattern has enhanced the ability to compare the status of patients with that of healthy subjects. Furthermore, it is now possible to clearly identify changes in performance due to a prescribed rehabilitation programme. Clinical examination of the pathological group using the clinical test referred to as the Neutral-Zero Method confirmed the graphically represented ROM. It has been found that the newly selected movement tests are quite suitable for use in the monitoring of the effectiveness of therapy. Three-dimensional analysis of arm movement is in the position to provide a suitable clinical test in the area of rehabilitation. With further investigations into the validity and reliability of the movements tests and screening protocol this methodology can do much to augment the ability to differentiate between individual deficits and thus enhance quality control in the area of rehabilitation.

5. Conclusion

In the future, quantitative three-dimensional motion analysis extremities will play a valuable role in quality control during diagnosis and treatment, as well as for the design of shoulder rehabilitation programs.

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Non-invasive assessment of jaw elevator muscle anatomy and implications on the sensitivity of amplitude and spectral surface emg variables to different electrode locations

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Abstract - *The aims of this work were 1) to identify the main innervation zone(s) in the superficial masseter, deep masseter, and temporalis muscles, 2) to estimate, from non-invasive recordings, the fiber length on these muscles, 3) to investigate the sensitivity of amplitude and spectral surface EMG variables to electrode location, 4) to analyse if this sensitivity is affected by the inter-electrode distance of the bipolar recording, and 5) to investigate the effect of inter-electrode distance on the estimated amplitude and spectral EMG variables. The study was conducted on 13 subjects. The innervation zones for the superficial masseter, deep masseter and temporalis muscles have been identified and their variability within subjects outlined. The variability of EMG variable estimates along the array may be very large and are reduced by increasing the interelectrode distance.*

1. Introduction

In this work we applied multi-channel surface EMG techniques for the investigation of the myoelectric activity of jaw elevator muscles. The aims were 1) to identify the main innervation zone(s) in the superficial masseter, deep masseter, and temporalis muscles, 2) to estimate, from non-invasive recordings, the fiber length on these muscles, 3) to investigate the sensitivity of amplitude and spectral surface EMG variables to electrode location, 4) to analyse if this sensitivity is affected by the inter-electrode distance of the bipolar recording, and 5) to investigate the effect of inter-electrode distance on the estimated amplitude and spectral EMG variables.

2. Materials and Methods

2.1 Subjects

The study was conducted on 13 subjects (9 males, 4 females, age (mean ± SD) 27.3 ± 2.7 years, overjet 3±1 mm, overbite 3±1, SpP^GoGn 22°±5°). Inclusion criteria were 1) no signs or symptoms of temporomandibular disorders (TMD), 2) no cross-bite, 3) no orthodontic treatment, 4) no prosthetic rehabilitation, and 5) no missing teeth.

2.2 Experimental protocol

The masseter muscle, deep and superficial bundles, and anterior temporalis of the right and left side were tested. Each muscle was analysed during a maximum voluntary clenching in the intercuspal position for ten seconds. The subject performed six clenching. A rest period of 2 minutes was given between each contraction to avoid fatiguing effects.

2.3 Signal recording

The EMG signals were detected from the surface of the skin using a linear array [1,2] of 16 silver electrodes (point electrodes, 1 mm diameter, 2.5 mm inter-electrode distance between centres). The skin was cleaned and slightly abraded before electrode placement. The array was fixed with adhesive tape. The portion of each muscle covered by the surface electrodes was 37.5 mm (16 electrodes with 2.5 mm inter-electrode distance) (Figure 1).

The EMG signals were detected in single differential mode to minimise line interference, amplified (16 channel surface EMG amplifier, EMG 16, LISiN – Prima Biomedical & Sport, Treviso, Italy), and filtered (3 dB bandwidth, 10-500 Hz), sampled at 2048 Hz, and converted to a numerical format using a 12 bit A/D converter. The single differential detection was performed between adjacent electrodes resulting in an inter-electrode distance of 2.5 mm.

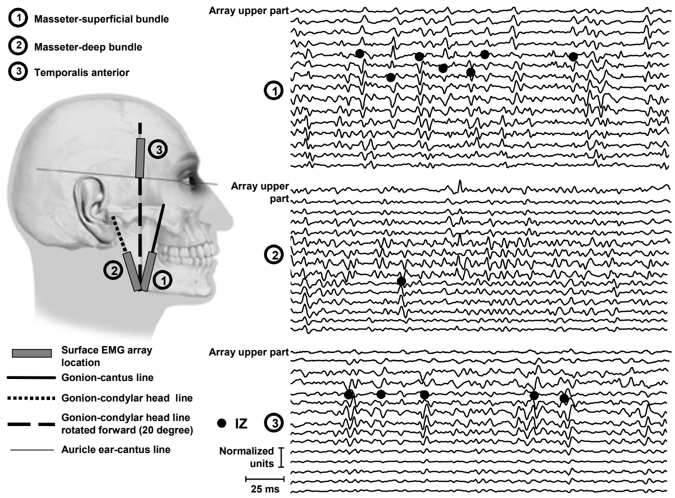

FIG. 1 : Schematic representation of the anatomical lines used for electrode location and examples of surface multi-channel EMG signals detected from the three muscles of one subject (right side). Within each array the signals are normalized with respect to the maximum value. The location of the innervation zones (IZs) of a few MUs is indicated by a black circle. Fiber length has been estimated for a subset of potentials which could be identified in the interference pattern and for which the two tendon regions could be detected by clear decrease in signal amplitude

2.4. Signal processing

The total variability Δ_{tot} was defined, for each EMG variable, as the percentage difference between the highest and the smallest EMG variable value estimated in all the considered electrode locations. This index indicates the percentage difference which can occur between EMG variable estimates when the electrodes are placed without specific criteria in the muscle region covered by the array (37.5 mm). The local variability Δ_{min} was defined, for each EMG variable, as the minimal percentage difference among all the percentage differences in EMG variable estimates from adjacent electrode locations. Thus, it indicates the difference which may occur in EMG variable estimates when the recording system is placed in the optimal location with a possible error of ±2.5 mm (the original inter-electrode distance). The location resulting in Δ_{min} was considered as the optimal one.

3. Results

The innervation zone locations and their variability within subjects are reported in Table I.

Tab. I : Range of the locations of the innervation zones (IZs) for the three muscles and the 13 subjects. The results from both sides are pooled together. In the case of the superficial and deep masseter, the IZ location is reported in percentage of the anatomical lengths introduced (see text and Figure 1 for details) while for the temporalis anterior the IZ location is indicated as an absolute distance.

	S1	S2	S3	S4	S5	S6	S7	S8	S9	S10	S11	S12	S13
Sup. Masseter (IZ location, %)	22.2 - 38.8	17.5 - 30.0	20.4 - 27.2	22.7 - 29.5	17.5 - 30.0	15.7 - 31.5	13.6 - 29.5	21.4 - 30.9	25.0-31.8	20.4 - 29.5	25.0 - 36.1	25.0 - 44.4	25.0 - 33.3
Deep Masseter (IZ location, %)	25.0 - 50.0	25.0 - 50.0	19.4 - 36.1	25.0 - 27.7	20.5 - 35.2	17.8 - 42.8	30.5 - 36.1	28.1 - 31.2	22.2-30.5	26.4 - 38.2	30.0 - 46.6	28.5 - 50.0	30.5 - 42.8
Temporalis anterior (IZ location, mm)	18.0 - 28.0	14.0 - 20.0	12.0 - 22.0	10.0 - 20.0	8.0-18.0	10.0 - 18.0	12.0 - 22.0	10.0 - 24.0	18.0	16.0 - 22.0	12.0 - 26.0	10.0 - 24.0	18.0 - 24.0

The estimated fiber lengths are reported in Table II. The percentage difference in EMG variable estimates along the array was very large (more than 100% of the estimated value for amplitude). Increasing the inter-electrode distance implied a significant reduction of the variability of estimates due to electrode displacements (Figure 2).

	N = 13 Fiber length from surface EMG (mm) Right side	N = 13 Fiber length from surface EMG (mm) Left side	N = 8 Fiber lenght from van Eijden et al. (1997) (mm)
Sup. Masseter	27.3 ± 2.4	27.0 ± 1.7	24.6 ± 4.1
Deep Masseter	21.8 ± 1.6	21.6 ± 0.8	18.0 ± 2.8
Temporalis	25.9 ± 2.3	26.6 ± 1.6	27.1 ± 3.8

Tab. II : Fiber length (mean ± SD) of the masseter and temporalis anterior muscles, obtained by the analysis of the multi-channel surface EMG signals and compared with the data obtained from 8 cadavers [3]. No statistical difference (Student t-test for dependent samples) was observed between the right and left side.

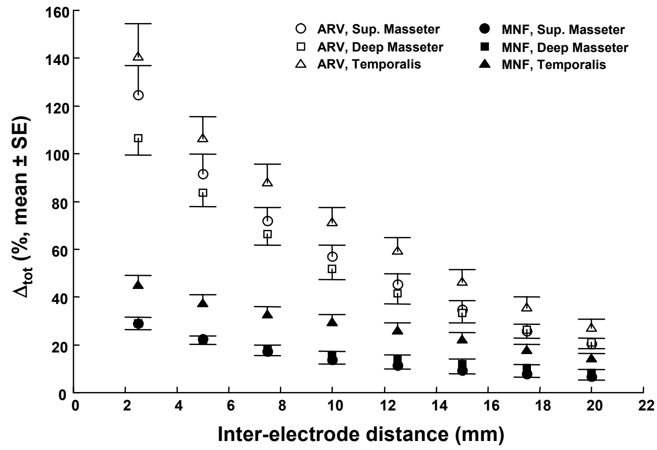

Fig. 2 : ARV and MNF percent variability with location (Δtot is the maximal percent difference between estimates obtained from different locations along the array) as a function of the inter-electrode distance for the three muscles investigated. Both sides are pooled together.

4. Discussion and conclusion

This study reported estimates of fiber length and innervation zone locations from multi-channel EMG recordings. The estimated fiber length was comparable with the results obtained from cadavers (van Eijden et al., 1997) (Table II). The possibility of estimating muscle fiber length has important implications in clinical research and for the issues of identifying the area over the skin corresponding to propagating potentials.

As a consequence of the short fiber length and of the large spread of innervation zone locations, the sensitivity of EMG variables to electrode location was very high when all the possible locations in the muscle portion investigated were compared. From Figure 2 it appears that, especially for small inter-electrode distances (2-5 mm), the percentage differences between estimates obtained in different

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