A thin, flexible multi-electrode grid for high-density surface EMG

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Lapatki B.G. 1,2, Van Dijk J.P. 2,3, Jonas I.E. 1, Zwarts M.J. 2,3, Stegeman, D.F. 2,3

1 Department of Orthodontics, School of Dental Medicine, University of Freiburg i.Br.,
Germany

2 Institute of Neurology, Department of Clinical Neurophysiology, University Medical
Centre Nijmegen, The Netherlands

3 Interuniversity Institute for Fundamental and Clinical Human Movement Sciences,
Amsterdam, Nijmegen, The Netherlands

Corresponding author:
Bernd G. Lapatki
Department of Orthodontics, School of Dental Medicine
University of Freiburg i.Br.
Hugstetter Str. 55
D-79106 Freiburg i.Br., Germany
e-mail: lapatkib@zmk2.ukl.uni-freiburg.de
phone: +49 761 2704851
fax: +49 761 2704852

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ABSTRACT

Although the value of high-density surface electromyography (sEMG) has already been proven in fundamental research and for specific diagnostic questions, there is as yet no broad clinical application. This is partly due to limitations of construction principles and application techniques of conventional electrode array systems. We developed a thin, highly flexible, 2-D multi-electrode sEMG grid, which is manufactured using flexprint techniques. The material used as electrode carrier (Polyimid®, 50µm thick) allows grids to be cut out in any required shape or size. One universal grid version can therefore be used for many applications reducing costs. The reusable electrode grid is attached to the skin using specially prepared double-sided adhesive tape, which allows the selective application of conductive cream only directly below the detection surfaces. To explore the practical possibilities, this technique was applied in single motor unit analysis of the facial musculature. The high mechanical flexibility allowed the electrode grid to follow the skin surface even in areas with very uneven contours, resulting in good electrical connections in the whole recording area. The silver-chloride surfaces of the electrodes and their low electrode-to-skin impedances guaranteed high baseline stability and a low signal noise level. The electrode-to-skin attachment proved to withstand saliva and great tensile forces due to mimic contractions. The inexpensive, universally adaptable and minimally obstructive sensor allows to extend the principal advantages of high-density sEMG to all skeletal muscles accessible from the skin surface and may lay the foundations for more broad clinical application of this non-invasive, 2-D sEMG technique.

KEYWORDS

Electrode array, single motor unit analysis, multichannel EMG, electromyography, facial muscles
INTRODUCTION

Surface electromyography (sEMG) is applied in many areas of muscle research and patient care, for instance in human movement sciences, rehabilitation, ergonomics and clinical neurophysiology. The single bipolar montage, commonly used in these fields, is of value for diagnostic purposes, but offers little information at a single motor unit (MU) level. To deal with this limitation, sEMG techniques have been extended with the design of linear and two-dimensional electrode arrays and grids (3,23,28,29,33,37). These multi-electrode grids cover a larger part of the muscle and add spatial information that is largely independent of the “classical” temporal information. If electrode grids with small electrode sizes and inter-electrode spacing (so-called “high-density sEMG arrays”) are used, it is possible to decompose the sEMG interference pattern for single MU analysis and the determination of firing events of individual MUs (see Refs. 10,18,24,36,38). Although single MU analysis is a key attraction of multichannel sEMG, spatial (or topographical) information from a muscle is also useful in other respects: [1] It allows the construction of higher order electrode montages for spatial filtering (7,15); [2] the muscle’s spatial functional properties can be studied and mapped, respectively (19,27,32); [3] grid areas with high signal amplitude can be selected (on-line or off-line) for more detailed inspection or analysis (20).

High-density sEMG has proven its value for fundamental research and is acknowledged for specific diagnostic purposes (8,39), but there is as yet no broad clinical application. A crucial first step on this road is the availability of sEMG electrode grid systems that are [1] flexible in their use, i.e. their applicability should include as many skeletal muscles as possible, [2] inexpensive (at least the sensor component), and [3] easy to apply and to maintain (including sterilisation or at least sufficient disinfection). Although successful progress regarding these criteria has
been made, sEMG grids available up to now still contain restrictions which mainly result from their construction principle; they usually consist of metal pins or bars (as single electrodes) mounted in apertures of a substrate sheet and integrated in a single container. A considerable disadvantage resulting from this construction is the fixed, i.e. unchangeable size of the array and the arrangement of the electrodes. Consequently, the examination of muscles which have largely different sizes and shapes requires a number of different grids. For large scale use, this might become ineffective and expensive. Other significant disadvantages of such container arrays are the unavoidably large outer dimensions (especially the large height of several cm) and the limitations regarding mechanical flexibility. In the examination of small muscles, bulky and relatively stiff electrode arrangements may hinder (or even alter) the functions to be studied (11). In uneven skin areas (e.g. in the face), grids with a limited mechanical flexibility do not completely follow the patient’s anatomy; consequently, a good electrode-skin-contact cannot be achieved in the entire recording area.

For currently available sensor systems, it is not only the electrode arrangement itself, but also the technique by which the sensor is attached to the skin, that contains some principle problems. In most techniques, skin attachment is realized by means of external fixations. These consist either of bands of Velcro which are tied round a limb (3,28) or medical plasters which are tightly drawn over the whole electrode containers and then adhered to the adjacent skin. In some areas (e.g. in the face), it is not practical or even possible to employ such an attachment technique. As a consequence, the application of these conventional sEMG grid systems is limited to certain muscles and/or positions. Moreover, external fixations unavoidably compress the soft tissue in the area of the sensor. This pressure might cause problems if [1] the array consists of sharply contoured electrode surfaces used to reduce electrode-to-
skin impedance in dry application (i.e. in application without conductive cream or gel), [2] the array is applied in areas where the skin is relatively thin and sensible, and [3] the underlying hard tissue has an uneven contour. These are factors causing uneven pressure distribution in the recording area, leading to electrode impressions on the skin and pain to the subjects and patients. In addition, there is a certain risk that sharp electrodes applied with pressure even slightly injure the skin. Together with the risk of insufficient cleaning and disinfection (which may occur in routine clinical applications), slight skin lesions may lead to transference of pathogens (although this infection risk can be considered small).

In this contribution, we describe the design and performance of a thin, highly flexible multi-electrode sEMG grid. The aim of this development was to provide a relatively inexpensive, easily adaptable and minimally obstructive sensor, thus laying the foundations for more broad clinical applications of high-density sEMG, and extending the use of this technique to all superficially located skeletal muscle groups. A crucial aspect connected with the application of such an electrode grid type was the development of a special skin attachment technique that yields firm sensor fixation and sufficiently low electrode-to-skin impedances without requiring external fixations. In order to explore and demonstrate the practical possibilities, this newly developed sensor technique was applied in the facial musculature. The facial musculature was chosen because of the large methodological demands in this area relative to others (5,21).

MATERIALS AND METHODS

Electrode grid design

The manufacturing process of the new multi-electrode sEMG grid principally corresponds to that of a standard flexprint, i.e. the detection surfaces and connecting
wires are chemically etched and electrochemically deposited on a highly flexible carrier material (Polyimid®, 50µm thick). The layout and design was made in cooperation with Digiraster Tetzner GmbH (Stuttgart, Germany). The printed rectangular grid consists of 7 columns of 13 electrodes each (Fig. 1A).

Fig. 1. A: Photograph of an sEMG multi-electrode grid (including the connection tails) manufactured using flexprint techniques. The basic grid design used in our application example consists of 7 by 13 electrodes, with an inter-electrode distance (IED) of 4mm (center-to-center) in both directions. The thin (50µm) electrode carrying material (Polyimid®) allows smaller grids to be cut out from the basic grid. In principle, grids of any desired electrode sizes and arrangements can be manufactured. B: Each electrode (1.95mm in diameter) consists of a copper body, which has been surface coated with pure-silver (99.99% Ag). The traces printed on the reverse side of the flexprint are visible in this photograph due to the translucency of the Polyimid® material. Perforations of 1.2mm in diameter (each centred between 4 electrodes) were made to facilitate skin attachment of the electrode grid. C: Illustration of the high mechanical flexibility of the multi-electrode grid. Total thickness of the flexprint (electrode, carrier material, traces, protection lacquer) is 470µm in the area of the electrodes and 150µm between them.
The inter-electrode distance (IED) is 4mm center-to-center in both directions. This value was chosen on the basis of the spatial version of the Nyquist sampling criterion (see Ref. 3), as well as the relation between the number of available recording channels and the required size of the recording area. The thin electrode carrying substrate allows smaller grids, of any desired size, to be cut out of the basic grid version with a common scissors (in our application example, we used trimmed grids of 5x12 and 6x10 electrodes, respectively, see below). The flexprint includes a regular pattern of 1.2-mm diameter perforations (through holes) each centred between 4 electrodes.

The grid's electrodes have the shape of a solid circle with a diameter of 1.95mm, protruding 300µm from the Polyimid® material. They consist of copper, which is surface coated with a pure-silver (99.99% Ag) layer (Fig. 1B). Before chloriding the electrodes, we roughen their outer silver surfaces using a fine glass fiber pen. Total thickness of the grid (electrode, carrier material, traces, protection lacquer) is 470µm in the area of the electrodes and 150µm between them. The high mechanical flexibility of the grid resulting from this minimal thickness and the material properties is demonstrated in Fig. 1C.

For electrical connection of the electrodes, traces of 80µm width are printed on the reverse side of the Polyimid® substrate (i.e. on the opposite side of the electrodes). Their position on the reverse side protects the traces against damage due to removal of the double-sided adhesive tape (used for skin attachment, see below) or cleaning of the grid's detection surface after the measurements. Printed trace lines are insulated with a flexible, non-conductive overlay lacquer. The traces of one electrode column (13 electrodes in our grid design) converge into groupings of parallel lines that lead to a 3mm wide and 50mm long tail (Fig. 1A). Every tail terminates at a connection end that has exposed electrically conductive surfaces and is dimensioned
to mate with an external connector (13FLZ-SM1-TB, J.S.T. Deutschland GmbH, Winterbach, Germany).

**Skin attachment of the electrode grid**

An electrode grid is attached to the skin using 100µm thick double-sided adhesive tape (1522 Medical double coated tape, 3M, St. Paul, MN, USA). This tape has been specially prepared by creating regular patterns of holes of 2.2mm in diameter to leave the electrode areas blank, and smaller holes (1.2-mm diameter) that topographically match with perforations of identical size in the electrode grid (see Fig. 2A).

The attachment procedure is as follows: after shaving the skin in the corresponding area (if necessary) and cleansing it with an alcohol-wetted swab, we fix the prefabricated double-sided adhesive tape in the correct position. In order to reduce electrode-to-skin impedance we then evenly apply conductive cream (Elektrodencreme®, Marquette-Hellige, Freiburg i.Br., Germany) in the whole area of the attached tape. The surplus cream is removed on the outer barrier foil of the tape using a dental cotton roll, leaving only a thin cream layer on the skin of the blank electrode spaces (Fig. 2A). Since we peel off the outer barrier foil after the cream has been applied, it is guaranteed that the tape’s outer bonding surface is kept free from electrode cream. The next step is to accurately fixate the electrode grid on the outer bonding surface of the attached double-sided adhesive tape. Proper positioning is facilitated by identical patterns of 1.2-mm perforations in the electrode carrier material and the adhesive tape (see Fig. 2B); the detection surfaces are exactly centred in the blank 2.2-mm electrode spaces, if the 1.2-mm perforations of the electrode grid and the tape exactly meet.
Fig. 2. A: Double-sided adhesive medical tape (1522 Medical double coated tape, 3M, St. Paul, MN, USA) attached to the skin with surplus electrode cream and dental cotton roll used for surplus removal. Perforations of 2.2mm in diameter were punched out to leave the detection surface areas blank. Smaller (1.2-mm) perforations were made to facilitate the skin attachment procedure. B: Attachment of the electrode grid on the double-sided tape after the outer protecting foil of the tape has been removed. The grid is accurately positioned on the adhesive tape if the 1.2-mm perforations in the electrode carrier material exactly meet the perforations of identical size in the adhesive tape.
After the measurement, the grid and adhesive tape are detached from the skin by carefully pulling at the connection tails. The tape can then be pulled off from the flexprint. It is advisable to clean the electrode grid immediately after detachment with an alcohol-wetted gaze for subsequent usage.

**Application example**

**Subjects**

This new technique was tested in the facial musculature on a group of 13 healthy subjects (males n=6, females n=7, mean age 27.2 years). Nine of them were trumpet students or professional trumpeters, who were expected to have good facial motor control. The participants had no known neurological or general health disorders. The Ethics Commission of the University Medical Centre Nijmegen (NL) approved the protocol.

**Setup for the measurements and data acquisition**

In each subject, we observed the upper and lower facial muscles in two separate sessions (either on the right or left side of the face). In the upper face, sEMG measurements were made using two 5 by 12 electrode grids positioned side by side (see Fig. 3). In the lower face, we used two grids of 6 by 10 electrodes that were also positioned side by side, but vertically displaced by one electrode row (see below in Fig. 6A). Thus, in both recording areas 120 channels were simultaneously sampled. By using grids of different shapes in the upper and lower face, we could optimally adapt the dimension of the recording area to the anatomy of the underlying musculature.

A single, flexible and lightweight cable (132 conductors Junflon PFA coaxial round cable, J14B0596-A, W.L. Gore & Associates GmbH, Pleinfeld, Germany), 0.7m in length and 8mm in outer diameter, was used to electrically connect the electrode grids to the amplifiers. It contains 132 individually shielded leads and additionally has
an outer shield. Especially for application in the face, the cable has been split in two parts behind the head. The two cords were separately guided to the face along movable arms, which are mounted on a headset (Fig. 3). The 3-dimensional flexibility of the arms allowed the connectors to be positioned in a manner that the tails of the electrode grid adopt a curved course guaranteeing maximal freedom of movement between the attached electrode grid and the connectors.

Fig. 3. Setup for high-density sEMG in the upper facial area. We used two trimmed electrode grids (each consisting of 5 by 12 electrodes) positioned side by side. The cable from the amplifiers was split in two parts behind the head, each containing the connections of one electrode grid. Both cords were guided along two 3-D movable arms that were fixed at the right and left sides of a headset. At the distal end of both arms, two hardprints were mounted serving as a strain relief. The hardprints were equipped with connectors having lockable slots that match with the connection ends of the electrode grids. We secured the borders of the electrode grids in areas that are exposed to moisture or high tensile forces using strips of elastic medical plaster (Fixomull Stretch®, Beiersdorf, Hamburg, Germany).
Signals were recorded monopolarly, referred to an electrode positioned on the dorsum nasi. A second reference electrode provided a common mode signal (CMS). Both reference electrodes used (Mühl, Freiburg, Germany) are 4mm in diameter and have been originally developed for conventional facial sEMG recordings (21). They consist of sintered Ag / AgCl and therefore have similar electrochemical properties as the grid electrodes. A driven right leg (DRL) electrode (see Ref. 26) was attached at the forehead. The 128-channel-system used for data acquisition (Mark-6, Biosemi Inc., Amsterdam, NL) has an input impedance > 100MΩ and a CMRR > 120dB. Signals are band-pass filtered (3.2 - 400Hz; high-pass: first order Bessel, low-pass: fourth order Bessel) and synchronously sampled at 2000Hz with a resolution of 0.5µV over a range of +/- 16mV (16 bits). The acquisition software allows the experiments to be controlled by on-line inspection of the mono- or bipolar data of selected electrode rows or columns. Electrode-to-skin impedances can be measured using a 20mV (p-p), 62.5Hz square wave signal over a 1MΩ resistance in series with two electrodes (i.e. the corresponding grid electrode and the reference electrode). The maximal current is 20nA. Such low current will not disturb the skin-metal double layer, which would make the impedance current dependent. See also Ref. (3) for more technical details regarding the recording system.

**Recording procedure**

At the beginning of each recording session, subjects were instructed and trained in performing selective contractions of the muscles to be examined. After placing and electrically connecting the two grids, we performed an initial impedance measurement. Surface EMG signals were then recorded while each muscle in the recording area was selectively activated at different levels. Since especially in the facial muscles, contractions at a constant level are difficult to perform, we implemented an sEMG amplitude feedback tool into the acquisition program that
visually displayed the activity level of a selected muscle to the subject via the PC monitor. Muscle activity level was determined by calculating the normalized mean value of the RMS values of selected bipolar signals (window duration for RMS calculation was 500ms, normalization was made to a maximal reference contraction of the muscle). Each recording session lasted for approximately 2 hours and was finished by a second impedance measurement.

**Data analysis**

Data analysis was performed off-line using algorithms programmed in Matlab, Version 6.5 (The Mathworks Inc., Natick, MA, USA). Impedance data (120 impedance values per measurement) were statistically evaluated by calculating boxplot percentiles separately for each subject, recording moment (initial and final impedance measurement), and measurement location (upper and lower face). For each of the five calculated boxplot percentiles (i.e. the 5th, 25th, 50th, 75th and 95th percentiles) we then determined the median values across the study group (separately for the initial and final recordings and the two measurement locations). Noise performance of the grid was evaluated by calculating RMS values of signals recorded while the muscles were kept at rest. We only evaluated measurements taken in the upper face about one hour after electrode grid application. Similar to the impedance evaluation, we calculated percentiles for the 120 values (per measurement and subject, respectively) first; from these individual percentile values, median values across the study group were determined.

For analysing the topography of facial MUs, we decomposed the sEMG interference pattern by [1] detecting peaks in the bipolar signals of three selected channels, [2] classifying these peaks according to their differential spatio-temporal amplitude and firing characteristics, and [3] averaging the bipolar data windows around the detected peaks for all 120 channels. Peak classification was based on the Ward’s clustering
algorithm using “Euclidean distances” between the spatial and the temporal peak characteristics (for more details see Ref. 18).

RESULTS

The evaluation of the impedance measurements (Fig. 4) indicate a statistically significant reduction of the electrode-to-skin impedance values during the recording sessions (Wilcoxon signed-rank test, p<0.01) from 163.7 and 137.6kΩ (median values of the 50th percentiles for the measurement in the upper and lower face, respectively) to 75.7 and 99.4kΩ. The difference between the initial and final impedance values appeared to be larger in the upper than in the lower face.

![Fig. 4. Results of the impedance measurements made at beginning and end of the recording sessions in the upper and lower face. Impedances were measured by applying a 62.5Hz square wave signal over a 1MΩ resistance in series with the reference electrode (which has a relatively low impedance value) and the corresponding grid electrode. The values given in these boxplot diagrams (from bottom to top 5th, 25th, 50th, 75th or 95th percentiles) are the median values across the study group calculated from the corresponding percentile values of the individual measurements (see also ‘Data analysis’ section).](http://jap.physiology.org/Downloaded from 10.220.33.2 on June 11, 2017)
Thin and flexible high-density surface EMG grid
Fig. 5A shows a subset of bipolar sEMG signals derived from a single vertical column of electrodes during a short contraction of the depressor anguli oris (DAO) muscle (the orientation of the selected electrode column approximately corresponds to the muscle fiber direction). The bipolar montage was constructed by subtracting the monopolar signals of consecutive electrodes. As recognized by Masuda et al. (22), the position of the motor endplate zone can be detected from the bidirectional propagation pattern of the action potentials, which also results in low amplitude at and opposite signal polarity on both sides of the endplate. Thus, in this recording the neuromuscular junctions are in the area of the 4th bipolar signal (from the bottom), i.e. they are located between the 4th and 5th monopolar electrodes.

In spite of considerable soft tissue movement during the onset period of the DAO contraction, the signals recorded with our new electrode grid showed very high baseline stability (see Fig. 5B, signal only band-pass filtered by hardware). The good signal quality can be assessed from Fig. 5C, which shows a data window of 100ms EMG at large amplification recorded prior to the DAO contraction. For signals recorded in the upper face at rest, we calculated a median noise level of $2.32 \mu V_{rms(RTI)}$ (RTI: referred to input). The corresponding median values of the 5th and 95th percentiles were 1.92 and 5.24 $\mu V_{rms(RTI)}$.

The median value in the study group (calculated from the 50th percentile values of the individual measurements) was 2.32 $\mu V_{rms(RTI)}$ for sEMG recordings taken in the upper face at rest.
Using the algorithm described above, we decomposed sEMG data recorded during selective contractions of different facial muscles at a moderate level, i.e. contractions of 20 - 40% of the maximal voluntary contraction. As a result of these calculations, we obtained templates or “fingerprints” showing the spatio-temporal characteristics of individual MU action potentials (MUAPs). Fig. 6A shows two electrode grids attached for recordings in the lower face. The Figs. 6B-F show examples of MUAPs belonging to different facial muscles underlying the attached grids, i.e. the depressor anguli oris (DAO), depressor labii inferioris (DLI), mentalis (MEN) and orbicularis oris inferior (OOI) muscles.

The template decomposed from data recorded during the selective contraction of the DAO muscle (Fig. 6B) shows a relative symmetric amplitude distribution. The endplates are located in the cranio-caudal center of the muscle. In contrast, the neuromuscular junctions of the MU belonging to the DLI muscle (Fig. 6C) are located inferior to the anatomical center of the muscle (note that the DLI muscle was completely covered by the two 6 by 10 electrode grids). The finding of asymmetrically located endplate zones and oblique fiber direction in this muscle was consistent in all subjects. The template calculated from data recorded during the MEN muscle con-

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**Fig. 6.** A: Two electrode grids attached to the lower face. The position and fiber direction of the underlying facial muscles are indicated by white arrows. B-F: Five different bipolar MU templates decomposed from data recorded during selective, moderate contractions of (B) the depressor anguli oris (DAO), (C) the depressor labii inferioris (DLI), (D) the mentalis (MEN), or (E+F) the orbicularis oris inferior (OOI) muscle. According to the fiber orientation of the corresponding muscle, the bipolar montage was constructed by subtracting monopolar signals of consecutive vertical (in the DAO, DLI and MEN templates, see B-D) or horizontal electrodes (in the OOI templates, see E+F). The grey bar in each template indicates the location of the innervation zone. Note that in the two OOI templates (E+F) the time and amplitude axes are exchanged (in order to facilitate endplate localization in the horizontal bipolar electrode montage). The 2-D orientation of these MUAPs (outlined by double arrows) is in good agreement with the anatomical position and fiber direction of the contracted facial muscles (1,4,30).
traction shows quite an asymmetric amplitude distribution in the vertical direction (Fig. 6D); above the endplate zone, the signal amplitude rapidly decreases in the cranial direction. In the OOI muscle, we found MUs with small territories and neuromuscular junctions in distinct locations (Fig. 6E+F).

**DISCUSSION**

Advantages of a new type of sEMG electrode grid have been demonstrated in a real applied situation. The flexibility of the new electrode grids, together with the possibility of mass/series production, are significant positive features for the development of clinical applications of high-density sEMG. Since the thin electrode carrier material allows grids of any desired size and shape to be cut out, one universal grid version can be used for many applications. If this basic grid is manufactured in large numbers, the unit price can be kept at relatively low level; for a grid size of 7 by 13 electrodes and an output of 1000 units, the manufacturer gives a rough estimate of € 50 / unit.

The performance of the new sensor introduced in this contribution becomes clear from the results of the impedance and noise calculations as well as from the presented signal examples. The electrode-to-skin impedances obtainable with our technique were found to be relatively low. Nearly all impedance values were far below 300kΩ (see Fig. 4). The latter value is – if obtained for the majority of electrodes – considered to be acceptable for such type of recording configuration (3). Impedance measurements with and without the use of electrode cream in three subjects revealed the great significance of electrode cream application regarding a good and stable electrical connection of the electrodes; impedance values with our non-serrated electrode design were found to be about 40 times higher in dry electrode application. The reduction of the electrode-to-skin impedance reduction
during the recording sessions is probably explained by the progressive absorption of the conducting cream. Since the cream was applied only in the areas of the detection surfaces, we avoided short-circuits between adjacent electrodes. Short-circuiting indeed is a problem when the cream is evenly applied in the whole recording area (25). The low impedance values also explain the lower noise level at rest when compared with signals recorded with gold-coated metal pin electrodes (3). In this respect it has to be taken into account that the calculated median noise level (at rest) of 2.32\(\mu V_{\text{rms(RTI)}}\) not only includes thermal noise of the electrodes, but also amplifier noise (0.8\(\mu V_{\text{rms(RTI)}}\) were specified from tests on the instrument) and some physiologic noise. Based on these values, thermal electrode noise plus physiologic noise was 2.18\(\mu V_{\text{rms(RTI)}}\). The good noise performance of the electrode grid may prove to be particularly advantageous in the detection of small and/or deeply located MU potentials. The good baseline stability, demonstrated by the recording during a short contraction, results from the chlorided pure-silver electrode surfaces; this material is known to have excellent metal-tissue properties over a broad frequency range (6,12,16,35).

Due to the exceptionally high mechanical flexibility of the electrode carrying Polyimid® substrate, good electrical connections of the single electrodes could be achieved even in areas with extremely uneven contours (e.g. in subjects having a particularly deep mentolabial sulcus). A flexible electrode arrangement (at least in one dimension) was already achieved in previous high-density sEMG arrays, by mounting metal pin electrodes either on springs (KC McGill, personal communication) or on a semi-flexible print supported by a cushion of plastic foam (3). Indeed, such array versions led to an improvement regarding a more even pressure distribution on the skin surface (the latter is important for minimizing the risk of poor electrical contacts in marginal parts of the recording area and for reducing the mechanical
deformation of the underlying muscle). However, the flexibility obtainable with such conventional techniques is limited (at least it is much less compared with that of polyimid® flexprints). Moreover, the crucial advantage of our new sensor is the combination of both high flexibility and minimal thickness. In investigations in which measured and modelled 2-D sEMG data are compared, it may be advantageous if measurements are taken with a rigid (i.e. inflexible) electrode array applied with pressure, because a curved spatial electrode arrangement (and also changes of the array’s shape during the contraction) would make modelling very complicated. These circumstances can also be realized with our high flexible electrode grid; its electrode arrangement can be kept flat by bonding a rigid plastic foil on the reverse side, and pressure can be exerted using medical plasters or Velcro tied around a limb.

The new technique in which the electrode grid was attached to the skin turned out to be very efficient. In the majority of measurements, the electrode-to-skin attachment proved to withstand moisture (i.e. saliva in the perioral region) as well as large and dynamic tensile forces. We observed a loss of electrical connection or instable electrode-to-skin impedance values during the measurements in a few electrodes at the borders of the grids, in a few subjects showing extremely contoured soft-tissue in the chin and lower lip region (i.e. a deep mentolabial sulcus). Contact loss is explained by these extreme conditions together with the insertion of the MEN muscle fibers in an obtuse angle directly in the mental dermis. Thus, the electrode-to-skin bonding of the grid was obviously exposed to very high tensile forces during maximal contractions of this muscle. An explanation for loss of electrical connection during measurements in the area immediately around the oral opening was the presence of saliva that may have dissolved the adhesive. The fact that contact loss mainly occurred in the chin and lower lip region, i.e. in the lower face, explains the less pronounced impedance reduction during the recordings in this part of the face when
compared with the upper face. The firm skin fixation of the flexible electrode grid arranged by means of double adhesive tape has the disadvantage that the orientation of the electrode grid cannot be changed after it is definitively attached to the skin. This is of significance, as in some muscles (e.g. in large limb muscles) it may be difficult to find the suitable sensor position and orientation (relative to the course of the muscle fibers) without monitoring signal amplitude distribution in the recording area. Possible solutions are to improve the criteria for guiding placement or to repeat the whole attachment procedure if inaccurate grid placement occurs. Another, probably more practicable possibility would be to perform a quick pre-measurement with an electrode grid pressed to the skin by hand. In this manner, the optimal orientation and position might be determined prior to definitive skin attachment. A final possibility that might be feasible is to correct the misalignment in data analysis using special alignment correction algorithms. Possible solutions still have to be explored.

By applying this technique to the facial musculature, we were not only able to demonstrate the good performance of our new electrode grid, but also to illustrate the general possibilities and advantages of the 2-D high-density sEMG technique. The topographical information of decomposed MUAPs makes it possible to determine the position of the motor endplate zones (14,25) and the area with highest signal amplitude (“go where the action is” principle). The results as presented here reveal some of the distinctive characters of facial MU topography; the occurrence of asymmetrically located endplate zones and neuromuscular junctions distributed over distinct areas of the muscle. These findings agree with those of histo-chemical studies (13). Knowledge of the location of neuromuscular junctions and areas of high signal amplitude is indispensable in establishing guidelines for placement of conventional electrode configurations (9,34). The decomposition of the sEMG
interference pattern into the contributions of single MUs appears particularly useful in
the complex facial muscle system consisting of many interweaving and overlying
muscular slips (1,4,30). The spatial profile of extracted MUAPs allows them to be
classified as belonging to certain facial muscle subcomponents. It is therefore
possible to map the highly variable facial muscle structure (17,31). From the clinical
point of view, this may be particularly useful [1] for describing characteristic
alterations on the MU level in neuromuscular disorders (examples of such diseases
affecting the facial musculature are facioscapulohumeral dystrophy and Möbius
syndrome), and [2] for observing regeneration and reinnervation of MUs after
peripheral nerve injuries or muscle transplantation. Another possible application,
based on the spatio-temporal information of decomposed sEMG data, is the
differentiation between the contributions of individual muscles to the sEMG
interference pattern in the examination of specific functions. This option may prove to
be an efficient strategy in the suppression of crosstalk, which is a special problem in
the facial muscle system (2,21). Our results demonstrate that decomposed
multichannel sEMG data even provide information regarding the 3-D orientation of
muscle fibers. The progressive decrease of the signal amplitude in cranial direction in
the MEN template can be explained by an increasing distance between the bio-
electric source and the electrodes (the MEN fibers course from the mental dermis in
dorso-cranial direction towards their origin at the mandibular bone in the depth of the
mental soft-tissues).

Although high-density sEMG is not yet used in daily clinical practice, our universally
applicable and relatively inexpensive sensor might bring this non-invasive technique
closer to the physician. Since sEMG electrodes cannot detect single fiber potentials,
they can not replace intra-muscular electrodes for clinical diagnostic purposes. In the
characterisation of single MUs an overlap exists between these two techniques.
Especially here, our new multi-electrode sEMG grid offers an attractive alternative to conventional needle or fine-wire electrodes, whereby the non-invasive sEMG character is particularly appealing in some recording areas (as shown here for the face), in children, and in long-term studies (36). At present, a number of pathophysiological mechanisms causing neuromuscular disorders have escaped the keen eye of the needle electromyographer due to the topographic character of the pathology (8); studying topographic aspects of neuromuscular diseases (on the level of both the muscle and MU) clearly belongs to the domain of high-density sEMG. In this respect, it is worthwhile to further explore clinical applications and possibilities of this technique.

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