PROSTHETICS/ORTHOTICS/DEVICES

A Self-Contained, Mechanomyography-Driven Externally Powered Prosthesis

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The measurement of the low-frequency (5–50Hz) "sounds" or vibrations produced by contracting muscles is termed mechanomyography (MMG). As a control signal for powered prostheses, MMG offers several advantages over conventional myoelectric control, including, nonspecific sensor placement, distal signal measurement, robustness to changing skin impedance, and reduced sensor costs. The objectives of this study were to demonstrate 2-function prosthesis control based on a triplet of distally recorded, normalized root mean square MMG signals and to identify necessary future research toward full clinical implementation of MMG signals in upper-limb externally powered prostheses. A novel self-contained MMG-driven prosthesis for below-elbow amputees was designed, implemented, and preliminarily tested on 2 subjects. This prosthesis was composed of specialized software and hardware modules that emulate a 2-site electromyography sensing system. Although the use of MMG signals for prosthesis control has been shown previously, we report, for the first time, successful control within a self-contained unit in unconstrained environments. Specifically, essential requirements for practical use, such as standardized sensor attachment, basic noise elimination, and miniaturization of the system, have been achieved. Both subjects were able to voluntarily open and close the prosthesis hand with no significant delays from intention to action (≈120ms). Quantitative analyses revealed 88% and 71% control accuracy for subjects 1 and 2, respectively.

Key Words: Amputees; Arm; Classification; Linear regression; Prostheses and implants; Prosthesis design; Rehabilitation.

MECHANOMYOGRAPHY (MMG) IS DEFINED as the measurement of the mechanical activity produced by contracting muscles. This activity is characterized by low-frequency (<50Hz) vibrations that produce displacements of approximately 500nm on the skin surface (ie, a sound volume of about 10dB 1mm away from the skin).[1] The origin and properties of MMG signals and their relation to muscle physiology have been documented in the literature. Applications in the field of sports physiology and rehabilitation engineering have also been explored.[2] When compared with conventional electromyographic signals (ie, the measurement of muscular electric activity), MMG signals have been found to be more accurate predictors of fatigue during muscle contraction.[3] Because of their propagation properties, sensor placement need not be specific and consistently reproducible. In fact, MMG signals can be recorded distal to the activating muscle, although with reduced amplitude[4]; something not achievable with their electromyographic counterparts. Distal sensor placement facilitates prosthesis manufacturing and reduces the risk of wire breakage. Furthermore, in contrast with electromyographic signals, MMG signals are not affected by changes in skin impedance.[2,5] Finally, MMG sensors are significantly cheaper. For example, at the time of writing, the coupled microphone-accelerometer sensor pairs (CMASPs)[6] used in this study are approximately one seventh the cost of standard electromyography sensors for prosthesis control. With the potential to overcome these practical issues that plague electromyographicically driven systems, the MMG signal is a favorable alternative for prosthesis control. In 1986, Barry et al[5] suggested the use of MMG as a control signal for externally powered prostheses. Using a single-channel, full-wave rectified MMG signal and a tristate control strategy, Barry[5] showed that intentional MMG signals can be used to open and close a free-standing prosthesis hand. The reported system successfully discriminated between wrist flexions and extensions while exhibiting robustness to changes in sensor placement and skin impedance. However, unresolved sensor-attachment challenges, low-frequency noise elimination issues, and the lack of miniaturization of the system precluded its practical use. Since that initial study, we have made several significant strides toward the implementation of a self-contained, MMG-driven, externally powered prosthesis for below-elbow amputees. In this article, we report our efforts to date toward implementing such a device.

METHODS

The MMG-driven prosthesis is composed of specialized hardware and software modules that emulate a conventional 2-site electromyography sensing system. This particular setup facilitates the use of existing electromyography-based control strategies in the prosthesis hand by eliminating the need for additional adjustments.

Hardware

The hardware (fig 1) consists of (1) a soft silicone socket (the MMG socket) with an array of embedded MMG sensors, (2) a hard socket and hard cover that act as the actual forearm and contain the power supply (6-V Ni-Ion battery[a]), (3) a miniaturized (diameter, 4cm) electromyography emulator board, and (4) a standard Otto-Bock hand.[a]

MMG sensors. MMG signals were transduced with the CMASPs[6] designed in our laboratory (fig 2). CMASPs can be built by combining a standard microphone enclosed within a

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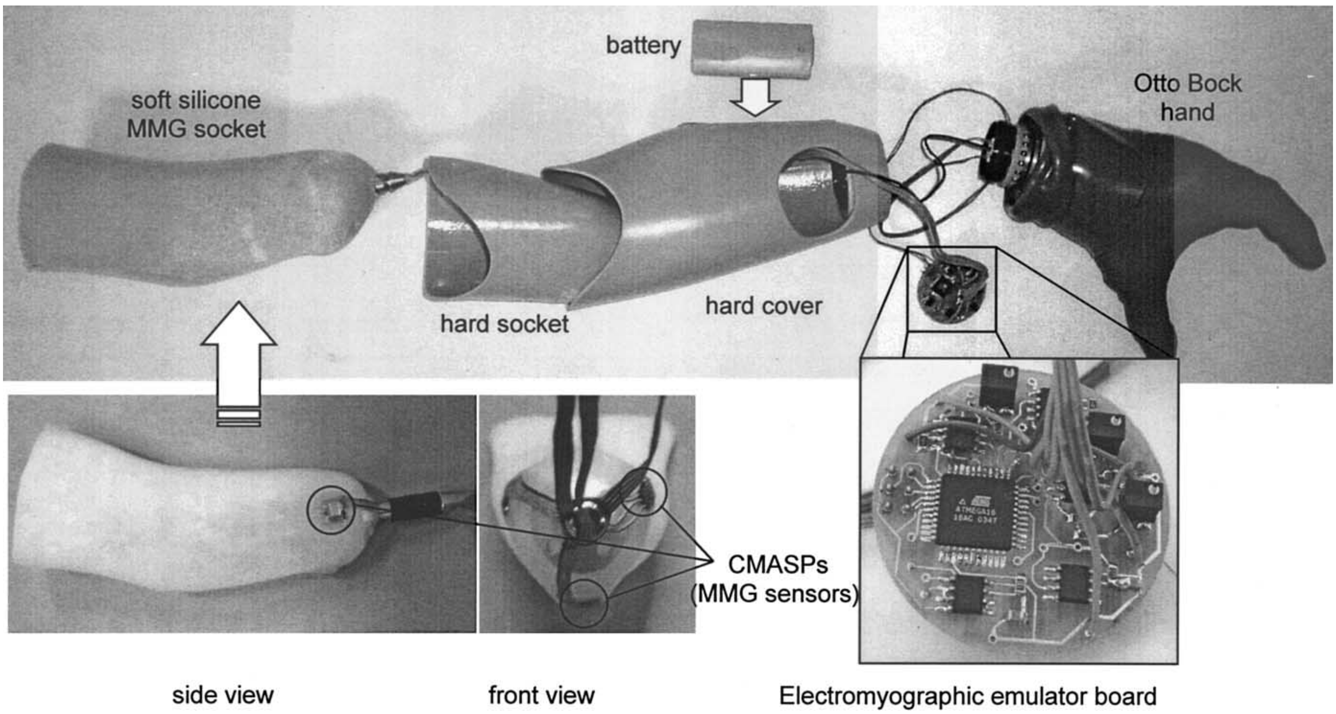

Fig 1. Hardware components of the MMG-driven prosthesis arm. The MMG socket (bottom left) contains the 3 CMASPs used to record sounds from contracting muscles. The electromyographic emulator board (bottom right) generates sound-based estimates of electric muscle activity.

sealed air chamber and an accelerometer that measures external interference. CMASPs are small (19×19×9mm), reusable, and optimized for MMG signal recording, and present significant advantages over standard microphones such as passive mechanical low-pass filtering and increased sensitivity. A detailed description of the manufacturing processes and properties of CMASPs can be found in Silva and Chau's report.[6] An array of 3 CMASPs was embedded into the soft silicone socket 1.5cm from the distal end. Each CMASP was placed at equally distant angles (ie, 120°) around the forearm.

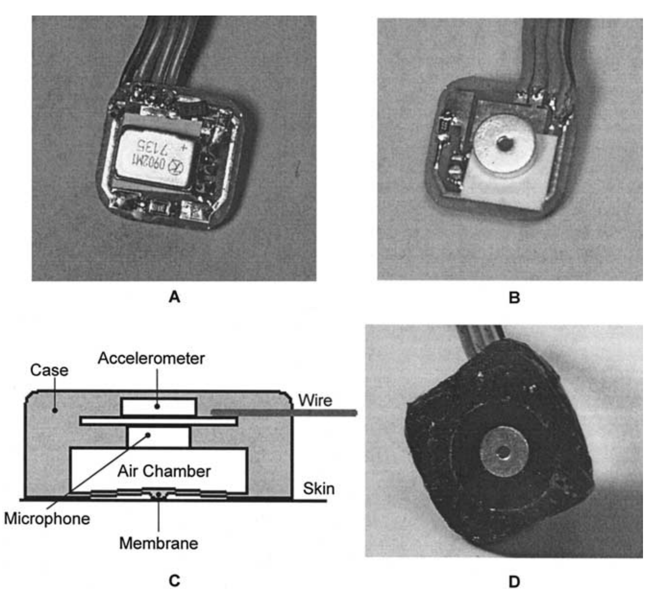

Fig 2. The CMASP.[6] (A) top view (accelerometer's side), (B) bottom view (microphone's side), (C) schematic diagram, and (D) bottom view after encasing (an additional membrane is later added to seal the air chamber).

MMG socket. Soft silicone sockets are an alternative suspension method normally used in passive and aesthetic prostheses. They are molded onto the subject's residual limb, providing a tight fit and, consequently, natural and comfortable suction-based distributed suspension. This enhanced suspension also ensures that the sensors are held tightly against the skin. Furthermore, CMASPs can be embedded within the soft silicone socket itself.

EMG emulator board. This miniaturized board, which sits in the space between the hard socket and the hard cover (ie, distal end of the prosthesis forearm), contains a signal-conditioning stage that includes instrumentation amplifiers and band-pass filters (bandwidth, 5–50Hz) for every CMASP embedded in the array. It also contains an 8-bit microcontroller programmed with the necessary algorithms that emulate electromyographic signals as inputs to the prosthesis hand.

Software

The electromyography emulation software (fig 3) consists of (1) a preprocessing module, (2) a feature extraction module, (3) a signal detection module, and (4) an electromyography emulation module.

Preprocessing. The preprocessing module digitizes the band-pass filtered signals from every CMASP in the array at a sampling frequency of 411Hz and performs a digital full-wave rectification. Then, fully overlapping 0.3-second long processing windows are saved into memory for every transducer in the CMASP array (ie, 3 microphones, 3 accelerometers). This

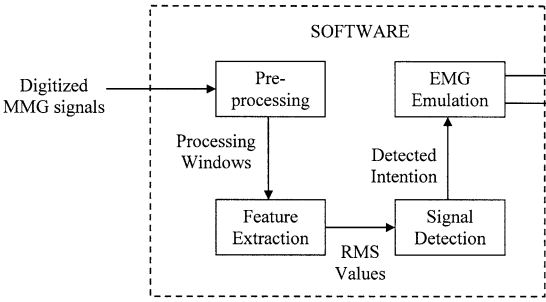

Fig 3. Schematic diagram of the software modules programmed in the electromyographic (EMG) emulator board. MMG-based estimates of electric muscle activity are obtained for the wrist flexors (EMG_f') and extensors (EMG_e'). These signals are then fed to the conventional prosthesis hand hardware. Abbreviation: RMS, root mean square.

window length represents the lowest frequency considered (ie, 3.3Hz) during MMG signal recording. Furthermore, these values coincide with the known useful bandwidth of MMG signals.[1-5]

Feature extraction. The feature extraction module estimates the root mean square (RMS) value of the processing window for each signal. These RMS values are then used in the signal detection module to determine the existence of a muscle contraction.

Signal detection. CMASPs facilitate the implementation of dynamic noise reduction algorithms for the elimination of external interference.[6] A simple example is a dynamic threshold algorithm proposed by Silva et al.[7] The dynamic threshold algorithm is based on the reported impedance mismatch between the microphone and accelerometer signals in a CMASP.[6] Because of its design and specific arrangement, the accelerometer in a CMASP is not capable of measuring skin vibrations. Consequently, increased accelerometer activity is assumed to be exclusively due to external interference (eg, limb movement during walking). Therefore, the accelerometer's RMS value is used as a dynamic detection threshold for the microphone signal.[7] In the MMG prosthesis reported here, we used this dynamic threshold algorithm to detect muscle contractions with the CMASP array. If a contraction was detected, the emulation module was engaged; otherwise no outputs were produced.

Electromyographic emulation. This module performs a weighted sum of the 3 microphones' RMS values. Silva et al[8] formulated this as a machine-learning regression problem and suggested a simple methodology for training and optimizing the corresponding classification and estimation algorithms. These suggestions constitute the electromyographic emulation algorithm implemented in the MMG-driven prosthesis. According to Silva,[8] the optimal muscle activity linear classifier/estimator can be found by performing a simple 2-step process. First, the user is asked to control a free-standing prosthesis hand using a conventional 2-site control strategy with Otto Bock electromyography surface electrodes[a] while wearing the MMG socket. Both electromyographic and MMG signals are recorded simultaneously. Then, during the algorithm training session, 60% of the recorded data is used to find, by regression, the vector $\theta=\{\theta_1, \theta_2, \theta_3\}$ that best recreates the activity measured by the electromyography (EMG) electrodes as described by the following equation:

$$EMG_f(t)-EMG_e(t)=\theta_1 MMG_1(t)+\theta_2 MMG_2(t)+\theta_3 MMG_3(t) \quad (1)$$

where $EMG_f(t)$ and $EMG_e(t)$ are the signals from the electromyography electrodes placed over the wrist flexors and extensors, respectively, θ_1, θ_2, and θ_3 are the regression coefficients, and $MMG_1(t)$, $MMG_2(t)$, and $MMG_3(t)$ are the estimated RMS values of simultaneously acquired signals from the 3 microphones in the MMG socket. Once the optimal vector θ has been found, the remaining 40% of the acquired MMG and electromyographic data can be used to evaluate the emulator's performance in an algorithm testing session.

After algorithm training, the right side of equation 1 (ie, the electromyographic emulation function) was programmed into the microcontroller on the electromyography emulator board. Therefore, an estimate $EMG_f'(t)-EMG_e'(t)$ of the original electromyographic signal was obtained for new MMG data previously unseen by the emulator.

Prosthesis Manufacturing

Three visits (30min each) are required for integrating and fitting the prosthesis to the user. During the first visit, circumference measurements are obtained along the length of the user's residual limb. Then, a negative mold of the limb is obtained with conventional plaster bandage. The mold is later used to create a plaster positive mold that will be used to fabricate the actual MMG socket. Once the positive mold is dry, it is modified according to the circumference measurements previously obtained and reduced by removing superficial material (2–4mm depending on limb density) to ensure a snug fitting MMG silicone socket. After reducing the positive, a 1-mm thick silicone layer is rolled over top of it. Three square openings of approximately 19×19mm are cut for the CMASPs around the mold, 120° apart from each other 1.5cm from the distal end. After the sensors and the shuttle pin base (or umbrella) are attached, Lycra cloth is added on top of the MMG socket to maintain a smooth surface and facilitate the donning process. Finally, vacuum is applied to remove air pockets and the MMG silicone socket is cured at 90°C for 12 hours.

During the second visit, a negative mold is obtained over the residual limb while the user wears the MMG socket. This mold is later used to obtain a plaster positive cast. The plaster cast is subsequently used to laminate the hard socket (including shuttle locking system) using 4 layers of nylglass[a] and 2 layers of perlon stockinette.[a] Two-part polyurethane is used to foam up the forearm, shaping the foam according to the user's sound arm. The lamination process is then repeated for the hard cover (prosthesis forearm) after which the wrist and battery box are glued. At this point, the prosthesis is ready for assembly. Finally, during the third visit, the electromyographic emulation algorithms are trained, tested and programmed into the electromyographic emulator board according to the previous descriptions.

Participants

Two adult subjects in the musculoskeletal program at the Bloorview MacMillan Children's Centre participated in a preliminary clinical study to evaluate the performance of the self-contained MMG-driven prosthesis. Subjects were recruited based on their level of amputation (both below-elbow amputees), availability, and ability to provide relevant feedback on their ease of control during a short on-site evaluation. Subject 1 (age, 26y) was a unilateral congenital amputee and subject 2 (age, 65y) was a unilateral traumatic amputee (his arm was severed as a result of an industrial accident). Both participants were long time electromyography-driven prosthesis users with 22 and 25 years of experience, respectively.

Evaluation

To facilitate prosthesis evaluation, we simultaneously recorded both electromyographic and MMG signals during an initial on-site trial while the subjects wore only the MMG soft socket. The subjects were asked to generate contractions to open and close their prosthesis hand in the same way that they operated their electromyography-driven prosthesis. The recorded electromyographic signals were then used to label the user intention of each muscular activity (ie, either "opening" or "closing" of the prosthesis hand).

As described earlier, MMG signals previously unseen by the emulator were used to quantify the prosthesis performance during the on-site trial. The rate of success was used as the performance (accuracy) measure. Rate of success (RS) was defined as the duration of "correct" prosthesis activity, t_C, over the total length of the trial, t_T, that is,

$$RS = t_C / t_T \tag{2}$$

The duration of "correct" prosthesis activity was defined as the summation of all instances when the muscle activity labels output by the emulation module agreed with the electromyographic labels. In other words, when the MMG-derived label of muscular activity coincided with the electromyographic label, the prosthesis was considered to be acting correctly. It is important to note that, although closely related, the rate of success is not a direct measure of true user intention. Rather, rate of success is a measure of agreement between MMG and electromyographic signals. In other words, we assume electromyography is 100% accurate and every disagreement between MMG and electromyographic signals is always an error of the MMG system. This, of course, is not always true; however, it simplifies the evaluation and accurately estimates the lower bound of the MMG-driven prosthesis performance.

After the quantitative assessment was completed, the subjects were allowed to wear the prosthesis on-site for a short period of time (20–30min) to gauge their ease of control and to make any relevant observations. Subject 1 completed an additional 2-week evaluation period by taking the prosthesis home to perform activities of daily living.

RESULTS

Quantitative Evaluation

Figure 4 shows a sample emulated electromyographic signal using MMG testing data (black trace) as compared with the measured electromyographic signal, $EMG_f(t) - EMG_e(t)$ (gray trace). Note the close agreement between the 2 traces. Positive MMGs always correspond to flexions and negative MMGs to extensions. Therefore, by considering the sign, 2 separate emulated electromyographic signals (ie, $EMG_f'[t]$ for flexions, $EMG_e'[t]$ for extensions) can be generated and sent to the prosthesis hand.

We obtained 88% and 71% in control rate of success for subjects 1 and 2, respectively. For subject 1, this represents an 18% enhancement compared with previous studies conducted with the same subject when only the MMG socket was tested (see Silva et al[8]).

Qualitative Evaluation

Both subjects were able to voluntarily open and close the prosthesis hand during the on-site evaluation period with no significant delays from intention to action (the calculated mean value of this delay was 120ms). However, the subjects initially reported a slight difference in the response of the MMG-driven prosthesis compared with their conventional electromyography-driven prosthesis. In particular, they described an increase in the difficulty of controling the prosthesis as the level of contraction increased. However, after several minutes of practice, the subjects were able to generate, with ease, consistent contractions exhibiting the proper amplitudes for the MMG-driven prosthesis.

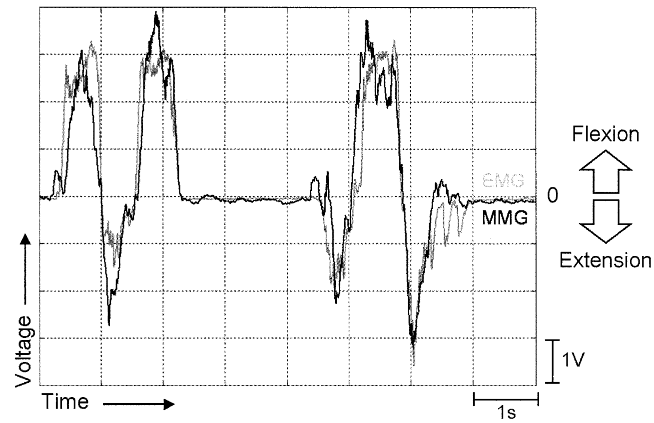

Fig 4. Comparison of measured (gray trace) and estimated (black trace) electric muscle activity signals. Measured electromyographic signals were obtained with conventional Otto Bock electromyography electrodes. Estimated signals were obtained from raw MMG signals acquired with the CMASPs within the MMG socket, and processed by the proposed emulation algorithms.

Subject 1 reported very reliable and repeatable control. After a 15-minute practice period, he succeeded in tying his shoelaces on repeated attempts, a task that requires extremely good control and one that poses a significant challenge for electromyography prosthesis users. This subject wore the prosthesis for an additional 2-week evaluation period. To our knowledge, this is the first report of a user wearing and operating a self-contained, MMG-driven, externally powered prosthesis outside of an institutional setting for an extended period of time.

Although subject 2 reported a fair degree of voluntary control, the prosthesis was not delivered for further evaluation because of a loss in control accuracy of the hand in certain limb positions. This diminished accuracy was most notable when weight was applied against the prosthesis suspension, such as when carrying loads or placing the arm above the head.

There were no reports of control issues relating to changes in sleeve positioning caused by sweat or voluntary adjustment, variations in sensor placement, and relative movement of the sleeve over the residual limb.

For subject 1, no control issues were identified during the on-site evaluation. However, a limitation similar to that reported by subject 2 was found during the 2-week evaluation period: the prosthesis became difficult to control when the user applied a considerable amount of force to the MMG socket (eg, trying to hold a knife while cutting meat). Moreover, the hand occasionally responded to very-low-frequency signal interference (<5Hz) such as the natural swinging of the arm during gait. However, elimination of typical environmental noise (eg, tapping on the prosthesis forearm, sudden movement, and loud sounds) was achievable.

DISCUSSION

The preliminary results obtained in this study reaffirm the potential of MMG signals for practical prosthesis control. Specifically, it can be inferred from the rate of success that information about muscle activity similar to that obtained by conventional electromyography sensors can be extracted from MMG signals. Furthermore, the use of multiple sensors has revealed an enormous potential for the detection and discrimination of intentional muscle activity, which may eventually facilitate the generation of multiple control outputs (ie, >2).

The reported limitations of the implemented prosthesis are most likely related to the nature of the mechanical coupling between the MMG socket, the CMASPs, and the residual limb. The microphone in a CMASP is extremely sensitive to changes in externally applied quasi-constant forces (ie, additionally applied weights). These changes are generally disregarded based on the information provided by the accelerometer in every CMASP. However, for low frequencies (<5Hz), the sensitivity of the accelerometers decreases due to the required low-pass filtering in the preprocessing stages, and signal discrimination between MMG signals and externally applied force becomes more challenging. This decreased low-frequency sensitivity likely accounts for (1) the occasional malfunction during gait, when low-frequency changes in the total inertia of the prosthesis resulting from natural swinging of the arm induce low-frequency variations in the total force applied on each CMASP; and (2) the lack of responsiveness when applying a constant force against the prosthesis (eg, cutting meat) and, in turn, on each CMASP.

However, the results show that, for subject 1, this heightened sensitivity was also advantageous. The integration of the MMG socket within the self-contained prosthesis actually enhanced the mechanical coupling. Prosthesis performance for this subject increased from 70% accuracy when wearing only the MMG socket[8] to 88% when wearing the fully integrated MMG prosthesis.

The reported difficulty encountered when applying significant force to the prosthesis was mainly due to the fact that, as muscle contractions become stronger, they produce resonant vibrations (ie, physiologic tremors) that cause generalized limb movement. These are detected by every accelerometer in the array as external interference. In response, the detection threshold is automatically raised, causing the user to experience increased difficulty in the generation of reliable control signals. However, the subjects were able to compensate for this reduced operating range of the MMG-based prosthesis. Furthermore, users may actually benefit from this reduced operating range because weaker contractions can be made to generate more forceful prosthesis outputs, reducing user fatigue in the long term. A future enhancement would be the addition of a tunable, nonlinear gain function (eg, sigmoid) to both the microphone and accelerometer. This tunable gain would facilitate control of sensor sensitivity and increase the operating range of the MMG-driven prosthesis.

Although the performance reported for the MMG-driven prosthesis is not sufficient yet for practical use, we are only beginning to tap into the vast potential of MMG-driven upper-limb prostheses. The signal algorithms presented in this report are the simplest ones conceivable. For instance, only 1 feature per signal is used to generate the corresponding outputs. A prudent selection of additional features or the deployment of nonlinear emulation algorithms will most likely enhance prosthesis performance. In particular, the use of frequency-related information is currently being investigated. Furthermore, a study of the mechanical interactions among the various components of the system has recently been completed.[9] This study has helped us obtain mathematical models and transfer functions that may more efficiently eliminate interference in the acquired signals and optimize mechanical coupling, thereby further maximizing CMASP sensitivity to MMG signals.

Additionally, the total compatibility of the miniaturized electromyographic emulator board with conventional electromyography-driven prosthesis devices facilitates the evaluation of new signal manipulation algorithms in ecologically relevant environments through quick and simple reprogramming. Finally, the emulation algorithm implemented facilitates user acclimatization by transferring most of the onus of learning to the machine.

CONCLUSIONS

Significant strides toward the clinical implementation of a self-contained MMG-driven externally powered prosthesis for below-elbow amputees have been achieved. By combining soft silicone sockets with CMASPs, we have provided a standardized and practical solution for the age-old problem of sensor attachment.

Despite the simplicity of the chosen implementation, it shows that an MMG-driven, externally powered prosthesis can be practically realized. This initial clinical study performed with the self-contained MMG-driven prosthesis has provided valuable insights into the potential and challenges of this technology in real-life settings. This information would have been impossible to obtain from tests with a free-standing prosthesis hand within a controlled institutional environment.

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