A SURFACE NEUROMUSCULAR ELECTRICAL STIMULATION DEVICE FOR UNIVERSAL CARTESIAN FORCE CONTROL IN HUMANS

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Abstract: *in recent years, neuromuscular Electrical Stimulation has found many applications both within the medical field and outside. While this technology has been widely recognized as a valid tool for rehabilitative and assistive applications, most solutions presented in the literature seem to focus on highly specific cases and facilitate very selective movements. In this article, we present a novel surface stimulation-based prototype which, coupled with an internally designed musculoskeletal model, allows to induce the output of generalized forces at the human end-effector in Cartesian coordinates. The control has been validated here through a 6-axis force-torque sensor coupled with a robotic manipulator. Thus, the measured forces at the user's end-effector were compared to the commanded forces. The results confirm that open-loop control of the output force is possible with an average correlation coefficient between commanded and measured force output direction greater than 0.7. This could eventually provide full, general purpose impedance control of the human neuromuscular system, which would allow to induce arbitrary movements in the peri-personal space.*

Keywords: *NMES, FES, surface electrodes, Cartesian control, wearable, validation*

Introduction

Neuromuscular Electrical Stimulation (NMES) is a technique, which is currently being applied both within and outside of the medical field [1] [2]. While many solutions involving NMES and, in particular, Functional Electrical Stimulation (FES) are present in the literature, most of them do not focus on general-purpose control, but rather on facilitating very specific force outputs and movements.
In this article, we present the MyoCeption, a wearable setup that allows, through NMES applied via adhesive electrodes, to control the force output of the user's end-effector in Cartesian space. The system also features the possibility of performing a twitch-based calibration procedure (in some respects similar to the one presented in [3] and [4]) in order to automatically adjust the musculoskeletal model to any given user.
This framework could be the base for a general-purpose impedance control of human limbs, which could be applied in both rehabilitation and assistance in activities of daily living for, e.g., patients affected by spinal cord injury, but also in VR or teleoperation scenarios.

Materials and Methods

The MyoCeption consists of a musculoskeletal model (see Fig. 1) used to compute the stimulation, and of a wearable setup (see Fig. 2) which can inject stimulation currents through surface electrodes. The system can provide amplitude-modulated, rectangular stimulation pulses with 16-bits resolution on up to 10 channels, with a pulse-width of 200μs, frequency ranging from 0.5Hz to 100Hz, and a maximum current amplitude of 70mA.

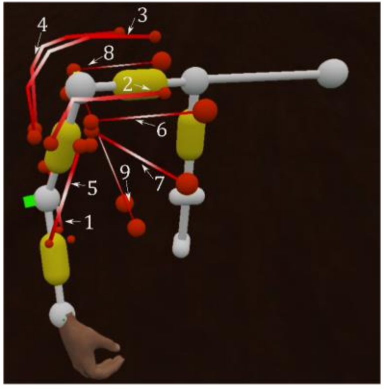

Figure 1: The MyoCeption's musculoskeletal model. The lines of action are marked with the stimulation channel

The MyoCeption Control Environment (MCE), which runs on a remote host, features the possibility to perform a calibration procedure where the twitch caused by a sharp stimulation signal is used to adjust the line of action corresponding to the stimulated muscle group within the musculoskeletal model, as well as the expected effect of the stimulation.
The calibration procedure relies on the assumption of coplanarity of the joint and the line of action of the stimulated muscle. The joint itself and the twitch vector define the plane on which the line of action should lie. The calibration procedure simply minimizes the distance of the origin and insertion point (that is to say, the most proximal and the most distal point) of the line of action from the aforementioned plane. In order to avoid the trivial solution where the origin and the insertion points coincide with the joint, these points of interest are expressed in cylindrical coordinates with the cylinder's axis coinciding with the skeletal link, and the coordinates over which this distance can be minimized are limited to the azimuth.
Furthermore, the magnitude of the twitch vector is used to infer the proportionality coefficient between the stimulation intensity and the induced torque at the joint.

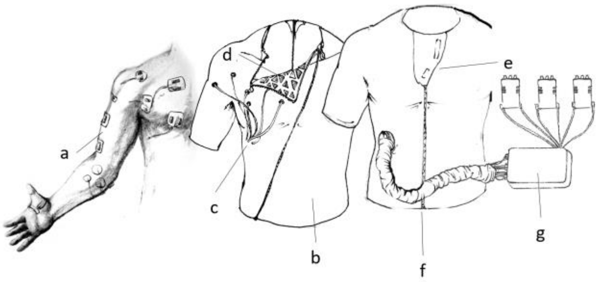

Figure 2: Main elements of the wearable stimulator. From left to right: adhesive electrodes applied to the user's skin, fitted with Velcro hooks on the outside (a). Inner compression jacket (b) featuring holes (c) to run the electrode cables through, fitted with Velcro loops on the inside (d). Outer jacket (e) grouping the cables in a single umbilical (f) connected to the control electronics (g), and providing further compression.

The musculoskeletal model computes the needed stimulation starting from a desired end-effector force vector $\vec{F}_{ee}$ by converting the desired force into torques at the joint level by means of the arm's Jacobian with respect to the *j*-th joint $J_{arm,j}$ according to the following equation

$$\vec{\tau}_j = J_{arm,j}^T \vec{F}_{ee} \qquad (1)$$

Each muscle group is associated with an expected torque output $\vec{\tau}_m$ at the joint level. This torque is simply the projection of the force f_m acting along the muscle's line of action through a cross product with the corresponding lever arms, according to

$$\vec{\tau}_m = f_m \frac{1}{N} \sum_{i=1}^{N} (\vec{p}_i - \vec{j}) \times \left(\frac{\vec{p}_i - \vec{p}_{i-1}}{||\vec{p}_i - \vec{p}_{i-1}||} \right) \qquad (2)$$

where $\vec{p}_i$ indicates the *i*-th of the *N+1* points lying on the line of action of the muscle group *m*, and $\vec{j}$ is the position of the joint.

The stimulation required to achieve the desired torque for an individual joint $\vec{\tau}_j$ is then computed by finding a combination of muscle torques of the form $\vec{\tau}_m$, which best approximates the desired joint torque vector. The type of approximation depends on the selected recruitment strategy. Here, the system used a nearest-neighbour recruitment, thus only stimulating the muscle group with the torque output $\vec{\tau}_m$ closest in direction to the desired joint torque $\vec{\tau}_j$. Other recruitment strategies could provide an optimal linear combination of stimulations on the available muscle groups. All solutions must respect the constraint that negative stimulation of the muscles is not possible. Admittance control or the employment of a proper pseudo-inverse of the muscular Jacobian can achieve this result. Search algorithms can also provide viable solutions.

The intensity of the amplitude-modulated stimulation depends on the expected magnitude of the induced torque. This expected effect has an initial value but can be overwritten through the calibration procedure.

The validation of the device was conducted as an internal test on 3 healthy participants (3 males, 34.3±12.7 years old, 1.76±0.09m, 77.3±6.67kg). All the participants signed an informed consent form and a data release form.

The participants were fitted with the MyoCeption device, as well as the BodyRig [5], a completely wearable IMU-based body tracker which requires no optical equipment. In this case, the BodyRig employed 5 IMUs. Through a simplified forward kinematic model of the human torso, clavicular-scapular joint, humerus, forearm and hand segments, the BodyRig was able to compute in real time the upper body configuration with high precision. The surface electrodes were applied in order to stimulate the *biceps brachii*, the *triceps brachii*, the *deltoid* superior, anterior, posterior, the clavicular and the sternocostal head of the *pectoralis major*, the *trapezoid scapular*, and the *latissimus dorsi*.

Participants sat in a predetermined position with their right arm coupled to a force-torque sensor attached to the DLR HUG system [6], as shown in Fig. 3.

The HUG allowed for easy realignment of the force-torque sensor if the arm had to be repositioned to better fit the size of the user, as the LWR arm made it possible to instantly know the absolute orientation of the force-torque sensor in space, and therefore to reconstruct the absolute direction of the measured forces and torques in the environment.

The participants were first asked to exert forces along 6 directions for 10 repetitions by following visual feedback. Thereafter, visual feedback was taken away. The uncalibrated MyoCeption was then fed desired force output vectors randomly in 6 different directions, with 2 different magnitudes, for 5 repetitions. The MyoCeption provided stimulation to induce a force output corresponding to the commanded forces.

Following this, the calibration procedure described above was performed for all stimulation channels, and the experiment with no visual feedback was then repeated with the calibrated MyoCeption.

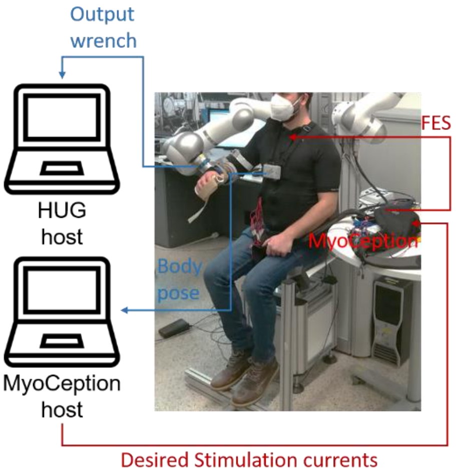

Figure 3: The experimental setup.

Results

In order to evaluate the performance of the feedforward force control of the MyoCeption, the Pearson correlation coefficients of the normalized commanded force vector with respect to the normalized measured force output were analysed. This was done to qualitatively characterize the ability of the system to induce force outputs in distinct directions as commanded.

Fig. 4 shows the correlation coefficients for the three participants during the conditions with visual feedback, force feedback with uncalibrated MCE, and force feedback with calibrated MCE, respectively, on three rows. Each of the leftmost three columns represent a subject, while the rightmost one represents the average across all subjects. For this figure, the Kabsch algorithm [7] was used in order to find the rotation around the vertical axis that best aligns the commanded and measured force output. This was done to compensate for possible errors in tracking the user's body pose.

Fig.5 shows the same correlation matrices, but with a three-axial correction through the Kabsch algorithm. In all cases, the asterisks indicate the level of significance: one asterisk indicates $p<0.05$, two asterisks indicate $p<0.01$, three asterisks indicate $p<0.001$.

Discussion

The results of this validation show that a simple feed-forward architecture through surface FES enables good directional control of the force output in Cartesian coordinates. When working in the condition shown in the first row in Fig. 4 and Fig. 5, the users were reacting to visual feedback, and were simply voluntarily pulling in the indicated direction. This condition is expected to have the clearest correlation between commanded and measured force direction and serves as a baseline comparison for the performance of the MyoCeption in inducing force output. The calibration procedure leads to a more consistent performance across different users (for this, compare for example the second and the third rows of matrices in Fig. 4) and to an overall better correlation, on average. Looking at the correlation coefficients after applying a 3-axial Kabsch correction as shown in Fig. 5, there is a noticeable improvement in the correlation coefficients between commanded and measured force outputs, when compared to the matrices shown in Fig. 4. In some cases, especially when no calibration was used, the needed Kabsch correction is not negligible, but this is still a good indication that the MyoCeption is able to provide force feedback in clearly distinct directions. It should therefore be theoretically possible to use a setup such as the one used in this experiment in order to calibrate the system before normal operation in order to compensate for the needed correction. Future work should focus on investigating this possibility, as well as on integrating appropriate sensors in order to close the loop of force control. Furthermore, the implementation of an impedance control loop should also be the subject of future research. The impedance control loop would be closed in position thanks to the body tracking system.

Conclusion

While the setup, in its current stage, implements a simple open loop control, in terms of force, the obtained results are promising. The MyoCeption is able to elicit force outputs in clearly distinct directions. The very next stage in experimentation would be the application of this force control within an impedance architecture in order to induce movements along a specific trajectory. In this case, monitoring the user's body pose would enable us to close the control loop in terms of end-effector pose.

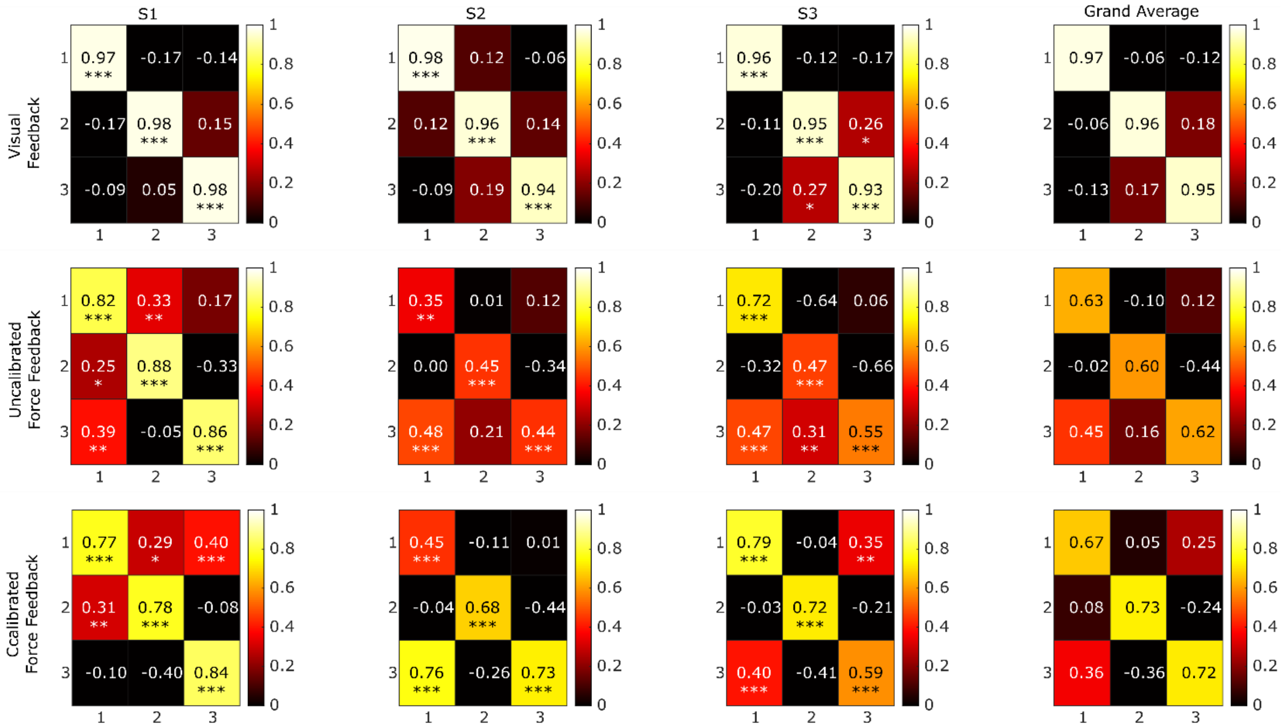

Figure 4: Pearson's correlation coefficients for the visual feedback condition with Kabsch correction around the vertical axis. The rows represent the components of the commanded force output, the columns those of the measured force output.

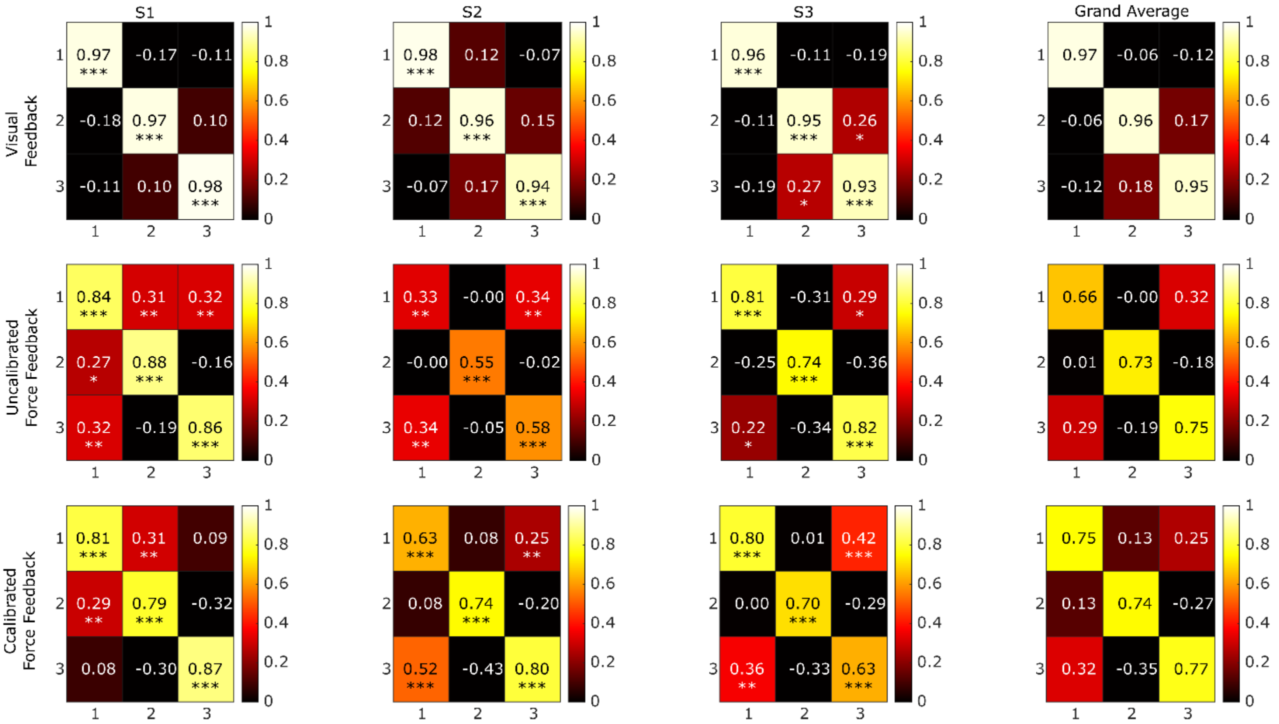

Figure 5: Pearson’s correlation coefficients for the visual feedback condition with 3D Kabsch correction. The rows represent the components of the commanded force output, the columns those of the measured force output.

Additionally, the integration of appropriate sensors in the system could be envisioned, which would allow to close the control loop in terms of Cartesian or joint forces as well. For example, a hybrid solution involving both a wearable device such as the MyoCeption and an exosuit, the tendons of which are fitted with load cells, would enable the system to monitor the exerted forces in real time.

Furthermore, additional analyses on the data acquired in this experiment are planned, with the goal of establishing whether a Machine Learning model would be able to compute the needed stimulation currents given a desired force output. If this is the case, a setup such as the one used in this experiment could be used in order to calibrate the MyoCeption to better fit any given user, prior to normal operation. This approach could improve the performance of the presented controller, even in the absence of sensors able to measure the exerted forces in real time during normal usage.

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