SYNERGISTIC CONTROL FOR AN ELBOW NEUROPROSTHESIS

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Abstract: *The proposed control method for the elbow neuroprostheses (NP) assumes that the most proximal non-paralyzed segment drives automatically the paralyzed segments of the upper-limb; in this case, the upper arm controls the forearm. The control uses production rules methodology: the rule-base comprises synergies formed automatically by inductive learning (IL). The IL automatically builds a decision tree based on many examples (kinematic data) and supervised learning. The examples selected for the learning have been recorded in able-bodied subjects while performing typical activities of daily living (eating with a fork, drinking from a can, writing, using a telephone receiver, and pouring from a container). The IL captures the synergy and expresses it in the form of IF-THEN rules: IF the shoulder flexion/extension velocity is given in the three consecutive times, THEN the elbow flexion/extension velocity is predicted. These rules are used for the coordination, being an interface between the volitional and actuator control levels. The actuator level is customized to the individual biomechanical and physiologic properties of the eventual user. This system can be part of both an implantable and/or a surface electrode NP. Initial tests with tetraplegic humans take place with the surface electrodes NP, using the Belgrade grasping system [13]. The results show that subjects can reach objects with required accuracy.*

Keywords: Control, Elbow joint, FES, Tetraplegia, Synergy

I. Introduction

For a tetraplegic subject restoring of reaching and grasping is among the most important objectives of rehabilitation [19]. Subjects with tetraplegia can benefit from a grasping system using Functional Electrical Stimulation (FES), but they must be able to control the shoulder and elbow movements.

Surface electrodes FES system for reaching has been integrated with the grasping system and clinically evaluated [8] leading to a therapeutic device for hand called NESS Handmaster [9]. This system used voice control and twelve channels of bipolar stimulation for the control of the elbow joint (two channels), hand and wrist.

The control of the elbow joint was integrated with the grasping system using percutaneous electrodes [7]; the recent version of the elbow extension neuroprosthesis (NP) relies on the implantable hardware known as the Freehand system [1, 2], and uses position triggered on-off control.

The Belgrade Grasping/reaching System (BGS) uses FES with surface electrodes for the control of grasping and elbow movement [13]. The arm movement have been decomposed into two processes: 1) movement of the hand in a plane formed by the upper and lower arm due to the shoulder (α) and elbow (β) flexion/extension (FE); and 2) the rotation of the plane of the arm around the shoulder joint (adduction/abduction, and humeral rotation).

II. Control principles for an elbow NP

The controller for elbow NP has a three level hierarchical structure (Fig. 1). The top level is voluntary; the user transmits commands to the coordination by using the adequate interface. The external control comprises two levels: the coordination uses the off-line prepared rules and distributes controls to the actuator level. The actuator control is model-based, sensory driven, and fully customized to the user [11, 16].

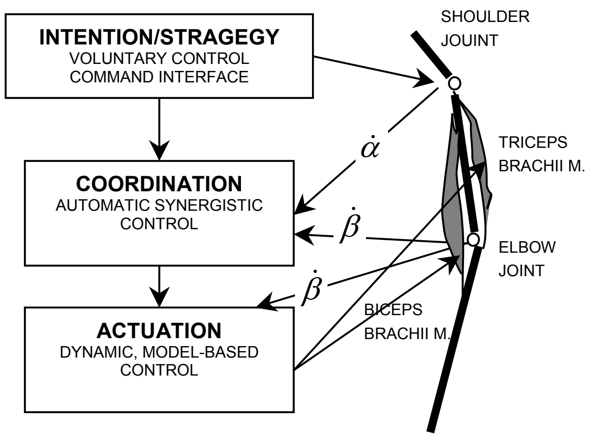

Fig. 1: Diagram of the organization of the hierarchical controller for an elbow neuroprosthesis. $\dot{\alpha}$ and $\dot{\beta}$ are the shoulder and elbow FE velocities respectively.

The coordination level of control should rely on the nature-like synergies, established during the growing and development of reaching, the musculo-skeletal substrate being built to support those, and the ease of relearning.

The synergistic control of the elbow NP was initially based on the scaling law between the FE at the shoulder and elbow joints determined in able-bodied

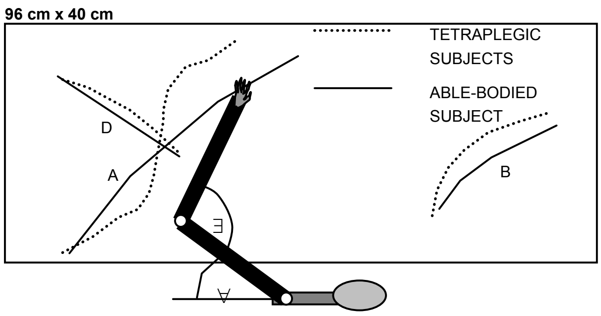

Fig. 2: Three recorded movements (A, B, and D) in able bodied subjects (full line) and the same three movements recorded when tetraplegic subjects used NP with scaling law based synergies (dashed line). α and β are the FE of the shoulder and elbow joints.

humans [13, 15, 17, 18]. Clinical trials with the BGS [12] showed that for some movements the scaling law is not adequate inducing not-acceptable errors in the positioning of the hand (Fig. 2). The desired paths (A, B, and D) recorded in able-bodied subjects are shown with full lines, and the hand trajectories recorded in tetraplegic subjects when using the elbow NP based on the scaling law are illustrated with dashed lines.

The study of positioning errors lead to the following findings: the scaling law is applicable for many domains of the workspace, yet there are movements, which do not fit this hypothesis. The scaling is equivalent to linearizing of synergies between the FE velocities at the joints. The study also showed that the workspace could be divided into zones, each zone being characterized by a single synergy.

This is equivalent to dividing the movements in classes; each class fitting a rather complex, yet characteristic synergy between FE at the shoulder and elbow joints [5]. The right panels (Fig. 3) show the movement from the initial position to the target (A, B, and D). Experimental findings suggest the following synergies: 1) a group where FE velocity vs. time at one joint has two peaks, while the other profile vs. time has a single peak (Fig. 3, right, top). τ_1, τ_2 are the intervals between the peaks; 2) a group where both FE velocities have one peak (Fig. 3, right, middle), and they are shifted with respect to each other for $\tau \neq 0$, the delay; and 3) a group where the bell-shaped velocity profiles vs. time have simultaneous peaks ($\tau = 0$) as shown at the bottom panels of Fig. 3, right. The presented joint velocities correspond to movement (A, B, and D) of the left arm in the horizontal plane below the shoulder, as indicated with full lines in Fig. 2. The left panels in Fig. 3 show the phase plots, having shapes of tilted figures eight; each comprising two leaf-like patterns. These graphs are obtained by plotting FE velocities of the elbow vs. shoulder joint. Left panels are the phase plots of the movement from the initial position to the target (the leaf patterns pointed with arrows) and return to the initial position (not shown in right panels).

III. Determination of synergies by inductive learning

In order to widen the class of controlled movement it was essential to capture highly nonlinear synergies. For that, a machine learning technique was applied. We selected the Inductive Learning (IL) method among several available machine learning techniques. The IL was already employed for control of walking assistive systems [3, 6], and used for determining the synergies of the elbow and shoulder joints for reaching tasks [4].

The IL used a training set that has been prepared from kinematics recorded in able-bodied humans. Able-bodied humans have been instrumented with joint motion sensors, and they performed the activities of daily living. The FE angles at the shoulder and elbow joints have been recorded and processed and the angular velocities calculated. The input data for learning are sets of FE velocities at the shoulder joint in three consecutive times $[t_i, t_i - 10\,ms, t_i - 20\,ms]$, while the output data

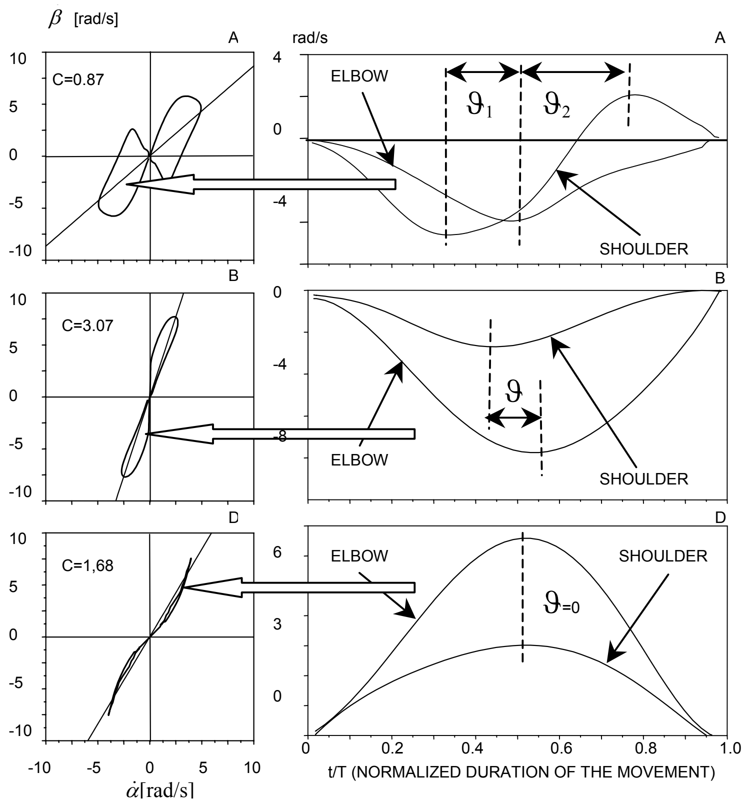

Fig. 3: The velocity space plots (left) and flexion/extension velocities vs. time (right) at the elbow and shoulder joints for three different movements A, B, and D shown in Fig. 2. Phase plots include the solid line (scaling law) described with the coefficient C. τ is the interval (delay) between the peaks at angular velocities.

were FE velocities at the elbow joint at time t_i. Fig. 4 shows the calculated FE velocities at the shoulder joint from recorded data, and two superimposed series: calculated FE velocities at the elbow joint from recordings and estimated FE velocities by IL [14]. The matching between the original and estimated FE velocities is very good. The difference between the desired and estimated trajectory, the error, has been extracted at the bottom of both panels. The mean error, that is the mean difference between the recorded and IL estimated FE velocity, is 0.12 radian/s, with the maximum error of 0.28 radian/s. The errors are generated because of the adopted multiple threshold learning strategy [10].

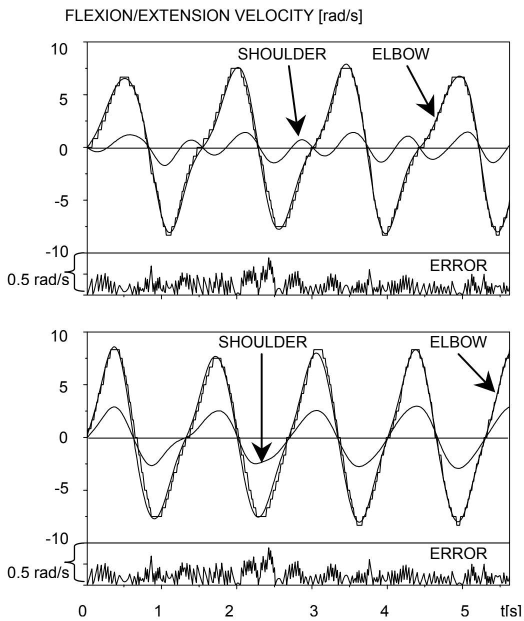

Fig. 4: Results of IL for determining the synergies between flexion/extension for two different movements. The desired and IL estimated FE velocities vs. time are superimposed, and their difference (error) shown at the bottom.

IV. Testing of the novel control

The novel control method has been tested using the modified BGS as an elbow NP (Fig. 5). The electronic stimulator delivers electrical charge to biceps and triceps brachii muscles via surface (or implanted) electrodes.

The interface for selecting the zone was a push button device. The subject was trained to push the adequate button depending on the zone that he wanted to move the hand with the contralateral hand. Each button was associated with the off-line synergy determined by means of IL. The stimulation profiles delivered to motorneurons via electrodes have been determined based on off-line simulation of elbow joint FE. The simulation has been customized to the eventual user [11]. The sensors used are the flexion/extension goniometers at the elbow and shoulder joints.

Once the command signal is sent to the coordination, the command signal to the programmable electronic stimulator is the desired angular velocity at the shoulder joint (Fig. 1), while the elbow angular velocity is used for feedback.

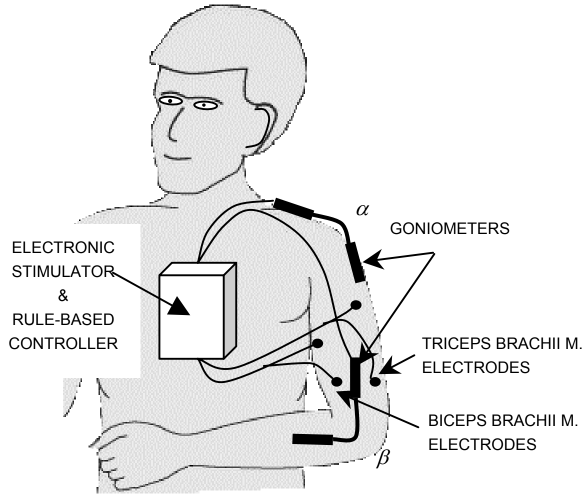

Fig. 5: Sketch of the NP for control of elbow movement in humans with disability of upper arm, with preserved shoulder functions and paralyzed elbow and hand.

This signal is updated in real-time based on IF-THEN rules determined off-line using the IL algorithm. The IF part of rule comprises the three consecutive values of the FE at the shoulder joint $[\dot{\alpha}(t_i), \dot{\alpha}(t_i - T), \dot{\alpha}(t_i - 2T)]$, $T = 10ms$, and THEN part of the rule is the shoulder FE velocity $[\dot{\beta}(t_i)]$. The sketch shows the Penny and Giles goniometers and surface electrodes used in our study, but other transducers including velocity encoders and implanted electrodes can improve the performance. The real-time control includes the sampled velocity feedback to correct the stimulation parameters [13].

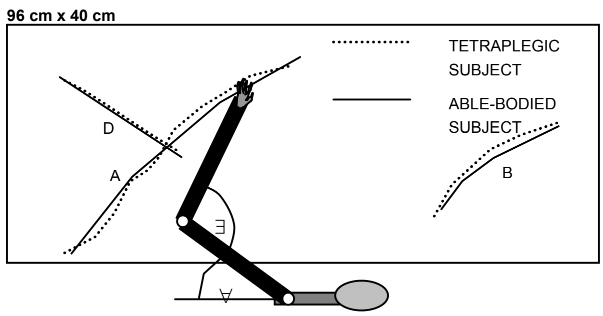

Fig. 6: Diagram showing the three movement (A, B, and D) recorded in an able-bodied subject (full line) and the recordings from the C4/C5 tetraplegic subject when using NP control based on IL determined synergies (dashed line).

Fig. 6 shows an example of using the described controller in a single subject (tetraplegic, C4/C5) having preserved shoulder functions, innervated (paralyzed, grade 0) triceps brachii muscle and grade 1 biceps brachii muscle. The same three movements shown in Fig. 2 are selected to demonstrate the effects of the novel synergy used for control. Recorded movements (full lines) from the able-bodied subject and IL synergy controlled movement from a tetraplegic subject (dashed lines) show much better matching compared with the scaling law vs. desired paths (Fig. 2). The results are average from 30 trials, after the tetraplegic subject was given the opportunity to practice in the laboratory setting for five days, 30 minutes each day. The repeated trials during the recording showed large variability in trajectories, but not in reaching the final position. The differences in hand paths shown in Fig. 6 are acceptable, since tetraplegic subjects can compensate for small errors by moving the trunk for a few centimeters.

V. Conclusion

This report describes a method to control forearm movement with NP. The control scheme is based on a hierarchical controller using a synergistic strategy at the coordination level and sampled feedback at the actuator level. Both levels use look-up tables as the knowledge base within a rule-based controller. The coordination look-up table is universal and determined in able-bodied subjects by IL. The other look-up table (actuation) is model-based and fully customized to the user.

This paper concentrates on the IL based method of determining rules for the coordination level of control, specifically addressing the differences of synergies between various regions of the workspace.

The lower, actuator level uses sampled feedback to correct for variation in dynamic and biologic conditions as well as unpredicted perturbations. Without these corrections it is not realistic to expect that the synergistic control would work correctly. This combined approach provides a tool to individually tailor the parameters for stimulation.

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