34th Annual International Conference of the IEEE EMBS
San Diego, California USA, 28 August - 1 September, 2012

Control of Robot Assistant for Rehabilitation of Upper Extremities

Miloš D. Kostić, Mirjana B. Popović, Dejan B. Popović, *Member, IEEE*

***Abstract*— The assisted movement in humans with paresis of upper extremities is becoming popular for neurorehabilitation. We propose a novel method for trajectory selection and assistance control. This paper presents simulation of a planar two degrees of freedom robot that assists horizontal movement of the hand. The control assumes that during the exercise the hand needs to follow healthy alike trajectories. The robot is assumed to provide minimal assistance and operate as a teacher of the movement.**

I. Introduction

The assisted exercise of task-related movements has become increasingly popular in the neurorehabilitation of patients diagnosed with upper extremities paresis [1]. However, this type of therapy requires intensive one-to-one sessions which are physically demanding and time consuming for therapists. One possible solution to facilitate the therapy would be the use of haptic robots as rehabilitation assistants [2]. This type of therapy has been receiving a lot of attention recently, because the rehabilitation robots improve the patients' motor abilities and boost the recovery process by providing them access to crucial kinetic and tactile information [3,4] with minimal supervision.

Rehabilitation which includes task related movements is one of the areas where such robots might have the largest potential as rehabilitation tools. One type of robots used in rehabilitation is a planar manipulandum, which can assist the movement by guiding the patient's hand in one plane (e.g. Braccio Di Ferro [5], InMotion Arm Robot [6]).

A widely accepted method of controlling assistance robots is impedance control [12]. The robot acts as the impedance, and the human as the source meaning that the robot produces forces according to the measured human motion. In this study we propose a method for robot assistance control, based on this principle, which is capable of accommodating the variability typical for healthy arm movements. According to this method, the robot does not impose one "ideal" movement strategy, but allows the subject to use its preferred strategy, and gradually improve it. This type of assistance is referred to as "gentle". The control assumes that during the exercise the hand needs to follow the healthy alike trajectory. Hence, the robot should provide assistance in order to guide the hand along the preferred trajectory.

When this type of movement is performed by healthy individuals, free to move their hand in different directions, there is a close to linear relation between the torque in elbow muscle and the torque in shoulder muscle [7,8]. The other characteristic is the movement kinematics with distinctive bell-shaped velocity profiles [9]. If assistance is provided so that the movement has healthy alike trajectory and dynamics, the natural elbow-shoulder synergies may be formed, as these develop in humans well before any other adult-like motor strategy in reaching [10]. In one of our previous studies [11], we introduced the method of selecting the trajectory through observing the kinetics of movements performed by healthy individuals.

II. Trajectory Selection and Mathematical Model of the Robot

The mathematical model of the robot that we use was developed for the model shown in Fig. 10 (right). Each of the segments in robot arm is characterized with its mass (m), length (l) and moment of inertia (J_C). The distance of center of the mass (C) from the pivot joint is annotated with d. M_S and M_E are joint torques, produced by motors. We suggest the force feedback; therefore, force sensors are included in interface between the hand and the robot Fig. 1. The angles of robot arms are α_1 and α_2, and they are assumed to be monitored by encoders mounted in the robot joints. The robot is interacting with the human at the end point. The interface force is annotated with F_W. The velocity and acceleration of the end point are vectors v_W and a_W.

We assume that the robot arms are rigid bodies, and that there is no friction in the robot joints. For purposes of simulation both arms are modeled as 42cm long rods with 1kg mass. The other assumption is that the movement of the robot arms is in the horizontal plane.

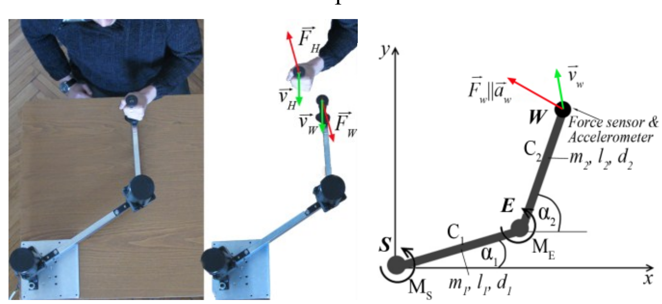

Figure 1: Left: Patient with the robot. Center: Decomposed system. Right: Mathematical model of the robot.

By using basic theorems of mechanics (the law of momentum and the law of angular momentum) we obtained the mathematical model of the robot arm given by (1) and

*This work was supported by the FP7 EC strep project HUMOUR, No. 231724, the Research project No. 175016, Ministry for Education and Science, Republic of Serbia, Belgrade and Swiss National Foundation, Berne (Project InRES IZ73Z0_128134)

Miloš D Kostić, PhD student, Mirjana B Popović and Dejan B Popović are with the School of Electrical Engineering, University of Belgrade, Bulevar kralja Aleksandra 73, 11120 Belgrade, Serbia (phone: 381-11-3218-348; e mail: thekostic@etf.rs).

Mirjana B Popović and Dejan B Popović are also with the Department of. Health Science & Technology, Aalborg University, Denmark.

(2). Where acceleration of points C_1 and C_2 ($\ddot{x}_{C1}, \ddot{y}_{C1}, \ddot{x}_{C2}, \ddot{y}_{C2}$) is determined by measuring changes in angles α_1 and α_2.

$$M_E = J_{C2}\ddot{\alpha}_2 - ((F_{Wx} + m_2\ddot{x}_{C2})d_2 \sin\alpha_2 + (F_{Wy} + m_2\ddot{y}_{C2})d_2 \cos\alpha_2 + + F_{Wx}(d_2 - L_2)\sin\alpha_2 + F_{Wy}(d_2 - L_2)\cos\alpha_2) \quad (1)$$

$$M_S = J_{C1}\ddot{\alpha}_1 + J_{C2}\ddot{\alpha}_2 - ((F_{Wx} + m_2\ddot{x}_{C2} + m_1\ddot{x}_{C1})d_1 \sin\alpha_1 + (F_{Wy} + m_2\ddot{y}_{C2} + m_1\ddot{y}_{C1})\cdot \cdot d_1 \cos\alpha_1 + (F_{Wx} + m_2\ddot{x}_{C2})((d_1 - L_1)\sin\alpha_1 + d_2 \sin\alpha_2) + (F_{Wy} + m_2\ddot{y}_{C2})\cdot \cdot((d_1 - L_1)\cos\alpha_1 - d_2\cos\alpha_2) + F_{Wx}(d_2 - L_2)\sin\alpha_2 - F_{Wy}(d_2 - L_2)\cos\alpha) \quad (2)$$

In presented simulation the workspace is divided into six zones by using the 2 distance attributes (short (S) and long (L)), and 3 directions (ipsilateral (I), frontal (F), and contralateral (C)) as shown in Fig. 2 (top panel). Each of the movements is represented by the starting point, target point and two lookup tables for acceleration components, termed Probability Tubes (PT), Fig. 2 (bottom panel). One acceleration component is always directed towards the goal point (a_D), while the other one is perpendicular, in positive direction (a_N).

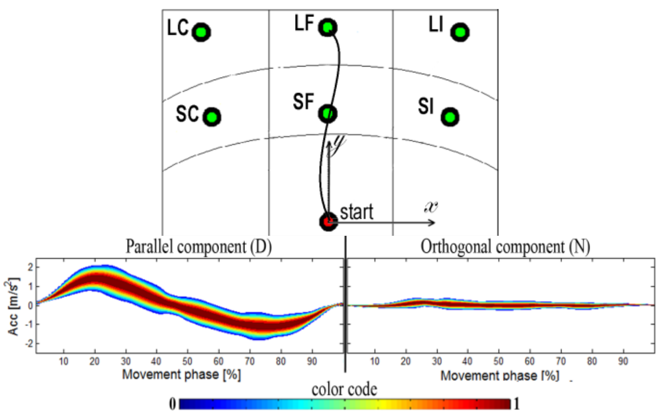

Figure 2: Top panel: The sketch of the division of the workspace into six zones. The acronyms are described in the text. Bottom panel: the Probability Tubes for the acceleration during the movement towards LF. Acceleration is decomposed to component in the direction of target point, and the orthogonal component.

The PT is a stochastic model for arm trajectory formation which determines the probability of hand acceleration at the specified phase of the movement. It is formed in experiments where healthy subjects performed reaching movements, while using the device in passive mod, as described in [11,13]. It was designed to capture movement dynamics while preserving the natural variability of healthy human movement, which is reflected in the probability profile. As function P= *PT(a(t), i)* it determines the probability that the movement is being performed correctly if at phase *i* the acceleration is *a(t).* Process of forming the PT and determining movement phases is presented in detail in our previous work [11].

III. Control Algorithm

The main feature of this method is the ability to provide assistance without imposing "golden standards". At every point of the movement the robot provides assistance in order to follow the reference values of acceleration components which are determined from the PTs. By using PT as the reference, subject is assisted to perform the movement in the correct manner; yet the system allows subject to deviate from "ideal" to its preferred kinematics. The control algorithm works in this mode as long as the deviation of acceleration is beneath a predetermined threshold. When the deviation is higher than the threshold, it means that the movement kinematics are outside the probability tube, and the difference is not a consequence of variability, but of wrong (if any) motor strategy. As a consequence of no preferred acceleration within such strategy, the assistive force is applied so that the "ideal" acceleration is achieved.

This type of control was realized through the following algorithm: if at the moment *t* the hand is at the phase *i* of the movement and has acceleration *a(t)* than the reference acceleration is

$$a_r(t) = PT^{-1}(PT(a(t), i) + \frac{1 - PT(a(t), i)}{k}, i), \quad k > 1; \quad (3)$$

where *a(t)* is current acceleration and *i* current phase. The factor *k* determines the level of allowed variability. Being that $PT^{-1}(x,i)$ has two solutions, $a_r(t)$ is the one closer to *a(t)*[11]. In cases when external forces are such that the endpoint acceleration is outside the certain range and guidance can no longer be considered as gentle, gradual increase of PT value is pointless. Then a_r is determined as $a_r(t)$= PT^{-1}(max).

In order to achieve the reference acceleration a_r the robot exerts assistive force on the hand. This force is a consequence of motor torques produced according to (1) and (2). In these equations a_r does not appear directly, but is represented through angular accelerations. The relation between a_r and angular accelerations is given in equation:

$$\begin{bmatrix} \ddot{\alpha}_1 \\ \ddot{\alpha}_2 \end{bmatrix} = \begin{bmatrix} L_1 \sin\alpha_1 & L_1 \cos\alpha_1 \\ L_2 \sin\alpha_2 & L_2 \cos\alpha_2 \end{bmatrix}^{-1} \times \times \begin{bmatrix} -a_{rD}\cos\gamma + v_D\dot{\gamma}\sin\gamma + a_{rN}\sin\gamma + v_N\dot{\gamma}\cos\gamma - L_1\dot{\alpha}_1^2\cos\alpha_1 - L_2\dot{\alpha}_2^2\cos\alpha_2 \\ -a_{rD}\sin\gamma - v_D\dot{\gamma}\cos\gamma - a_{rN}\cos\gamma + v_N\dot{\gamma}\sin\gamma - -L_1\dot{\alpha}_1^2\sin\alpha_1 - L_2\dot{\alpha}_2^2\sin\alpha_2 \end{bmatrix} \quad (4)$$

where γ is the current bearing of the target point.

Variability of the acceleration can lead to a case in which velocity v_W is different than zero at the end of the movement. In order to avoid this, last 5% of the movement is not controlled by the PT, but deterministically, leading the hand to a standstill.

IV. Simulation

Described relations and rules were implemented in a MATLAB simulation where we calculated moments M_S and M_E for different scenarios. Subject's activity was represented by simulated force signals. Simulation was performed for reaching movements, shown in Fig. 2 (top panel). Four different cases were considered, each depicting a characteristic scenario which can be expected in clinical use.

In the first case, we simulated performance of a healthy subject. Here the movement is performed correctly along the trajectory, and no assistance is needed; thus, the interface force is simulated as zero. In this case subject should not "feel" the inertia, so robot actuation has to compensate for it.

One common case is a spastic subject performing movements with the system. There, subject is unable to follow the required dynamics pattern, and opposes the movement with an uncontrolled spastic force. This was simulated with strong force ($\vec{F}_W$) acting against the movement. The amplitude of $\vec{F}_W$ is such that it hinders gentle assistance during the entire movement.

The case which is in the focus of interest of this method is when the system is providing assistance to a patient who is trying to follow the correct movement kinematic. Due to unperfected motor control the arm and manipulandum movements are not ideally synchronized, resulting in appearance of the force. In general case this force changes direction and amplitude, as the subject browses the acceleration space within the PT. This was simulated in the third scenario, where force is a signal with pseudo random amplitude and direction, with sub tremor frequency, up to 4Hz. The force was simulated with amplitude small enough to apply gentle assistance.

The fourth considered scenario simulates a situation in which the patient is inactive. In this case the patient is letting the robot provide the entire movement, while he/she is mostly "riding along". This was simulated with small amplitude of force which opposes the movement (inertia of the persons arm). The simulated force was small in a sense that gentle control could be applied.

V. Results

Results of all four simulated scenarios are shown on the representative example, movement LF, which is marked in the Fig. 2 (top panel). Results of other 20 simulations are not presented in this paper. Each simulation result is presented by the performed trajectory and velocity profiles of the hand, and motor moments applied during the movement.

The trajectory is presented in a subject centered coordinate system, with the origin positioned in the midpoint of subject's sternum. Beginning and the end of the movement are marked with magenta asterisk and green circle, respectively. Velocity is shown with respect to time. Moments of each motor are presented individually, in respect to time.

VI. Discussion

A system which moves the hand, like a planar manipulandum, introduces additional dynamics during the movement. This effectively changes the motor task, manifesting as deviation of preferred trajectory, velocity profile or changes in muscle activation pattern. This can deteriorate quality of motor learning, since the motor task presented to the subject is not the same as the one without the manipulandum [13].

In the proposed method compensation of the system inertia is integrated in the control algorithm. In the simulation shown in Fig. 3A there is no external force, meaning that motors' moments actively compensate only inertial forces, which appear in radial movement with acceleration profile following reference, as the one given in Fig. 2 (bottom panel).

In absence of external disturbance motor moments are in approximately linear relation, as can be noticed in the right panel of Fig. 3A. This principle is biomimetic, since it is present in healthy arm movements, as described by [7,8], and can be explained as natural organizing principle which, at the level of muscle torques, decreases the degrees of freedom. Here we can also notice that when the phase of deceleration begins, motors start acting in the opposite direction. The activity in deceleration phase of the movement is higher, since it is shorter than acceleration phase. Highest activity of inertial force suppression near the end of deceleration phase is in accordance with our previous work [13] where we found that inertial forces create greatest disturbance in this phase (marked with green * in Fig. 3A).

In case when subject does not have the required motor skill to perform the task, his hand movement will not be synchronized with the manipulandum. This may result in three characteristic scenarios depicted in Fig. 3 B-D.

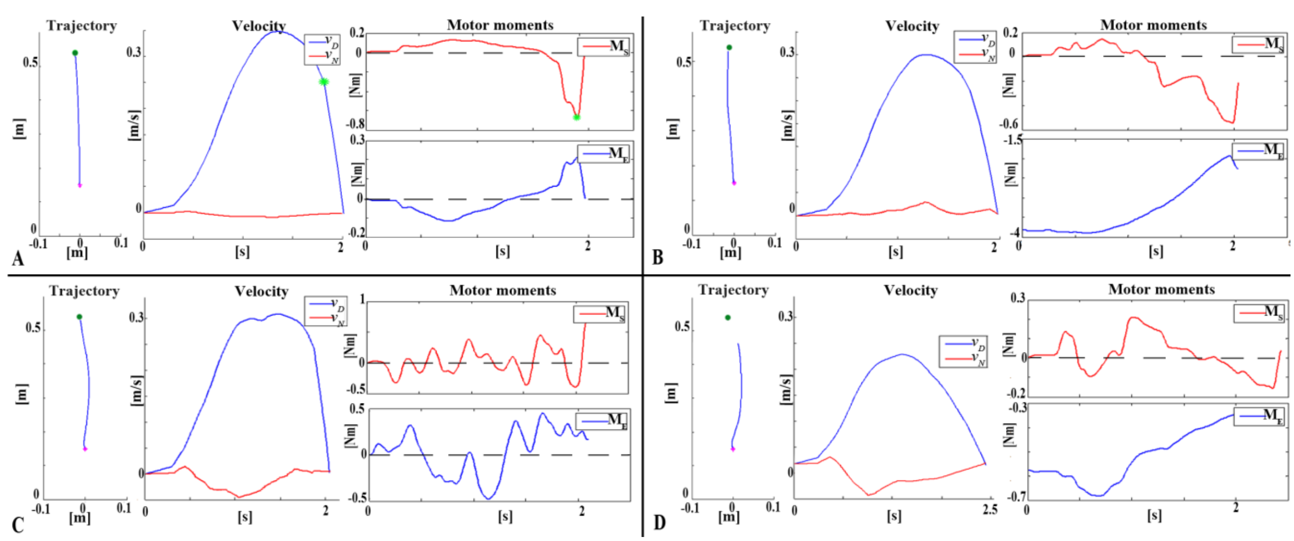

Figure 3. Trajectory velocity and motor moments obtained during simulation of: A) the movement without external force; B) the movement with high external force; C) the movement with variable external force; D) the movement with small opposing external force.

In the case of simulated spastic patient, shown in Fig. 3B, the disturbance is too large for gentle assistance, and the "ideal" movement is imposed. This was performed successfully as both trajectory and velocity v_D are almost the same as in the Fig 3A. The influence of $\vec{F}_W$ in direction perpendicular to the movement is very small, so the control allowed some variability in that direction, leading to different v_N profile, as well as the slight deviation of the trajectory to the left. The moment M_E is significantly different than the one in Fig. 3A, as the bulk of the force $\vec{F}_W$ had to be compensated by it. As moment arm of $\vec{F}_W$ in respect to point S is close to zero, motor M_S mostly compensates the inertial forces.

In the simulation of a patient who is trying to learn the movement, the control algorithm applied gentle assistance, helping the subject to achieve the preferred movement dynamics. Because of the assistance principle proposed here, motor moments in this simulation do not cancel out the effects of the external force, but rather reshape it to achieve the preferred movement dynamics. Motor moments graph (Fig. 3C) shows continuous adaptation of the system to subject's movement. Amplitudes of these moments are relatively low, suggesting compliance of the system.

There are noticeable differences between the trajectory of the movement performed in preferred manner, and the "golden standard" shown in two previous figures; yet it can still be considered as correctly executed movement. Trajectory in this simulation shows slight curvature. This is a consequence of movement component perpendicular to shortest distance direction, which appears even with healthy subjects, though they generally have straight line trajectories in radial movements [9]. Deviation from "ideal" case can also be noticed in the velocity graph. Velocity in the shortest distance direction, v_d, has a bell shaped profile, different than the one in the previous cases, but still within the domain of normal, healthy movement. Graph of velocity v_n reflects the trajectory deviation.

Allowing variance in acceleration of the hand in the way presented in this paper has one potential drawback. If the hand is moving too slow during the entire movement, but with the acceleration within the threshold values for gentle assistance, the robot will tolerate the lack of acceleration as deviation. In this case the hand does not generate high enough velocity in the acceleration phase, and it loses it too fast in the deceleration phase. This leads to a shorter movement, and failure to reach the goal point, as presented in Fig. 3D. The level of motor activation in Fig. 3D suggests that the control algorithm recognized the measured acceleration as tolerable variation and did not impose "ideal" acceleration. This is obviously a method specific error which reveals a potential flaw of the algorithm.

Subject's active participation in training is crucial for motor learning [3], and it should be insisted on. On the other hand failure to reach the goal point can happen only if the subject is passive. Therefore if the shortened movement is used as indication of insufficient subject activity these flaw becomes advantage of this method, over noncompliant methods of assistance which are prone to allow "slacking".

VII. Conclusion

Allowing variability in movements during training is essential for learning a new motor skill. Control algorithm based on the Probability Tube, presented in this paper, enables such variability, while providing assistance for healthy like movements. This is a general algorithm which applies on any type of planar point-to-point movement for which the PT is provided. The PT was designed with the idea that additional movements can be added by therapists in clinical conditions, making this a practical control algorithm for robot assisted rehabilitation of upper extremities.

References

[1] C. J. Winstein, S. L. Wolf "Task-oriented training to promote upper extremity recovery," Stroke Recovery & Rehabilitation.New York: Demos Medical Publishing 2009; 267-290.

[2] J. V. G. Robertson, A. Roby-Brami "Augmented feedback, virtual reality and robotics for designing new rehabilitation methods," In: Rethinking physical and rehabilitation medicine Springer: Paris, 2010; 223-245.

[3] G. Kwakkel, B. J. Kollen, H. I. Krebs "Effects of robot-assisted therapy on upper limb recovery after stroke: a systematic review," Neurorehabilitation and Neural Repair 2008; 22: 111-121.

[4] G. B. Prange, M. J. A. Jannink, C. G. M. Groothuis-Oudshoorn, H. J. Hermens, M. J. IJzerman "Systematic review of the effect of robot-aided therapy on recovery of the hemiparetic arm after stroke," Journal of rehabilitation research and development 2006; 43: 171.

[5] M. Casadio, V. Sanguineti, P. G. Morasso, V Arrichiello "Braccio di Ferro: a new haptic workstation for neuromotor rehabilitation," Technology and Health Care 2006; 14: 123-142.

[6] InMotion Arm Robot, Interactive Motion Technologies, Massachusetts, USA. http://interactive-motion.com/, 12.02.2012

[7] G. L. Gottlieb, Q. Song, D. A. Hong, G. L. Almeida, D. Corcos "Coordinating movement at two joints: a principle of linear covariance," Journal of neurophysiology 1996; 75: 1760-1764.

[8] M. B. Popovic, D. B. Popovic, R. Tomovic "Control of arm movement: reaching synergies for neuroprosthesis with life-like control," Journal of Automatic Control 2002; 12: 9-15.

[9] Flash T, Hogan N. The coordination of arm movements: an experimentally confirmed mathematical model. The journal of Neuroscience 1985; 5: 1688-1703.

[10] F. T. J. M. Zaal, K. Daigle, G. L. Gottlieb, E. Thelen "An unlearned principle for controlling natural movements," Journal of neurophysiology 1999; 82: 255.

[11] M. D. Kostic and D. B. Popovic, "Action representation of point to point movements: Classification with probability tube," IEEE 19th Telecommunications Forum TELFOR, 22/11- 24/11/2011, Belgrade, Serbia, p: 43-46, DOI: 10.1109/TELFOR.2011.6143888

[12] N. Hogan "Impedance control: An approach to manipulation, Parts I, II and III," Journal of Dynamic Systems, Measurement, and Control 1985; 107: 1-23.

[13] M. D. Kostic, D. B. Popovic and M. B. Popovic, "Influence of planar manipulandum to the hand trajectory during point to point movement," 12th International Conference on Rehabilitation Robotics, 29/06 - 02/07/2011, Zurich, Switzerland, p: 1-4, ISBN: 978-1-4244-9861-1