# Controlling upper-limb prostheses with body compensations

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***Abstract*— With their advanced mechatronics, myoelectric upper-limb prostheses now have many motion possibilities. Yet, the latter are not fully employed because of the inconvenient control and prostheses wearers often use their device as a rigid tool while achieving hand positioning and orientation with compensatory movements. In this paper, we propose to take advantage of this natural human behaviour to control prosthesis motions: the user is in charge of the end-effector while the devices role is to correct the human posture when necessary. We here apply this concept to control a prosthetic wrist pronosupination. A Rolyan clothespin test performed by two transradial amputees shows that the proposed control is as efficient as myoelectric control while requiring no learning.**

## I. Introduction

With the latest advances of mechatronics, the motion possibilities of upper-limb (UL) prostheses have increased a lot [1], [2] but they are not fully exploited because the control of these devices remains highly challenging.

The most widespread approach directly connects electromyographic signals generated by the user to the movement of the prosthesis. It includes conventional myoelectric control (on/off or proportional), integrated into most of the commercially available devices, and pattern-recognition based myoelectric control. Performant enough when controlling one degree of freedom (DOF), these two schemes become limited with an increasing number of DOF [3]. The muscular fatigue and the mental burden induced by theses control approaches often lead prostheses users to employ their device as a rigid tool and move their end-effector with compensatory movements (see [4] e.g.). Body compensations are indeed efficient to achieve many tasks, but they are to be avoided since they cause musculoskeletal disorders [5].

In this paper, we propose to take advantage of this natural reaction of prostheses users to control the device: the end-effector task (placing and orientating the hand) will be achieved by the human subject while prosthesis motions will correct the human posture. This will be performed by servoing prosthetic joints motions to the user's compensatory movements. Note that this scheme is valid to control intermediate joints like wrist, elbow and shoulder, but not for hand since no body compensation substitutes grasping functions.

This paradigm has already been explored and validated for the control of wrist pronosupination with able-bodied subjects [6]. In the work presented here, we extend this study to transradial amputated people, on a different task.

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## II. Materials and Methods

*Control law*

The control law implemented to servo the prosthetic wrist pronosupination to the user's compensatory motions is the same as the one presented in [6]; we briefly recall it below. To supplant wrist pronosupination, amputated subjects tend to use trunk and arm compensatory motions to change the hand orientation [4]. To avoid an artificial mapping between these motions and prosthesis' motions, we directly measured the forearm rotation around its longitudinal axis (elbow-hand axis) and servoed the wrist motion to it.

First, the rotation between $\theta_{fa}$, the current posture of the lower-arm, and $\theta_0$, an objective posture for the subject to go back to, is computed:

$$\epsilon(t) = \theta_{fa}(t) - \theta_0 \tag{1}$$

$\theta_0$ is set here as the initial forearm posture but can change if necessary, depending on the task. $\epsilon$ is then used as the input of the control law, which pilots the prosthetic wrist angular velocity, $\dot{\theta}$:

$$\dot{\theta}(t) = \begin{cases} 0 \text{ if } |\epsilon(t)| < \epsilon_0 \\ \lambda(\epsilon(t) - sign(\epsilon(t) - \epsilon_0)\epsilon_0) \text{ otherwise,} \end{cases} \tag{2}$$

where $\epsilon_0$ is a deadzone threshold (here set to 5 deg) and $\lambda$ is a scalar gain (set to 2 $s^{-1}$) that tunes the rate of correction. These values were chosen experimentally to ensure stability.

*Experimental set-up*

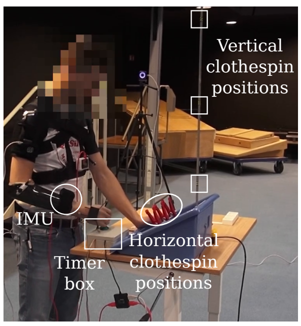
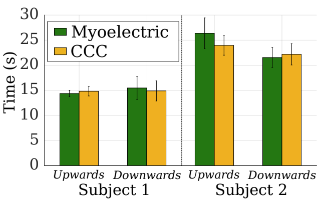


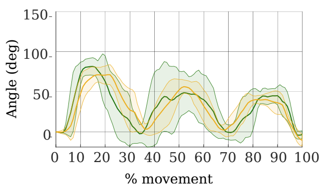

Fig. 1: Set-up of the refined Rolyan Clothespin test with transradial amputees.

Two transradial amputees, regular myoelectric-prosthesis users, participated to this experiment. They were asked to perform the refined Rolyan Clothespin test [7], both with their usual myoelectric control (MYO) and with the proposed control scheme, later called Compensations Cancellation Control (CCC). There were no specific training session and the Rolyan was achieved 5 times with each control mode. For the latter, the participants' wrist rotator was replaced with one from the laboratory, including an encoder, and controlled with an external Raspberry Pi 3©. As CCC cannot be implemented for the hand, the usual myoelectric control of each participant was conserved for the grasping function. For this experiment, $\theta_{fa}$ was obtained with an Inertial Measurement Unit located on the prosthetic forearm (see Figure 1). The protocol of this experiment was approved by the Université Paris Descartes ethic committee CERES. All participants gave their written informed consent in accordance with the Declaration of Helsinki.

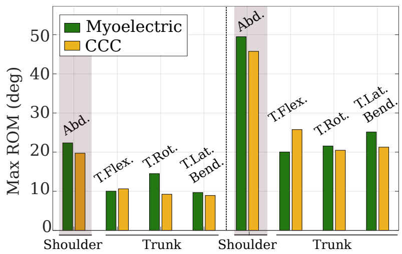
## III. Results




(a) Time of the task




(b) Example of wrist motions




(c) Shoulder and trunk maximum Range of Motion

Fig. 2: Rolyan assessment (mean and standard deviation over the 5 trials for each mode and subject). (a) Task performance; (b) wrist motion (wrist pronosupination for downwards pins relocation, by one subject) and (c) body compensations.

*Task performance*

Figure 2(a) shows the time, averaged over trials, obtained with the two control modes, for both subjects. The times are clearly similar between the direction of the motion (upwards –horizontal-to-vertical– or downwards –vertical-to-horizontal–) and the control modes. We can also notice that the standard deviation is small, for both MYO and CCC.

*Joints motions*

The example of wrist angular trajectories of Figure 2(b) illustrates that CCC gives trajectories very close to the ones given by MYO. Considering shoulder and trunk, it has to be checked that, with CCC, the use of compensatory motions as input of the controller does not enhance them. The maximum Ranges of Motion (over trials and pins location) of shoulder abduction (Abd.) and trunk angles (flexion, rotation and lateral bending – T.Flex., T.Rot., T.Lat.Bend.–) with CCC are thus compared to the ones with MYO. Yet, as prosthesis users may employ undesirable compensatory motions to achieve the task with MYO, it was also verified that compensations of these participants were not troublesome (angles values and holding time) [8]. In Figure 2(c), we see that compensatory joints motions are not higher with CCC than with MYO and stay in acceptable ranges.

## IV. Discussion

The different metrics assessing Rolyan test all show that CCC was quickly mastered and was as good as MYO, while it was totally unknown from the participants, who discovered it during the experiment: the time of the task is similar, with a small standard deviation, wrist angular trajectories are close and body compensations are not heightened. Both participants also appreciated the absence of co-contraction with CCC and reported that this control mode induces less muscle fatigue than MYO. These results confirm what has been obtained with able-bodied subjects, on a different task.

Yet, as seen Figure 2(c), the subjects called different shoulder and trunk strategies (Subject 2 moved them much more than Subject 1). We are thus intending to perform an extensive study with more amputated participants to have a larger overview of users' strategies. The definition of the objective posture also merits further attention, to confirm that a same objective can be used for multiple arm movements.

## V. Conclusion

The new control scheme for UL prostheses presented in this paper proposes to use the device to control the user's posture, while letting him/her in charge of the end-effector positiong and orientation. Tested with two transradial amputees, it was as good as conventional myoelectric control without any specific learning, while potentially reducing the muscle fatigue. Following these promising results, the concept developed here could be extended to the control of more than one DOF. We think that it could also be adapted to other rehabilitation devices like UL exoskeletons.

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