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Kinematic characterization and validation of an upper limb rehabilitation device

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Abstract
Rehabilitating patients with upper limb poststroke paresis is a difficult challenge for both therapists and patients. In fact, to be as effective as possible and enable long-term restoration of arm's motoric function, therapy must be introduced early and combined with frequent rehabilitation sessions. Active rehabilitation devices are establishing themselves as a likely solution, as they not only provide highly repetitive movements enabling to reinforce the connections between the visual and the motion sensory systems, but also allow alleviating the workload of physiotherapists. The conceived design configuration of an upper limb rehabilitation device, comprising 14 degrees of freedom, of which 8 are active, is thoroughly analysed in this work. The ranges of motion are validated against human arm movements using functional model representations. The sensors and their approximate positioning are considered as well. An in-depth analysis and characterization of the kinematic performances of the device is performed by using the Denavit-Hartenberg parameters. The simplified model of the rehabilitation device, and the resulting kinematic tree, are then implemented in the robotic operating system, enabling the simulation of its behaviour. The compatibility of the device for the rehabilitation of daily life activities is also validated by using a workspace analysis based on the functional representation of the arm.

Upper limb rehabilitation, mechatronics device, kinematic analysis, workspace validation, ROS implementation

1. Introduction

Strokes, typically caused by obstructing or rupturing blood vessel, are one of the major causes of impaired brain areas related to fine upper limb movements. The recovery of the patients is conditioned by the severity of stroke, the related brain injury, patients' age, as well as the volume and timing of the rehabilitation process [1]. Using active devices in the rehabilitation process, especially considering earlier therapy introduction, critical for positive outcomes, as well as of the ability to provide a large number of intense movements, tailored individually to each patient, enhances considerably rehabilitation quality. In fact, the available body of evidence proves that autonomous exercise, combined with dynamical repetitive task-oriented training, is a promising and efficient rehabilitation modality [2], as it allows reinforcing the connections between the visual and the motor sensory systems. Proper interactions between the patient and the rehabilitation device are then of utmost importance for successful rehabilitation. Upper limb active rehabilitation devices are usually classified in two groups: wearable exoskeletons, which provide support and assistance in motion control of the entire arm, and end-effector devices, that are used in controlling the trajectory of the hand only. An innovative exoskeleton-type device, whose basic concept was previously proposed [3], is hence elaborated in detail in this work. To define the respective functional and technical requirements, which is an essential step in modelling its interactions with the patient, in the whole design process a functional model of the human arm is used, hence enabling a credible representation of the rehabilitation device. By using a 15 degrees of freedom (DOFs) biomechanical model of the arm based on experimental data developed in [4], which includes an accurate representation of muscles and joints, an optimization of the mechanical design and of the control subsystem is enabled as well. This enables simulating and testing several feasible device variants.

The aim of this work is, therefore, to validate the proposed design concept of an upper limb rehabilitation device in terms of the necessary number, placement and alignment of the DOFs and of the patient-device interfaces. An in-depth kinematic analysis of the device is also performed via simulations and studies enabled by implementing its model in the robot operating system (ROS). Some associated prototyping undertakings are described as well.

2. Design of an innovative rehabilitation device and peripherals

The model of the conceived innovative rehabilitation device, fitted to the functional representation of the human arm, is shown in Fig. 1. The model of the device is implemented in the open source 3D creation suite *Blender* in combination with the *Robot Designer* add-on, developed as part of the *Neurorobotics* platform (NRP) that allows blending the brain, body and robot models with their environment [5]. The patient-device interaction is achieved herein along the shoulder-elbow-wrist-hand interaction chain. The devised rehabilitation device itself comprises rigid girders, coloured in blue, mutually connected by joints enabling relative movements between them, each corresponding to one DOF. Two different joint types are used: prismatic joints enabling linear, and revolute joints permitting rotational motions. The joints can then be active (actuated) or passive. The combination of the diverse joint types allows attaining the following functional variants: the revolute active (RA), the revolute passive (RP), and the prismatic passive (PP) joint sets.

2.1. Rehabilitation capability and patient-device interactions of the designed device

The key steps in designing a smooth and functional rehabilitation device, is aligning the complex human joints to the revolute robotic joints, and compensating the misalignments between

the rigid exoskeleton girders and the soft human tissue. Since one revolute joint cannot entirely capture the complex motion patterns of the corresponding human joint, nor the joint coupling effects, the number of revolute joints has to be increased, resulting in bulky and expensive devices. In the herein employed conformation, a 15 DOFs skeletal model of the human upper extremity (UE), with the functional joint kinematics devised in [4], is then used (Fig. 1). The UE model incorporates 3 DOFs for the shoulder kinematics (elevation plane and angle, and shoulder rotation), 1 DOF for the elbow (flexion), 3 DOFs for the wrist (forearm rotation, flexion and deviation), a well as 4 DOFs for the index finger and 4 DOFs for the thumb. The coupling among the joints is considered, while the ball-and-socket joint is used to represent the articulation between the scapula and the humerus. What is more, regression equations are used to define the shoulder girdle movement. The used bones' dimensions are, in turn, based on the typical 50th percentile male setup.

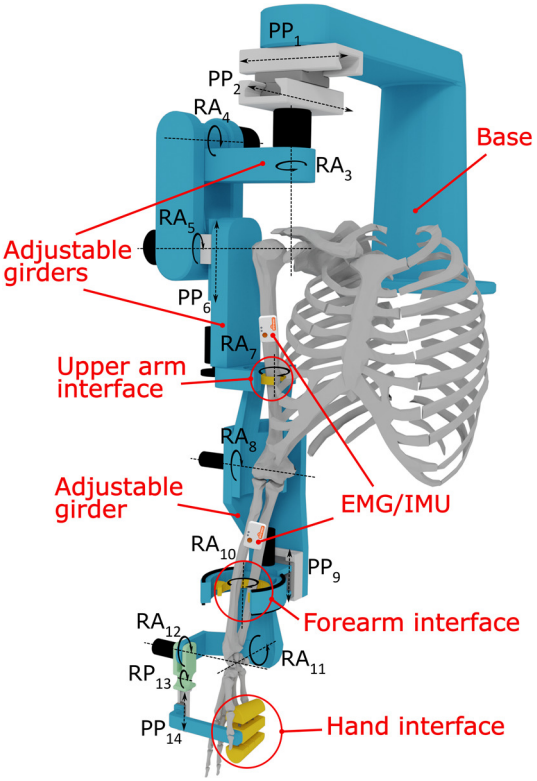

Figure 1. Model of the device fitted to functional arm representation.

To design a self-aligning device [6], minimize the number of active DOFs, expand the ranges of motion (ROMs), and enable smooth movements, the introduction of additional PP joints for misalignment compensation is mandatory. The conceived innovative device comprises thus 14 joints (i.e., 14 DOFs), of which 8 are active, while 6 are passive. Every actuator (depicted in Fig. 1 in black) drives then one RA joint, with the respective axes of revolution (represented by dashed lines) positioned so as to enable the rotary motion of the human joints. Preliminary in-house measurements on differently aged human subjects, allows establishing that in activities of daily life (ADLs) the output velocity of the actuators should be in the 1 rad/s range, providing an important boundary condition for their dimensioning. The whole device is then fixed to a strong base that supports its weight, while it is connected to the arm via three interfaces.

The different joint types used in the proposed rehabilitation device, with the respectively used designating numbers, are thus specified in the first two columns of table 1. The main assisted UE motions related to each joint are outlined in the third column, bearing in mind that for certain UE motions more than one joint are employed. As depicted in Fig. 1, two grey PP joints (PP_1 and PP_2) are fixed here at the interface to the base, enabling the compensation of the lateral and the posterior/anterior translations of the shoulder girdle. Three RA joints (RA_3, RA_4 and RA_5) enable, in turn, the shoulder protraction/retraction, abduction/adduction and flexion/extension motions. Joint PP_6 compensates the linear misalignments that may occur between the upper arm interface and the arm itself. Joint RA_7 allows the internal/external rotation of the shoulder, while joint RA_8 enables the flexion/extension of the elbow. Joint PP_9 is used to compensate the misalignments between the elbow and the wrist axes, while the revolute joints RA_{10}, RA_{11} and RA_{12} allow the forearm pronation/supination, the flexion/extension of the wrist and the radial/ulnar deviation. The compliant RP_{13} joint is the only used revolute passive joint, and is used, together with PP_{14}, to compensate the misalignments between the joint RA_{12}, assisting in the radial/ulnar deviation, and the hand interface.

To define the ROMs of the device and compare them to the factual arm's ROMs, the neutral position (i.e., baseline) needs to be defined first. The neutral position of the arm, illustrated in Fig. 1, is set when the humerus is parallel to the thorax vertical axis, the elbow is fully extended, the forearm and the hand lie in the sagittal plane, while the third metacarpal of the wrist is aligned with the longest axis of the forearm [4]. The limits of the angular and linear ROMs of the devised rehabilitation device (listed in table 1 in the 4th and 5th column), and of the arm (6th and 7th column in table 1), are then calculated relative to this baseline. It can hence be deduced that the designed device allows attaining a suitable assistance for most of the upper arm motions, as well as compensating unintentional translations and interface misalignments. The design of the device induces, however, some motion constraints in the shoulder overhead motions, i.e., in the abduction/adduction and the flexion/extension ROMs, where ~ 60° would additionally be needed, providing hence room for future improvements. Given the foreseen position of the RA_{12} actuator, a motion "deficit" of about 15° appears also in the flexion/extension of the wrist. Employing a different motion transmission element between this actuator and the link, could enable to position it differently, thus permitting the utilization of the whole ROM of the wrist.

As for the mentioned three arm-device interfaces, the first one is fixed to the upper arm for the shoulder interactions, the second is used for the forearm interactions, while the third is aimed at the hand interactions (cf. Fig. 1). The upper arm and forearm interfaces are then designed as straps for fastening the device to the arm, concurrently serving as the ideal place for the placement of the force and torque sensors supporting the foreseen flexibility of the control. Using an ergonomic approach, and taking into account the often-occurring hand spasticity [7], the hand interface is, in turn, designed as a 5 DOFs spring device that can be easily grasped. This tree-part component encompasses a base, a rear part designed for thumb placement, which is connected to the base via three springs, and a front part with segments used for the placement and the rehabilitation of each of the remaining fingers. The tension of each of the used springs can be adjusted here to accommodate different grasping strengths. Such a configuration can be used to rehabilitate the ADLs such as pinching and grasping.

2.2. Control, adaptability and safety requirements

Following and intertwining with the mechanical design, the control sub-system of the rehabilitation device must also be addressed. As the respective design criterion, the challenging adaptive assistance, where the patient is encouraged to contribute to the performed movement as much as possible, while the rehabilitation device provides minimal assistance, is thus selected. The device should also allow memorising the session-to-session recovery model and adapt it to the recovery

dynamics of each patient. To meet such requirements, an adaptive control approach, relying on decision making based on numerous sensors' measurements, must be incorporated in the design. To enable the sensing of patient's muscle activity, and acting upon it, two wireless surface electromyography (EMG) sensors will hence be used, one on patient's upper arm and the other on his forearm. To allow an accurate determination of arm's position and orientation, the EMG sensors will additionally be integrated with an altimeter and a multi-DOFs inertial measurement units (IMU) comprising an accelerometer, a gyroscope and a magnetometer. Initial tests with an EMG/IMU *Shimmer3* integrated device, coupled to the *Consensys* interfacing software [8], prove that such a configuration is fully compatible to the foreseen application. The thus envisioned mounting positions of such sensing units is depicted in Fig. 1.

Table 1. Definition of the used joints, of the rehabilitation device and upper extremity ROMs, and of the DH kinematic parameters of the device.

Joint i	Type	Enabled UE motion or compensation	Device ROMs		UE ROMs [4, 9]		DH kinematic parameters			
			Min.	Max.	Min.	Max.	θ_i/°	α_i/°	r_i/cm	d_i/cm
0	Fixed	Base	-	-	-	-	-	-	-	-
1	PP	Compensates SH girdle lateral translations	- 5 cm	5 cm	-	-	0	90	36.68	31.40
2	PP	Compensates SH girdle anterior/posterior translations	- 5 cm	5 cm	- 0.29 cm	0.62 cm	- 90	- 90	5.69	d_2
3	RA	SH girdle protraction/retraction	- 120°	120°	- 15°	20°	- 90	90	0	d_3
4	RA	SH abduction/adduction	- 90°	120°	0°	180°	θ_4	90	14.72	3.83
5	RA	SH flexion/extension	- 90°	120°	- 60°	180°	θ_5 - 90	- 90	15.09	0
6	PP	Compensates SH interface misalignments	- 5 cm	5 cm	-	-	θ_6 - 90	- 90	0	- 3.67
7	RA	SH internal/external rotation	- 120°	120°	- 20°	90°	90	0	9.14	d_7 + 17.18
8	RA	EL flexion/extension	- 10°	150°	0°	130°	θ_8 + 90	- 90	0	12.22
9	PP	Compensates EL-WR misalignments	- 2.5 cm	2.5 cm	-	-	θ_9	- 90	6.4	- 4.31
10	RA	FA pronation/supination	- 110°	110°	- 90°	90°	180	0	9.46	d_{10}
11	RA	WR flexion/extension	- 55°	120°	- 70°	70°	θ_{11} + 90	90	7.63	10.31
12	RA	Radial/ulnar deviation	- 30°	60°	- 25°	10°	θ_{12} - 90	- 90	0	6.24
13	RP	Compensates WR-HD misalignments	- 5°	5°	-	-	θ_{13} + 180	0	3.73	0
14	PP	Compensates WR-HD misalignments	- 2.5 cm	2.5 cm	-	-	θ_{14} + 90	90	0	0
15	Fixed	HD interaction	-	-	-	-	0	0	10.49	d_{15}

PP: prismatic passive, RA: revolute active, RP: revolute passive, UE: upper extremity, SH: shoulder, EL: elbow, FA: forearm, HD: hand, WR: wrist.

In order to enable the rehabilitation of the majority of stroke patients, it is necessary to equip the conceived device with several variable length links. The respective girders are hence configured to be adjustable to each patient. The most important factor in dimensioning the variations in the respective lengths is the age related to the incidence of stroke. In fact, despite occurring among children, the yearly incidence of stroke is limited in this case to 1 to 2.5 cases in 100 000 children, which is substantially lower than the 30 to 120 cases per 100 000 adults aged 35 to 44. After the age of 55, each decade results in a rapidly increasing stroke probability, giving rise to an extremely high yearly incidence of 670 to 970 cases per 100 000 adults aged 65 to 74 [10]. The conceived device is, thus, ideated so as to be suited for the usage by the working-age and elderly population. In the device design depicted in Fig. 1, three girders are therefore made adjustable in length: the first one positioned close to the base for aligning the RA_3 joint (shoulder girdle), while the remaining two are intended for the coarse adjustment to differently sized patient's upper arms and forearms. The mentioned passive prismatic joints PP_6 and PP_9 are, in turn, employed for the respective fine adjustments.

Several safety layers are, finally, to be incorporated in the designed device, including software limits based on human ROMs, electrical limit switches and mechanical hard stops.

2.3. Kinematic analysis of the proposed rehabilitation device

The kinematics of the conceived rehabilitation device, necessary to describe the relative positions of the bodies in motion and aimed at being used in motion planning algorithms and control, is described using two methods. Both methods are based on kinematic chains consisting of links connected in a parent-child relationship, where predefined types of joints are used to enable relative movements. Each link has then an assigned coordinate system (i.e., local frame), whose position is defined relative to its parent's frame. The joints constrain, in turn, the motion of the respective links to a translation along one axis (prismatic joints) or a rotation around one axis (revolute joints). The first employed method, the *Unified Robot Description Format* (URDF), stores the kinematic chain in a tree structure, together with the 3D models of the corresponding robotic links. This method allows also the addition of sensors to be used in robotic simulators, primarily for implementation in the mentioned ROS framework – a collection of tools and libraries used in creating the complex robotic NRP platform employed to develop simulations, while integrating and interlacing the human muscosceletal and the rehabilitation device models. The URDF format is quite versatile and widely used in many robotic applications, but at an expense of a rather intricate kinematic characterization. To succinctly describe the kinematics of the ideated rehabilitation device, an approach using the Denavit-Hartenberg (DH) convention [11] can thus be used. The numbering of the joints with a running variable i, as outlined in table 1 and in Fig. 1, where 0 indicates the base-frame of the device, while 15 is associated with the hand interface frame treated as the end effector, is hence used. The convention then dictates that the i^{th} joint connects with the $(i\text{-}1)^{th}$ and the i^{th} link, while the reference frames are assigned to the joints according to the following convention: (1) if joint i is prismatic, the z_i axis is assigned to its translation axis, while, if i is a revolute joint, the z_i axis is assigned to its axis of revolution; (2) the x_i axis is assigned to the common normal of the z_i and z_{i-1} axes, with its direction pointing from the z_{i-1} towards the z_i axis; (3) the y_i axis is selected so as to create the right-handed coordinate system.

Since the device base is fixed, all three axes (x_0, y_0 and z_0) are positioned here arbitrarily. The option presented in Fig. 2, with z_0 pointing upwards, is hence chosen. Assigning then the local frames to all joints, four DH parameters (θ_i, α_i, r_i and d_i), necessary for the basic definition of the kinematic chain, have to be measured. The parameter θ_i represents in this frame the angle between the axes x_i and x_{i-1} measured around the z_{i-1} axis and, since for the revolute joints z_{i-1} is the axis of revolution, θ_i characterizes the rotation of the joints. Parameter α_i indicates, in turn, the angle between the z_{i-1} and z_i axes around the common normal axis x_i. The remaining two parameters describe linear distances, with r_i indicating the distance between the z_{i-1} and z_i axes along the common normal x_i (for revolute joints, r_i is

thus equivalent to the radius of rotation), while d_i is the linear displacements between the x_{i-1} and x_i axes along z_{i-1} (in the case of prismatic joints, d_i characterizes, hence, their linear motion). Following these conventions, the DH parameters of the device are shown in Fig. 2, while their values are presented in a concise form in last four columns of table 1. It is important to note here that in table 1 the variable angular position of the revolute joint θ_i, as well as the variable linear position of the prismatic joint d_i, actually describe the $(i\text{-}1)^{th}$ joint and can be associated with the respective ROMs. For example, θ_4 can thus vary between the RA_3 minimal and maximal values of +/- 120°.

Since the number of the DH parameters is quite big, the following simplifications are then adopted to provide in Fig. 2 a visualization that is as clear as possible: (a) the origins of the frames are omitted to avoid cluttering the used schematics; (b) two or more bodies that are connected using fixed links are considered as a whole with one corresponding frame (although the respective URDF simulation file contains a more detailed kinematic description); (c) the zero length DH parameters are omitted from the schematic representation.

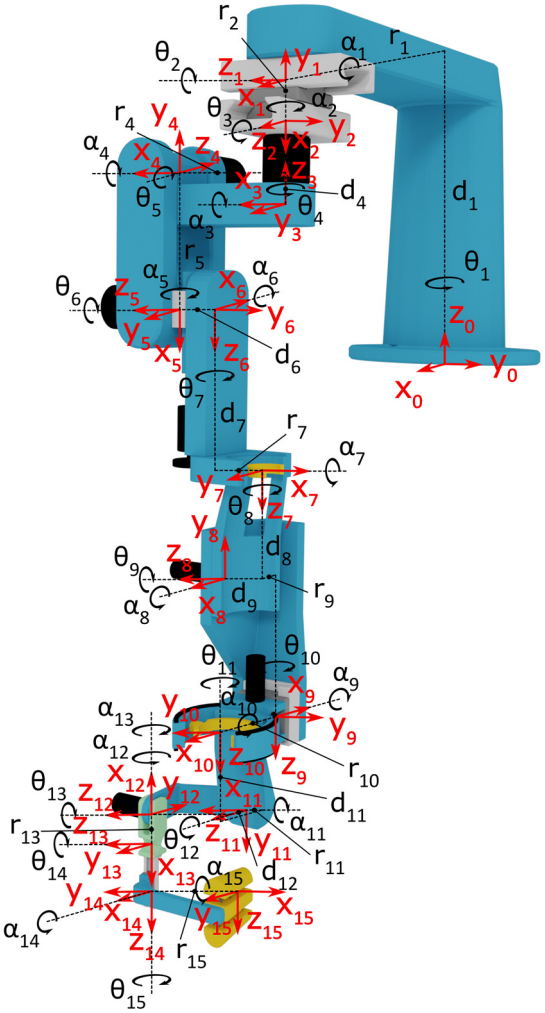

Figure 2. DH kinematic chain with the associated parameters.

Using the thus acquired DH parameters, homogenous transformation matrices and the corresponding kinematic equations can be setup to represent the relative positions of the links and the overall position of the conceived rehabilitation device. Since the DH parameters correspond to the physical lengths of the links of the device, they also serve as an outline for detailing iteratively further the device itself. The resulting rehabilitation system actually comprises therefore two mutually interacting systems: the human kinematic chain, previously described and developed in [4], and the kinematic chain of the ideated rehabilitation device, presented in this section.

3. Conclusions and outlook

An original design framework of a self-aligning rehabilitation device is detailed in this work. The main design goal, apart from multi-layered safety, is the usability of the device in the smooth and purposeful patient-device interaction, planned to instil a proper rehabilitation mechanisms resulting in the positive rehabilitation outcomes. What is more, the rehabilitation outcomes being aimed at, should be measurable and should results in a significant progress in terms of patient sensory and motion capabilities. The biomechatronic design approach is hence followed, based on the full functional representative model of the human agent used as basis for the iterative design process, as well as for the respective validation phases. The distribution of the DOFs across the shoulder-elbow-wrist-hand interaction chain is thus optimized, allowing to reduce the number of DOFs from 16 DOFs used in the initial concept [3], to 14 DOFs of the herein proposed set-up, providing considerable functional and implementation advantages. The size and position of the critical components is then determined, the device kinematics is detailly examined in the ROS framework, the functionality of the conceived device is validated, as are the rehabilitation ROMs. The dimensional adaptability of the device to a prospective diverse patient population is also taken into account, allowing to finalize the functional model of the device. Force and torque, as well as surface EMG/IMU sensors, and their indicative positioning, are also considered, providing the basis for their integration in an adaptive control approach, thus enabling to fully embrace patients' active participation.

The final rehabilitation device design configuration, although covering the overall ROMs in most of the human arm motions, imposes some remaining constraints in shoulder overhead and wrist motions, providing room for further improvements via the detailing of the device parts, coupled to structural optimization routines. The complete human muscosceletal and robotic sub-systems, as well as the corresponding interactions and dynamics, will hence be studied iteratively further using the developed NRP platform, enabling also the selection of the optimal actuators with the respective motion transmission elements. To facilitate the attainment of the low weight and low cost of the device, most of the resulting passive parts will, in turn, be 3D printed and prototyped using diverse materials, with special care devoted to the dimensioning of the passive compliant joint RP_{13}. Future work will also include the integration of the sensors in the control loops, the corresponding filtration of the output signals and the conforming data extraction, as well as special emphasis on the resulting data-based decision making procedures.

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