Extended Abstract

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Towards the optimal design of a passive upper limb exoskeleton compensating gravity

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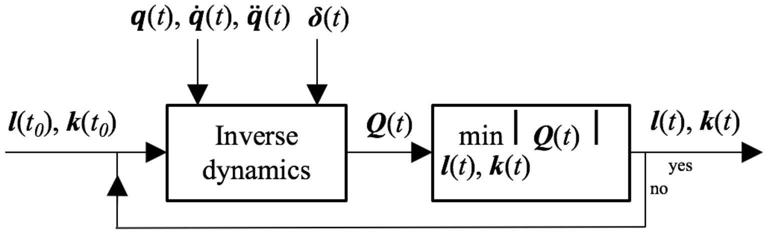
Context: A large proportion of the population undergoing rehabilitation is affected by neuromuscular disease. It is even the major population in paediatrics. Although these diseases are very variable, they usually result in muscular weakness and coordination problems. These disorders limit these patients in most daily activities, such as walking, drinking/eating, and communicating. In this context, it has been recognized that portable assistive exoskeletons would provide the necessary complement of muscular forces in their daily life to increase their level of autonomy and social participation [1]. But today, most portable exoskeletons for people with neuromuscular disease are dedicated to the lower limbs: e.g. Re-Walk, the first FDA-approved portable exoskeleton. Contrarily at the upper limb, there is still no motorized portable exoskeleton to assist people with neuromuscular disease in their daily life. Further, prototypes are barely emerging, primarily aimed at healthy adults for either workplace, sports, or military applications, currently leading to designs that could not be adapted to patients with neuromuscular disease, due to their different control strategy and to the problem of engine oversizing [2]. A solution could be the development of an upper limb exoskeleton compensating for gravity in a passive way, i.e. with passive elements, such as springs, and without a system of actuators, acquisition cards and batteries. The objective of this study is to identify the optimal design of a passive portable upper limb exoskeleton compensating gravity for patients affected by neuromuscular disease.

Methods: We propose to define the optimal design problem as follows: for a given geometrical model of the "upper limb – exoskeleton" system (developed with Robotran software [3]), given dynamical parameters, $\boldsymbol{\delta}$, and upper limb kinematics, $\boldsymbol{q}$, $\dot{\boldsymbol{q}}$, $\ddot{\boldsymbol{q}}$, the optimal design can be the one (Fig. 1) that provides at each instant, t:

- the best geometric parameters, $\boldsymbol{l}(t)$, e.g. the positions of the passive elements and mechanism dimensions
- the best parameters of the passive elements, e.g. either the spring stiffness, $\boldsymbol{k}(t)$, or neutral length, $\boldsymbol{L_0}(t)$

that minimize the absolute joint torques at the upper limb. The optimization constraints are that the global kinematic error will remain below a certain threshold of tolerance, namely a few mm. This global kinematic error is defined [4] at each instant as the difference between the actually measured Cartesian coordinates of the upper limb joints for a given motion and the Cartesian coordinates of the upper limb multibody model joints following this motion. The underlying idea is that on the one hand a sufficiently faithful fit of the human motion by the model would be a criterion of comfort of use of the exoskeleton, but that on the other hand these motions should not fit exactly, within a few millimeters, if this allowed the optimization to converge better.

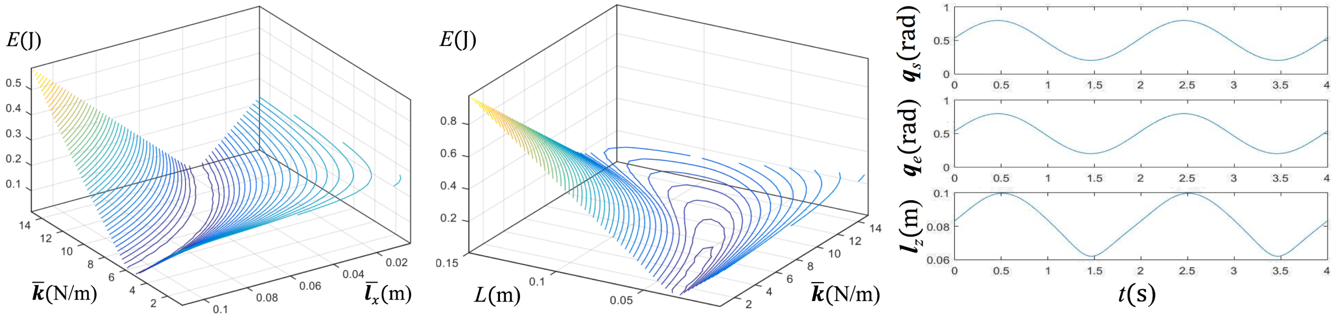


$\boldsymbol{l}$: geometric parameters ; ($\boldsymbol{l_0}$: initial)
$\boldsymbol{k}$: parameters of the passive elements ($\boldsymbol{k_0}$: initial)
$\boldsymbol{q}$, $\dot{\boldsymbol{q}}$, $\ddot{\boldsymbol{q}}$: upper limb kinematics (relative generalized coordinates and derivatives)
$\boldsymbol{\delta}$: dynamic parameters, including segment masses, centres of mass and inertias
$\boldsymbol{Q}$: upper limb joint torques

Fig. 1. Process defining the optimal design, performed at each instant, t.

Further, let us note that currently, we try to identify the optimal solution variants without worrying too much about the physical realization. In a future step, we will analyze if it is necessary to constrain the optimization to obtain physically feasible mechanisms. For example, we currently leave the optimization free to identify whether the best solutions require a spring with variable stiffness, $\boldsymbol{k}(t)$, over time. If this spring stiffness variation does not have much influence on the final performance, then we will use the best value of constant/average stiffness, $\bar{\boldsymbol{k}}$. Contrarily, if this spring stiffness variation is paramount, then we will study the physical feasibility, either by using/developing the appropriate spring, or by discretizing the problem by using a spring coupled to a clutch.

Results: To simplify the interpretation, the illustrated results are based on a simple 2 d.o.f. upper limb model, composed of one arm and one forearm articulated around lateral hinge joints and linked by a spring. More complex geometrical models of the "upper limb – exoskeleton" system will be presented at the conference. Fig. 2 A-B present the mechanical energy at the upper limb elbow joint according to: A. spring constant stiffness, $\bar{\boldsymbol{k}}$, vs. lever arm constant position of the spring at the arm level, $\bar{\boldsymbol{l}}\boldsymbol{x}$, and B. spring constant natural length, L, vs. spring constant stiffness. Fig. C. presents the optimal temporal variation of the position of the spring along the arm.



A. B. C.

Fig. 2. A and B. Mechanical energy, E, at the elbow joint according to the spring constant stiffness, $\bar{\boldsymbol{k}}$, the lever arm constant position of the spring at the arm level, $\bar{\boldsymbol{l}}\boldsymbol{x}$, and the spring constant natural length, L. C. Optimal variation of the position of the spring along the arm, $\boldsymbol{l}_z$, for given kinematics (shoulder, $\boldsymbol{q}_s$, and elbow, $\boldsymbol{q}_e$) and constant natural length, L, and stiffness, $\bar{\boldsymbol{k}}$, of the spring.

Discussion and conclusion: The preliminary results from Fig. 2A and B show that for a minimum expended mechanical energy, portable assistive exoskeletons could tend to be smaller in size, by reducing the lever arm and increasing the stiffness (Fig. 2A), and fixing the natural length (Fig. 2B). This suggests that it would be possible to develop further assistive exoskeletons in the form of exosuits, thinner to follow more accurately the natural action lines of muscles. Fig. 2C shows that the position of the spring along the arm only evolves of a few cm, without any constraint on this variable on the optimization. Further, let us note that by combining the two parameters of longitudinal position, $\boldsymbol{l}_z$, and lever arm, $\boldsymbol{l}_x$, of the spring on the arm, we were able to generate various trajectories of the spring fixation on the arm, which we currently analyze: especially when this trajectory is circular or quasi-elliptic for given constant natural length, L, and stiffness, $\bar{\boldsymbol{k}}$, of the spring, this trajectory can be achieved by a four-bar mechanism with spring, which confirms the interest for this solution in other gravity compensation mechanisms in the literature. To conclude, generally this process can be used as a tool for optimal design, to customize for each subject the best geometric and dynamic parameters of a portable upper limb exoskeleton. As a perspective, the exoskeleton model could be extended to a tridimensional upper limb model [4], by replacing the four-bar linkage e.g. by an RSSR linkage, composed of two revolute (R) and two spherical (S) kinematic pairs.

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