

# Design of a Hybrid Adaptive Support Device for FES Upper Limb Stroke Rehabilitation

Giuseppe Cannella, Dina S. Laila and Christopher T. Freeman

**Abstract** A novel design of a low cost non-powered orthosis device for home-based upper limb stroke rehabilitation is proposed. The design allows the device to be integrated with a dual robotic and electrical stimulation control scheme. This enables exploitation of the motor relearning principles which underpin both robotic therapy and Functional Electrical Stimulation (FES) based stroke rehabilitation. This work focuses on the mechanical design of the non-powered orthosis, based on gravity balancing theory and provides preliminary dynamic simulations of the 3D CAD model.

**Keywords** Passive orthosis • Gravity balancing theory • FES • Stroke rehabilitation • Multibody dynamics

## 1 Introduction

Stroke patients can relearn how to use impaired parts of their body by progressive goal-based physical exercise programmes. New therapies often employ expensive and complex robotic devices to train patients. Another widely used technology in the field of stroke rehabilitation is Functional Electrical Stimulation (FES), which contracts muscles through short electrical pulses in the same way as those produced by the central nervous system [3].

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In this research it was chosen to combine FES with an adjustable low cost support device, controlling both with a model-based controller. This represents a novel non-conventional therapy and the overall device can be defined as hybrid adaptive low cost support for home-based stroke rehabilitation. The design of the mechanical support device is based on gravity balancing theory [4]. This theory allows a low cost arm support to be designed to carry the weight of the patient's arm during training throughout their whole range of motion, allowing them to use their residual muscle strength to perform the rehabilitation task. Moreover, the system is integrated with an adjustment balancing mechanism which allows variation of the support given to a patient's arm according to their inertial characteristics. These project specifications are addressed in this paper which develops a 3D CAD model of the mechanical support, whose dynamic performance is evaluated in Sect. 4.

## 2 Non-Conventional Rehabilitation Therapies

In non-conventional rehabilitation therapy the manual assistance of the physiotherapist is augmented or entirely replaced by a rehabilitative device. These specific devices are usually classified as: robotics, powered and non-powered orthoses. Robotic devices and powered orthoses consist of complex robots and active mechanical systems operating under computer control. Their design features mean they can be personalised for each patient's anatomy, but this often results in a complex and expensive device. On the other hand, non-powered orthoses are much simpler devices that provide gravitational support to the specific limb that needs to be trained, helping the patient to perform rehabilitation tasks. Their design and usage complexity is lower compared to that of robots and active orthoses. The only restriction for non-powered orthoses is that they need to be actuated by the patient's neuromuscular effort, since they only provide gravitational support. Hence they can only be employed for the rehabilitation of subjects with residual force in their muscles. Two examples of existing non-powered orthoses are the Wilmington Robotic Exoskeleton (WREX) [7] and the Armon Orthosis [5, 6]. WREX is a body-powered orthosis that provides gravitational support to the patient's arm thanks to a light exoskeleton mechanism that is counterbalanced by means of elastic bands. The Armon Orthosis is non-powered and designed to be mounted on a wheelchair or on a table, providing gravitational support to the patient's arm. It exploits gravity balancing theory, combining a parallelogram linkage and zero-free length springs.

In this research, we propose a novel design that combines gravity balancing based non-powered orthotic support and muscular stimulation by means of FES, in order to increase the effectiveness of therapy without sacrificing the low cost home-based philosophy.

### 3 Mechanical Design

In Fig. 1 the mechanism chosen to meet the device specifications for the purpose of this research is shown. It comprises a planar 2 degrees of freedom (d.o.f.) kinematic chain made of a parallelogram linkage combined with two zero-free length springs, where each spring balances each d.o.f. of the mechanism.

Zero-free length springs are useful components within gravity balanced systems since they have linear characteristics; the force acting on the spring is directly proportional to its deflection, and this feature simplifies the mechanical analysis of the systems in which they are employed. Moreover, the parallelogram linkage in this particular configuration allows the device to be in contact with just a small area of the forearm of the patient, identified by the Combined Center of Mass (CCM). Performing both a potential energy and moment equilibrium analysis [1] yields the following balancing conditions for the mechanism:

$$\begin{cases} (m + m_4)gL + m_3gr_{m3} + m_1gr_{m1} = k_2ar \\ mgL + m_4gr_{m4} - m_2gr_{m2} - m_3gr = k_1ar. \end{cases} \quad (1)$$

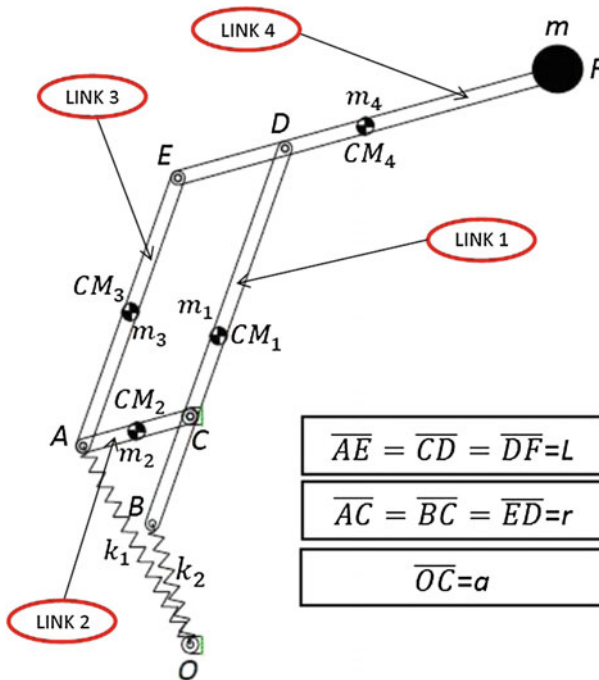


Fig. 1 Kinematic chain of the arm support



**Fig. 2** 3D CAD model and subcomponent labels

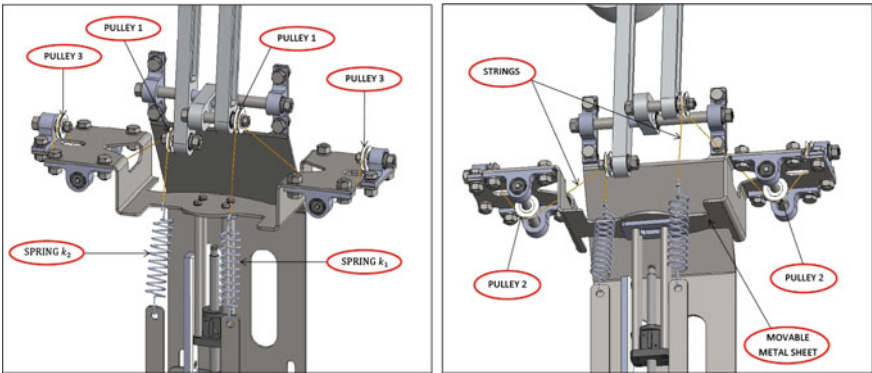
This system is a linear mechanical model, which will later simplify the analysis. Moreover, the rehabilitative device has to be adaptive and the above system contains all the variables of the mechanism that can be varied in order to adjust the mechanical behaviour of the system according to different patient anatomic characteristics and their progress in the rehabilitation programme. Among all the inertial and spring parameters,  $a$  was chosen to be varied. This choice allows the adjustment mechanism to be placed below the linkage and it does not directly affect the inertia parameters of the links.

In Fig. 2 the 3D CAD model of the orthosis is shown. It consists of a parallelogram linkage, with a light weight interface connection for the user's arm (Fig. 3), a balancing mechanism (Fig. 4), an adjustment system (Fig. 5) and a frame. In total it has 5 d.o.f. divided as follows: 2 d.o.f. for the interface connection, 2 d.o.f. for the parallelogram linkage and 1 d.o.f. for the vertical axis of rotation of the mechanism. For more details about the 3D CAD model refer to [1].

Deep groove ball bearings were selected to be mounted in the joints of the device in order to reduce friction and thereby improve the mechanical performance of the overall system. The linkage links are made of aluminium in order to reduce the overall inertia of the mechanical device. The same approach has been used for the interface connection in order to produce a light weight component which does not introduce additional dynamics to the overall linkage. The lower part of the linkage is connected to the springs via a string-pulley arrangement. The linear system (1) results from the assumption of zero-free length springs, however the proposed solution allows them to be replaced with normal extension springs, significantly reducing the cost of the device. The balancing behaviour of the device

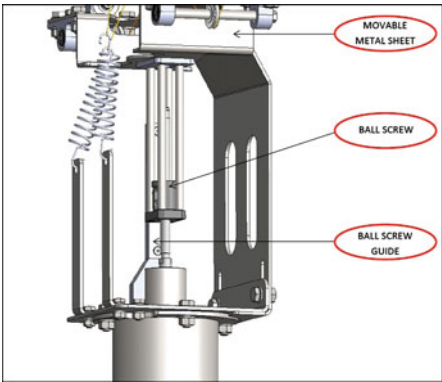


**Fig. 3** Detailed view of the two revolute joints of the interface connection



**Fig. 4** Detailed views of the gravity balancing mechanism. *Left* view from above. *Right* view from below

**Fig. 5** Detailed view of the adjustment system



can be varied by an adjustment system comprising a ball screw connected to an electric motor. This electromechanical solution will be controlled by a model-based FES controller which automatically varies the amount of support needed by the patient by changing the value of  $a$ . The placing of the balancing mechanism and the adjustment systems between the linkage and the 5th revolute joint allows the overall inertia of the system to be reduced about the vertical axis of rotation. Since the device may need to be moved between different environments, four casters with brakes have been installed at its base.

## 4 3D CAD Model Dynamic Simulation

In this section the dynamic simulation of the 3D CAD model is carried out in SolidWorks using the Motion Analysis module. The study is performed by analysing the displacement of the centre of mass of the interface connection, when a specific payload is carried by it. For the sake of simplicity the 2 d.o.f. of the interface connection are locked during the simulations. The device can adapt its balancing behaviour for a range of different payloads. It was chosen to perform four simulations where the values of the payload and the distance  $a$  have been calculated according to the linear relationship (1): first simulation  $m = 0$  kg and  $a = 3.5$  mm; second simulation  $m = 1$  kg and  $a = 19$  mm, third simulation  $m = 2$  kg and  $a = 34.5$  mm, fourth simulation  $m = 3$  kg and  $a = 50$  mm, which corresponds to the maximum arm weight. The stiffness of the two springs is calculated by a Matlab code based on the gravity balancing condition (1), and the values are  $k_1 = 4048$  N/mm and  $k_2 = 5035$  N/mm, which are kept constant during all four simulations. Since the joints have been designed using bearings, the static friction is  $\mu_s = 0.0024$  and the dynamic friction is  $\mu_d = 0.0012$ . The simulations are run for a duration of 20s. The coordinates of the centre of mass of the interface connection are shown in Figs. 6, 7 and 8. The results show how the dynamic behaviour of the device improves when the payload increases. The reason is that the stiffness of the springs is a fixed parameter that has been calculated for the case of the maximum payload,  $m = 3$  kg. Moreover the oscillatory trend of the centre of mass is due to the lack of damping and to the fact that the stiffness of the springs and the parameter  $a$  were set using a static gravity balancing analysis. These confirm that the device is able to support a payload with only a minor level of oscillation, but indicate that a controller is required in order to overcome this drawback. It should be also taken into account that on the real prototype, these oscillations will be mitigated by the additional natural damping introduced by the patient's arm.

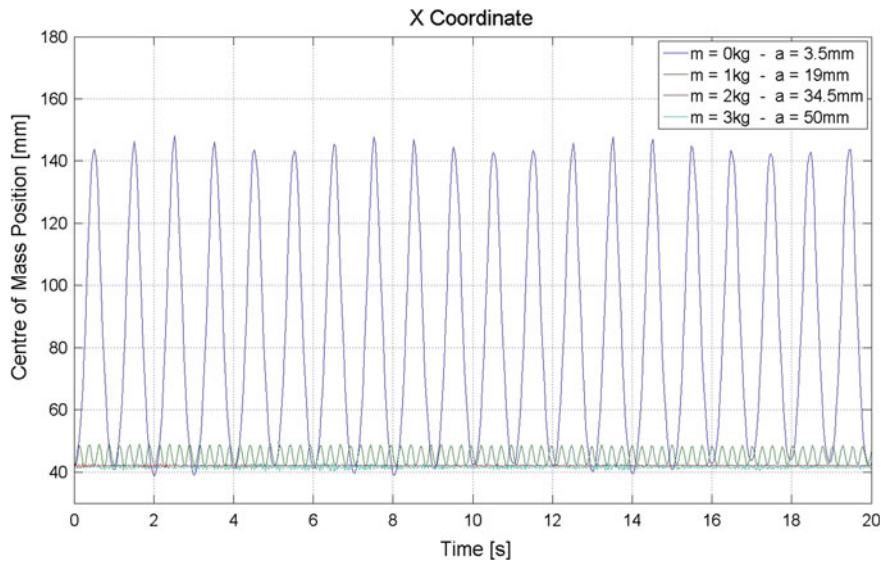


Fig. 6 X coordinate of the position of the centre of mass

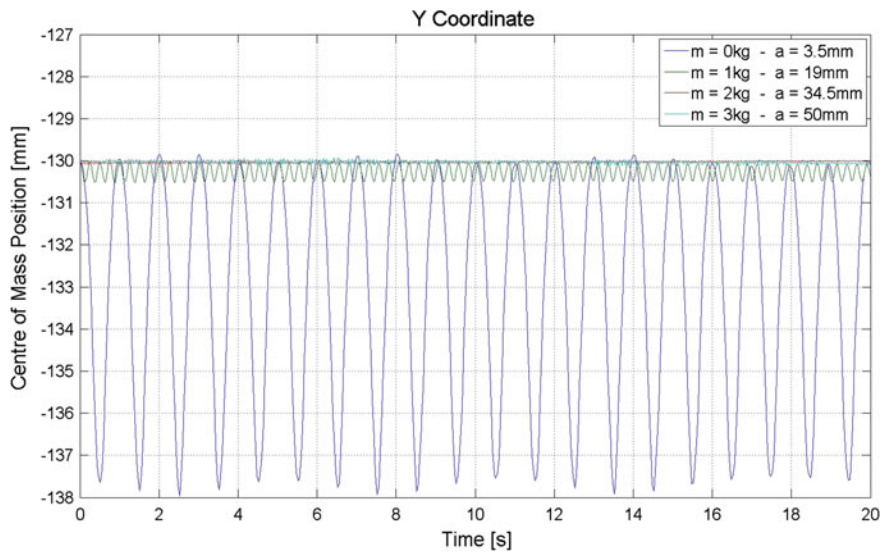
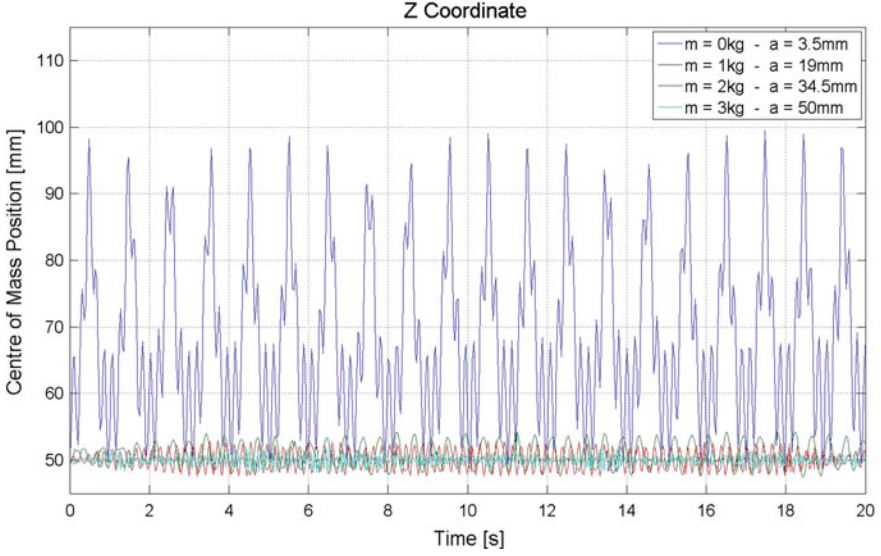


Fig. 7 Y coordinate of the position of the centre of mass



**Fig. 8** Z coordinate of the position of the centre of mass

## 5 Hybrid Control Scheme

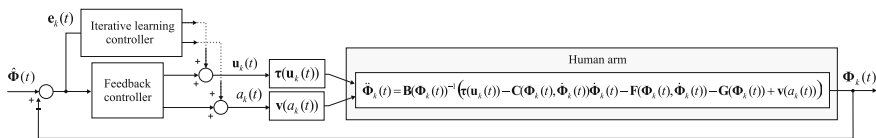
During therapeutic application, the support device will be used simultaneously with FES applied to clinically relevant muscles groups in the patient's upper limb. This will enable a greater level of support than either can provide when acting alone, and is expected to translate into more effective clinical outcome measures. This section presents details of the hybrid control scheme that will be employed to assist upper limb tasks using a combination of mechanical and FES modalities.

From [3], a dynamic model of the human arm with FES applied to a subset of joints is given by

$$B(\Phi(t))\ddot{\Phi}(t) + C(\Phi(t), \dot{\Phi}(t))\dot{\Phi}(t) + F(\Phi(t), \dot{\Phi}(t)) + G(\Phi(t)) = \tau(u(t), \Phi(t), \dot{\Phi}(t)) \quad (2)$$

where  $B(\cdot)$ ,  $C(\cdot)$  are inertial and Coriolis matrices respectively,  $F(\cdot)$  and  $G(\cdot)$  are friction and gravitational vectors respectively, and  $\Phi(t)$  contains the joint angles of the human arm. Vector  $\tau(u(t))$  comprises the moment produced through application of FES signal  $u(t) = [u_1(t), \dots, u_n(t)]^T$  to actuate  $n$  muscles, and  $\Phi(t)$  contains the anthropomorphic joint angles. Application of gravity balancing to this arm model produces a moment,  $v(a(t), \Phi(t))$ , about the anthropomorphic joint axes which is added to the right hand side. In addition, the dynamic properties of the mechanical support can also be incorporated into the terms  $B$  and  $C$  if deemed significant (see [3] for details).





**Fig. 9** Hybrid control scheme

The control task is to select gravity support parameter  $a(t)$ , and FES signal  $u(t)$  in order to assist the user in tracking a joint reference signal  $\hat{\Phi}(t)$ . To achieve this, consider the quadratic minimisation problem

$$\min_{u,a} J(u,a), \quad J(u,a) = \|\hat{\Phi} - \Phi\|^2 + \|a\|_{W_a}^2 + \|u\|_{W_u}^2 \quad (3)$$

in which  $\|x\|_W^2 = x^T W x$ , and weighting matrices  $W_a$ ,  $W_u$  are positive definite. Selection of FES signals  $u(t)$ ,  $a(t)$  to minimise (3) over the task duration can be computed using any iterative minimisation procedure. In the rehabilitation setting the user has multiple attempts at performing the task and hence (3) can be solved in an iterative manner using data collected over these attempts. Denoting each attempt with the subscript  $k$ , this naturally leads to an Iterative Learning Control (ILC) scheme, with an appropriate control scheme shown in Fig. 9. Note that explicit feedback and ILC structures are derived in [2, 3].

The support,  $\tau$ , provided by FES acts around each joint axis, while the gravity balancing support,  $v$ , acts against the gravity term  $G(\Phi)$ . In particular, if the system mass were concentrated solely at the end-effector, then  $a(t)$  can be stipulated such that  $G(\Phi) - v(a(t), \Phi(t)) = \beta(t)G(\Phi)$ ,  $\beta \in [0, 1]$ . Hence  $a(t)$  solely regulates support against gravity by acting on the term  $G(\Phi)$  while FES assists the remaining human arm dynamics. This effectively partitions the control system and prevents/suppresses interaction between controlled variables. Simpler control strategies are also possible by employing constraints within (3), for example if  $a_k(t) = a_k$  is fixed during each attempt then the level of gravity balancing is only updated between task attempts.

## 6 Conclusion and Future Work

The kinematics and the 3D CAD model of the device have been analysed. Then the dynamic analysis of the 3D CAD has been carried out and confirms satisfactorily dynamic behaviour with only minor oscillation of the arm support for realistic upper limb masses.

The next step for this research program is to complete further dynamic analyses and FEA of the 3D CAD model. Then the first prototype of the support will be manufactured and tested. To integrate the device with FES, parameters in the model

of the supported stimulated upper limb (2) will be identified, and the proposed optimal control strategy will be employed to implement a hybrid control system which controls FES and mechanical support in order to balance voluntary effort and performance to maximise therapeutic effect. Once the overall design is completed, the device will be tested with both unimpaired and neurologically impaired subjects to evaluate its performance and optimize its function.

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