Development of an Assistive Motorized Hip Orthosis

Kinematics Analysis and Mechanical Design

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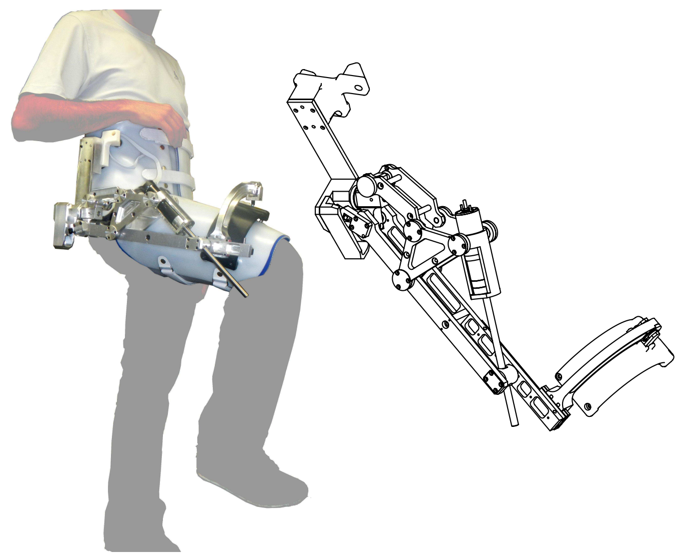
***Abstract*—With the increase of life expectancy, a higher number of elderly need assistance to maintain their mobility and their independance. The hip joint is crucial for walking and is problematic for a large number of aged people. In this paper we present a novel design of a motorized hip orthosis to assist elderly people while walking, stair climbing and during the sit-to-stand transistions. The kinematics was developed based on biomechanics considerations. To be able to achieve a large assistance rate, velocity and torques of the hip joint were studied from the literature. In order to fit with these requirements, an amplification mechanism inspired by excavators was developed and implemented. Comfort considerations were also taken into account and a custom interface was designed with the collaboration of a professional orthopaedic technician. First tests with the prototype showed that the workspace is sufficent for walking, for stair climbing as well as for sit-to-stand transitions. The assistance rate can go up to 30% for a 70 kg subject during walking at a cadence of 100 steps/min. The comfort is guaranteed despite the important weight (4.3 kg) of this first prototype.**

***Index Terms*—exoskeleton; elderly; hip**

I. Introduction

Decent mobility is crucial for elderly people, both from the physical point of view as well as for psychological aspects. More specifically, the aptitude to walk is one of the key points to be able to stay at home independently and to keep a fair physical condition [1]. Considering that the population of people aged 65 and more will increase from 7% in 2000 to 16% by 2050 [2], the need for devices to assist walking is then logically growing.

Exoskeletons have proven to be effective in several walk related situations such as assistance and rehabilitation for spinal cord injured patients or neurological diseases victims [3], [4], [5], [6]. Paraplegic, tetraplegic or stroke patients could therefore benefit from these devices. Studies with exoskeletons designed specifically for the elderly have been conducted too. For example, the EXPOS developed at Sogang University showed promising results [7]. The mechanical design of the EXPOS is different from most of other exoskeletons given that it is composed by a very light "tendon-driven exoskeleton" (less than 3kg in total) and by a walker which contains the heavy parts i.e. the actuators, the drivers, the controller and the batteries. Another interesting study was performed on elderly patients using a hip orthosis developed by Honda [8]. The idea was here to compensate for the redistribution of torques and powers during walking which happens with ageing [9], [10]. Actual improvements on the walking ability could be established.



(a) (b)

Fig. 1. (a) Photo of the motorized hip orthosis worn by a subject. (b) CAD model of the assistive orthosis. The amplification mechanism inspired by an excavator and actuated by a ball screw enables a large range of motion with a torque depending on the position.

This paper describes a new 6 degrees of freedom (DOF) orthosis to assist the movements of the hip in the sagittal plane while minimizing the effects on the other articulation rotations (see fig. 1). Only one DOF is actuated, the five others are passive. The presented mechanism enables assistance during walking but also during stair climbing or the sit-to-stand transitions. The later is crucial for mobility since elderly people often have difficulties during this phase. The targeted users are elderly people with reduced strength and because of that a reduced mobility. This device will be used to test the effects of single articulation assistive orthosis on elderly's walking, stair climbing and standing up capabilities. The design details of our orthosis are presented and its capabilities are assessed.

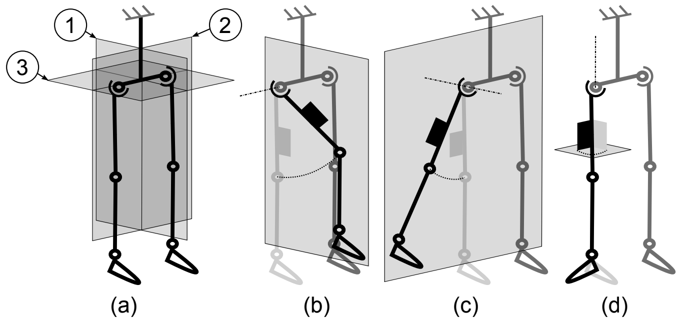


Fig. 2. Possible rotations of the hip joint. (a) Rotations can be in the sagittal plane (1), in the frontal plane (2) or in the transverse plane (3). (b) The flexion/extension is the rotation in the sagital plane. (c) Adduction/abduction is in the frontal plane. (d) Internal/external rotation is the movement around the axis of the leg. When there is no flexion/extension nor adduction/abduction, the internal/external rotation is in the transverse plane.

II. Design specifications

A. Kinematics

In our study we assume that the hip joint is well approximated by a spherical joint whose position is aligned with the head of the femur. Three rotations around this point can therefore be assumed for the design of the device kinematics (see fig. 2). Overconstraint in the system formed by the hip articulation and the mechanism is proscribed in order to limit parasitic forces and torques on the wearer which may lead to discomfort.

The orthosis must also be able to mimic correctly the movements of the user. According to Roaas et al. [11] the range of motion of the 3 rotations are usually assumed to be:

- flexion/extension: –10° to 120°
- adduction/abduction: –30° to 40°
- internal/external rotation: –35° to 35°

During walking these ranges can be diminished as presented in [12].

B. Torque and Dynamics

The rotational velocity of the hip during walking can also be deduced from [12]. Since the stride frequency is usually less than 1Hz [13] we can assume that the maximum rotational speed is about 140°/s with the angle varying between -15° and 30°. The exoskeleton must therefore be able to move at this speed in this area.

When standing up, most of the hip joint torque is acting in the sagittal plane (extension). During walking, this component of the torque is also prevailing (flexion/extension) [14]. An assistive orthosis should therefore focus on this component, ideally without constraining the other two. The torque peak value in the sagittal plane during level walking is usually assumed to be around 0.8 Nm/kg (standardized by the mass of the subject) [14]. When standing up, this value can go up to 1 Nm/kg when the hip flexion angle is around 70° [15]. The motorized orthosis does not need to provide this torque in its whole since it is only an assist device and the goal is not to take over the complete movement. Nevertheless to guarantee that a wide range of assistive rate can be provided, this order of magnitude needs to be respected.

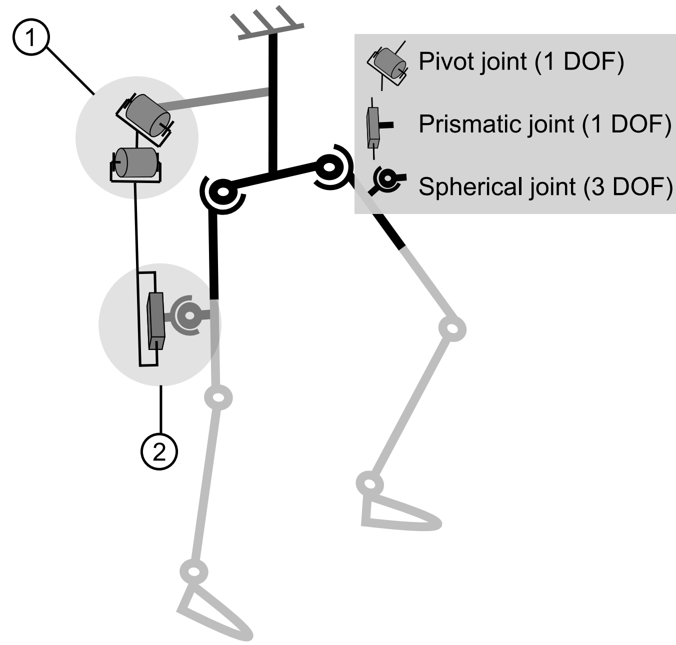


Fig. 3. Kinematics. The mechanism is attached to the pelvis (1) through two rotational joints (pivot joints with one DOF each) and to the thigh (2) by means of one prismatic joint (one DOF) and one spherical joint (three DOF).

C. Contact forces at interface with the user

As explained by Jarrasse et al. [16], large contact surfaces at the interface with the wearer are required to limit the tissue deformations which are unpleasant. The body tolerance to torques is also limited and pure forces are much better accepted.

III. Design implementation

A. Kinematics

The mechanism being placed in parallel with the body and therefore forming a kinematic loop, additional DOF are required. Indeed, since we want to keep the original mobility of the hip joint, the Chebychev-Grübler-Kutzbach criterion imposes:

$$M = 6 \cdot (N - 1 - j) + \sum_{i=1}^{j} f_i$$

where, M is the mobility of the system, N is the number of link (including the reference link, e.g. the pelvis), j is the number of joints and each joint possesses a freedom f_i. In our case, M, N and j are all equal to 3 and f_{hip} (f_i for the hip joint) is as well equal to 3. Therefore $f_{mechanism}$ (f_i for the two links of the mechanism, one attached to the pelvis and the other one to the thigh) must be equal to 6.

We decided to implement a design with two rotational DOF located at the pelvis junction. The four other DOF are composed by one prismatic joint and one spherical joint. These last joints are located at the connection with the thigh. The kinematics is presented in fig. 3. With this kinematics, the flexion/extension movement of the leg is directly linked with the angle of the second rotational joint in the kinematic chain of the mechanism. The internal rotation is made possible by the fact that two spherical joints are placed next to each other in the kinematic chain (one from the mechanism and one from the hip). The last rotation (i.e. the abduction/adduction) is more challenging to handle. Due to the chosen kinematics, the adduction/abduction is allowed only if the flexion angle is small. Indeed when the second rotational joint lifts (i.e. when the flexion angle increases), the axis of rotation of the leg rotates in the sagittal plane and the three axes are not orthogonal anymore. When the flexion angle is close to 90°, the thigh is aligned with the axis of the first pivot joint thus causing a singularity (see fig. 4 (a)). The first consequence is that one DOF of the leg gets locked (adduction/abduction). Notice that this effect does not appear on normal walking range but rather in sitting positions (flexion angle typically >45°). The second consequence is that the mechanism gains one DOF relatively to the wearer's body. This parasitic movement is not desired and therefore we designed a cam system (see fig. 4 (c)) to progressively lock it when the flexion angle gets greater than 30°. The actual range of motion of the first pivot joint as a function of the second joint is presented in fig. 4 (b). The shape of the cam is fairly complexe since the rotation that needs to be constrained depends on the second rotational joint. The movement of the cam follower therefore describes a curve in 3D. Fig. 4 (d) shows the movement of the cam follower during flexion with maximum permitted abduction.

B. Amplification mechanism

In order to assist efficiently both sit-to-stand and walking, a mechanism with a varying transmission ratio was designed (see fig. 5).

The maximal torque will be required during sit-to-stand when the flexion angle is between 70°and 80° [15]. The goal is to have a torque as large as possible in this area with a motion range and a velocity being able to fit with human gait. We developed a back drivable mechanism actuated by a 60W motor and a spindle drive from Maxon. The peak force provided by this actuator is 1 kN and its maximum speed is about 250 mm/s on a travel of 200 mm. In order to have a sufficient workspace with enough velocity and torque, a mechanism inspired by excavators was studied. The maximum velocities and the possible torques as function of the position are presented on fig. 6. It can be observed that the maximum torque is higher in the area that will be used during sit-to-stand movements. On the other hand a higher velocity can be reached in the walking range of motion.

C. Interface with the user

The interface with the user is ensured by an orthosis designed by a professional orthopaedic technician (see Fig. 7). The interface was molded on a subject in order to perfectly fit with the different body parts. Despite this customized design, the orthosis was tested on several subjects and showed an excellent comfort (yound and healthy males between 1.7 m to 2 m high). The contact with the user applies on large surfaces as required by our specifications. The proposed kinematics imposes only forces at the thigh connection which limits the discomfort. Indeed the last link is a ball joint which cannot transmit torques. Meanwhile the pelvis can be subject to torques. However the interface being very large, no skin deformation can be observed. Thanks to this interface, donning and doffing are relatively easy for an experimental device. Nevertheless, improvements would still be needed for enabling an elderly user to be autonomous with the orthosis.

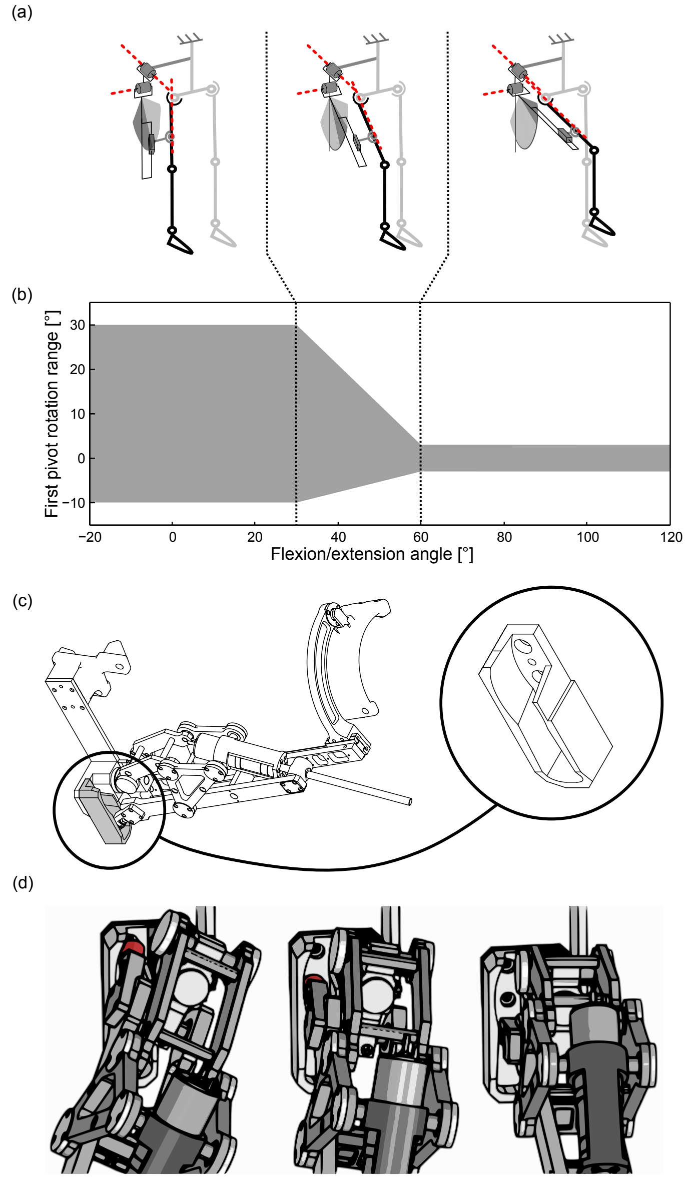


Fig. 4. Management of the adduction/abduction angle. (a) When the flexion angle increases, the leg tends to align itself with the first pivot joint. When the flexion angle is around 90°, there is a singularity. (b) To avoid the singularity, the angle of the first joint was constrained as shown in the graph. (c) To do so, we designed a cam system to prevent the rotation of the mechanism when the flexion angle increases. (d) The movement of the cam follower on the cam is shown at different flexion angles.

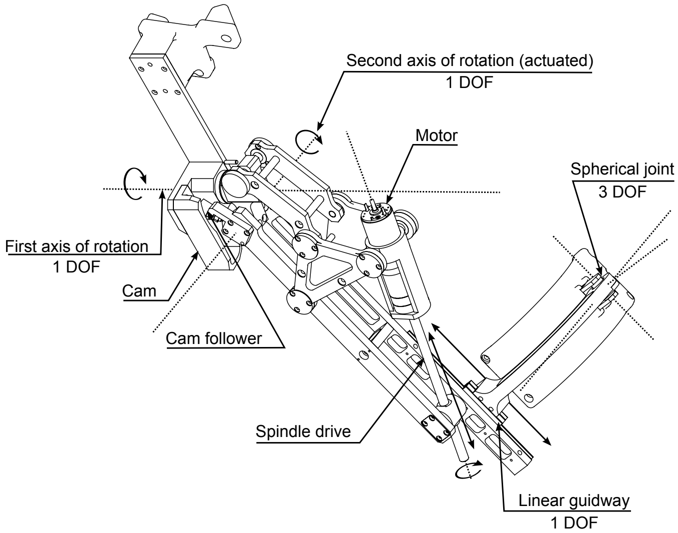


Fig. 5. The motor torque is amplified by means of the spindle drive and by a mechanism similar to the excavators.

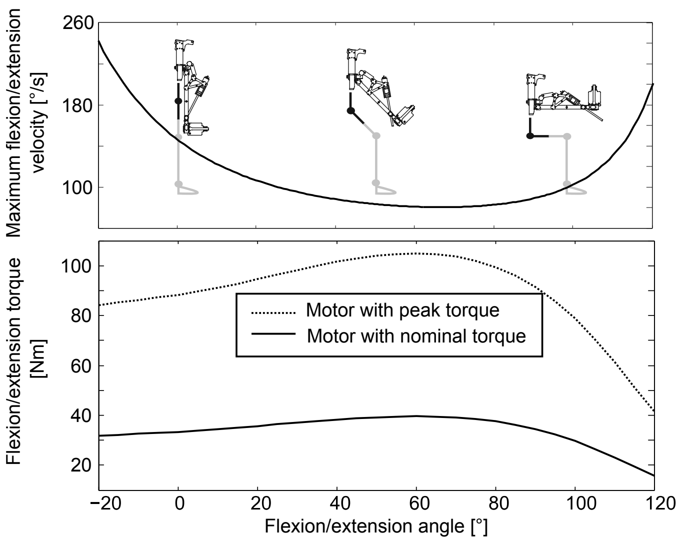


Fig. 6. Mechanism capabilities. The upper graph shows the maximum hip rotational velocity with the actuator recommended speed. The actuator is able to rotate at twice this speed. The lower graph shows the maxium torques at the hip with motor nominal and peak torques.

Fig. 7. Pelvis and thigh interfaces. These parts were molded on a subject by a orthopaedic technician.

IV. Testing protocol

To test our motorized orthosis, a typical flexion/extension trajectory (between –15° and 30°) is evaluated (see fig. 8(a)). Different cadences between 60 steps/min and 120 steps/min are considered. The flexion/extension angles correspond to typical young and healthy subject data. Therefore they are largely sufficent for being used with elderly people (notably with the higher frequencies).
Our pc-based real-time controller runs a control loop at 1 kHz. No load is applied on the orthosis during the experiment and the input torque is recorded. The amount of torque required to make the orthosis follow real walking trajectories can thus be recorded and compared to our model. Gravity, frictional, and dynamic effects are therefore quantified.

V. Results

With the implemented control, the orthosis follows precisely the input trajectory. The required torque is presented on fig. 8(b). We could compare our model with the real required torque and thus identify the friction effects. From this figure and from the motor specifications we can deduce that an additional torque of 42.1 mNm RMS can be provided at this frequency. Indeed the motor can produce a 85.6 mNm continuous torque. The additional torque the orthosis can provide typically represents 30% of assistance for a 70 kg person. The maximum required speed during the trajectory is 80% of the nominal motor speed (see fig. 8(c)). Other frequencies were also evaluated. The RMS torque during the trajectory was measured in order to test the limits of the orthosis. The results are presented on fig. 8(d).
The sit-to-stand transitions being performed in a short period of time a larger torque (about 2.5 times nominal torque) can be produced by the motor. By means of the high transmission ratio (maximum when the flexion angle is around 70°), 100% of the hip torque required to stand up can be provided.
The mechanism is therefore perfectly adjusted to demonstrate the effects of assistance at the hip level while walking as well as during sit-to-stand transitions. Nevertheless, since the device is carried by the user, the important weight due to the robust design could be a problem. Indeed, the device for the right leg (with the interface) weighs 4.3 kg.

VI. Conclusion and Future work

In this paper we have presented a motorized hip orthosis. Flexion/extention can be assisted while abduction/adduction and internal/external rotations are passive. Its design is based on kinematics and dynamics considerations developed from the literature. A substantial torque in the sagittal plane can be provided thanks to a variable transmission ratio. This ratio is maximal when the flexion angle of the hip is about 70°). This corresponds to the position where a maximal torque is required during a sit-to-stand transition. Therefore, an important rate of support is achievable for walking as well as for sit-to-stand which is of major importance for elderly assistance. A wide range of motion can be mimicked by the mechanism thus allowing the user

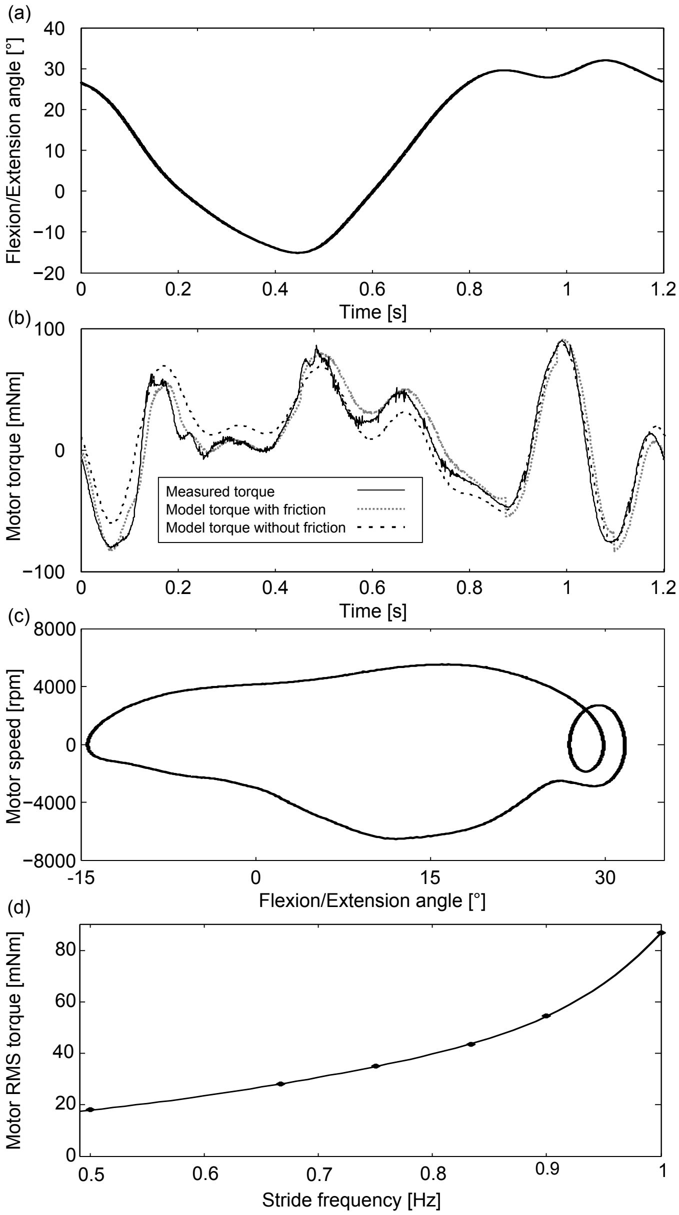


Fig. 8. (a) Typical walk trajectory we used for testing the mechanism capabilities. In this case the frequency is 0.83 Hz, which correspond to 100 steps/min. (b) The corresponding torque during the trajectoy was recorded and compared to our model. The torque RMS value is 43.5 mNm which corresponds to 50% of the nominal motor torque. (c) The motor speed during the trajectory is maximum 80% of its nominal value. (d) RMS value of the motor torque during the trajectory at different frequencies. The maximum frequency is around 1 Hz. In that case no assistance can be provided.

to move naturally during walking. The combination of flexion (more than 60°) with adduction/abduction is however constrained by a cam mechanism for practical reasons. First tests have shown that gravity, friction and dynamic effects are well compensated. First impressions while wearing the orthosis in a transparent mode (i.e. by compensating the gravity, friction and dynamical effects) are very encouraging despite the important weight of the device. Future work will focus on strategies to assist walking, stair climbing and sit-to-stand transitions. Tests on elderly subjects will be conducted in order to validate the usefulness of this device.

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