Development of a Human Assistive Robot to Support Hip Joint Movement During Sit-to-stand Using Non-linear Springs

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The literature concerning human assistive robots typically focuses on “wearable” devices, with the aim of reducing the muscular effort required of patients during movements. This paper describes the design of an orthosis for assisting patients during sit-to-stand (STS). The newly developed device generates a hip joint torque and reduces the muscle activity required of the wearer. The device makes use of non-linear springs called stiffness adjustable tendons (SATs), to simulate the behavior of human tendons, and to exploit their ability to store energy when in motion and to return it at a later time. A series elastic actuator (SEA) was adopted to create the device. A position reference is designed to realize an assist control without a force sensor. EMG sensors are used to verify the effective reduction of muscle activity required of the wearer during the STS.

Keywords: human assistive robots, series elastic actuator (SEA), sit-to-stand, non-linear springs

1. Introduction

Human assistive robots, i.e. systems with actuation capabilities that assist human motions, have been intensively developed in recent years. Mechatronic technologies play significant roles in applications that improve the quality of life. Recently, power assistive devices have been developed in the form of wearable robots, “exoskeletons” or “orthosis”, for assisting physically impaired people or for augmenting human power(1)-(3). Typically, the term “exoskeleton” is used to describe a device that augments the performance of a healthy wearer, whereas the term “orthosis” is typically used to describe a device to assist a person with a limb pathology.

The objective of this paper is to carry out a preliminary development of an orthosis for helping people during the STS, i.e. the movement of standing up from a chair to an upright posture (Fig. 1). In particular, the target is to realize a device that can generate a hip joint torque so that the human effort required during STS is greatly reduced. This has an immediate impact on the improvement of the quality of life for unhealthy subjects, as STS requires a peak joint torque which is greater than the one required for other movements, such as stair climbing or walking.

As an additional requirement, in the project developed, elastic elements in the connection between actuator and limb have been used. This is for taking advantage of their properties to store energy while in motion and make it available at a later time, thus reducing energy consumption and costs. The solution adopted for the orthosis presented here is to realize a series elastic actuator (SEA), i.e. a system in which a compliant element is placed between the gear train and driven load, to intentionally reduce the stiffness of the actuator(4)-(7). This paper utilizes non-linear springs(8) as a compliant element. Since the stiffness increases as the tension increases, the elastic element can simulate the behavior of human tendons.

A final design requirement, in order to keep the overall cost of the device low, is to develop a device that does not make use of force sensors. Therefore, position profile is decided in the aforementioned through analysis of STS, which is utilized as a position command to the developed power assist device for STS.

It is worth noticing that, being an explorative project, the device is tested only on healthy people, trying to identify strengths and weaknesses of the design.

In the following, Section 2 will briefly introduce the mechanical design of the orthosis. The analysis of the STS is presented in Section 3, in which some literature results on the hip force developed during STS will be combined with

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the measurements of the hip joint angle, in order to produce a desired force profile and, in turn, a non-linear spring displacement depending on the hip angle. Section 4 reports the experimental results obtained with assistive force limited up to 18% of the maximum required. Reduction of the muscular activity is evidenced in the EMG recordings. Section 5 presents some conclusions and gives some directions for future research.

2. Orthosis Design

2.1 Orthosis Design

Figure 2 shows the completed orthosis designed. The Series Elastic Actuator (SEA) consists of: a brushless motor, a planetary gearbox, an elastic element, aluminum components and a hip corset.

In detail, the brushless motor is a “Yaskawa Electric Corporation” SGMAS-01A2A41, the planetary gearbox of 1/33 gear ratio is a HPGP-14A-33-J6ABL by “Harmonic Drive Corporation”. The elastic element is composed of three non-linear springs in parallel, the aluminum components are designed by using the software SolidWorks and, finally, the hip corset is made by the company “Arizono Orthopedic Supplies Co. LTD.”.

The system is easy to wear thanks to the hip corset, which also provides a suitable support for the aluminum mechanical structure, the actuator, the gearbox, and the elastic elements. The aluminum components, constituting the rigid bodies of the SEA, are shown in details in Fig. 3.

Fig. 2. Human assistive robot designed

(a) (b) (c)

Fig. 3. Aluminum structure designed: a) frontal view, b) side view, c) top view

2.2 Working Principle

The working principle of the device is explained through explanation of each parts using Fig. 4. \( \tau \), \( \theta \), and \( r \) show torque, angle, and radius of pulleys, respectively. Subscripts \( M \), \( L \), and \( H \) show motor, load pulley, and hip, respectively. Relation between the motor and the load pulley is given by the gear ratio \( R = 1/33 \).

\[
\begin{align*}
\theta_L &= R \theta_M \quad \text{(1)} \\
\tau_L &= R^{-1} \tau_M \quad \text{(2)}
\end{align*}
\]

The length of the wire wrapped around the load pulley \( x \) and the force of the wire \( F_b \) are given by the radius of the load pulley \( r_L = 10 \text{ mm} \).

\[
\begin{align*}
x &= r_L \theta_L \quad \text{(3)} \\
F_b &= r_L^{-1} x \quad \text{(4)}
\end{align*}
\]

The motion of the wire is caused by not only the motion of the hip joint angle \( \theta_H \) but also the elongation of the springs \( l \). Here, \( \theta_H \) is the angle between the trunk and the thigh. When hip angle is moved by \( \theta_H \), the load pulley wraps the wire by the arc length of hip pulley \( r_H \theta_H \).

\[ x = r_H \theta_H + l \quad \text{(5)} \]

where \( r_H = 60 \text{ mm} \) is the radius of the hip pulley. Here, the elongation of the springs \( l \) also becomes the motion of the wire \( x \). The hip angle \( \theta_H \) can’t be detected because there is no encoder in the hip pulley. Moreover, \( \theta_H \) can’t be detected from the encoder in the motor because \( l \) affects \( x \) as shown in (5).

As shown in Fig. 4, the projection of \( F_b \) along the axis radial to the thigh \( F_r \) produces an assistive hip torque \( \tau_H \), to help the subject during the STS.

\[
\tau_H = B F_r = B F_b \sin \theta_b \quad \text{(6)}
\]

where \( B \) is the distance between the center of the hip pulley and the fix point where the force is applied. \( \theta_b \) is the angle of the direction of the force applied on the wire. Since \( \sin \theta_b = r_H/B \), \( \tau_H \) becomes as follows:

\[
\tau_H = r_H F_b \quad \text{(7)}
\]

2.3 Elastic Element

The elastic element shown in Fig. 5 is composed by three non-linear springs, called stiffness adjustable tendons (SATs), two aluminum bars, one fixed and one movable, and a wire that is connected to the moving bar and to the motor shaft by means of the load pulley. The maximum force for each spring is 100 N, so the maximum force applicable on the elastic element is limited to 300 N.

Figure 6 shows the characteristics of the non-linear springs.
The percentage of force error caused by the elongation error $\Delta l$ is dependent on the elongation of springs. Therefore, if the spring is linear, the equation becomes as follows

$$\frac{\Delta F_b}{F_b} = \frac{K\Delta l}{l} = \frac{\Delta l}{l}$$

where $a_1$, $a_2$, and $a_3$ are constants depending on characteristics of springs. Stiffness $K$ is derived by differentiating (8) on the elongation $l$.

$$K = \frac{\partial F_b}{\partial l} = a_1 e^{a_1l}$$

A force error $\Delta F_b$ caused by an elongation error $\Delta l$ is evaluated because force is controlled by elongation of springs.

$$\frac{\Delta F_b}{F_b} = \frac{K\Delta l}{F_b} = \frac{a_1 e^{a_1l} \Delta l}{a_1 e^{a_1l} + a_3}$$

If $a_1 e^{a_1l} \gg a_3$, it is approximated as

$$\frac{\Delta F_b}{F_b} \approx a_2 \Delta l$$

If the spring is linear, the equation becomes as follows

$$\frac{\Delta F_b}{F_b} = \frac{K\Delta l}{Kl} = \frac{\Delta l}{l}$$

The percentage of force error caused by the elongation error $\Delta l$ is dependent on the elongation of springs $l$ in the case of linear springs. For example, influence of $\Delta l$ is large when $l$ is small. On the other hand it is constant in the case of non-linear springs. Therefore, $\Delta F_b/F_b$ is not affected by working point of non-linear springs.

3. STS Analysis

3.1 Hip Angle Measurement Method  
In order to properly design the device to assist a user during STS, it is important to correlate the hip angle with the required torque. In turn, by knowing the hip joint torque necessary during STS, it is possible to find the corresponding force on the wire, generated by the actuator. In the orthosis developed, in order to keep its cost low, no hip joint angle sensors have been used, so an alternative method for measuring such angle has been developed. If the applied force to the springs is null, i.e. $F_b = 0$, the elongation of the springs $l = 0$. From (1), (3), and (5), the hip angle $\vartheta_H$ can be measured by the motor encoder $\vartheta_M$ if $F_b = 0$.

$$\vartheta_H = \frac{r_H}{r_L} \vartheta_M$$

The scheme of the force control is shown in Fig. 7. The objective of such force control is to keep the wire in tension during the movement without deforming the springs (i.e. with zero-elongation), thus not providing any help to the subject. To implement this control, the force reference is the minimum constant value necessary for keeping the wire in tension and it is compared with the force measured by a force sensor, directly mounted on the wire. The force sensor used in this development stage can measure a maximum force of 80 N and will be removed in the final version of the device, being unable to cope with the maximum assistive force required in STS.

The PI controller used in Fig. 7 is given by

$$F_r = K_p (\vartheta_H - \vartheta_M) + \frac{1}{T} \int (\vartheta_H - \vartheta_M) dt$$

The percentage of force error caused by the elongation error $\Delta l$ is dependent on the elongation of springs $l$ in the case of linear springs. For example, influence of $\Delta l$ is large when $l$ is small. On the other hand it is constant in the case of non-linear springs. Therefore, $\Delta F_b/F_b$ is not affected by working point of non-linear springs.

3.2 Hip Angle Measurement Result During STS
So, the procedure to find the hip joint angle profile vs. time makes use of a simple experiment, which consists of three steps: a calibration phase of 5 seconds to put the wire in tension, the Sit-To-Stand motion (which usually lasts within 5 seconds), and an upright posture for 2 seconds. The subject tries to reproduce the STS with the same speed and the same movement in all experiments.

Figure 8 shows the experimental results obtained by using the force control explained above during the STS. The hip angle profile obtained with the proposed control is in good accordance with those found in literature (e.g. in (9)). The circle shows the initial phase, when the subject starts to lean forward, approaching the trunk to the thigh, stretching the
elastica; for this reason, the system responds by rotating the motor counterclockwise, loosening the wire. Then, in correspondence to the reverse of motion, we have the so-called Seat-Off instant. After this point, in order to reach the final position, the subject moves the trunk away from the thigh and the control responds by turning the motor in clockwise wrapping the excess wire, so the position of the motor increases with approximately linear trend, until the subject reaches the conclusion of the movement. During the last two seconds of the experiment, the position measured by the encoder is constant, this is because the subject has concluded the movement, the elastic element is in tension, the force er- coder is constant, this is because the subject has concluded the movement. During the last two seconds of the experiment, the position measured by the encoder is constant, this is because the subject has concluded the movement, the elastic element is in tension, the force error is about zero, namely the motor does not have to produce torque thus it remains stationary in the final position.

**3.3 Hip Joint Torque** As for the hip joint torque produced by a subject during STS, it is important to observe that the available instrumentation does not allow an in-depth analysis about the sit-to-stand directly on the orthosis designed. For this reason, we resorted to the results reported in (10) about the same movement, in which the results are normalized to the height and weight of the subjects. In the paper, the joint torques during sit to stand for healthy subjects and people with Parkinson’s disease have been recorded.

According to the recordings in (10), the hip joint torque versus time during STS for healthy subjects, is as shown in Fig. 9. This curve is approximately equal at a triangle with the maximum value close to the Seat-Off instant. In this paper, only the positive torque, obtained by stretching the wire, is considered. In fact, the actuator used in the designed orthosis can generate only positive torques, so the, in the initial phase, when the hip joint torque is negative, the actuator will not be activated, keeping the commanded torque at zero.

Then, it is possible to consider the torque necessary during the STS with a triangular profile and a maximum value near at the Seat-Off instant (after two seconds from the start of the experiment) and with the base that considers the 5 seconds of movement (which is a typical duration of a seat-off motion in an unhealthy subject).

It is important to observe that to produce the hip torque \( \tau_H \), the only way is to generate a force on the wire \( F_b \). Once the mechanical parameter of the structure, as radius of the hip pulley \( r_H \), is known, it is possible to obtain the force necessary on the wire during STS from (7). Like the hip torque, such force can be approximated at a triangle function, with the maximum value at about 2 seconds, equal at 1.7 kN; this value considers the results obtained from the data found in literature (10), by considering a subject with a height of 1.73 m and the weight of 72 kg.

At this point, by combining the force and the corresponding hip angle of Fig. 8, it is possible to obtain directly the force with respect to the hip angle during the STS, by considering as start point the instant of Seat-Off. Figure 10 shows the result obtained. This figure represents the force developed by a human during STS, as a function of the hip angle. It also represents the maximum assistance required for helping unhealthy people during the STS. Here, the time scales of Fig. 8 and Fig. 9 are adjusted in order to synchronize the motion because Fig. 8 is for persons who needs assistance and Fig. 9 is from healthy subjects. Moreover, it is assumed that the hip joint torque shown in Fig. 9 is invariable for motion speed.

**4. Design and Implementation of the Supportive Action**

**4.1 Control Method and Assistance Amount** Given that the target of the project was the design of an orthosis with a limited cost and the minimum number of sensors, we developed a simplified control strategy, in order to obtain the desired assistive force during STS. In practice, the idea is to offset the position profile obtained with the zero-elongation control (Fig. 8) by a quantity that will generate, at each position, an elongation of the non-linear springs such that the desired torque is provided at the hip joint.

To test the idea, we need to compare the desired force profile with the actual one, but due to some hardware limitations, this could be done only at reduced levels of assistance. In fact, in the experimental setup developed, the maximum force on the wire applicable is limited by three non-linear springs, so that it can support a maximum of 300 N, that corresponds to 18% of the maximum force necessary. Then, in the following, we will show experiments, reporting the results obtained with an assistance limited at 18% of the maximum force.

**4.2 Position Reference Generation** Since Fig. 10 is force of the wire to generate all hip joint torque, the vertical axis is multiplied by assist ratio. Corresponding displacement of non-linear springs is obtained from the characteristics of the single non-linear spring shown in Fig. 6 to generate the force of the wire. Here, the force is tripled because three non-linear springs are in parallel. To generate the motor angle command, the displacement is added to the hip joint angle shown in Fig. 8. Using a polynomial fitting, the smooth motor angle command is obtained as shown in Fig. 11.

As shown in (11), the percentage of force error is proportional to the elongation error \( \Delta l \) and the proportional gain \( a_2 \).
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Fig. 11. Position pattern for the motor. The position without help is in blue, the position with help to 18% is in red

\[ \theta_{M}^{ref} = P(s) + PI(s) T_M \]

Fig. 12. Motor position controller during STS

is constant regardless of working point of non-linear springs. The assist force is increased and decreased if the motion of the person is late and advanced for the hip joint angle shown in Fig. 8, respectively. When linear springs are utilized, the increase ratio of assist force is varied depending on the elongation of springs \( l \) as shown in (12).

4.3 Reaction Force Observer  To test the effectiveness of the assistive action designed, the subject is asked to perform the STS, when the motor is under position control as shown in Fig. 12 and the reference is the position profile calculated.

The P and PI controllers are given by

\[ P(s) = 212, \quad PI(s) = 0.0035 + \frac{0.74}{s}. \]

Here, the position controller is designed so that satisfactory tracking for the position reference is realized. It is important to observe that in this case the force sensor was removed, so it was not possible to measure the force on the wire during the STS, obtained by implementing a position control.

For this reason, a Reaction Force Observer - RFOB has been implemented\(^\text{[1]}\). In order to estimate the force on the wire. Thanks to the RFOB, it is possible to obtain an estimate of the actual force on the wire. To test the effectiveness of the RFOB, we validated it by comparing its estimates with the actual force measurement, when the 5% assistance is implemented. The result is shown in Fig. 13, where the comparison of the measurement and the estimate confirms that the RFOB provides the actual force mostly during STS, i.e. from 1.5 sec to 4.5 sec. Therefore, the RFOB is utilized as a tension force during STS to evaluate assist force.

4.4 Assistance of 18% during STS  Using the RFOB, we estimated the assistive force provided by the proposed strategy, based on the modified position reference. As a result, Fig. 14 shows the force estimated with the RFOB during STS of 18% assistance. The force is not exactly the same as ideal one, Fig. 10, this is because there are two main issues to consider: i.e. the lack of synchronism between the subject and the control system (in fact, there is not a feedback signal that allows the system to obtain information on the state of the subject) and the involvement of the patient (in fact, he/she can arbitrarily apply different muscular effort in each experiment). Despite this, the signal has a triangle shape, with the maximum value near 300 N which corresponds to 18% assist of maximum hip joint torque. Moreover, the waveform is similar to the one of human hip joint torque shown in Fig. 9.

To validate the analysis done and to verify the actual decrease of the muscle activity of the subject during the STS, it is acquired the signal from an EMG sensor placed above the knee. The results by the EMG sensors are shown in Fig. 15. From this figure it is possible to observe a decrease of the muscle activity with the help by the device designed. The maximum value, for the iEMG signal, measured with the help by the device is about 0.05 V which corresponds at 20% of the maximum value for the blue line (0.25 V).

The EMG sensors allow to validate the analysis made so far. The position profile, used with the aim of helping the subject during the STS, produces the desired effect: the healthy
subject, produces less muscle effort during movement with the support of the device designed.

5. Conclusion and Future Research

This paper shows the wire driven hip joint assistive robot with the non-linear elastic element, which is designed for STS. From analysis of STS, position profile for the motor which drives the wire is decided to realize 18% assist of STS from the characteristics of the non-linear springs. Position control is performed by using the encoder in the motor. The results obtained show a decrease in muscle activity of the subject during the STS, when using the designed orthosis and using the proposed position-based control.

To obtain more help from the device, it is possible to increase the number of springs in parallel, or to change the elastic element used with one having greater strength and allowing a greater maximum applicable force. Another important issue to consider is the availability of a feedback signal, to know the status of the subject, also for safety reasons. In addition to these structural aspects, about the analysis done, it will be important to consider an instrumentation that allows detailed analysis of the STS directly on the device designed. Moreover, the effectiveness of the power assist device with the non-linear springs should be investigated.

References


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