Abstract
In this paper a robotic assistive device, aimed at assisting physically impaired individuals when rising from a sitting to a standing position, is presented. We investigate the standing up motion of aged people who requires to power support and typical standing up motion by nursing specialist. It has been shown that the contribution to the joint torques by the gravitational torques is dominant during sit-to-stand (STS) motion. To this end, a semi-active semi-passive assistive device is proposed. Design includes two parts: (a) Passive part uses auxiliary parallelograms to identify the center of mass of the system and appropriate springs connected to device to reduce the required joint torque. (b) Active part includes actuator with appropriate controller to provide the remaining joint torque. Using this device, users need insignificant force for STS movement; in particular, it improves the quality of life for individuals with difficulties in sitting down and standing up. This device is able to be portable and connected to a wheelchair.

Keywords: sagittal plane, rehabilitation robot, assistive device, STS movement,

Introduction
Instability and falling are among the most serious problems associated with aging. It has been stated that 28-35% of community dwelling people over the age of 65 years and 42-49% in people over the age of 75 years will experience at least one fall [1]. In addition to elderly people, physically impaired persons and paraplegics often have difficulty when rising to standing position. In addition a sit-to-stand movement requires muscle strength higher than that of other daily activities like walking, jogging or jumping. Therefore they need to use assistive device for daily tasks. These problems define necessity to studying rehabilitation devices for STS motion.

There are many rehabilitative devices designed and manufactured for STS motion. As an example a powered knee-orthosis intender for assisting the sit-to-stand task is presented in [2]. Both pneumatic muscles and dc motor actuators are used in this design study. This design reduces the required joint torque at the knees but not at the ankles and hip. It can reduce the required joint torque by half at slow and natural STS trials and by 2/3 at fast trials and not all the required joint torque. Another active assistive device is a walking-aid and STS assisting device to help impaired by three wheels and two linear actuators [1]. This device uses to rehabilitative exercises after surgery of lower limb. It balances the stability of patients during sit-to-stand transfer and walking. In another work, a number of researchers developed a force assistive device for standing up motion in which prevents the decreasing of physical strength by using the remaining physical strength of patients [3]. They use force control during lifting up the trunk and use position control in other phases of STS motion. In addition to powered orthosis, functional electrical stimulation (FES) has been employed by many investigators to artificially activate skeletal muscles and augment the muscles capability [4-9]. Especially FES has been used to help a paraplegic patient to stand up from a wheel chair [6]. In another research, Donaldson and Yu proposed a strategy to account for voluntary upper body effort in the control of FES-supported standing up [7]. One of the limitations of these completely active devices is that they have low safety in compared to passive ones and another limitation feature of these machines is that they move patients through predetermined movement patterns rather than allowing them to move under their own control. Hence, a passive gravity-balanced assistive device designed in which provides required torques by using springs and supporting during standing up [10]. Because of supporting weight and springs with high stiffness, this passive device is unable to be portable and connected to a wheelchair.

In this paper, a semi-passive semi-active assistive device is designed to take advantage of both active and passive devices using combinations of springs and actuators.

Modeling
A sagittal plane (normal walking plane) three link-segment model is used in study to model STS. Link segment model represents the human body as a set of rigid links connected by revolute joints on 2D plane as shown in Figure 1 with assumption of bilateral symmetry.
Three links $l_l, l_u, l_H$ represent lower leg, upper leg, and HAT (head, arm and trunk) segments of human body.

Figure 1: A 3-DOF planar model of the human body in STS motion

In this study, we analyze standing up motion for elderly people. The experimental time histories of joint angles $\theta_a$ (ankle), $\theta_k$ (knee) and $\theta_h$ (hip) for STS motion of an elderly subject in terms of percent of movement pattern are given in [3]. They can be approximated for a STS motion in 6 s, by the following polynomials:

$$\theta_a = 0.0160t^5 - 0.1578t^4 - 0.3886t^3 + 9.1207t^2 - 34.6563t + 50.0657t^2 - 35.6012t - 60.4066$$

$$\theta_k = 0.0113t^5 - 0.2931t^4 + 2.8716t^3 - 13.4103t^2 + 31.2381t - 38.1200t^2 + 23.7712t + 100.4604$$

$$\theta_h = 0.0062t^5 - 0.1266t^4 + 0.9992t^3 - 3.7204t^2 + 5.9573t^3 + 0.0037t^2 - 12.6853t + 70.3759$$

The joint angles, the joint angular velocities and the joint angular accelerations of these polynomials are shown in Figures 2-4. The time for STS motion for an old person is between 5 s to 6 s. We simulate the STS motion in 6 s; the simulation results are considered during 0.5-5.5 s where the joint accelerations are not too high.

Figure 2: Joint angles for standing up motion $\theta_a$ (ankle), $\theta_k$ (knee), $\theta_h$ (hip)

Figure 3: Joint angular velocities for standing up motion $\dot{\theta}_a$ (ankle), $\dot{\theta}_k$ (knee), $\dot{\theta}_h$ (hip)

Dynamic of model during STS motion

The joint torques at the ankle, knee and hip joints as written in Eq. (4) have the following components during motion: (1) an inertial torque due to the inertial forces of the segments ($\tau_i$), (2) a passive elastic torque due to stiffness of the muscles ($\tau_e$), (3) a gravitational torque due to gravity ($\tau_g$).

$$\tau_a = \tau_{ag} + \tau_{ai} + \tau_{ae}$$
$$\tau_k = \tau_{kg} + \tau_{ki} + \tau_{ke}$$
$$\tau_h = \tau_{hh} + \tau_{hi} + \tau_{he}$$

Figure 5: Dynamic model of the human body

Figure 5 shows a three link planar model used in deriving the dynamic equations of human body for this motion. Here, $l_l, l_u, l_H$ represent the location of center of mass of lower leg, upper leg and HAT segments and $I_l, I_u, I_H$ are the moments of inertia of each segments where all expressed with respect to their center of mass.

Half of the upper body mass is considered for the mass of HAT ($m_H$). $m_l$ and $m_u$ are the mass of lower and upper leg.

Inertial and gravitational joint torques can be determined using the dynamic model. Lagrange equations for this model are:

$$\frac{d}{dt} \left( \frac{\partial T}{\partial \dot{\theta}} \right) - \frac{\partial T}{\partial \theta} + \frac{\partial V}{\partial \dot{\theta}} = Q$$

$$T = \frac{1}{2} \sum_{i=1}^{N} (m_i v_i^2 + I_i \omega_i^2) = \frac{1}{2} \dot{\theta}^T J(\theta) \dot{\theta}$$
\[ \omega_t = \dot{\theta}_t, \omega_z = \dot{\theta}_z, \omega_{\bar{t}} = \dot{\theta}_{\bar{t}} + \dot{\theta}_{\bar{z}} \]

\[ v_t = \dot{\theta}_t e_t \]

\[ v_z = \dot{\theta}_z e_z \]

\[ v_{\bar{t}} = \dot{\theta}_{\bar{t}} e_{\bar{t}} + l_{\bar{t}} \ddot{\theta}_{\bar{t}} + \dot{\theta}_{\bar{t}} e_{\bar{t}} \]

\[ I(\theta)\ddot{\theta} + l(\theta, \dot{\theta})\dot{\theta} - \frac{1}{2} \frac{\partial^2 V}{\partial \theta^2} + \frac{\partial V}{\partial \theta} = \begin{bmatrix} \tau_1 \\ \tau_2 \\ \tau_3 \\ \tau_4 \end{bmatrix} \]

Using Lagrange equations, a closed-form system of equations for three-link planar model, in state-space representation [12], was developed as

\[
\begin{bmatrix}
\tau_{\text{a}} \\
\tau_{\text{g}} \\
\tau_{\text{p}} \\
\tau_{\text{g}}
\end{bmatrix} = \begin{bmatrix}
I_1 & I_{12} & I_{13} & 0 \\
I_{12} & I_{22} & I_{23} & 0 \\
I_{13} & I_{23} & I_{33} & 0 \\
0 & 0 & 0 & 0
\end{bmatrix} \begin{bmatrix}
\dot{\theta}_1 \\
\dot{\theta}_2 \\
\dot{\theta}_3 \\
\dot{\theta}_4
\end{bmatrix} + \begin{bmatrix}
\nu(\theta, \dot{\theta})_1 \\
\nu(\theta, \dot{\theta})_2 \\
\nu(\theta, \dot{\theta})_3 \\
\nu(\theta, \dot{\theta})_4
\end{bmatrix} + \begin{bmatrix}
\tau_{\text{a}} \\
\tau_{\text{g}} \\
\tau_{\text{p}} \\
\tau_{\text{g}}
\end{bmatrix}
\]

\[ I_{11} = I_1 + I_2 + I_3 + m_1 l_1^2 + m_2 (l_2^2 + l_2 l_3 + l_1 l_3) + m_3 l_3^2 + 2 m_2 l_2 l_3 + l_1 l_2 c_{\theta_2} + l_1 l_3 c_{\theta_2}
\]

\[ I_{12} = I_1 + I_2 + m_2 l_2^2 + m_2 (2 l_3 c_{\theta_2} + l_2 c_{\theta_2}) + m_3 l_3^2 + 2 l_2 c_{\theta_2} + l_1 l_3 c_{\theta_2} + 2 l_2 c_{\theta_2} / 2
\]

\[ I_{13} = I_1 + m_2 l_2 l_3 c_{\theta_2} + m_2 l_2 l_3 c_{\theta_2} / 2
\]

\[
\begin{align*}
I_{22} &= I_1 + I_2 + m_1 l_1^2 + m_2 (l_2^2 + l_2 l_3 + l_1 l_3) / 2 \\
I_{23} &= I_1 + I_2 + m_2 l_2 l_3 / 2 + l_1 l_3 c_{\theta_2} \\
I_{33} &= I_1 + m_2 l_2 l_3 c_{\theta_2} / 2
\end{align*}
\]

\[ v(\theta, \dot{\theta})_3 = -(m_1 l_1 l_3 + m_2 l_2 c_{\theta_2} + m_3 l_3 c_{\theta_2}) \dot{\theta}_2 + \frac{\ddot{\theta}_3}{2} - m_1 l_1 l_3 c_{\theta_2} \dot{\theta}_2 - m_2 l_2 l_3 c_{\theta_2} \dot{\theta}_2
\]

\[ v(\theta, \dot{\theta})_4 = -(m_1 l_1 l_3 + m_2 l_2 c_{\theta_2} + m_3 l_3 c_{\theta_2}) \dot{\theta}_2 + \frac{\ddot{\theta}_4}{2} + m_1 l_1 l_3 c_{\theta_2} \dot{\theta}_2 + m_2 l_2 l_3 c_{\theta_2} \dot{\theta}_2
\]

\[ \nu(\theta, \dot{\theta})_1 = m_1 l_1 l_3 \dot{\theta}_2 + \frac{\ddot{\theta}_1}{2} - m_1 l_1 l_3 \dot{\theta}_2 - m_2 l_2 l_3 \dot{\theta}_2
\]

\[ \nu(\theta, \dot{\theta})_2 = m_2 l_2 l_3 \dot{\theta}_2 + \frac{\ddot{\theta}_2}{2} - m_2 l_2 l_3 \dot{\theta}_2 - m_2 l_2 l_3 \dot{\theta}_2
\]

\[ \nu(\theta, \dot{\theta})_3 = m_2 l_2 l_3 \dot{\theta}_2 + \frac{\ddot{\theta}_3}{2} - m_2 l_2 l_3 \dot{\theta}_2 - m_2 l_2 l_3 \dot{\theta}_2
\]

The passive elastic joint torques due to the muscles is approximated using exponential forms presented in Eq. (11)-(13) [11].

\[ \tau_{\text{a}} = \exp(2.0111 - 0.0833(\theta_2 - 90) - 0.00900\dot{\theta}_2)
\]

\[ \exp(-9.9250 + 0.2132(\theta_2 - 90)) - 1.970\]

\[ \tau_{\text{g}} = \exp(1.0372 + 0.0400(\theta_2 - 90) - 0.0494\dot{\theta}_2)
\]

\[ \exp(-1.1561 - 0.002(\theta_2 - 90) + 0.0254\dot{\theta}_2 + 0.0036\dot{\theta}_2) \exp(2.5 - 0.25\dot{\theta}_2) + 1\]

\[ \tau_{\text{p}} = \exp(2.1080 - 0.0160\dot{\theta}_2 - 0.0195\dot{\theta}_2)
\]

\[ \exp(-2.1784 + 0.070\dot{\theta}_2 + 0.01349\dot{\theta}_2) - 15.24\]

It may be noted that the joint angles and their time derivatives presented in Figure 2-4 are used to derive the inverse dynamics.

Using the geometric and inertia parameters of a normal subject with 168 cm height and 64 Kg weight presented in Table 1 [13], the components of the ankle, knee and hip joint torques for the abovementioned subject are derived and depicted in Figure 6-8. As shown, passive elastic joint torques are low and not important during motion especially during slow speed. Since STS motion is considered as a slow motion, the gravitational joint torques is the most dominant in STS motion, as shown in these Figures.
Table 1: The geometric and inertia parameters for a normal subject

<table>
<thead>
<tr>
<th>Link</th>
<th>Length(m)</th>
<th>Mass(Kg)</th>
<th>C.M(m)</th>
<th>Moment of inertia(Kg/m²)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Lower limb</td>
<td>0.41</td>
<td>3.9</td>
<td>0.18</td>
<td>3.3317</td>
</tr>
<tr>
<td>Upper limb</td>
<td>0.41</td>
<td>6.4</td>
<td>0.18</td>
<td>0.0352</td>
</tr>
<tr>
<td>HAT</td>
<td>0.79</td>
<td>21.7</td>
<td>0.49</td>
<td>0.1122</td>
</tr>
</tbody>
</table>

**Assistive device designs**

Consider two designs for this assistive device: (a) Two springs and two actuators for every leg, (b) Three springs and one actuator for every leg.

The second design is described in this section by using the following steps. The first design is also obtained by similar procedures. However, for the first design, it is assumed that \( K_i = 0 \) and thereby two actuators are required for remaining joint torques at the hip and knee.

1-Locating the center-of-mass (COM)

The location of COM for the three-link of the human body is shown in Figure 9. The parameters \( u, l, d \) and \( H \) are used to form parallelograms in order to identify the COM [14]. These parameters are obtained by equating the following two equations representing the location of the system COM.

\[
r_{oc} = m_i l_i e_i + m_a (l_a e_a + l_v e_v) + m_{hs} (l_h e_h + l_v e_v) + l_v e_v + l_h e_h + l_v e_v + l_h e_h
\]

2-Spring selection

Attaching three springs to the system as shown in Figure 9, the total potential energy of the system consist of gravitational and elastic energies due to the springs. Its expression is given as

\[
V = \frac{1}{2} K_1 x_1^2 + \frac{1}{2} K_2 x_2^2 + \frac{1}{2} K_3 x_3^2 - Mg\cdot r_{oc}
\]

where, \( x_1 = \overrightarrow{C\cdot C} \), \( x_2 = \overrightarrow{C\cdot C} \), \( x_3 = \overrightarrow{C\cdot O} \)

\[
V = \frac{1}{2} K_1 x_1^2 + \frac{1}{2} K_2 x_2^2 + \frac{1}{2} K_3 x_3^2 - Mg\cdot r_{oc}
\]

1. \( V = \frac{1}{2} K_1 x_1^2 + \frac{1}{2} K_2 x_2^2 + \frac{1}{2} K_3 x_3^2 - Mg\cdot r_{oc} \)

With the use of Eq. (15), the gravitational joint torques are derived as:

\[
\tau_{oc} = \frac{\partial V}{\partial \theta} = (-K_1 d_i - K_2 (l_i - d_i))d_{hs} \frac{d}{d_h} \frac{d}{d_h}
\]

\[
\tau_{sp} = \frac{\partial V}{\partial \theta} = (K_1 d_i - K_2 (l_i - d_i))d_{hs} \frac{d}{d_h} \frac{d}{d_h}
\]

On setting Eqs. (18) and (20) to zero, the springs stiffness are selected as

\[
K_i = \frac{Mg}{d}
\]

Using these springs, the amount of knee joint torque, i.e., Eq. (19) is also reduced. In this analysis, it is assumed that the undeformed length of spring is zero. Nonzero free length spring can be used behind the pulley where the spring force can be transmitted through the strings.

As an example, the parameters of a normal subject given in Table 1 are used to design the assistive device. By inserting these parameters into Eq. (14) and the results obtained into Eq. (21), we obtain:

\[
d_i = 0.3826(m), d_a = 0.3073(m), d_{hs} = 0.3252(m)
\]

\[
K_1 = 0.356(kN/m), K_2 = 1.066(kN/m), K_3 = 4.970(kN/m)
\]

3-Using actuator

The remaining joint torque in the knee, shown in Figure 10, is provided by the use of an actuator with an appropriate controller.
Based on the required joint torque at the knee, maxon DC motor, model RE36-118798, has been selected to meet the basic requirement of STS motion in the knee for our design.

Model parameters of the motor are as follows: rotor inertia $I_r = 69.9 \text{ gcm}^2$, electrical inductance $L_e = 0.2 \text{ mH}$, electrical resistance $R_e = 1.11 \Omega$, torque constant $T_K = 36.4 \text{ AmNm}$, and the speed constant $K_m = 263 \text{ rpm/V}$.

The standard model for a dc motor can be given as

$$ J_m \ddot{\theta} + B_m \dot{\theta} = r \tau_i $$

where $J_m$ and $B_m$ are actuator inertia and damping ratio of the mechanical component of the motor. Also, $\tau_i$ is disturbance torque. The torque generated by the motor can be predicted using motor parameters and Eq. (23).

### 4-Controller design

In this paper, the controller that we apply on the actuator in knee is PD controller based on separated joint control method. In this method dynamic of system is considered as load effected on motor and our objective is the control of angular performance of the knee joint.

Figure 11 shows the block diagram for this control method.

According to this block diagram $\theta_m$ is obtained from Eq. (24) as:

$$ \theta_m(t) = \frac{k_e}{r} \left( \frac{k_p}{r} \theta_d(t) + \left( k_p + k_d \right) \dot{\theta}_d(t) + k_d \ddot{\theta}_d(t) \right) $$

where $k_e = \frac{k_m}{R_e}$ and $k_p$, $k_d$ are proportional and derivative gains and $\dot{\theta}_m$ is angular velocity of the motor shaft.

We can express Eq. (24) in time space as follows:

$$ J_m \ddot{\theta}_m + (B_m + k_d \dot{\theta}_d) \ddot{\theta}_m + k_p \theta_m(t) = \left( k_p + k_d \right) \dot{\theta}_d(t) + k_d \ddot{\theta}_d(t) - r \tau_\theta(t) $$

Moreover, the relation between motor angle and joint angle according to reduced ratio of gear head that is equal to $r = 0.01$ can be written as:

$$ \theta_m(t) = \frac{\theta_d(t)}{r} $$

By substitution of Eq. (26) in Eq. (25), expression of Eq. (24) based on angle of knee joint and its time derivatives are obtained as:

$$ J_m \ddot{\theta}_m + \left( B_m + k_d \dot{\theta}_d \right) \ddot{\theta}_m + k_p \theta_m(t) = \left( k_p + k_d \right) \dot{\theta}_d(t) + k_d \ddot{\theta}_d(t) - r^2 \tau_\theta(t) $$

where $\theta_d(t)$ and its time derivatives are actual data of knee joint and desired trajectory of knee joint $\theta_d(t)$ is a prescribed function of time.

Using this equation, we can derive actual trajectory of angle of knee joint under effect of designed PD controller and simulate the motion.

According to Eq. (26), less value of reduction ratio of gear head results in better performance of control system by reducing the effect of disturbance and dynamic load.

Results of joint angle and angular velocity for knee joint using controller are given in Figure (12), (13). As shown, the actual trajectory of joint angle and angular joint velocity are in a very good agreement with the desired ones.
Feasible designs

Results of two designs, explained in previous section, are shown in Table 2. As shown, second design is better than the first one because it uses one actuator instead of two ones and both of them use springs with the same stiffness.

Table 2: Result of two selections for design

<table>
<thead>
<tr>
<th>2 springs, 2 actuators</th>
<th>$K_1 = 0.356\text{ (N/m)}$</th>
<th>$K_2 = 4.970\text{ (N/m)}$</th>
</tr>
</thead>
<tbody>
<tr>
<td>3 springs, 1 actuator</td>
<td>$K_1 = 0.356\text{ (N/m)}$</td>
<td>$K_2 = 4.970\text{ (N/m)}$</td>
</tr>
</tbody>
</table>

Conclusions

In this paper, we presented a semi active semi passive device to help elderly and impaired subjects during STS motion. The results showed that this device can be portable and is capable of connecting to wheelchair because of its lightness and low stiffness springs and also motor and its gearbox are light. Impaired subjects can use it with insignificant forces during STS motion. The fabrication of this design will be the subject of future work.

References