A Novel Variable Impedance Compact Compliant Ankle Robot for Overground Gait Rehabilitation and Assistance

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Abstract

This paper presents the modular design and control of a novel variable impedance compact compliant wearable ankle robot (powered Ankle-Foot Orthosis, AFO) for overground gait rehabilitation and assistance mainly for stroke patients. Design of AFO, its construction, kinematics, working principle, actuation, control etc. are described. A novel variable impedance compact compliant series elastic actuator (SEA) is designed to actuate the AFO with variable impedance and compliance. The actuator design consists of a servomotor, a ball screw, a torsional spring connecting the motor and the ball screw via a pair of spur gear, and a set of translational springs connecting the ball screw nut to the output link. Two types of springs of the actuator with different stiffness provide variable impedance and compliance to the AFO based on the range of operational force. The translational springs with low stiffness are used to handle the low force operation that reduces friction, impedance and impact. The torsional spring, being in the high speed range, has high effective stiffness and improves bandwidth in large force operation when the translational springs are fully compressed. Application of the variable impedance compact compliant actuator makes the AFO compact, and it may enhance user-friendliness. Simulation results confirm the effectiveness of the AFO design. Competitive advantages of AFO over its existing counterparts are described and its future extensions are discussed.

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Keywords: Ankle-Foot Orthosis; Gait Rehabilitation and Assistance; Series Elastic Actuator; Variable Impedance; Force Control; Modular Design

Nomenclature

\begin{itemize}
  \item \(a\) shank length
  \item \(b\) foot length
  \item \(x\) linear displacement of actuator output
  \item \(F_1\) motor input force
  \item \(F_2\) output force due to deflection in spring
  \item \(J_1\) moment of inertia of the motor
  \item \(J_2\) moment of inertia of the ball screw
  \item \(m_1\) equivalent mass of the motor
  \item \(m_2\) equivalent mass of the ball screw
  \item \(k_1\) equivalent translational spring constant of the torsional spring
  \item \(k_2\) spring constant of the translational spring
  \item \(b_1\) viscous damping for motor
  \item \(b_2\) viscous damping for ball screw
  \item \(\theta\) angular displacement between foot and shank
\end{itemize}

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1. Introduction

According to the world health organization (WHO), cardiovascular disease is the leading cause of death globally and is projected to remain so. On average, more than 17.1 million people die a year from cardiovascular diseases, representing 29% of all global deaths and occurring almost equally in men and women. Of these deaths, over 5.7 million occur due to stroke. These diseases have no geographic, gender or socio-economic boundaries and the number will increase to about 24 million by 2030, and the largest increase in number of deaths will occur in the south-east Asia region [1].

Stroke is the leading cause of permanent disability in the USA. Over 700,000 Americans per year suffer a stroke, and more than half of them survive with disability [2]. A similar picture is also seen for Europe, Australia and Japan. Each year, there are over 920,000 stroke cases in Europe [3]. Stroke is Australia’s second single greatest cause of death [4], accounting for about 10 percent of all deaths. It is the leading cause of long-term disability in adults, representing 25 per cent of all chronic disability [5]. Despite decades of declining mortality from stroke since the 1960s, stroke remains the third most common cause of death in Japan. On average, 1 in 5 men and women of middle age has risk of experiencing stroke in their remaining lifetime in Japan [6]. According to WHO, the old people (people over 65 years old) will increase by 88% for coming years, and the old people are particularly prone to suffer stroke disease [2]. Stroke disease causes death and disabilities, and increases health care anxieties, rehabilitation and health care costs.

Physical or occupational therapy is the primary means of rehabilitation and recovery for people suffered from stroke. Stroke damages neurons that can only be cured by rehabilitation therapies where modern medicine sometimes cannot regenerate stroke patient’s neurons. It has been proven that stroke victims receiving frequent physical therapy have much greater chance of recovery than other forms of treatments[7]-[8]. Human brain is capable of self-organization or plasticity that offers an opportunity for motor recovery. Repeated physical therapy can establish new neural pathways or unmask dormant pathways that control volitional movement, and thus can maximize motor performances and minimize functional deficits [9].

Generally, for manual physical therapy, one or more therapists interact physically with the stroke patient, and assist and encourage the patient to follow a number of repetitive exercises. However, manual rehabilitation is not precise, it is non-reproducible, it is slow, and it adds burdens to human therapists. The repetitive nature of therapy makes it possible to provide the therapy by properly designed robotic devices. Rehabilitation robots may reduce the burden of human therapists or even may replace the therapists, can measure data on patients while training them and thus help record patient progress, it is highly reproducible, precise, ensures consistency in therapy activities and thus can improve the therapy effect on the patients, which is better than that of the manual therapy [10]. This is why, robot-assisted rehabilitation or robot-aided orthopaedic or neurological therapies for stroke patients and other patients suffering from paralysis, paraplegic disease, cerebral palsy, hemiplegic problems, polio etc. have recently become very active areas of research [10]-[18].

Rehabilitation robots for both upper limb (arm) training [17]-[18] and gait training [10]-[16] are being developed. The upper limb systems seem to be easier. However, the gait training seems to be more important, but it is complicated and researches on these systems have not got much attention yet. Again, most of the currently available lower limb gait rehabilitation systems are set on treadmill [13], which is not movable and natural. The systems usually have hip and knee joints, but there is no actuated ankle joint [10], except in [12], which is not so practical. These systems are also not so robust to adapt with changing environments. The existing systems are expensive, large, heavy and most of them do not include body-weight supports. Balance is also not so good for the existing systems, and these systems do not have good appearance. Sizes of the existing systems are not generalized to accommodate patients of different sizes, ages etc. Most of the systems do not carry the therapist when serving the patients. The interfaces with the patient of the currently available systems are also not so convenient and the motion output methods and mechanisms are also not so good. Patient’s up-down, front-back and left-right motions are important. But, most of the currently available systems cannot provide these motions. Again, most of the lower limb rehabilitation systems are still not commercially available and could not produce social impacts due to incompleteness and impracticality in their design [10]-[18].

Integrated robot-aids are usually larger than the particular requirement of the patient and hence some of the facilities may remain unused, these aids are heavy and costly etc., they may create local effects on the unexercised limbs or muscle groups etc. These problems with the integrated robot-aids motivate towards the modular design where different groups of muscles (e.g. spatial arm movements), limbs (e.g. wrist, fingers, legs) etc. are exercised separately using specific robotic devices [2], [9]. However, most of the currently available systems do not have modular design advantages [10]-[18].

Among different modules, the ankle seems to be the most important one because the ankle plays a central role in normal gait, contributes to propulsion, shock-absorption and balance in normal gait, the foot-drop disease due the stroke occurs in the ankle etc.[9]. However, the present integrated robot-aids either do not include powered ankle-foot orthosis or the orthosis is on treadmill instead of over ground. Again, the ankle robots should have low friction, low and variable mechanical impedance, they should be back drivable etc. to resemble their natural counterparts. However, the present ankle robots do not satisfy these requirements entirely.

Hence, this paper presents the novel design of a compact and modular powered Ankle-Foot Orthosis (AFO). In this paper, the design, construction, kinematics, actuation system, control mechanisms etc. of the AFO are analysed. The
novelties and competitive advantages of the model over its existing counterparts are explained, and future extensions of the model are discussed.

2. Compliant actuation system for the AFO

Even few years ago, robotic actuation was designed using heavy actuation devices with stiff position/velocity and torque control methods. The transmission mechanisms were also rigid and non-back-drivable [19]. These robotic systems may provide good precision, speed, repeatability etc., and thus may be good for conventional heavy industrial autonomous applications where the working environments are usually well defined and the systems have less or no interactions with humans [19]. However, industry is now shifting towards automated production of small batch, customized, and short-life-cycle products. Domestic uses of robots are also increasing, and in both industrial and household setups, the robots need to adjust with versatile unstructured environments as well as with human demands [20]. However, as the robots and the humans come closer, the safety and human-friendliness issues become more important. In the industry, sometimes the interactions and coexistences between humans and robots are ignored by not allowing humans and robots share the same workspace. However, these barriers between humans and robots have already been started to decline with the turn of the 21st century because in the near future many aspects of our lives and activities will need to be encompassed by tasks performed in cooperation with robots and humans [21].

Stiff actuation is not suitable for unstructured environments and for the workplaces where human’s safety is a vital issue such as the AFO. Robot researchers have identified the traditional stiff actuation methods as one of the main hindrances in the way of safe and friendly human-robot systems [19]. It has now become clearly evident that the stiff actuation methods have limitations related to efficiency, controllability, sensing, stability, back-drivability, adaptation to changing situations, impedance adjustment, inertia etc. that are the reasons behind the unsafe and unfriendly interactions with human users [19]. One way of increasing the safety level of stiff actuation is to use some special types of software techniques based on impedance/admittance regulation and joint torque control [22]-[23]. However, delays at all stages of the systems have made this effort difficult.

On the other hand, natural systems especially humans are robust as they follow variable impedance, soft and compliant actuation [24]. It is believed that the actuation methods developed for rehabilitation robots mimicking their human counterparts may match the motion, safety and energy efficiency performances of humans and thus may provide better cooperation to their human partners in human-robot cooperation tasks [25]. However, robot’s competence in these areas is still inferior to that of their biological/human counterparts [26]. The above situations have driven the need of research on variable impedance, soft and compliant actuators for safe and human-friendly rehabilitation robotics applications [27]. It was expected that soft and compliant actuators might improve control, sensing, safety etc. of the rehabilitation robots and make the robotic systems behave like natural systems [26]-[27].

To address the aforementioned issues, a wide range of novel variable impedance compliant actuation systems have been developed [28]-[30]. Impedance was modulated by modulating stiffness, damping and inertia. Low stiffness increases safety and the stiffness modulation is achieved by the means of force/position feedback, improvement in hardware such as redundancy in degrees of freedom, improving kinetics, adding extra actuator or spring, using pneumatic or artificial muscles etc. The series elastic actuators (SEAs) are the early development towards the realization of actuators with inherent compliance [26], [31], [32]. However, the SEAs still have many limitations that make them unsuitable for use in the AFO. A novel design concept for SEAs is necessary for the AFO so that the design could overcome the existing limitations of the SEAs and provide required variable impedance and compliance to the AFO.

3. Series elastic actuators (SEAs)

SEAs use a fixed compliant element between a high impedance actuator and the load. There are two types of series elastic actuators (SEAs) - linear and rotary. The compliant elements used are linear springs or viscoelastic materials usually combined with ball screw reduction drives (linear system) or planetary gearbox and cable assemblies (rotary system) [19]. Major components of SEAs are: motor, transmission, spring, controller and sensor. A position sensor measures the deflection, and the force output is accurately calculated using Hooke’s law. A control loop then servos the actuator to the desired output force [26]. Basic requirements of SEAs are that force/mass and power/mass ratios should be high and the actuator must not be damaged and instable due to falls, impact, disturbances etc. [26]. These are also the requirements for human-friendly robots. SEAs were developed with the belief that they would provide advantages such as high force/torque fidelity, low impedance, low friction, good force/torque control ability, reduction in shock impacts and backlash, energy storage, robustness to load changes, stability and safety, high torque/mass and power/mass ratios etc. [26],[31],[32]. Although current SEAs provide a part of these advantages, these designs have still limitations.

The major limitation in SEAs is the fixed nature of the required compliance. Compliance largely depends on spring constant [26]. Soft spring produces high fidelity of force control, reduces stiction, but also limits the bandwidth at high force
range. On the other hand, stiff spring produces larger force bandwidth, but force fidelity becomes low and stiction becomes high [26]. These limitations also affect the control performances [33]. Again, a fixed spring produces a fixed compliance and the spring needs to be changed if a change in the compliance is required. Moreover, the conventional SEAs are bulky, error prone, less power efficient if not designed using ball screw, limited to small passive deflection ranges, mechanical design is complex and the system is loosely integrated etc. [19]. This is why, the applications of SEAs to compact and multi-degree of freedom human-interactive robotic systems such as the AFO still remain as challenging tasks.

4. Construction of AFO

A physical prototype of the AFO was developed. Figure 1 shows the prototype novel modular powered Ankle-Foot Orthosis (AFO). The design consists of a series elastic actuator (SEA) to actuate the ankle-foot system. The design possesses a plethora of novelties. The novelties were incorporated aiming to overcome the limitations of the existing gait rehabilitation systems [10]-[18]. Novelties of AFO design are as the following:

1. **Novel actuator**: the actuator designed for AFO is a compact Series Elastic Actuator (SEA) that is compliant, back drivable, variable impedance and force-controllable. Compared to the conventional rotary geared motor actuator, it is more powerful, light weight and highly efficient. It also ensures safety for the human machine-interaction due to the low mechanical impedance and compliance [26]-[33].

2. **Adaptive shared control methodology**: compared to normal position-based control for rehabilitation robots, the proposed control method is a shared control structure that provides torque assistance that is just sufficient based on the user performance. The controller also includes on-line learning for adaptation to individual user characteristics and different stages of the rehabilitation process [22]-[23].

3. **Light-weight structural design using advanced materials**: the design uses carbon fiber composite material for its main structure and space grade aluminum alloy for the actuator design that make it light-weight, enough strong and convenient [10]-[18].

4. **Modular design**: the powered AFO is an ankle module of an integrated rehabilitation system and it can be easily attached to the knee module of the knee-ankle-foot orthosis depending on the needs of the patients and thus this design provides the advantages of the modular design [2], [9].

![Fig.1 CAD model of the prototype AFO for gait rehabilitation and assistance.](image-url)
5. Kinematics of AFO

The kinematics for AFO for ankle-foot motion is shown in Fig. 2 as a module for the ankle-foot system. In Fig. 2, \( a \) is the shank length, \( b \) is the foot length and \( x \) is the linear displacement of the actuator output. The kinematics for the ankle-foot motion can be described using Eq. (1) as follows. The actuator produces \( x \), and \( \theta \) changes due to the change in \( x \). Hence, the relationship between \( x \) and \( \theta \) can be found as Eq. (2). In fact, the objective is to control \( \theta \) for the ankle-foot orthosis to assist the rehabilitation. Relationships between the derivatives of \( x \) and \( \theta \) can also be determined using these equations.

\[
x^2 = a^2 + b^2 - 2ab \cos \theta \quad (1)
\]
\[
\theta = f(x) \quad (2)
\]

![Fig.2 Kinematics for the ankle-foot motion.](image)

6. Actuator design for AFO

6.1. Construction of the actuator

The actuator for novel AFO (Fig.1) is also specifically designed and constructed with high-level novelties. A novel SEA with novel working principle is designed. The novel actuator design consists of a servomotor, two springs (passive elastic elements), a ball screw and an output link (prospective robot link). A torsional spring is placed in series between the servomotor and the ball screw. The ball screw nut is attached to the output link in series through another set of translational springs. Figure 3(a) shows the schematic diagram of the prototype actuator. This design is to provide a back drivable actuation system mimicking muscle like properties beneficial for joint actuation, and to provide force feedback to control the amount of force in operation.

As it is shown in Fig.3 (a), the servomotor with encoder is attached to the torsional spring, which is attached to a spur gear and the gear is attached to another rotary encoder. The two rotary encoders measure the deflection of the torsional spring and the motor torque is calculated based on the spring constant. The gear is then attached to the ball screw. The rotational speed is reduced using appropriate gear ratio. The ball screw converts rotational motion to translational motion.

A set of translational springs is attached to the nut of the ball screw. When the nut moves, a deflection is produced in the springs, which is measured by a linear encoder. The output force is calculated based on the spring constant and the linear deflection following Hooke’s law. The shuttle assembly is connected to the other end of the joint. Linear springs added on both ends of the ball screw nut are connected to the shuttle that helps achieve back drivable actuation in the green arrow directions. The ball screw nut moves independent of the shuttle. Key components of the force feedback are torsional spring, and two high resolution rotary encoders. Torsional spring is incorporated with predefined spring constant and the encoders measure the rotational difference between the motor (input) and the spur gear (output).
To achieve the bi-directional loading of the torsional spring, it is packaged inside an assembly containing two opposite winding torsional springs loaded independently with respect to the rotational direction. The output pin is used to attach to the output link. The hinge point is used to tie the actuator to any suitable frame. The translational springs are soft and small. Although the effective spring constant is very big, the size of the torsional spring is very small as it is at the high speed and low torque range. Therefore, overall size of the actuator is smaller than that of the existing actuators and the weight is also very low [26]-[33]. Figure 3(b) illustrates the exploded view of the physical actuator.

The actuator assembly main parts are machined out of aluminium (6061T6) taking the advantage of mechanical properties of the material (weight and strength). Shuttle assembly and ball screw nut shares the same linear guide for constraint in movement other than the intended stroke direction. Selected motor (Maxon DC brushless motor, EC 4-pole 120 Watt 36V) is universal in all joints applications. It is used due to its lightweight (0.175kg), low moment of inertia, favourable power to weight ratio and compactness.

The actuator is designed to be able to provide up to 60Nm assistive torque at human joint. To have a high resolution force sensing, a minimum of 1024ppr rotary encoder is integrated in the system. For mechanical transmission, the ball screw is used for high efficiency and the spur gears are connected to compact the actuator assembly. The ball screw selected from EichenbergerGewinde AG has a pitch of 2mm/rev and can output up to 1500Nforce. The translational springs have spring constant of 24 N/mm and a working stroke of 12mm. They are used to operate in the range of about 25% of the full force. The total mass of the actuator is less than 0.85kg.

6.2. Working principles of the actuator

In order to analyse the actuator performances at the output end, which produces linear output force, the actuator is modelled as a system consisting of translational elements only by converting the rotary elements to equivalent translational elements. The actuator model for the equivalent translational motion is shown in Fig.4 (a). In this model, $F_1$ is motor input force, $m_1$ as in Eq. (3a) is equivalent mass of the motor, $p$ is the pitch of the ball screw, $m_2$ as in Eq. (3b) is equivalent mass of the ball screw, $k_1$ as in Eq. (3c) is considered as the equivalent translational spring constant of the torsional spring $k_t$, $k_2$ is spring constant of the translational spring, $b_1$ and $b_2$ are the viscous damping for motor and ball screw respectively, and $F_o$ is the output force. In this design, $k_1$ is selected to be much bigger than $k_2$ (i.e., $k_1 \gg k_2$). For the prototype design, $k_1$ is several hundred times of $k_2$. Despite high effective stiffness, size and weight of the torsional spring is still very small because it is in the high speed range.

Due to big difference of the two spring constants, at low force range, the model can be reduced to as shown in Fig. 4(b). It means that the torsional spring behaves as rigid and does not work as a spring, and only the translational spring works. Hence, the output impedance and compliance are due to only the translational spring and they depend on the spring constant $k_2$. As the allowable stroke for the translational springs is small, at high force range, the translational springs become compressed and only the torsional spring works, the model reduces to as shown in Fig.4(c). Hence, the bandwidth of the actuator is very high at the high force range due to big spring constant $k_1$ of the torsional spring.

Therefore, the whole performances depend on force range and on the difference between the spring constants $k_1$ and $k_2$. Thus, the actuator changes its output impedance characteristics and dynamic bandwidth according to the force range without requiring a change in the hardware (springs). The actuator can thus achieve a high force bandwidth due to the torsional spring, low output impedance characteristics and small non-linear friction due to the soft translational springs. This force-range-based variable impedance may increase human-friendliness of the actuator for the AFO.
Fig.4 (a) Working principle of the series elastic actuator for translational motion, (b) principle for low force, (c) principle for high force.

\[ m_1 = J_1 \left(\frac{2m_2}{P}\right)^2 \]  
\[ m_2 = J_2 \left(\frac{2m_2}{P}\right)^2 \]  
\[ k_1 = k_t \left(\frac{2m_2}{P}\right)^2 \]  

7. Dynamics responses of the actuation system for AFO

Closed-loop model for the SEA for ankle-foot motion is shown in Fig.5. The dynamic motion equations for the low force condition are derived as Eqs. (4)–(6) based on Fig.4(b) and Fig.5. When the force is small, the high stiffness torsional spring, motor and the ball screw can be considered as a single mass as \( m_1 + m_2 \). \( F_2 \) is the force on the translational spring \( k_2 \) and it can be calculated by Hooke’s law as Eq. (5). It is the output force on the load. \( x_1, x_2, x_3 \) are displacement as in Fig.5.

\[ F_1 = b_2(x_2 - x_3) + k_2(x_2 - x_3) + (m_1 + m_2)x_2' \]  
\[ F_2 = k_2(x_2 - x_3) \]  
\[ (F_d - F_2)(k_p + k_d) = F_1 - F_2 \]  

The transfer functions based on Eqs.(4)–(6) for three conditions are as follows: (i) open loop transfer function with load end fixed as given in Eq.(7), (ii) closed loop transfer function with load end fixed as given in Eq.(8), and (iii) output impedance with load end fixed as given in Eq.(9), where \( F_d \) is the desired force. A PD controller expressed as \( k_p + k_ds \) was used for the model, where \( k_p \) and \( k_d \) are proportional and derivative gain respectively. These three transfer functions completely specify the linear dynamic characteristics of the actuator system [26].

\[ T_2(s) = \frac{k_2}{(m_1 + m_2)s^2 + b_2s + k_2} \]  
\[ T_2(s) = \frac{k_2}{(m_1 + m_2)s^3 + b_2s^2 + k_2} \]  
\[ \frac{T_2(s)}{x_3(s)} = \frac{-m_2k_2 + m_kx_2}{(m_1 + m_2)s^3 + b_2s^2 + k_2} \]  

Similarly, for large force, the low stiffness spring is compressed completely and the ball screw and load are integrated as a single mass. So the actuator can be only activated by the torsional spring in high force range. The dynamic equations for high force conditions according to Fig.4(c) and Fig.5 are derived as Eqs.(10)-(13). The transfer functions for the open loop, closed loop force control and the output impedance with the load end fixed are shown in Eqs.(14)-(16) respectively.

\[ m_1\dot{x}_1 = F_1 - k_1(x_1 - x_2) - b_1(\dot{x}_1 - \dot{x}_2) \]  
\[ m_2\dot{x}_2 = -k_1(x_2 - x_1) - b_1(\dot{x}_2 - \dot{x}_1) \]  
\[ F_2 = k_t(x_1 - x_2) \]  
\[ F_1 - F_2 = (F_d - F_2)(k_p + k_d) \]  

\[ m_1\ddot{x}_1 = -k_1(x_1 - x_2) - b_1(\dot{x}_1 - \dot{x}_2) \]  
\[ m_2\ddot{x}_2 = -k_1(x_2 - x_1) - b_1(\dot{x}_2 - \dot{x}_1) \]  
\[ F_2 = k_t(x_1 - x_2) \]  
\[ F_1 = F_2 + (F_d - F_2)(k_p + k_d) \]  

\[ m_1\ddot{x}_1 = -k_1(x_1 - x_2) - b_1(\dot{x}_1 - \dot{x}_2) \]  
\[ m_2\ddot{x}_2 = -k_1(x_2 - x_1) - b_1(\dot{x}_2 - \dot{x}_1) \]  
\[ F_2 = k_t(x_1 - x_2) \]  
\[ F_1 = F_2 + (F_d - F_2)(k_p + k_d) \]
\[
\frac{F_2(s)}{F_1(s)} = \frac{k_1}{m_1 s^2 + b_2 s + k_1} \\
\frac{F_2(s)}{x_2(s)} = \frac{k_d k_2 s + k_p k_1}{m_1 s^2 + b_1 s + k_d k_1 s + k_p k_1} \\
\frac{F_2(s)}{x_2(s)} = \frac{-m_1 k_1 s^2}{m_1 s^2 + b_1 s + k_d k_1 s + k_p k_1}
\]

Figure 5 Closed-loop model of the actuator system for low (upper), and high (lower) force condition.

Figure 5 shows the diagram for the closed-loop system for the actuator model for low and high force conditions. Values of \( m_1, m_2, b_1, b_2, k_1, \) and \( k_2 \) for MATLAB simulation for the model were selected based on the prototype actuator for AFO (Table 1). The step response results are shown in Fig.6 (a) for small (100 N) and in Fig.6 (b) for large (1000 N) force. For small force, the PD controller’s parameters were \( k_p=2.0, \ k_d=0.02 \) and it is seen that the overshoot is small and the settling time is small (0.28s). For large force range, the PD controller’s parameters are \( k_p=0.14, \ k_d=0.0002 \). The overshoot is zero and the settling time is small (0.35s). The results show satisfactory performances for the actuator for the ankle-foot motion for AFO [34]. However, the control will need to be designed in such a way that a switch occurs automatically from low to high force condition based on the situation or demand of the input force, and the system is stable during the switch.

Table 1. Parameters and their values used for simulation

<table>
<thead>
<tr>
<th>Values of hardware parameters for rotational motion</th>
<th>Values of hardware parameters for equivalent translational motion</th>
</tr>
</thead>
<tbody>
<tr>
<td>( J_1=8.91 \times 10^{-5} \text{kg.m}^2 )</td>
<td>( m_1=J_1 \ (2\pi \rho)^2 =8.78 \text{kg} )</td>
</tr>
<tr>
<td>( J_2=642 \times 10^{-5} \text{kg.m}^2 )</td>
<td>( m_2=J_2 \ (2\pi \rho)^2 =6.33 \text{kg} )</td>
</tr>
<tr>
<td>Torsional spring constant ( (k_t) ) = 1 N/m.rad</td>
<td>( k_t=k_t \ (2\pi \rho)^2 =9.86 \times 10^6 \text{N/m} )</td>
</tr>
<tr>
<td>Translational spring constant ( (k_d) ) = 24x10^3 N/m</td>
<td>( k_d=24\times10^3 \text{N/m} )</td>
</tr>
<tr>
<td>Pitch of the ball screw ( \rho ) = 2 x 10^{-3} m</td>
<td>( p=2 \times 10^{-3} \text{m} )</td>
</tr>
<tr>
<td>( b_1=700 \text{Ns/m} )</td>
<td>( b_1=700 \text{Ns/m} )</td>
</tr>
<tr>
<td>( b_2=900 \text{Ns/m} )</td>
<td>( b_2=900 \text{Ns/m} )</td>
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![Fig. 6 The step responses for (a) low force (100N), and (b) high force (1000N).](image-url)
The bode plot is shown in Fig.7. It is seen that when the force is small, the bandwidth is 20Hz, while the force is big, the bandwidth is 914Hz. It means that the higher stiffness torsional spring can raise the force bandwidth of the actuator. If $k_p$ increases, the bandwidth also increases, and if $b_1$ increases, the bandwidth reduces.

The impedance’s cut off frequency is 13Hz when the force is small, and when the force is big the cut off frequency is 102Hz, as shown in Fig.8. $k_p$ determines the bandwidth. When $k_p$ is big, the cut off frequency is also big. When $b_1$ becomes big, the cut off frequency becomes small.

It is seen in the bode plots in Fig.7 that the frequencies when the magnitudes deviate from zero are about 70 rad/s and 100 rad/s for the low and high force respectively. Again, the phases do not cross -180 degree for low and high force range. As 70 rad/s is smaller than 100 rad/s, it indicates that the system is stable. Again, the system does not show any tendency of approaching towards 0dB and -180 degree, which also indicates the system’s stability [34]. As the plots show, there is no tendency of resonance. The difference between $k_1$ and $k_2$ is very big that produces a big difference between the natural frequencies of low and high force condition. This also prevents resonance [34]. Dynamic responses may be further optimized and customized for the real applications by changing the values of the parameters in Table 1, and $k_p$ and $k_d$.

8. Conclusions and future works

Electro-mechanical design, novelties in the design and the kinematics for the AFO are presented. Then, the design, configuration, materials selection, working principle, dynamics and dynamics responses for the control of the novel variable impedance compact compliant series elastic actuator for the ankle-foot motion for the AFO are presented. The AFO design consists of novel actuator that is built with a servomotor, one torsional spring, one set of translational springs, a ball screw and an output link. The main novelty is that the actuator model of AFO has two types of springs (torsional and translational) at different speed ranges so that it can produce variable impedance based on force ranges without requiring a change in hardware, which may enhance human-friendliness in AFO applications. The other novelties are adaptive shared control, light-weight structure due to especially selected materials and modular design. Though the prototype AFO with novel actuator has not been tested yet physically, simulation results of dynamics and control analyses of the design have been proven satisfactory, which indicate that the prototype physical AFO built on the proposed design will also produce...
satisfactory performances in real rehabilitation applications. In near future, the prototype AFO with the proposed design will be clinically tested using stroke patients to verify the performances.

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References