Retractable Prosthesis for Transfemoral Amputees Using Series Elastic Actuators and Force Control

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Abstract—We present a highly functional and cost-effective prosthesis for transfemoral amputees that uses series elastic actuators. These actuators allow for accurate force control, low impedance and large dynamic range. The design involves one active joint at the knee and a passive joint at the ankle. Additionally, the socket was designed using mirroring of compliances to ensure maximum comfort.

I. INTRODUCTION

Transfemoral amputation involves the loss of two joints: knee and ankle. Both of these joints play crucial roles in enabling efficient human motion. The former is a rotational joint that permits flexion and extension of the leg as well as a slight internal and external rotation [1]. The latter generates most of the power needed to walk and is critical for shock absorption and balance.

In order to allow smooth movements, avoid gait asymmetries, and prevent falls, a transfemoral prosthesis must be able to address the following needs:

1) Provide enough power for the foot to safely clear the ground.
2) Adapt to the user’s walking pattern and speed.
3) Function in different terrains and environmental conditions.
4) Provide robust footing and adequate shock absorption.
5) Comfort.

For 50% of transfemoral prosthetic users, gait is not automatic. In other words, they must think every step they make [1]. Nowadays, most high-technology prostheses are too expensive for the average user and commercial designs are not functional enough to allow complex locomotor activities. Thanks to our material, sensor, and actuation selection, the design presented in this paper provides an optimal balance between cost and functionality. Hence, this prosthesis for transfemoral amputees could potentially be commercialized in low-income countries where high-tech bionic devices are not an option.

The prostheses that are currently being used in sub-developed countries are passive and uncomfortable [2]. Therefore, even though our design involves only one active joint and a basic sensory-actuator system, it will undoubtedly be a major improvement with regard to existing devices of a similar price. The prototype is shown in Fig. 1.

II. RELATED WORK

A. Actuators

There has been considerable progress in developing actuators for robotics in the past few decades. In particular, Series Elastic Actuators were developed by Pratt and Williamson [3], [4] who directly measured the strain of a spring in series with transmission and actuator output. Since the spring deforms a significant amount, the fidelity compared to typical strain gauge structures for force control is much higher. However, their motion bandwidth is rather small, but biomimetic actuators can trade off small motion bandwidth for good force control.

The highest performance force controlled actuator has been a brushless DC motor rigidly connected to a ball screw which drives the linear motion. The socket presented in the figure is a simplified representation of our final design. See section V.

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The highest performance force controlled actuator has been a brushless DC motor rigidly connected to a robot link, also known as direct-drive [5]. These actuators eliminate friction and backlash, typical of motors with transmissions. To compensate for the loss of transmission, direct-drive actuators must be large in order to achieve adequate torque. This means increased motor mass and cost.

Although, Series Elastic Actuators may not have as large a dynamic range as comparable direct drive actuators, the weight savings are well worth the trade off for our intended weight sensitive application.

B. Robots

Several biomimetic robots using Series Elastic Actuators have been constructed and demonstrated. For instance, Spring Turkey [6] and SpringFlamingo [7] developed by MIT use linear drive Series Elastic Actuators.

Other laboratories and companies have also worked on walking robots. One of the most remarkable and well known being Boston Dynamic’s Big Dog. A quadruped robot designed to achieve animal-like mobility on rough and
rugged terrain, terrain too difficult for any existing vehicle. It uses low-friction hydraulic cylinders regulated by two-stage aerospace-quality servovalves for actuation. Each actuator has sensors for joint position and force [8].

C. Force Control

In highly unstructured environments, force controlled robots that can comply to the surroundings are desirable [9]. An ideal force controllable actuator would be a perfect force source. In a perfect force source, impedance is zero (completely back drivable), stiction is zero, and bandwidth is infinite.

By adding Series Elasticity to these conventional systems, a force controllable actuator with low impedance, low friction, and good bandwidth will result.

Thanks to their ability to closely approximate a pure force source, Series Elastic Actuators lead to a much more accurate force control than other traditional technologies such as [9]:

- Direct drive actuation.
- Current control with a geared actuator.
- Current control with low-friction cable drive transmissions.
- Load cells with force feedback.
- Fluid pressure control.

D. Socket

Scientific studies have been conducted to quantify attributes that may be important in the creation of more functional and comfortable lower-limb prostheses. The prosthesis socket, a human-machine interface, has to be designed properly to achieve satisfactory load transmission, stability, and efficient control for mobility. The biomechanical understanding of the interaction between prosthetic socket and the residual limb is fundamental to such goals [10].

Some early designs of the prosthetic socket, such as the plugfit, took the form of a simple cone shape, with very little rationale for the design. With an understanding of the residual limb anatomy and the biomechanical principles involved, more reasonable socket designs, such as the patellar tendon bearing (PTB) transtibial socket, and the quadrilateral transfemoral suction socket were developed following World War II [11], [12].

By the 1980s, the so-called hydrostatic weight-bearing principle and the total surface bearing (TSB) concept were introduced. Examples include the silicone suction socket [13] and ICEROSS [14], as well as those incorporating the use of interfacing gel-like materials.

The most radical of new prosthetic developments is certainly direct skeletal attachment of limb prostheses through osseointegrated implants. This method completely obviates the need for the prosthetic socket through percutaneous titanium fixtures that transfer load from the prosthesis directly to the skeletal bone. While it may seem that osseointegration renders moot any discussion of prosthetic interfaces, even this radical advance in the state of the art only changes the location and type of the interface problem. New challenges arise from the metal/bone and metal/skin interfaces [10].

Fig. 2. Human walking gait through one cycle, beginning and ending at heel strike. Percentages showing contact events are given at their approximate location in the cycle. Taken from [15].

Fig. 3. Exploded view of the prototype showing all of the components.

III. Approach

The design aims to enable the user to walk with ease at a low price. In order to do so, understanding the biomechanics of human walking is crucial, see Fig. 2. Our main innovation is using linear actuation instead of rotatory for the knee joint. This significantly lowers the price and mechanical complexity while maintaining high functionality. One of the few downsides of Series Elastic Actuators is their relatively small range of motion. However, we can see in Fig. 2 that we usually do not lift the foot off the ground more than a couple centimeters in each stride. Additionally, thanks to the retractability feature, this prosthesis can be easily adapted to users with a wide range of heights. In other words, it is a ”one fits all” design and, therefore, extremely cost-efficient. The exploded view in Fig. 3 helps get a better grasp of how the mechanism works and the different parts that it comprises.

IV. Series Elastic Actuators

Series Elastic Actuators provide many benefits in force control of robots in unconstrained environments. These benefits include high force fidelity, extremely low impedance, low friction, and good force control bandwidth. Series Elastic Actuators employ a novel mechanical design architecture which goes against the common machine design principal of stiffer is better. A compliant element is placed between the gear train and driven load to intentionally reduce the stiffness of the actuator. A position sensor measures the deflection, and the force output is accurately calculated using Hooke’s Law (F=Kx). A control loop then servos the actuator to the desired output force. The resulting actuator has inherent shock tolerance, high force fidelity and extremely
Fig. 4. Schematic diagram of a Series Elastic Actuator. A spring is placed between the motor and the load. A control system servos the motor to reduce the difference between the desired force and the measured force signal. Taken from [9].

Fig. 5. CAD rendering of the Series Elastic Actuator used in the knee joint.

Note that Series Elastic Actuators are topologically similar to any motion actuator with a load sensor and closed loop control system. In fact, the main components of this actuation system are:

1) Motor
2) Transmission
3) Spring
4) Sensor
5) Controller

Our actuator design can be seen in Fig. ?? In the following sections we will study different design parameters that must be taken into account when choosing each of the components.

A. Motor Selection

When selecting a servomotor there are many parameters we must take into account. It is worth highlighting the following:

- Rated Voltage: the voltage it is most efficient while running.
- Operating current: average amount of current the motor is expected to draw under a typical torque.
- Average Power = Rated Voltage x Operating Current
- Stall current: maximum amount of current the motor will ever draw, and hence the maximum amount of power. It is measured by powering up the motor and then applying enough torque to force it to stop rotating.
- Stall torque: torque required to stop the motor from rotating.
- Operating torque: the torque the motor was designed to provide. Usually it is the listed torque value (it is typically intended to be applied at 1 cm from the shaft).
- Power requirements = Load Torque x Speed
- Operating speed: refers to the maximum speed at which motor can turn: measured in seconds per 60. If it is 0.5 seconds per 60, then it will take 1.5 seconds to turn 180.

Since our mechanism uses a servomotor coupled to a ball-screw (see section IV-C) we used the following approach:

1) Calculate the torque (1) and speed (2) requirements of the ball-screw:

\[ \tau_{BS} = \frac{F \times L}{2 \times \Pi \times \eta} = 1.18 (N \times m) \quad (1) \]

\[ \omega_{BS} = \frac{V_L}{L} = 3600 \text{ (rpm)} \quad (2) \]

Where:

- \( F = \text{Force} = 300 \text{ lb (1334 N)} \to \text{Assuming the person’s weight is 200lb and that during walking each leg experiences 1.5 x body weight [16].} \)
- \( L = \text{Screw Lead} = 5 \text{ mm/rev} \to \text{Linear displacement of nut for one revolution of screw} \)
- \( \eta = \text{Efficiency} = 0.9 [17] \)
- \( V_L = \text{Linear Speed} = 18000 \text{ mm/min} \)

2) Select a motor whose operating speed and torque exceed the ball-screw requirements: We chose the MOOG BN34 55 EU 02LH fabricated by Moog Inc. The specifications can be seen in Table I. We decided to use a brushless DC motor because it has been shown to minimize the motor friction seen through the transmission [18]. Keeping friction and motor saturation low is always desirable. Additionally, we used a frameless motor configuration where the motor magnets are mounted directly onto an extended ball-screw shaft instead of using a coupling, gears, or a belt drive. The main goal of this was to keep transmission dynamics at a minimum, as it is one of two limiting factors in using high feedback gain [19]. The other limiting component to achieve high feedback gain is the sensor. See section IV-B.

B. Sensors: Linear Potentiometers

The sensor needs to directly measure the spring deflection. This insures that the feedback measurement is a representa-
tion of true force. Noise in the sensor is also very detrimental to operation. We use a linear potentiometer to measure spring deflection.

C. Transmission: Ball–Screw Selection

During operation, the servomotor directly drives the ball screw, converting rotary motion to linear motion of the ball nut. When the motor rotates, the ball nut moves up or down the screw depending on the direction of motor rotation. The ball nut pushes on the springs, which help with shock absorption and allow fine force control.

When choosing a ball screw the most important specifications are the lead and the diameter. The lead determines how fast the prosthetic will be able to retract and the diameter is proportional to the load it will be able to support.

We selected a ball screw manufactured by Thomson Linear. First we chose a 5mm/rev lead because it was the most popular for robotic applications. Additionally, it allows sufficiently fast retractability. Given an effective linear travel, ELT, of 108mm (measured using SolidWorks) and a motor speed, $\omega_m$, of 79.83 rev/sec we can calculate that time as follows:

$$\text{time} = \frac{\text{ELT}}{L \times \omega_m} = \frac{108(\text{mm})}{5(\text{mm/rev}) \times 79.83(\text{rev/s})} = 0.27s$$

(3)

As we can see, our prosthesis can move more than 10cm in less than 3 tenths of a second. This exceeds the requirements needed for normal walking or even running.

Once we had chosen the lead, we fixed the nut size to 24mm because this ensured it would support a 350lb load, which is good enough for our 200lb person hypothesis. Finally, we selected a 24mm dia ball screw to match the nut.

D. Spring Selection

Choosing the spring constant, $k_s$, for the elastic element requires special attention because we must balance two competing requirements:

1) Large force bandwidth requires a high spring constant.
2) Minimizing nonlinear friction and impedance requires a low spring constant.

It is important to note that biomimetic actuators can trade off small motion bandwidth for good force control. This makes springs an ideal element to include in the design [18].

The following guidelines can be used when selecting a spring constant:

1) Define an operational bandwidth, $\omega_o$, for which the actuator will need large forces → this places a lower bound on $k_s$
2) Insure that the controller gains can be raised to acceptable levels of stiction and impedance reduction → this places an upper bound on $k_s$

Empirically, it has been shown that 315 kN/m offers an optimum balance between force bandwidth and impedance in these kind of actuators [18]. Therefore, we selected this value for our springs.

V. Socket Design: A Variable-Impedance Prosthetic Socket

Surveys have shown that amputees complain about their prosthesis being uncomfortable [20], [21]. It is not uncommon for amputees to develop skin problems on the residual limb, such as blisters, cysts, edema, skin irritation, and dermatitis [22], [23]. Discomfort and skin problems are usually attributed to a poor socket fit.

The basic principles for socket design vary from either distributing most of the load over specific loadbearing areas or more uniformly distributing the load over the entire limb [10].

Even the most rigorous scientific analyses to date have focused in large part on socket designs based on historical use and proven clinical adequacy[10], not on numerical analysis. Modern instrumentation and computer modeling have allowed us to discover what had only previously been the implied conditions inside prosthetic sockets. However, the most recent advances in the understanding of stresses experienced at the limb/prosthesis interface have not yet fundamentally altered clinical practice [24].

For all prosthetic socket designs, the optimal load distribution should be proportional to the ability of the body to sustain such stresses, without crossing the thresholds of pain or skin breakdown [10].

The CAD/CAM technology for the prosthetic socket may make the socket design and manufacture process more effective and objective. However, the current computer-aided design and manufacturing (CAD/CAM) systems cannot offer any expert suggestion on how to make an optimal socket design.

We decided to use mirroring of compliances. This technique consists on creating a socket with varying stiffness in such a way that it mirrors the biomechanics of the underlying tissue. We chose this approach because it has been proven that an inversely proportional relationship between residual limb stiffness and the corresponding socket wall stiffness at each spatial location across the residual limb surface leads to reduced contact pressures over the fibula and tibia anatomical landmarks during walking and quiet standing [25]. This is extremely important because socket interface pressure is a major reason for sores, pain, and discomfort in sockets[26].

To create this socket there are several steps that must be followed [25]:

1) MRI Imaging: MRI data of the amputee’s residual limb can be used to estimate and map body stiffness and anatomical landmarks directly to the prosthetic sockets wall stiffness. See Fig. [3]
2) Inverse mapping: An inverse linear equation is used to map bone tissue depth to socket material stiffness properties. Regions where the body was stiffest interfaced with the most compliant material, whereas regions where the body was softest interfaced with the least compliant material. Empirical data has led to the following equation: $Y = 0.0382 \times X + 1.0882$ where $Y$ is the Youngs Modulus of the printing material and $X$
is the bone tissue depth.

3) 3D printing: Polyjet Matrix 3D printing technology is used to seamlessly integrate variable durometer materials into the socket design to achieve intrinsic spatial variations in socket wall impedance while maintaining structural integrity.

VI. MATERIALS

Material selection had both cost and function in mind during the design process. Most components were either found commercially or easily manufactured with traditional manufacturing techniques.

- Aluminum 6061: This material was used for parts that need to be custom machined. Aluminum 6061 provides strong structural properties, low cost, and easy machinability. Listed below are the parts that use Aluminum 6061 which can be shown in Fig. 3:
  - Motor Mounting Plate
  - Carriage Plates
  - Base Plate
  - Foot Attachment
- Stainless Steel (Variety of Grades): This material is found in commercially found parts. The grade of the stainless steel varies depending on the product. By using commercially available parts, costs are generally lowered. The parts made from stainless steel include which can be seen in Fig. 3:
  - Screw
  - Linear Rails
- ABS Plastic: ABS is a very common grade of plastic. It is heavily used in the 3D printing industry. ABS provides a low cost and durable option to custom, intricate parts. The parts using ABS plastic can be seen in Fig. 3:
  - Socket
  - Foot

Since the socket is to be of variable impedance, 3D printing will be utilized. In addition, the foot can be readily 3D printed with different designs to provide the ultimate level of comfort for the user. Once determined which is the best model, the foot can be easily produced using an injection molding manufacturing process to reduce cost if scale up occurs.

- Rubber: A variant of rubber will be used for the sole of the foot. This will allow traction to be generated similar to a sneaker.
- Teflon: Teflon coated linear bearings are to be used as a cheap, maintenance free alternative to conventional linear bearings.

VII. FINITE ELEMENT ANALYSIS

After several iterations of the design were completed, a finite elemental analysis was used to determine the structural integrity of the robot. Both the magnitude and direction of walking forces were approximated during the simulation. Fig. 7 shows the forces to be simulated throughout a typical walk cycle. Each of the forces (or sum of forces) were simulated at 1.5x the arbitrary body weight of 200 lbs.

Several assumptions, which aren’t necessarily accurate, were made during the FE simulations. These assumptions include a global bond between all connections, essentially making the whole robot a rigid body.

Analyzing the simulations, the current design has the potential to fail depending on the validity of the assumptions. Testing would ultimately have to be conducted to ensure that the model presented is safe. Assuming structural failure, the highest concentrations would be seen at the linear rails between the base plate and foot attachment. This makes logical sense as this portion of the design is under intensive moment loads. Possibly remedies include geometry changes, material changes, architecture changes, etc.

Fig. 8, Fig. 9 and Fig. 10 show the results of the simulation. The color bar shows the Von Mises stress, which is used as an indicator to know to predict yielding of materials. A material is said to start yielding when its von Mises stress (which can be computed using the Cauchy stress tensor) reaches a critical value known as the yield strength. At that precise point the material will begin to deform plastically (yield), that is, it will undergo non-reversible changes of shape due to the applied forces.

VIII. CONCLUSIONS

We presented a prosthesis for transfemoral amputees whose general specifications can be found in Table II. Overall, the use of a series elastic actuator coupled with force control provides a potential for a low cost, natural option to transfemoral amputees. With further testing, this design could show promise amongst less financially fortunate patients. Utilizing cheap sensors, cheap materials, and commercialized products the financial burden of purchasing an assistive robot may be overcome.

Analyzing the specs in Table II, the design shows promise to be a comfortable, natural, and versatile alternative to a conventional peg-leg.
Fig. 7. Simplified force model used during the finite element analysis. Taken from [16]

Fig. 8. Finite element analysis of stress distribution during opposite heel strike. See Fig. 2 for walking cycle notation.

Fig. 9. Finite element analysis of stress distribution during heel strike. See Fig. 2 for walking cycle notation.

Fig. 10. Finite element analysis of stress distribution while standing for a 200 lb person.

TABLE II

| Prosthesis Specifications | Variable impedance |
|---------------------------|--------------------|
| Socket Design             | User Specific or Mass Produced |
| Foot Design               | 9 Kg                |
| Overall Weight            | 665 mm              |
| Effective Travel          | 108 mm              |
| Sensors                   | Linear Potentiometers |

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