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Passive Ankle Joint Actuation in a Prosthetic Limb Using a Spring Mechanism

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***Abstract* - A major challenge in lower-limb prosthesis design is achieving functional mobility while minimizing prosthesis mass. Although hydraulic actuators provide effective active control, their relatively high mass limits broader application, especially in transtibial and transfemoral prostheses. This paper presents a passive ankle joint actuation concept based on a preloaded linear extension spring as a light-weight alternative to a hydraulic actuator. Required ankle joint torques for stand-ing and walking are analyzed, spring forces are calculated, and a suitable spring is dimensioned. The results demonstrate that stable standing and walking can be achieved using a single preloaded spring, providing a viable interim solution until lighter active components become commercially available.**

Keywords: Lower-limb prosthesis, ankle joint, passive actuation, spring mechanism, light-weight prosthetic design.

I. INTRODUCTION

Modern prosthetic design trends aim to enable amputees to achieve near-natural gait patterns while minimizing prosthesis mass and complexity. Excessive prosthesis weight increases metabolic cost, reduces comfort, and negatively affects long-term user compliance.

Active hydraulic actuation represents an effective solution for joint control in transfemoral prostheses, particularly at the knee joint. However, current hydraulic components remain relatively heavy, often resulting in prosthetic limbs that exceed the mass of the patient’s healthy leg. Until lightweight hydraulic systems become more accessible, hybrid solutions combining active and passive actuation are of significant interest.

In this work, a compromise approach is proposed: the knee joint is actively driven using a hydraulic actuator, while the ankle joint is passively actuated using a mechanical spring. The spring replaces an existing hydraulic cylinder at the ankle joint, maintaining identical mounting geometry while substantially reducing system mass.

II. BIOMECHANICAL REQUIREMENTS OF THE ANKLE JOINT

The ankle joint plays a critical role in human locomotion by providing stability during stance and generating propulsive power during gait. From a biomechanical perspective, the ankle primarily functions as a torque-generating joint, regulating forward progression and maintaining postural balance. Consequently, accurate estimation of ankle joint moment requirements is essential for the design of prosthetic ankle mechanisms.

During static standing, the ankle joint contributes to postural stability by counteracting small perturbations of the body’s center of mass. These corrective actions are primarily achieved through low-level plantarflexion and dorsiflexion torques generated around the ankle joint. Although the required moments during standing are relatively small compared to dynamic activities, they are continuously present and therefore critically influence prosthetic comfort and stability. Insufficient ankle stiffness or torque capacity during standing can result in instability, increased reliance on compensatory hip strategies, and elevated metabolic cost.

During walking, the biomechanical demands on the ankle joint increase significantly. The ankle undergoes a characteristic cycle consisting of controlled plantarflexion following heel strike, dorsiflexion during mid-stance, and rapid plantarflexion during push-off. The push-off phase is particularly important, as it contributes substantially to forward propulsion and walking efficiency. Peak ankle moments during walking occur during late stance and are predominantly plantarflexive. These moments must be adequately supported by the prosthetic ankle to ensure smooth rollover and natural gait dynamics.

In contrast, running introduces substantially higher ankle joint moments due to increased ground reaction forces, higher angular velocities, and greater energy storage and release. Peak plantarflexion moments during running can exceed those observed during walking by a factor of two or more. However, since the present study focuses on a lightweight, passive ankle joint intended primarily for standing and walking, running is not considered as a primary design condition beyond its minimal initial torque range.

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Based on reported biomechanical measurements of ankle joint kinetics, representative torque ranges can be defined for different activities:

- **Standing:** 1–5 Nm, primarily associated with postural stabilization
- **Walking:** 5–15 Nm, with peak values occurring during plantarflexion at push-off
- **Running:** 15–40 Nm, dominated by high plantarflexion moments

These values provide a practical design envelope for the passive ankle mechanism. For the purposes of this study, emphasis is placed on the upper bound of the standing torque range (approximately 5 Nm) to ensure adequate static stability, and on the upper bound of the walking torque range (approximately 15 Nm) to accommodate dynamic gait requirements.

It is important to note that ankle joint torque is influenced by multiple factors, including walking speed, body mass, stride length, and terrain. Furthermore, inter-individual variability among amputees can be significant. Therefore, the selected torque ranges should be interpreted as representative values rather than absolute limits. The proposed spring-based ankle mechanism is designed to provide sufficient torque within this envelope while maintaining mechanical simplicity and minimizing system mass.

By focusing on the dominant plantarflexion moments required for standing and walking, the biomechanical requirements are well aligned with the functional objectives of a passive ankle joint. This approach enables a targeted and efficient mechanical design while avoiding unnecessary over-dimensioning that would compromise weight and usability.

III. FORCE REQUIREMENTS FOR THE PASSIVE SPRING

In a passive ankle joint actuated by a mechanical spring, the joint torque required for functional activities is generated indirectly through the linear force produced by the spring acting at a finite distance from the ankle joint axis. Therefore, a clear relationship between ankle joint torque and spring force must be established in order to properly dimension the spring and evaluate its suitability for prosthetic use.

The ankle joint is modelled as a single rotational degree of freedom with a fixed axis of rotation. The extension spring is assumed to be connected to the joint mechanism at a constant effective lever arm distance r from the joint centre. Under these assumptions, the moment generated about the ankle joint by the spring force can be expressed as:

$$M = F \cdot r$$

where M is the ankle joint moment, F is the spring force, and r is the effective lever arm. The spring force is assumed to act approximately perpendicular to the lever arm over the relevant range of motion, allowing geometric nonlinearities to be neglected in the first-order analysis.

For the present design, the lever arm is defined as:

$$r = 0.025\text{m}$$

This value reflects the geometric constraints of the prosthetic ankle mechanism and corresponds to the attachment location previously occupied by a hydraulic cylinder.

3.1 Standing

During static standing, the ankle joint must generate sufficient moment to maintain postural stability and counteract small disturbances. Based on the biomechanical torque range defined in Section 2, ankle moments during standing are assumed to vary between 1 Nm and 5 Nm.

The corresponding spring forces are obtained by rearranging the torque equation:

$$F = \frac{M}{r}$$

For the minimum standing torque:

$$M_1 = 1\text{Mm} \Rightarrow F_1 = \frac{1}{0.025} = 40\text{N}$$

For the maximum standing torque:

$$M_2 = 5\text{Mm} \Rightarrow F_2 = \frac{5}{0.025} = 200\text{N}$$

These results indicate that a spring force of approximately 200 N is required to ensure stable standing under worst-case static conditions. Consequently, the spring must be preloaded to at least this force level to avoid slack and to provide immediate resistance to ankle rotation.

3.2 Walking

Walking imposes significantly higher demands on the ankle joint, particularly during the push-off phase, where plantarflexion moments reach their peak values. Based on the biomechanical analysis presented in Section 2, walking ankle moments are assumed to range from 5 Nm to 15 Nm.

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The corresponding spring forces are calculated as follows:

For the minimum walking torque:

$$M_3 = 5Mm \Rightarrow F_3 = \frac{5}{0.025} = 200N$$

For the maximum walking torque:

$$M_4 = 15Mm \Rightarrow F_4 = \frac{15}{0.025} = 600N$$

These values show that the spring force must increase from its preload value of 200 N during early stance to approximately 600 N during late stance and push-off. This force variation must be accommodated smoothly to ensure natural ankle motion and user comfort.

3.3 Consideration of Running Loads

Although running is not a primary target activity for the proposed prosthetic design, it is informative to consider its lower torque boundary. Minimum plantarflexion moments during running are comparable to peak walking moments, corresponding to spring forces on the order of 600 N. Peak running moments, however, would require spring forces exceeding 1,600 N, which are beyond the intended operating range of the proposed mechanism.

Accordingly, the present design explicitly targets standing and walking activities, while running is excluded from the primary design envelope in order to avoid excessive spring stiffness, mass, and mechanical stress.

3.4 Selection of the Operating Force Range

Based on the calculated force requirements, the operating range of the spring can be defined as 200–600 N. This range satisfies the following functional conditions:

- **Stable standing:** achieved through a preload force of 200 N
- **Controlled walking:** achieved through progressive force increase up to 600 N
- **Mechanical simplicity:** enabled by a single linear spring

By selecting a spring that operates entirely within this force range, the ankle joint can provide adequate resistance during both static and dynamic conditions without the need for active control or additional damping elements.

This force-based analysis establishes the foundation for the subsequent spring design and dimensioning presented in the following section.

IV. SELECTION OF OPERATING RANGE

The selection of the operating range for the passive ankle spring is a critical design step, as it directly determines the functional capabilities, comfort, and safety of the prosthetic ankle joint. Unlike active systems, where control algorithms can adapt joint behaviour to changing conditions, a passive system must rely entirely on its mechanical characteristics. Therefore, the operating force and torque ranges must be carefully chosen to balance functional performance with mechanical simplicity and minimal mass.

Based on the biomechanical analysis presented in Section 2 and the force calculations described in Section 3, it is evident that the ankle joint experiences substantially different loading conditions during standing and walking. Stable standing requires relatively low but continuously present joint moments, while walking demands higher, time-varying moments that peak during the push-off phase. Consequently, the selected operating range must accommodate both static and dynamic conditions without compromising stability in either regime.

4.1 Lower Bound of the Operating Range

The lower bound of the spring operating range is governed by the requirements for static stability during standing. As shown previously, maintaining postural stability under worst-case standing conditions requires an ankle joint moment of approximately 5 Nm, corresponding to a spring force of 200 N at the selected lever arm distance.

If the spring force were allowed to fall below this level, the ankle joint would exhibit insufficient resistance to small perturbations, potentially leading to excessive ankle rotation, reduced confidence during standing, and increased reliance on compensatory movements at the knee or hip. To prevent such behaviour, the spring must be preloaded to a force level that guarantees immediate torque generation even at zero or minimal ankle displacement.

For this reason, the preload force is selected as:

$$F_{min} = 200N$$

This preload ensures that the ankle joint remains mechanically engaged at all times and provides a baseline stiffness comparable to that of the biological ankle during quiet standing.

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4.2 Upper Bound of the Operating Range

The upper bound of the operating range is determined by the maximum ankle joint moment required during walking. Peak plantarflexion moments during late stance have been identified as approximately 15 Nm, corresponding to a spring force of 600 N.

Designing the spring to exceed this force would increase stiffness and mass without providing significant functional benefit for the intended use case. Moreover, excessive stiffness could impair gait naturalness by limiting ankle range of motion and increasing impact loads transmitted to the residual limb.

Accordingly, the maximum operating force is selected as:

$$F_{max} = 600N$$

This value represents a practical upper limit that allows for efficient energy transfer during push-off while maintaining acceptable mechanical stresses within the spring and surrounding structure.

4.3 Exclusion of High-Impact Activities

While running and high-impact activities can generate ankle joint moments well above those observed during walking, incorporating these conditions into the design envelope would require substantially higher spring forces and stiffness. Such an approach would conflict with the primary objective of minimizing prosthesis mass and complexity.

Given that the proposed prosthetic concept targets everyday mobility tasks such as standing and walking, high-impact activities are deliberately excluded from the operating range. This design choice reflects a common trade-off in prosthetic engineering, where functional adequacy for the majority of daily activities is prioritized over extreme performance scenarios.

4.4 Functional Implications of the Selected Range

By selecting an operating force range of 200–600 N, the passive ankle joint achieves the following functional characteristics:

- **Continuous engagement:** ensured by spring preload during standing
- **Progressive resistance:** provided during walking as ankle rotation increases
- **Energy storage and release:** enabled within the elastic deformation of the spring
- **Weight efficiency:** achieved by avoiding over-dimensioning

Importantly, the selected range allows both standing and walking to be supported by a single linear spring, eliminating the need for multiple elastic elements or complex mechanical linkages. This simplicity enhances reliability and reduces maintenance requirements.

4.5 Design Consistency and Transition to Spring Dimensioning

The definition of the operating range establishes a clear and consistent boundary condition for the subsequent spring design. The preload force defines the initial extension of the spring, while the maximum force determines the required stiffness and allowable stress levels. These parameters form the basis for the detailed spring dimensioning and material selection presented in the following section.

By grounding the operating range in biomechanical requirements and mechanical feasibility, the proposed passive ankle joint design achieves a balanced compromise between functionality, simplicity, and lightweight construction.

Analysis of the calculated forces indicates that:

- Stable standing requires a joint torque of approximately **5 Nm**, corresponding to a spring force of **200 N**.
- Walking requires torque up to **15 Nm**, corresponding to a spring force of **600 N**.

Therefore, both standing and walking can be supported using a single spring with a force range of **200–600 N**. Stable standing is only possible if the spring is **preloaded** to the minimum force of 200 N.

V. SPRING DESIGN DIMENSIONING

The purpose of the spring design process is to translate the selected operating force range into a physically realizable mechanical component that satisfies functional, geometric, and structural constraints. For a passive ankle joint, the spring must provide sufficient force to generate the required joint moments while maintaining linear behaviour, acceptable stress levels, and compact dimensions suitable for prosthetic integration.

Based on the analysis presented in the preceding sections, the spring is required to operate within a force range of 200–600 N, corresponding to ankle joint moments needed for stable standing and walking. In addition, the spring must be capable of preload, exhibit predictable force–displacement behaviour, and fit within the existing ankle joint mechanism originally designed for a hydraulic actuator.

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5.1 Selection of Spring Type

Among common spring types (compression, torsion, and extension springs), an **extension spring** is selected for the present application. This choice is primarily motivated by the existing mechanical layout of the prosthetic ankle joint, where a hydraulic cylinder was previously installed and operated under tensile loading.

An extension spring offers several advantages in this context:

- It naturally supports **preloading**, which is required to ensure stability during standing.
- It can be mounted using simple pin connections, directly replacing the hydraulic actuator.
- Its force–displacement behaviour can be accurately approximated as linear over the required operating range.
- It allows compact axial integration within the ankle mechanism.

For these reasons, the spring is modelled as a linear extension spring with an initial preload force.

5.2 Force–Displacement Relationship

The force generated by a preloaded linear extension spring is given by:

$$F = F_0 + k \cdot \Delta x$$

Where F_0 is the preload force, k is the spring stiffness, and Δx is the additional extension beyond the preload length.

From Section 4, the preload force is selected as:

$$F_0 = 200\,N$$

and the maximum operating force is:

$$F_{max} = 600N$$

The force increase required over the working stroke is therefore:

$$\Delta F = F_{max} - F_0 = 400N$$

To achieve this force increase over a controlled and mechanically reasonable displacement, the working extension is assumed to be:

$$\Delta x = 20\,mm$$

This displacement represents a compromise between compactness and moderate stiffness, avoiding excessive forces for small ankle rotations.

5.3 Determination of Spring Stiffness

The required spring stiffness is calculated as:

$$k = \Delta F / \Delta x$$

Substituting the selected values:

$$k=20N/mm$$

This stiffness value ensures that the spring force increases smoothly from 200 N during standing to 600 N during walking, providing progressive resistance throughout the ankle range of motion.

5.4 Geometric Parameters and Assumptions

To dimension the spring, the following geometric and material assumptions are adopted:

- Mean coil diameter: D=20 mm
- Number of active coils: n=8
- Shear modulus of spring material:

$$G=79000N/mm^2$$

These values are consistent with standard extension spring designs and allow the spring to remain compact while achieving the desired stiffness.

5.5 Calculation of Wire Diameter

The stiffness of a helical spring can be expressed as:

$$k = \frac{Gd^4}{8D^3n}$$

Solving for the wire diameter d yields:

$$d = \sqrt[4]{\frac{8knD^3}{G}}$$

Substituting numerical values:

$$d = 3.37mm$$

To ensure manufacturability and provide a margin of safety, a standard wire diameter of:

$$d = 3.5mm$$

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is selected.

5.6 Stress Analysis

The maximum shear stress in the spring wire under axial load is calculated using:

$$\tau = \frac{8FD}{\pi d^3}$$

At the maximum operating force F=600 N, the resulting shear stress is:

$$\tau = 715.2 MPa$$

This stress level remains below the allowable shear stress for high-quality spring steel, confirming that the spring operates safely within elastic limits under the intended loading conditions.

5.7 Preload and Total Extension

The preload displacement required to achieve the initial force of 200 N is:

$$x_0 = \frac{F_0}{k} = \frac{200}{20} = 10mm$$

The total extension of the spring during operation consists of:

- Preload extension:$x_0 = 10\ mm$
- Working extension:$x_1 = 20\ mm$

Thus, the total extension is:

$$x_{total} = x_0 + x_1 = 30\ mm$$

5.8 Spring Length and Integration Constraints

The free (unstretched) length of the spring is selected as:

$$L_0 = 60\ mm$$

Accordingly:

- Spring length at rest (preloaded):

$$L_1 = L_0 + x_0 = 70mm$$

- Spring length at maximum extension:

$$L_2 = L_0 + x_{total} = 90mm$$

These dimensions are compatible with the available space within the prosthetic ankle mechanism and allow direct replacement of the original hydraulic actuator without structural modification.

5.9 Summary of Spring Design Parameters

The finalized spring characteristics are as follows:

- **Type:** Extension spring
- **Stiffness:** 20 N/mm
- **Preload force:** 200 N
- **Maximum force:** 600 N
- **Preload extension:** 10 mm
- **Working stroke:** 20 mm
- **Material:** EN1027-1 (SH) spring steel
- **Shear modulus:** 79,000 N/mm²

5.10 Design Implications

The designed spring satisfies all functional requirements for passive ankle actuation during standing and walking. It provides sufficient torque, maintains acceptable stress levels, and integrates seamlessly into the existing prosthetic architecture. By avoiding over-dimensioning and focusing on the relevant biomechanical load envelope, the spring design supports the overarching goal of reducing prosthesis mass while preserving functional performance.

A linear extension spring with preload is assumed. The force–displacement relationship is:

$$F = F_0 + k\Delta x$$

Where:

- F_0 is the preload force
- k is the spring constant
- Δx is the additional extension

5.11 Spring Constant

The force difference between standing and walking is:

$$\Delta F = F_2 - F_1 = 600\ N - 200\ N = 400\ N$$

The corresponding displacement is assumed to be:

$$\Delta x = 20\ mm = 0.020\ m$$

Thus, the spring constant is:

$$k = \Delta F/\Delta x = 400/0.020 = 20{,}000 Nm = 20\ N/mm$$

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VI. DISCUSSION

The proposed passive ankle actuation system provides sufficient torque for stable standing and walking while significantly reducing mass compared to hydraulic solutions. Although dynamic adaptation and energy return are limited relative to active systems, the simplicity, reliability, and low weight make this approach attractive for hybrid prosthetic designs.

VII. CONCLUSION

This study demonstrates that a preloaded extension spring can effectively replace a hydraulic actuator at the ankle joint for standing and walking activities. The designed spring meets biomechanical torque requirements while maintaining acceptable stress levels. This solution represents a practical and lightweight interim approach until advanced low-mass active actuators become widely available.

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