TESIS DOCTORAL

COMPENDIO DE PUBLICACIONES

A NOVEL, SYNERGY-BASED APPROACH FOR THE DESIGN OF CABLE-DRIVEN EXOSUITS FOR WALKING ASSISTANCE

UNA NUEVA FILOSOFÍA DE DISEÑO BASADA EN SINERGIAS PARA EXOTRAJES DE ASISTENCIA A LA MARCHA ACTUADOS POR CABLE

DOCTORADO EN INGENIERÍA MECÁNICA Y DE ORGANIZACIÓN INDUSTRIAL

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"Ninguna ciencia, en cuanto a ciencia, engaña; el engaño está en quien no la sabe."

- Miguel de Cervantes

1 Resumen

El creciente envejecimiento de la población en todo el mundo y en particular en occidente, lleva a un cada vez mayor número de personas que requieren algún tipo de asistencia a la marcha. La esperanza de vida en España, por ejemplo, se ha duplicado en apenas cuatro generaciones. A este respecto existen numerosas soluciones, entre las que destacan los exoesqueletos, dispositivos con elementos rígidos que el usuario porta y que, en general, conducidos por motores, apoyan la marcha de una u otra forma. Como alternativa a estos dispositivos ha surgido una variante mucho más ligera y cómoda para el usuario: el exotraje o exoesqueleto vestible. Este, provisto únicamente de elementos textiles y de un sistema de actuación, ofrece una solución más cómoda y económica a quienes, estando capacitados para andar, precisan de un dispositivo que asista su marcha o, alternativamente, desean ampliar sus capacidades motoras. La ausencia de elementos rígidos obliga al usuario a poder sostenerse por sí mismo, sin restringir ni obstaculizar en modo alguno la marcha natural del sujeto. Una variante recurrente de exotraje de apoyo a la marcha es aquel que consta, por un lado, de un sistema de actuación compuesto por los motores, sus controladores y sus sistemas de transmisión de potencia, generalmente situados en una mochila y que transmiten una fuerza a través de cables, que se encuentran vinculados a las articulaciones que se desea actuar a través de los llamados puntos de anclaje, de naturaleza textil en la mayor parte de los casos. Cabe esperar que sea preciso un motor por cada articulación y pierna a actuar: allí donde se requiera asistencia en cadera, rodilla y tobillo, serán necesarios seis motores, el principal elemento contribuyente al peso y al precio del sistema.

Esta tesis doctoral surge de la necesidad de reducir el precio y peso de los exotrajes de asistencia a la marcha, así como la de aumentar su grado de comodidad. La investigación se inspira en el desarrollo de una nueva filosofía de diseño de exotrajes que permita reducir drásticamente el número de actuadores necesarios para producir la asistencia deseada y, por ende, proporcionar una solución mucho más económica y al alcance de un número mucho mayor de personas. La investigación consta, por tanto, de dos pilares fundamentales. Por un lado, el desarrollo de un modelo dinámico inverso que permita predecir las fuerzas y pares a aplicar para asistir la marcha del usuario, permitiendo

por tanto el dimensionado del sistema de actuación y de transmisión de potencia y suponiendo el principal sustento del esquema de control del dispositivo. Por otro, la descripción detallada de una nueva filosofía de diseño que permita alcanzar los objetivos propuestos, sirviendo de guía para el planteamiento de nuevos exotrajes que asistan algunas de las articulaciones del miembro inferior durante la marcha humana. Es en esta última donde radica la clave que llevará a la mejora de la tecnología actual en los campos de coste y peso: el diseño mecánico basado en sinergias.

Tras un análisis estadístico de la marcha humana se obtiene, contrariamente a otros conjuntos de movimientos, como agarrar o sostener un objeto, que la marcha tiene una naturaleza cíclica y varios de sus parámetros presentan claras similitudes. El análisis cinemático realizado de los ángulos de flexión articular en cadera, rodilla y tobillo, entre otros, arroja resultados que presentan un notable grado de similitud entre ellos. Esta similitud es también observable en la evolución de las fuerzas en los cables requeridas para aplicar una fracción del par natural desarrollado por las articulaciones a lo largo del ciclo de la marcha, y en las propias extensiones de los cables. Estas similitudes hacen posible abordar, por medios estadísticos, la reducción de dimensionalidad del problema manteniendo un elevado grado de similitud con el problema original. Dicho de otro modo, el número de curvas de actuación necesarias para actuar las articulaciones deseadas, y por tanto, el número de motores, se puede reducir tanto como la propia dimensionalidad del problema. Así, se ha conseguido plantear un diseño que permite la actuación simultánea de hasta seis grados de libertad (tres articulaciones por pierna) con un solo motor.

Todo ello lleva a la principal aportación de esta tesis doctoral al desarrollo actual de exotrajes para el apoyo a la marcha humana: la descripción detallada, rigurosa y matemática de un método de diseño innovador que tiene como objetivo final la significativa reducción del peso y precio de estos sistemas, así como hacerlos más cómodos y accesibles a un mayor número de personas.

2 Abstract

The growing tendency of worldwide societies, but particularly in the West, to aging leads to a growing number of individuals that might require any kind of walking assistance. Life expectancy in Spain, for instance, has doubled in barely four generations. As such, there exist numerous solutions, among which exoskeletons must be highlighted as wearable devices with rigid elements that assist, generally driven by motors, human gait in any way. As an alternative to these devices, a new variant, much lighter and cheaper for the user has been proposed: wearable exoskeletons, or exosuits. They, made only of textile elements aside from the actuation system, offer a more comfortable and economic solution to those who can walk, but need an assistive device or, alternatively, wish to increase their motion capabilities: the lack of rigid elements forces the subject to sustain themselves, yet at the same time does not restrict or impede their natural gait in any way. A common variety of exosuits consists of an actuation system, including the motors and the power transmission system, generally included in a backpack and transmitting force via cables anchored to the joints in the so-called anchor points, in most cases textile elements located at the joints that will be actuated. A motor might be required for each target joint and leg: if all hip, knee, and ankle are to be actuated, six motors will be required, which will be the main contributors to the overall system weight and price.

This work arises from the need of reducing such weight and price, as well as increasing the wearability of walking-assistance exosuits. The document focuses on the development of a new design approach for exosuits that allows a heavy reduction in the number of actuators required to provide the desired assistance and, thus, offers a much more affordable solution, at the reach of many more potential users. The thesis is, thus, supported by two main cornerstones. On one side, the development of an inverse-dynamics model capable of predicting the forces and torques to be produced by the system to assist human gait, establishing a base for the system design and for its control. On the other side, stands the detailed description of the novel design approach that provides the desired improvements, serving as a guide for future exosuit designs for the lower limb. The latter is where lies the main contribution of this work, which

will try to improve current technology with regards to cost and weight: the mechanical design based on synergies.

Statistical analysis of human gait demonstrates a rather evident fact: as opposed to other movements, such as grasping an object or holding it, walking has a cyclic nature and presents clear similarities between several of its parameters. A kinematic study performed on the joint flexion-extension angles for hip, knee, and ankle, among others, yields a notable degree of similarity between them. The same happens in the cable forces during gait required to provide a certain fraction of the total joint torque, or in the cable extensions themselves. These similarities make it possible to approach, by statistical means, a dimensionality reduction problem while keeping a high degree of correspondence with the original variables. In other words, the number of required actuation curves, thus the number of required motors, can be significantly reduced, as much as the problem's dimensions. Therefore, a design allowing the actuation of six simultaneous degrees of freedom (up to three joints per leg) has been proposed with only one motor.

All of it leads to the main contribution of the work to the current walking-assistance exosuits design: the detailed, rigorous, and mathematical description of a novel method of design targeting the significant reduction in the weight and price of these devices, while also improving their wearability and making them more accessible.

3 Agradecimientos

Es una tarea ardua y por lo general poco agradecida, la de alcanzar las metas que hoy se exijen para la consecución del título de doctor. Un camino angosto que difícilmente puede ser recorrido con la sola guía de uno mismo. Es por ello que, una vez se llega a puerto, resulta preceptivo reconocer que tal logro no hubiera sido posible sin el apoyo y la aportación de un grupo a menudo numeroso de personas, que de un modo u otro han hecho posible no solo los méritos del trabajo que aquí se expone, sino del que escribe.

Al profesor Francisco Javier Alonso Sánchez, codirector de esta tesis y catedrático de Ingeniería Mecánica en la Universidad de Extremadura le agradezco especialmente su constante guía y apoyo en la travesía, tan académica como laboral y personal en que hace tres años me atreví a aventurarme. Numerosas de las ideas que aquí se defienden, y gracias a las cuales se han alcanzado los objetivos propuestos, hubieron de ser refrendadas y contrastadas con su criterio académico y ampliadas, si no corregidas, por sus frecuentes aportaciones y sabio consejo.

Comparto con el profesor Ignacio Herrera Navarro, catedrático del área de Mecánica de los Medios Continuos y Teoría de Estructuras de la Universidad de Extremadura, desde que, a comienzos de la década pasada, comencé mi andadura en la Universidad, una visión común de los retos y desafíos que esta Institución presenta. A través de sus lecciones terminé de afianzar mi vocación académica por la Ingeniería Mecánica y la Teoría de Estructuras y me mantuve en la incesante búsqueda de la verdad que a todo investigador, a todo crítico o pensador y quizás a toda persona, ha de guiar en su camino. Como codirector de esta tesis ha reforzado su papel como mentor en cuantas áreas tuvo la oportunidad, facilitando la labor que posibilita este trabajo, los que hubo antes, y a buen seguro los que habrá en el futuro.

A Juana María Mayo Núñez, catedrática de Ingeniería Mecánica de la Universidad de Sevilla y tutora de mi tesis doctoral, le agradezco su positiva valoración de mi trabajo a lo largo de los años y las ideas que durante nuestros encuentros intercambiamos. A los profesores Francisco Romero y Rafael Agujetas les agradezco su habitual consejo y ayuda para resolver los muchos baches que encontré en el camino, así como sus comentarios y aportaciones que ayudaron a mejorar enormemente la calidad de este trabajo. Al doctor Rafael Lorente Moreno le agradezco su participación como especialista del área de medicina y traumatología.

A mis compañeros Javier Bermejo y Ashwin Jayakumar, doctorandos del área y miembros de la tripulación de DÉDALO, con quienes he compartido la mayoría de horas en el camarote de Biomecánica, les agradezco y valoro su trabajo que, de manera conjunta con el mío, ha hecho y hace posible el llegar a cuantos puertos nos proponemos. Yo llego hoy al puerto deseado, pero en su travesía seguiré como siempre al pie del cañón.

Fue Luis Fernando Sánchez Domínguez del I.E.S Gabriel y Galán, mi profesor de Física en la ESO y Bachillerato, quien me introdujo en el campo de la mecánica, entre muchos otros, y quien me planteó tantos retos y desafíos como para descubrir que mi vocación era superarlos. Blanca Nieves Garrido Tello me advirtió de peligrosos obstáculos que hube de sortear. Don Nicolás Moriano Domínguez, del C.P. Los Arcos, puso tal empeño, dedicación y cariño en su trabajo que quienes lo disfrutaron no lo olvidan.

Les doy gracias a mi padre y a mi madre, quienes desde temprana fecha plantaron en mi mente la semilla del saber y su contraparte, el aprendizaje, del deseo por la superación personal y la ambición que, junto con la humildad, son los principales motores de mi actividad académica y laboral, personal y cotidiana. A mis hermanas que desde que tengo memoria me acompañan y viven mis malos y buenos momentos,

y a todas aquellas personas que de cualquier manera han moldeado, como el agua a la roca, a quien hoy tiene el placer de presentar esta tesis doctoral.

Daniel Rodríguez Jorge

4 Acknowledgements

Achieving the individual goals that lead to the presentation of a doctoral thesis is often a rather difficult, and sometimes unrewarding task. A treacherous journey that would be extremely difficult to go through alone. That is why, once the finish line is crossed, one must recognize that such an accomplishment would not have been possible without the aid and support of numerous individuals, who in one way or another, have made possible not only the merits of the project itself, but also, the merits of the author of these lines.

I especially wish to thank Professor Francisco Javier Alonso Sánchez, co-director of this thesis and Full Professor of Mechanical Engineering at the University of Extremadura for his constant guidance and support throughout my journey, academically, professionally and personally, since the moment I set off three years ago. Many of the ideas described and defended here, thanks to which the established goals were achieved, were verified with his vast academic experience and expanded upon, when not corrected, by his frequent reviews and expert advice.

I share a common point of view with Professor Ignacio Herrera Navarro, Full Professor of Continuum Mechanics and Theory of Structures at the University of Extremadura, on the current challenges faced by universities. Since I started my own University adventure at the beginning of the past decade, my vocation for Mechanical Engineering and Theory of Structures was reinforced through his lectures, and I started a constant search for the truth that must guide every researcher, critic, and perhaps every human being. As a co-director of this thesis, he has honoured his role as a mentor in every way he could, facilitating this research, the ones conducted before, and surely those to pursue in the future.

I am grateful to Professor Juana María Mayo Núñez, tutor of this thesis, for her positive evaluation throughout the years and the ideas we exchanged in our meetings. I thank professor Francisco Romero and professor Rafael Agujetas for their frequent advice and help to solve the many problems I encountered on the path, which greatly enhanced the quality of this work. I appreciate the participation of Dr. Rafael Lorente

Moreno as a specialist in the field of Medicine and Traumatology.

I thank and highly value the work of Javier Bermejo and Ashwin Jayakumar, Ph.D. candidates and members of DÉDALO's crew, with whom I shared many hours in the Biomechanics cabin. Thanks to this combined effort, it has been made possible to reach all the destinations we sail towards. Although I arrive at my harbour today, I will be with them at the frontlines, supporting them on their own journeys.

It was Luis Fernando Sánchez Domínguez, Physics teacher at the Gabriel y Galán high school, who introduced me to the fields of Mechanics and many others, who challenged me so often, that I ended up realizing that overcoming those challenges was my vocation. Blanca Nieves Garrido Tello warned me when dangerous obstacles approached me. Don Nicolás Moriano Domínguez, teacher at Los Arcos public school, dedicated so much effort, work, and passion to his job, that those who enjoyed it never forget it.

I am thankful to my parents, who from the very beginning planted in me the seed of learning and knowledge, the desire for self-improvement, and the ambition that, along with being humble, serve as drivers in my daily academic and professional activities. I also thank my sisters, who stood by my side from as early as I can remember, through the ups and downs of life,

and to all who in any way have shaped, just like water to a rock, the person who today has the pleasure of presenting this Ph.D. thesis.

Daniel Rodríguez Jorge

Contents

5 Introduction

Soft exosuits are a successful new type of walking-assistance device, mainly because of their higher degree of wearability and being lighter and cheaper than their bulkier, heavier, and more costly equivalent exoskeletons. In general, the acting force is transmitted from the actuation unit/s to the textile elements wrapped around the limbs surrounding the target joint/s, the so-called anchor points. The main transmission elements are cables, whether it is plain steel cables or Bowden cables, which are able to transmit forces via the relative movement of an inner cable located inside a hollow cable housing. The latter are more comfortable but add extra friction forces and non-linearities to the system. This setup does not necessarily restrain human movement, although the user must be able to walk on their own.

A first goal when designing an exosuit implies characterizing the forces and torques that are going to be involved in the actuation during gait. For this purpose, a 3D inverse dynamics model is proposed and integrated with a model for a cable-driven actuation to not only predict the required motor torque but also ensure that the actuation is providing the desired joint torque and cable force. Joint torques are to be shared between the user and the exosuit for different design configurations, focusing, as a first step, on both hip and ankle assistance, later expanding the actuation to the knee and for both legs. The model will support the design of the exosuit regarding aspects such as the location of the anchor points, whose position will greatly determine the torque and power needs, and also the cable system design, and the actuation units. The model will also be the cornerstone of the control scheme used during the exosuit's operation, being key to determining the actuation requirements at a given time, provided the instantaneous kinematic data acquired, for instance, via IMUs.The inverse dynamics analysis will be tested on a target population of ten adults, both male and female, obtaining their kinematic and ground contact force data from a public database. The obtained joint reactions and cable forces will be compared with those in the source database and the literature and prove the model to be accurate and adaptable to any subject, provided their anthropometric data. The model also allows the free positioning of the anchor points and predicts cable forces and extensions.

In general, a walking-assistance exosuit is expected to actuate all hip, knee, and ankle in both legs, which may lead to a design involving up to six actuation units. That might result in a rather heavy and expensive device. There have been several attempts to reduce the actuators and optimize the cable set up in other types of assistive devices, such as arm exoskeletons or hand gloves, among which the synergy-based model must be highlighted. In this work, an innovative design philosophy based on postural synergies is proposed for lower-limb exosuits, further developing this approach in the field and proposing a systematic way to apply and optimize exosuit designs. This is presented using a mathematical model that leads to an exosuit's actuation system design from a certain set of postural synergies, that is, the statistical study of the kinematic variables during gait. The similarities between those variables may provide the guidelines to reduce the number of total required actuators and, thus, the overall system weight and price. Traditionally, elevation or joint angles[1] have been used to reduce the problem dimensionality, but studying the cable extension instead leads to a heavier reduction. Thus, cable extensions are chosen as the target for the statistical analysis via principal component analysis (PCA), mainly due to the high resulting cumulative variance that is obtained with few principal components. This results in a general design proposal for economic, light exosuits for the lower limb.

[1]Elevation angles are defined as the angles between each segment and a global reference frame, whereas joint angles are computed as the angles between two consecutive segments. A more detailed description for the lower limb can be found in [1]

6 State of the Art

Currently, there is an increasing trend in the numbers among the elderly population in all societies, with a special focus in Western countries: according to the World Health Organization, more than 500 million people are affected by any kind of motion pathology worldwide [2]. Those who need walking assistance, whether due to age or other causes (spinal cord injury or stroke, among others) are expected to significantly increase in number in both Europe and every other area with elderly population growth. This trend is not expected to be reversed in the close future, specifically in Europe where the European population over 65 may very well grow by around 30% by the year 2080 [3]. Moreover, motion pathologies may have psychological and social consequences mainly due to the feeling of isolation and dependency that causes negative emotions associated with increased risk of health problems such as heart disease, depression, and cognitive decline. Lack of self-esteem, anxiety, or phobias may occur [4]. Most of those potential patients may not be able to afford appropriate motion and psychological treatment.

In recent years, several solutions have been proposed to improve the mobility of humans, or even increase their capabilities, among which exoskeletons and exosuits for any kind of assistance are found. The interest in exosuits has increased substantially in the past few years, as shown in [5], where a constant increase trend in related publications is reported, as shown in figure 1.

Exosuits produce force transmission to assist the human gait and are made mostly of textile elements, drastically reducing their weight and making them much more suitable, for instance, for elderly people who see their mobility reduced. The use of wearable exoskeletons can help such persons to improve their life quality by increasing their independence. The exosuit range of applications has notably grown during the last decade as a natural evolution of traditional exoskeletons, which offer a rigid solution with constrained degrees of freedom, while exosuits get rid of such inconveniences in exchange for a lower transmitted joint torque. Since exosuits dispense with rigid tutoring bars, their supporting role is assumed by the human body itself and cables behave as muscle-tendon actuation units, capable of transmitting forces to different body

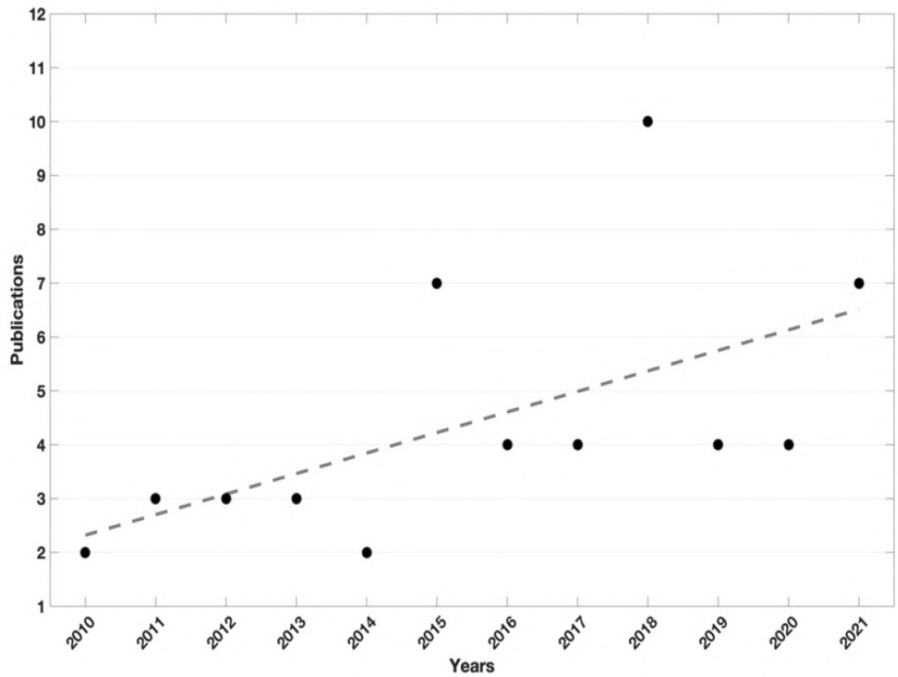

Figure 1: Trend in the number of publications related with upper and lower-limb exosuits in recent years according to [5]. Research is focused on human-exoskeleton interaction.

segments. The design of walking-assistance exosuits for humans can bring noticeable benefits to those who see their walking capacities reduced due to age or pathologies. Traditional exoskeletons are rigid, bulky, and generally heavy, and are gradually giving way to wearable exoskeletons or exosuits. Even though these are relatively new to the rehabilitation field, they offer a reliable and comfortable way to increase the user's moving capabilities without restricting movement or generating incompatibilities between the user and the exosuit's degrees of freedom [6]. The human skeleton itself becomes the support structure. The main advantages of these systems are the lower weight and the improved kinematic compatibility with the user, as opposed to exoskeletons [6]. Lacking tutoring bars not only decreases the weight of the device but also significantly improves its wearability. In [7] there is an excellent comparison between the conceptual schemes for exoskeletons, exosuits, and their hybrid solution, as in figure 2.

As a mandatory first step in the design and control of such systems, the calculation of net joint torques throughout the gait cycle must be conducted. In [8], this fact is underlined and a model for joint torque prediction based on Lagrangian mechanics is

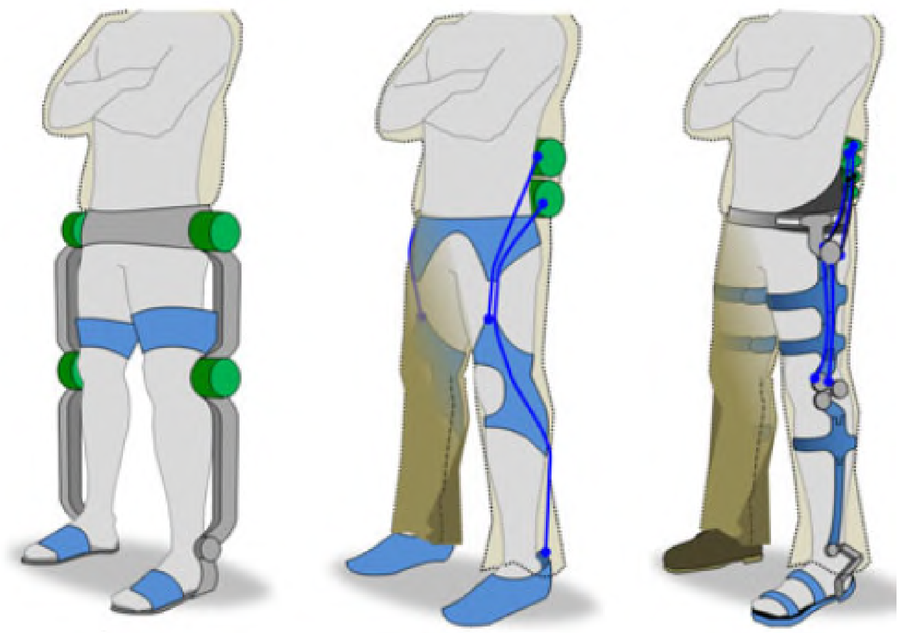

Figure 2: Comparison between rigid exoskeletons (left), exosuits (middle) and a hybrid proposal in [7] (right).

developed in the context of attending to a series of gait phases with different foot support conditions. The proposed model has seven segments and six joints, thus modeling both legs and the torso, obtaining anthropometric data from databases or equations as in [9], while the angular velocities are filtered via a kinematic Kalman filter. Equations are derived separately for each gait phase, which can be classified in different ways, as proposed in [8], separating single and double-support phases, as in figure 3.

In order to approach the problem from a mathematical perspective, two different points of view are especially interesting, namely, inverse dynamics models [10,11] which take kinematic data as input variables, and models where the problem is directly solved based on, for example, Lagrange Mechanics [12, 13]. While approaching the problem via the Lagrange Equations is less demanding in computational terms than the application of dynamic equations in multibody systems [12], inverse dynamics is an excellent method to estimate reaction forces and torques at each joint during gait without solving any differential equations. The results shown in [12], for instance, show how discontinuities can be found wherever there is a change in the gait phase, although the authors still provide a faster solution than forward dynamics, independent from the coordin-

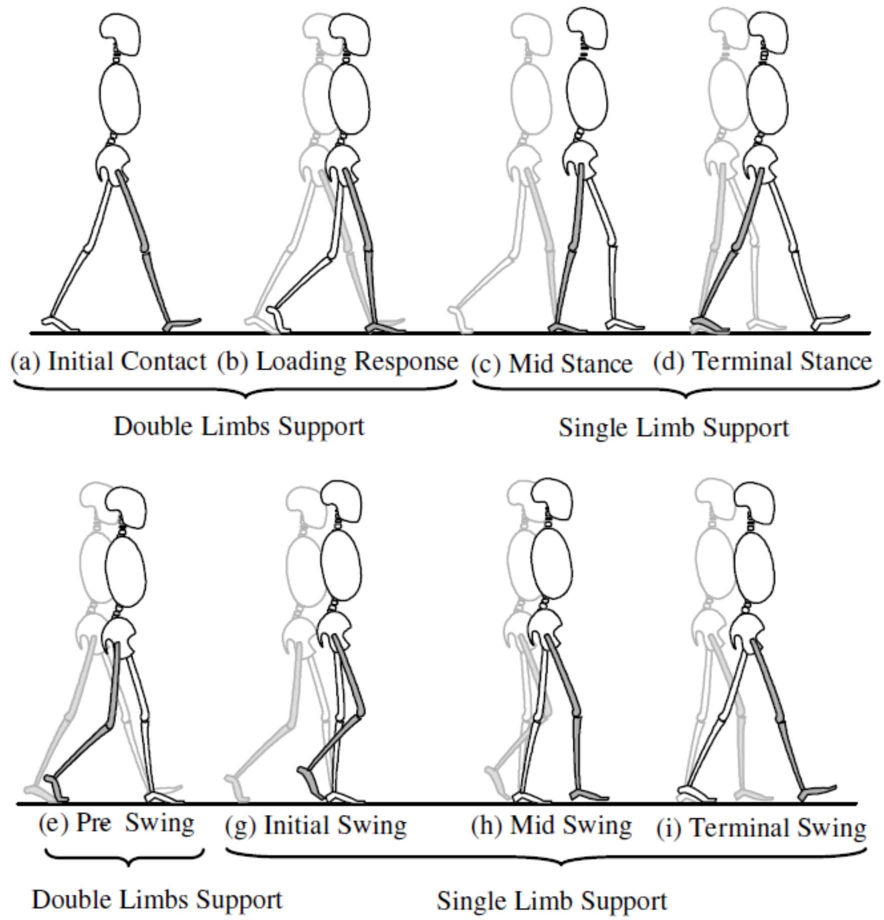

Figure 3: Gait phases in [8].

ate axes and requiring less gait information beforehand, such as the ground contact force. In [13], a complete description of Lagrangian dynamics is presented. Now when it comes to inverse dynamics models, the whole body model described in [10] gives good results when predicting sagittal forces and torques, although magnitudes related to out-of-plane motions did not yield precise results. The authors highlight many of the possible sources of error, such as the skin motion artifacts or the modeling of anthropometric data per subject (segment mass and inertia). The latter proved crucial, since the authors reported in [10] massive changes in their results when varying certain segment parameters. The model in [11] is rather hybrid, comprising a full-body 2D model and inverse dynamics elements to predict the force distribution. In [14], an inverse dynamic model is presented, with a focus on accelerometry to avoid measuring the ground contact forces during gait. Thanks to a multiple IMU per body segment configuration, similar results to those obtained via classical inverse dynamics plus force platform are achieved, although joint forces and torques are underestimated by about 20% due to cumulative IMU measurement errors. Applying the equations of motion to each segment, and joining all external forces, including ground contact forces, into a single variable vector, allows for a solution independent from the force platform but only dependent on a set of IMUS. An empirical analysis is performed on the hip using four triaxial accelerometers EGAXT-*-10, ENTRAN on a 72kg subject. The measured error of around 20% could be worsened when studying consecutive body segments instead of a single one, due to error propagation.

Results obtained by the Newton-Euler equations via inverse models offer a good correlation with those from multibody analysis, as proved in [15]. Here, the authors present an inverse dynamics model, assuming the gait kinematic data and the ground contact force as known quantities. Prediction of the joint torques and forces during gait is the main goal in seeking to aid the design of an exosuit for gait assistance. Results comparing inverse dynamics with a classic multibody methodology are included in [15] and shown in figure 4.

Figure 4: Comparison for joint reactions between an inverse dynamics approach and a traditional multibody methodology, as in [15].

A similar inverse dynamics model was presented in [10] for the whole body. Kinematic data such as position, acceleration, and segment angles can be measured using inertial measurement units and/or markers with a camera setup, as in [3], where a lower limb exoskeleton is proposed to reduce the energy cost of walking between 10% and 20%. Ground contact forces can be measured via sensors embedded in the shoes, as presented in [16], where a combination of air bladders and air pressure sensors are used to measure forces. This is a particularly relevant aspect in exosuits, since the ground contact force and its center of pressure greatly determine the results of the inverse dynamics analysis. Figure 5 shows the scheme for the "smart shoes" proposed in [16].

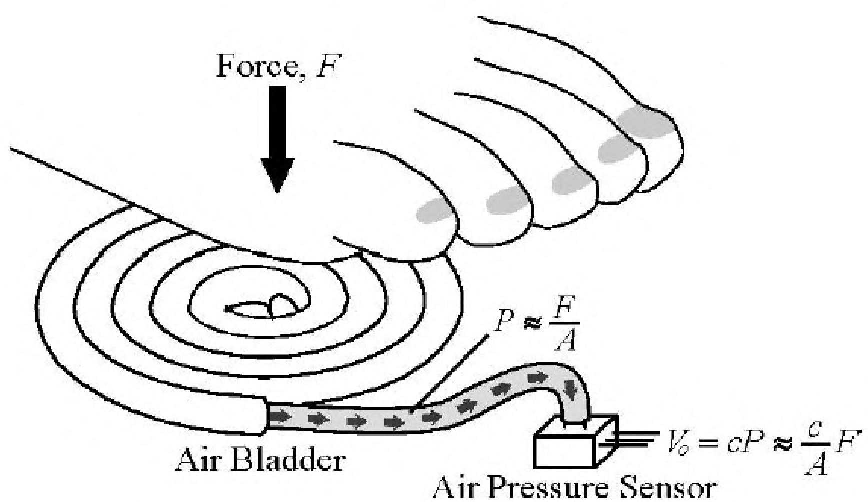

Figure 5: Ground contact force measurement system proposed in [16].

Inverse dynamic models have been extensively used when designing exosuits or exoskeletons for upper limb assistance, as in [17–19]. Inverse dynamics in the field of lower-limb assistive exosuits can be found in the recent literature, as in [3]. 3D inverse dynamics models, as in [10], make it possible to study the effects of the actuation at a certain joint upon its abd/adduction and rotation torques. Avoiding excessive undesired joint torques outside the sagittal plane may be a key design factor when approaching the positioning of the anchor points or the transmission system, among others. As the actuation depends on the results directly provided by the inverse dynamics model, the influence of gait and anthropometric data for a certain subject on the design or operation of the exosuit can be quantified as well.

As previously stated, there is an increasing number of exosuits under development. While there exist certain flexible exoskeleton proposals to assist the elderly, as in [7], soft exosuits with no support structure, as proposed in [20], provide the best-case scenario for near-zero suit-human interaction. For instance, there is a vast amount of recently developed assistive devices for the upper limb, including the arm itself and the hand [17, 21–24], most of which are soft exoskeletons or exosuits, and also for the lower limb, such as in [7, 25–27]. In [21], a slack enabling actuator is proposed for a cable-driven glove, getting rid of the pretension thanks to the proposed mechanism, shown in figure 6, along with the pulley-feeder. Furthermore, the number of actuators was also reduced thanks to the multi-pulley array.

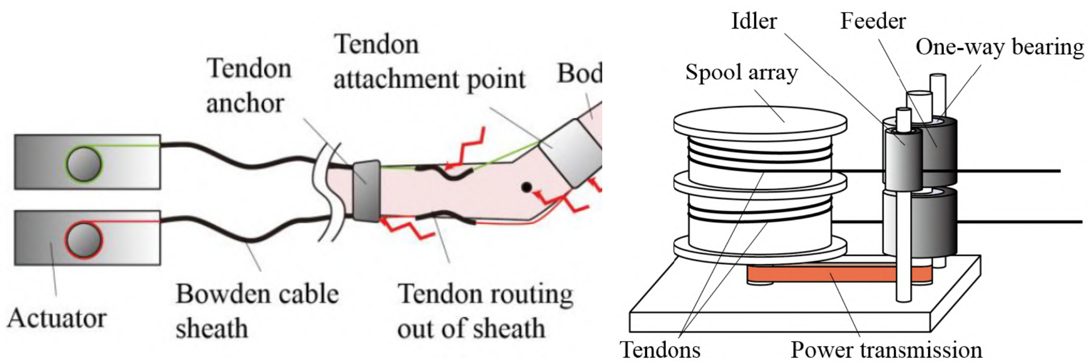

Figure 6: Slack enabling mechanism and pulley-feeder set up proposed in [21].

In [17], a cable-based exosuit for the upper limb was proposed using Bowden cables and a simplified model for the arm as in figure 7. The simplified model allows to easily obtain the cable elongations for both elbow flexion and extension, while a feeder and one-way clutches assembly enable the correct functioning of the system. The use of 2D models like the one in [17] provides accessible solutions for the force and torque characterization in wearable devices but does not provide information about out-of-plane motion and, perhaps, undesired reactions.

In [22], the advantages of exosuits over traditional exoskeletons for upper-limb assistance after neuromuscular disorders are highlighted: from the increased portability and cost to ergonomics and autonomy. Authors aim at proposing a low-cost device, a task they achieve thanks to their underactuated approach: instead of trying to actuate every single degree of freedom in the human hand, which would result in a system with several actuators, they focus on the hand's postural synergies to move 9 degrees of freedom with only one motor. In [24], an exosuit for elbow flexion-extension assistance is

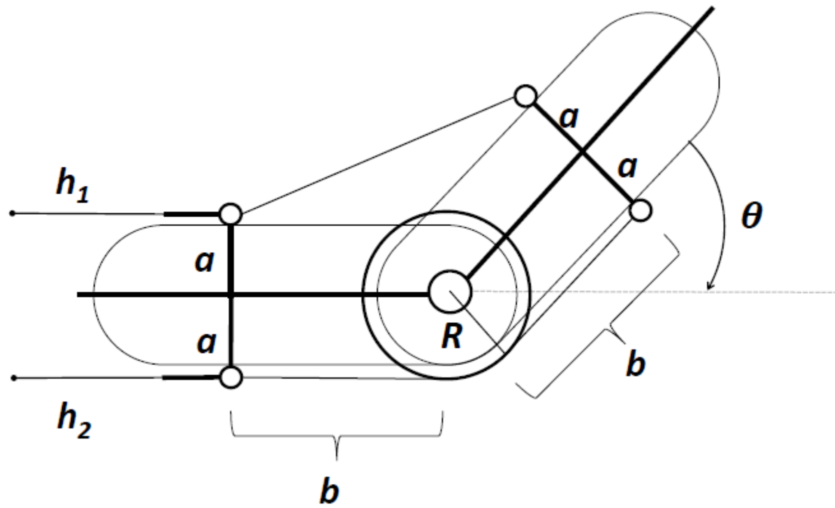

Figure 7: Simplified model for the elbow to assist the design of upper-limb assistance wearable devices proposed in [17].

proposed, with the actuation system located in a backpack and making use of Bowden cables. A scheme is shown in figure 8.

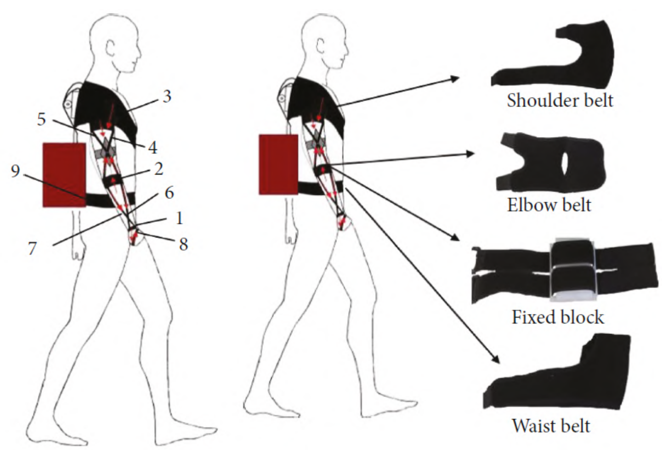

Figure 8: Actuation scheme for an upper-limb assistance exosuit in [24]

Exosuits for lower-limb assistance are also a research target in recent years. For instance, a platform to assist multiple joints was developed in [25], managing to provide up to 15% of the biological torque using only textile elements and cables. In [28], for instance, a practical application of lightweight exosuits for the lower limb is presented, enabling active rehabilitation in patients with various gait disorders. According to the authors, training via their proposed lower-limb exosuit was feasible, even though patients required some time to adapt. Another use that has been explored for lower limb exosuits is reducing the metabolic cost and muscle fatigue of the wearer. In [29], a light exosuit for hip assistance of around only 2.5 kg was developed and tested, showing how the metabolic consumption could decrease by around 8%. However, it is also shown how this decrease has to take into account the increase in metabolic cost inherent to wearing an exosuit. Similar studies are presented in [30, 31], all aiming at reducing the metabolic cost of walking, proving that whether it is for rehabilitation, assistance or simply to reduce the cost of walking in healthy subjects, weight and wearability take a key role in the design of lower-limb exosuits. Authors in [31] report lower metabolic costs using their assistive, soft exosuit. The exosuit is able to actuate multiple joints, as in [26], where a multi-joint soft exosuit capable of assisting both ankle and hip flexion is proposed. The achieved cable forces of up to 300N and 150N in the ankle and hip respectively manage to produce around 20% of each joint's biological torque. Joint synergies are considered to reduce the weight by only using two actuators for both joints and legs with two separate actuation paths, as in figure 9. Each load path is actuated with the same actuator, with a transition period to change between them.

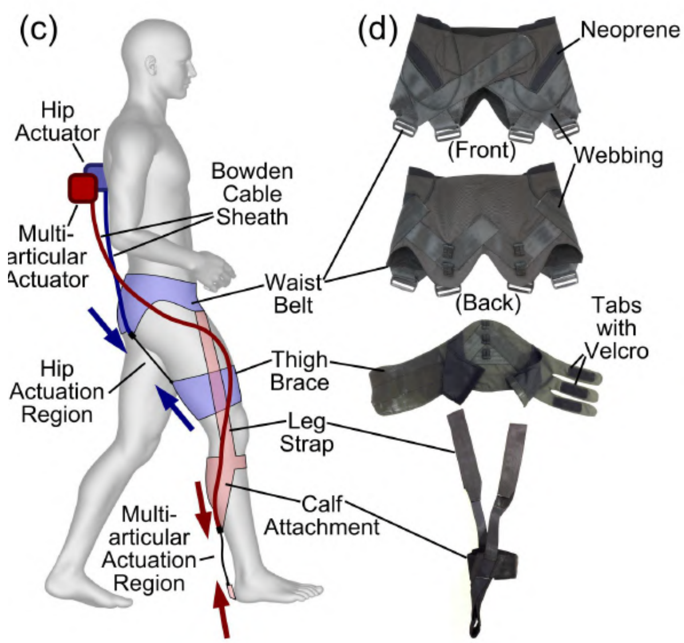

Figure 9: Multi-articular load paths for a lower-limb soft exosuit proposed in [26]

The exosuit was tested in outdoor environments and in a population of 20 individuals walking on a treadmill. It is stated that the use of the exosuit did not substantially disrupt the normal gait. Authors admit that heavy systems can increase the metabolic cost of walking, thus emphasizing the importance of decreasing the total system weight. Further development in the synergy-based approach may lead to an even greater reduction in the required number of actuators. Focusing on hip assistance, their team published in [27] a fully-portable exosuit proposal that achieved 30% of the total torque in the sagittal plane. The whole actuation system is located in a backpack, and the force is transmitted via cables to the anchor points located in the thighs, as in figure 10.

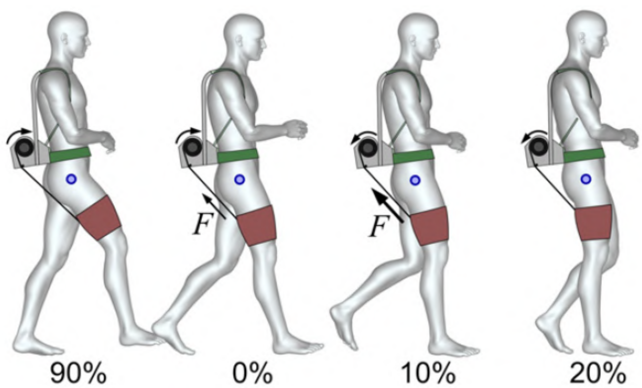

Figure 10: Actuation scheme for the hip-assistance exosuit proposed in [27]

This work is also relevant for considering the human-body contribution to system stiffness, managing to translate the human-suit interaction, mainly located at the hip and shoulders, as in figure 11, to a mathematical expression ready to be included in the control scheme. A more precise prediction of the cable extensions during gait is also possible thanks to their model.

For the chosen subjects, the authors reported cable forces of up to 150N to achieve the 30% joint torque, resulting in notable human-suit elongations, an effect that should be considered when approaching the system's control. Authors in [7] propose a hybrid solution that tries to cope with the disadvantages of both rigid exoskeletons and exosuits.

Reviewing the state of the art for walking-assistance exosuits reveals their most visible

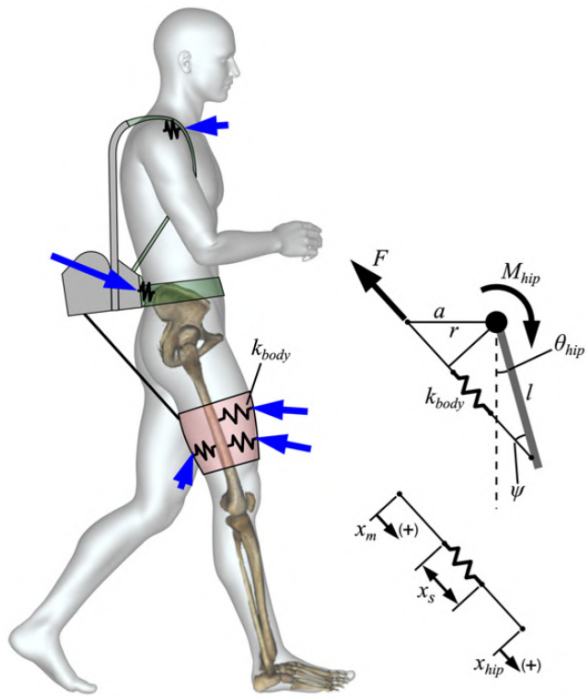

Figure 11: Human-suit interaction and equivalent stiffness model in [27]

limitation: the transmitting force. They will only be able to assist around 30% of each joint's total required torque, at best. This barrier comes from the way power is transmitted in a cable-driven actuation system and the high contact forces caused by the human-cable interaction. Thus, exosuits' major benefit is providing additional support during everyday activities, such as walking, grabbing, and reaching objects or stabilizing gait under pathological conditions, or for the elderly. The problem of reducing the price and weight, and increasing the device's wearability has also been the main concern in the development of the field. Many attempts have been made to reduce as much as possible the number of actuators, as in [18]. Here, several degrees of freedom of the arm are actuated using a single motor, thanks to a combination of clutches and brakes as in figure 12.

Traditionally, a single actuator was used to actuate each degree of freedom, although new designs opting for a lower number of actuators, key elements regarding the total weight, have been proposed, as in [32]. Furthermore, there exists a new design philosophy for exosuits that assist upper or lower limb joints: the so-called synergy-based

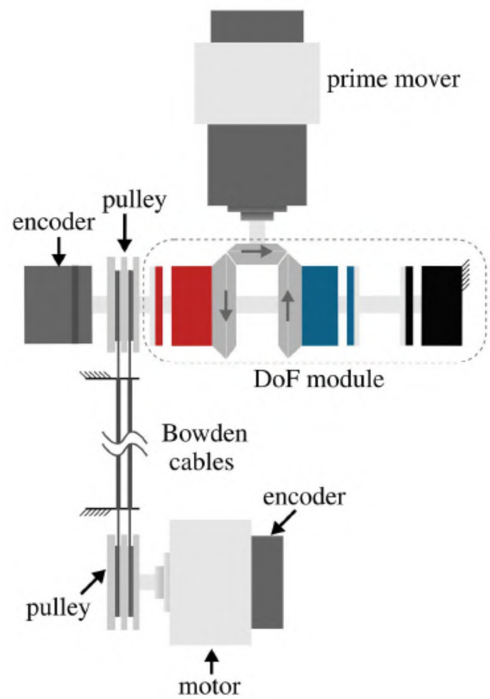

Figure 12: Transmission scheme in an underactuated exosuit for arm assistance proposed in [18]. Red and blue elements are clutches, black element is a brake.

approach. Using statistical analysis to determine the best way to approach the design of a mechanical system itself is not new, and similar applications such as hand assistance can be found in the literature. In [33], five fingers are assisted with only two actuators, significantly decreasing the weight and price of the device. In [34], a synergy-based actuation scheme for hand assistance is proposed, leading to the design of a soft assistive device for daily use. Lagrange equations yield the required motor torque and the tendon force and perceived stiffness at the joints. The underactuated device is complemented with a clutch, so that hand posture can be maintained at very low power consumption. The general actuation scheme is that of figure 13, where the pulley radii are calculated based on the hand postural synergies.

A testbench is also presented to check the motor's capability to produce the desired torque and velocity and to measure tendon forces when actuating based on the first postural synergy exclusively. The testbench in [34] features a 25W DC brushless motor (Maxon EC-max) with an encoder MR, 512 CPT), as well as a planetary gear with a

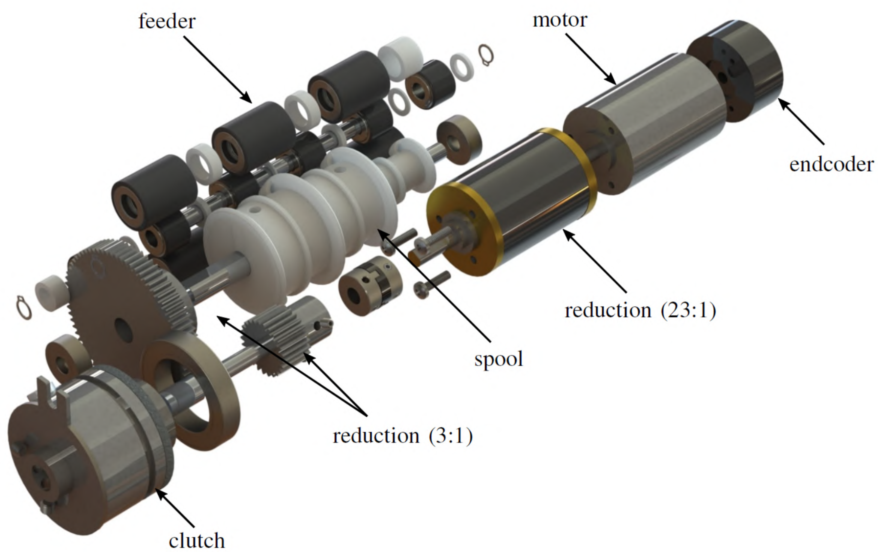

Figure 13: General actuation scheme for a hand-assistive, synergy-based soft device [34]

reduction of 23:1. A simplified scheme for the testbench actuation was shown, as in figure 14

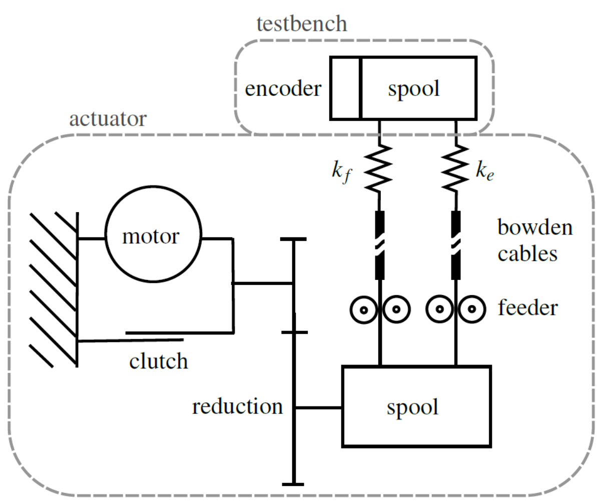

Figure 14: General actuation scheme for a hand-assistive, synergy-based soft device [34]

Synergies themselves are well known and have been commonly used for the past 30 years to decrease the dimensionality of any problem, including human movement. In applied biomechanics, and more specifically, in the design of exosuits, it is particularly relevant as a way of studying the coupling between degrees of freedom for motor control [35–39]. In such a field, for instance, kinematic synergies are investigated in [39] for 36 human movements, including walking, jumping, or sitting down/standing up. The authors report common basic patterns do exist among all human movements and they are able to reconstruct them with a high degree of variance from just a few of those patterns. The authors state that their work is aimed at inspiring the design and development of artificial limbs.

Now, since weight and wearability are determining factors when it comes to assisting human gait in pathological and older users, it is the task of this thesis to further expand the application field of synergy analysis and propose a low-weight, low-cost exosuit concept to assist hip, knee, and ankle during gait. That is, based on a new approach based on kinematic synergies, a novel design is proposed to significantly reduce the number of required actuators while still assisting both legs and multiple joints, thus reducing the price and weight of the device and increasing its wearability. It was certainly those benefits, lower weight, and price, as well as improved wearability, that inspired this work: extending the synergy-based approach to lower limb assistance, a field that benefits humankind on many levels.

7 Objectives

To develop a new design proposal for a walking-assistance exosuit, there are certain steps that must be accomplished before reaching a working, functional proof of concept. At first, a mathematical model to predict forces and torques based on the gait parameters is required. Such a model must be able to, given the assistance requirements and the gait kinematic information, predict the biological joint torques and obtain the expected cable forces and motor torques during gait. Having that model will provide, on one hand, a way to design the actuation unit and the rest of the exosuit's elements, and on the other hand, a basis for the control scheme.

In this work, an inverse dynamics model is chosen as a viable, cost-efficient method to predict forces and torques at the joints during gait and, consequently, determine the key variables involved in the actuation, including the cable forces and elongations, the torques in the actuation unit and power requirements, etc.

Once the pre-design is done, a novel design approach is presented and developed in this work, aiming at lower cost and weight, and more wearable exosuits. The synergy-focused proposal allows the position control to follow a reconstructed, simpler gait pattern, thus heavily reducing the number of required actuation units. In this work, the general approach is defined and presented to be used for any kind of lower-limb exosuits, providing a solution that may be useful for many.

8 Thesis content

This thesis includes a compilation of **three** research papers, in which the complete process to reach the proposed design is presented. The three of them were published between 2021 and 2022 in internationally recognized journals, and received positive feedback from editors and reviewers, who described them as "interesting and [...] useful for more efficient future designs for lower-limb exosuits". Said papers are:

- PAPER I: Rodríguez Jorge, D., Jayakumar, A., Bermejo García, J., Romero, F., and Alonso, F.J. 2021 "Modelo dinámico inverso para el apoyo al diseño de un exoesqueleto vestible de asistencia a la marcha". *Revista Iberoamericana de Ingeniería Mecánica* **25**, no. 2, pp. 75–84

- PAPER II: Rodríguez Jorge, Daniel, Javier Bermejo García, Ashwin Jayakumar, Rafael Lorente Moreno, Rafael Agujetas Ortiz, and Francisco Romero Sánchez. 2022. "Force and Torque Characterization in the Actuation of a Walking-Assistance, Cable-Driven Exosuit" *Sensors* **22**, no. 11: 4309. https://doi.org/10.3390/s22114309

- PAPER III: Jorge, D. R., Bermejo-García, J., Jayakumar, A., Romero-Sánchez, F., and Alonso, F. J. (June 30, 2022). "A Synergy-Based Approach for the Design of a Lower-Limb, Cable-Driven Exosuit." ASME. *J. Mech. Des.* October 2022; **144**(10): 103302. doi: https://doi.org/10.1115/1.4054768

9 Results and discussion

As a first step in the design of exosuits, an inverse dynamics model was proposed with six degrees of freedom per segment. The model takes the subject's kinematic data and ground contact forces during gait as inputs, and provides the joint reaction forces and torques. In PAPER I, the inverse dynamics model was combined with a simple geometric model for each joint to obtain preliminary results of cable forces and motor torques during gait. These results allowed a 95.6 mNm motor with a 1:33 reduction to be selected to enable actuation of 30% of the total hip joint torque. A notable gait phase comprised between around 25% and 75% yields compression forces at the cables, thus making actuation only possible outside of such phase. PAPER II greatly extends the results and applications of the presented model, introducing a general exosuit proposal for all hip, knee, and ankle actuation at both legs and extending the target population to a ten-individual population of older adults selected from the data available in [40]. The exosuit proposal contemplates an actuation unit located in a backpack, from which three separate actuation systems emerge: one for each target joint. Steel cables were used to transmit forces to the anchor points around the joints. Bowden cables are commonly used for the design of exosuits, yet with the consequent non-negligible efficiency loss due to non-linear friction between the cable and the sleeve [41, 42]. The population subjects were walking over the ground at comfortable speeds, to better match with the real conditions under which walking-assistance exosuits are assumed to work, instead of walking at constant speed on a treadmill. Joint torques were computed for all subjects during comfortable gait, yielding accurate results when compared with the data published in the same database.

Along with a fully 3D exosuit cable scheme, the model is able to provide cable forces and torques in the actuation unit, once the percentage of joint torque to be actuated is known. The model can also serve as the core of the control scheme, changing the markers used in for the tested population by inertial measurement units, for instance, to provide real-time kinematic data. Cable forces are dependent not only on the subject but also on the position of the anchor points: for a fixed position, maximum forces from around 40N to 150N are found to produce the same hip torque in different subjects.

The model also provided the required motor torques to actuate the hip and ankle for the population, whose peaks ranged between 20 to 60 mNm for the hip and from 60 to 115 mNm for the ankle approximately. Cable extensions are also obtained with and without considering the human body stiffness and its interaction with the suit, reaching similar results to those published by researchers developing other assistive exosuits, as in [27]. Parametric studies to analyze the influence of the involved parameters are also possible, such as the boundary conditions (anchor points location) or the anthropometric data. Results for the applied joint torque at each joint as functions of the anchor points location are shown. Initially, the hip and ankle are selected as target joints for the actuation, because they are the main power contributors during gait, as per [26, 43].

The obtained results are used in PAPER III to propose an innovative, low-cost low-weight design proposal for exosuits. Cable extensions obtained through the inverse-dynamics model for the selected population showed a rather impressive but expectable result: all cable extensions for hip, knee, and ankle at both legs followed very similar patterns throughout gait. That makes them eligible to perform a dimension-reduction algorithm based on principal component analysis (PCA) to potentially reduce the number of required actuators. The PCA study that was conducted highlights which type of synergies are more beneficial when it comes to approaching the design of a walking-assistance exosuit and stresses how much it might reduce the problem's dimensionality. A synergy-based study on walking variables yields interesting results: it has already proven very useful for arm and hand movement, and gait has a more cyclic nature than the latter. Furthermore, a general approach to how to translate postural synergies into actuation systems and mechanisms is presented, introducing a general transmission device that would allow for a general, synergy-based actuation. It can be useful for scenarios where more than one actuator is needed and provides a simple way to actuate a number of joints (to be chosen by the designer in each case) with a lower amount of motors. The PCA conducted on the target population yielded a 94% cumulative variance with only one PC when targeting six degrees of freedom, namely hip, knee, and ankle at both legs. Results were tested using a developed 3D model for a sample exosuit and on a testbench, designed taking into account the power requirements and forces predicted by the model developed beforehand and using steel cables.

Even though a general design proposal was presented, cable extension synergies proved to be an excellent way to reduce the problem's dimensionality: with only one single motor, gait was reasonably reconstructed. Furthermore, a testbench was built to reproduce gait at the hip, knee, and ankle for both legs using a single motor, using a transmission system with clutches. It was used to prove that the gait was just slightly worse reproduced with one motor than when using more while cutting the overall system price and weight considerably.

10 Conclusions

An innovative, synergy-based design philosophy was proposed for walking-assistance exosuits along with an inverse dynamics model to work as its control core and predict cable forces and motor torques. Cable extensions were obtained for a selected population and used as target variables in a PCA aiming at reducing the number of required actuators. A symbolic analysis of kinematic variables was conducted using PCA, which led to an expression for pulley radii and motor control schemes achieving around 94% of the total variance with only one actuator for six joints and a given population of elderly subjects walking comfortably over the ground. A general transmission system was conceptualized to actuate any number of joints with any number of simultaneous actuation units, along with an exosuit design based on cable-driven actuation. Reducing the number of actuators in such a drastic manner opens the possibility of actuating both legs with only one motor, once the actuation phases during gait are determined. Results were tested on a testbench designed to reconstruct the cable extension function for one joint (hip or ankle) in an actuation scheme built to actuate both joints simultaneously: removing one actuator, the one corresponding to the second and last principal components (PC), did not have a great impact on the result, as predicted by the statistical model. The design proposal is useful and applicable to walking-assistance or rehabilitation exosuits that are driven by cables and actuate any of the lower-limb joints and simplifies their transmission systems while also reducing the number of motors. Thus, this work helps improve the wearability and affordability of such assistive devices and is a step forward in making them accessible to all that need them.

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Appendix

The main content of this thesis is based on the following journal papers:

- PAPER I: Rodríguez Jorge, D., Jayakumar, A., Bermejo García, J., Romero, F., and Alonso, F.J. 2021 "Modelo dinámico inverso para el apoyo al diseño de un exoesqueleto vestible de asistencia a la marcha". *Revista Iberoamericana de Ingeniería Mecánica* **25**, no. 2, pp. 75–84

- PAPER II: Rodríguez Jorge, Daniel, Javier Bermejo García, Ashwin Jayakumar, Rafael Lorente Moreno, Rafael Agujetas Ortiz, and Francisco Romero Sánchez. 2022. "Force and Torque Characterization in the Actuation of a Walking-Assistance, Cable-Driven Exosuit" *Sensors* **22**, no. 11: 4309. https://doi.org/10.3390/s22114309

- PAPER III: Jorge, D. R., Bermejo-García, J., Jayakumar, A., Romero-Sánchez, F., and Alonso, F. J. (June 30, 2022). "A Synergy-Based Approach for the Design of a Lower-Limb, Cable-Driven Exosuit." ASME. *J. Mech. Des.* October 2022; **144**(10): 103302. doi: https://doi.org/10.1115/1.4054768

A complete copy of each paper is shown on the following pages. PAPER III is included in its accepted version.

Revista Iberoamericana de Ingeniería Mecánica. Vol. 25, N.º 2, pp. 75-84, 2021

MODELO DINÁMICO INVERSO PARA EL APOYO AL DISEÑO DE UN EXOESQUELETO VESTIBLE DE ASISTENCIA A LA MARCHA

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(Recibido 15 de septiembre de 2020, para publicación 7 de enero de 2021)

Resumen – En el marco del desarrollo de dispositivos para la asistencia de la marcha, destacan por su elevado grado de vestibilidad y menor peso y coste, aquellos conformados simplemente por los actuadores y una serie de puntos de anclaje localizados en las proximidades de las articulaciones a actuar. El esfuerzo se trasmite desde los actuadores hasta los puntos de anclaje a través de cables, aumentado el grado de confort del usuario al prescindir de los tutores que comúnmente se encuentran en exoesqueletos. Se aborda en este trabajo un análisis dinámico inverso, para obtener los pares articulares necesarios a lo largo de la marcha, como paso previo al dimensionamiento de los componentes del exoesqueleto a diseñar. Se plantea un modelo biomecánico plano con siete segmentos corporales (tres por pierna y el tronco) y los pares en cadera, rodilla y tobillo como principales acciones. Posteriormente, estos pares se distribuyen entre el sujeto y el exoesqueleto actuado por cable (*exosuit*) para distintas configuraciones de diseño. Este modelo, permite guiar el diseño del exoesqueleto en lo que respecta a factores tales como la posición de los puntos de anclaje, el número de actuadores preciso o la disposición de los cables, entre otros. A la hora de encontrar un diseño adecuado para la actuación y transmisión en el exoesqueleto, será posible, por ejemplo, tomar en consideración sobre qué articulación desea actuarse con mayor intensidad en función de las necesidades del proyecto, o incluso determinar si no ha de actuarse en absoluto sobre alguna de ellas. Podrá analizarse, además, la influencia de dichos parámetros de diseño sobre los parámetros de la transmisión y estudiarse la efectividad de los diseños propuestos y los encontrados en la literatura.

Palabras clave – Exoesqueleto vestible, actuación por cable, análisis dinámico inverso, marcha humana.

1. INTRODUCCIÓN

El diseño de exoesqueletos para asistir la marcha en seres humanos puede conllevar notables beneficios para aquellas personas que, por su edad o patologías, vean impedidas en cierta medida sus capacidades motoras. Los exoesqueletos tradicionales, rígidos y por lo general pesados, están dando paso a los exoesqueletos vestibles o *exosuit*. Aunque son relativamente nuevos en el ámbito de la rehabilitación, suponen una forma fiable y cómoda de aumentar las capacidades de movimiento del usuario, sin restringir su movimiento ni generando incompatibilidades entre los grados de libertad de exoesqueleto y usuario, siendo su propia estructura rígida la correspondiente al dispositivo de rehabilitación. Estos, además, hacen posible la transmisión de fuerzas para asistir la marcha, pero se componen mayoritariamente de elementos textiles, que reducen drásticamente su peso y los hacen mucho más accesibles a personas de la tercera edad que vean reducida su movilidad. Y es que la población que, por edad u otras causas, necesitan asistencia en la marcha, continuarán aumentando en España y su entorno conforme la población mayor lo haga, estimándose que para 2080 el 29,1% de la población europea superará los 65 años [1]. El uso de exoesqueletos vestibles en estas personas puede ayudar a mejorar su calidad de vida, aumentando su grado de independencia. En los últimos años se han planteado varias soluciones para mejorar la movilidad de las personas a través de exoesqueletos vestibles, ya sea para las extremidades superiores, [2] y en las inferiores, [3, 4].

Este artículo se enmarca en el desarrollo de un exoesqueleto vestible para las extremidades inferiores, que aspira a asistir la marcha en personas de la tercera edad.

Como paso imprescindible para el diseño y control de estos sistemas, destaca el cálculo de los pares articulares a lo largo del ciclo de la marcha. En [5] se destaca este hecho y se plantea un modelo para la predicción de pares articulares atendiendo a una serie de subfases en el ciclo de la marcha, atendiendo a las distintas condiciones de apoyo del pie. Se plantean, además, diversos puntos de vista a la hora de plantear el problema desde el punto de vista matemático, destacando dos: los modelos dinámicos inversos [6, 7], que parten de datos cinemáticos conocidos, y los basados en las ecuaciones de la mecánica de Lagrange [8, 9], que no precisan de los datos de contacto con el suelo y son menos exigentes computacionalmente que el mero planteamiento de las ecuaciones dinámicas en modelos multi cuerpo [8]. Los resultados que arrojan las ecuaciones de Newton-Euler en modelos dinámicos inversos ofrecen una buena correlación con los que se extraen de análisis multi cuerpo o *multibody* [10]. Aquí se presenta un modelo dinámico inverso apoyado en los datos cinemáticos de la marcha y la fuerza de contacto con el suelo, para la estimación de pares y fuerzas articulares de cara a apoyar el diseño de un exoesqueleto vestible para la asistencia a la marcha.

2. DESARROLLO DEL MODELO

El objeto de este capítulo es la derivación de las ecuaciones dinámicas que permitan la obtención de las fuerzas y pares articulares a partir de los datos cinemáticos de la marcha y la evolución de la fuerza de contacto entre el pie y suelo. Basta para ello con la aplicación, respecto de un sistema de referencia inercial y para cada uno de los tres ejes y segmentos que forman la parte del cuerpo que se desea modelar, de las ecuaciones:

$$\sum \mathrm{F_i} = m_\mathrm{i} a_\mathrm{i} \tag{1}$$

$$\sum n_\mathrm{i} = \mathrm{I_{z,i}} \ddot{\theta}_\mathrm{i} \tag{2}$$

Siendo el objeto la obtención de las fuerzas y pares articulares en tobillo, rodilla y cadera, se opta por un modelo simple tridimensional de la pierna como la composición de tres segmentos, unidos entre sí y cada uno de ellos solidario a un sistema propio de referencia local, que servirá para definir los vectores de posición que para el planteamiento de las ecuaciones sean precisos. Los enlaces articulares admiten movimiento

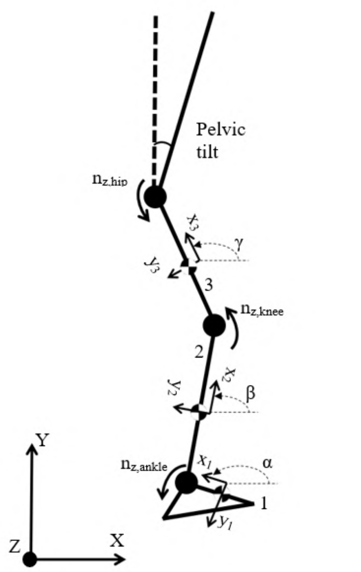


Fig. 1. Esquema simplificado de los tres segmentos considerados para el miembro inferior.

y giro en los tres ejes, pero producen reacciones en los seis posibles grados de libertad. La Fig. 1 muestra los elementos más esenciales de dicho modelo, obviando para mayor claridad las fuerzas articulares y los pares distintos del par en *z* (de importancia capital para el proyecto que nos ocupa).

Una vez todas las variables geométricas del problema estén definidas en base a los sistemas de referencia locales, bastará premultiplicar aquellas que sean vectoriales por la matriz de cambio de base $\mathbf{R_i}$:

$$\mathbf{R_i} = \begin{pmatrix} \mathrm{Cos}(\theta_i) & \text{-}\mathrm{Sen}(\theta_i) & 0 \\ \mathrm{Sen}(\theta_i) & \mathrm{Cos}(\theta_i) & 0 \\ 0 & 0 & 1 \end{pmatrix} \quad (3)$$

donde θ_i es α, β o γ, según corresponda. Así pues, las ecuaciones (1) y (2) son un conjunto de 18 ecuaciones diferenciales (9 de fuerza y 9 de par, seis en total para cada uno de los segmentos) que pasan a ser algebraicas si los datos de posición de cada punto (centros de gravedad de cada segmento y articulaciones), ángulos de cada segmento y sus aceleraciones lineales y angulares son conocidas. Se incluyen a continuación las ecuaciones 1 y 2 particularizadas para cada segmento.

- Segmento 1 o del pie.

$$\mathrm{F_{a,i}} = m_1\left(\mathrm{a_{1,i}}\text{-}\mathrm{g_i}\right)\text{-}\mathrm{F_{r,i}} \qquad i=1,2,3 \quad (4)$$

$$n_{a,i} = I_{1,i}\ddot{\theta}_1\text{-}\left(\mathbf{r_{ga,P}}\times\mathbf{F_r}\right)\big|_i\text{-}\left(\mathbf{r_{ga,A}}\times\mathbf{F_a}\right)\big|_i \qquad i=1,2,3 \quad (5)$$

donde Fr,i es la componente *i* (*x*, *y* o *z*) de la fuerza de contacto entre la planta del pie y el suelo. El subíndice a indica *ankle* o tobillo. El vector $\mathbf{r_{ga,P}}$ une el centro de gravedad del pie con el punto de aplicación de la fuerza de contacto con el suelo. El vector $\mathbf{r_{ga,A}}$ une el centro de gravedad del pie con la articulación del tobillo.

- Segmento 2 o de la parte inferior de la pierna.

$$\mathrm{F_{k,i}} = m_2\left(\mathrm{a_{2,i}}\text{-}\mathrm{g_i}\right)+\mathrm{F_{a,i}} \qquad i=1,2,3 \quad (6)$$

$$\mathrm{n_{k,i}} = \mathrm{I_{2,i}}\ddot{\theta}_2\text{-}\left(\mathbf{r_{gk,k}}\times\mathbf{F_k}\right)\big|_i+\left(\mathbf{r_{gk,A}}\times\mathbf{F_a}\right)\big|_i+\mathrm{n_{a,i}} \qquad i=1,2,3 \quad (7)$$

El subíndice *k* indica *knee* o rodilla.

- Segmento 3 o de la parte superior de la pierna.

$$\mathrm{F_{h,i}} = m_3\left(\mathrm{a_{3,i}}\text{-}\mathrm{g_i}\right)+\mathrm{F_{k,i}} \qquad i=1,2,3 \quad (8)$$

$$\mathrm{n_{h,i}} = \mathrm{I_{3,i}}\ddot{\theta}_3\text{-}\left(\mathbf{r_{gh,h}}\times\mathbf{F_h}\right)\big|_i+\left(\mathbf{r_{gh,k}}\times\mathbf{F_k}\right)\big|_i+\mathrm{n_{k,i}} \qquad i=1,2,3 \quad (9)$$

El subíndice *h* indica *hip* o cadera. La unión de las ecuaciones anteriores definidas respecto del sistema inercial de la Fig. 1, da lugar al sistema de ecuaciones a resolver. Un aspecto de especial relevancia es el modelado del desplazamiento del centro de presiones de la fuerza de contacto *Fr*. En [11] se incluye un diagrama, Fig. 2, que permite estimar la posición de dicho punto a lo largo de la marcha. De esta forma, es posible definir el vector $\mathbf{r_{ga,P}}$ como sigue:

$$\mathbf{r_{ga,P}} = \left(\mathrm{r_{ga,Px}C_1,r_{ga,Py}C_2,r_{ga,Pz}C_3}\right) \quad (10)$$

de manera que los coeficientes *Ci* permitan adaptar la posición del centro de presiones para cada instante de la marcha. Esta forma de proceder garantiza que, siempre que exista continuidad entre dichos coeficientes, existirá continuidad en la solución. Esta aproximación, sin embargo, resulta más restrictiva que otras que modelan la fuerza de contacto como la aplicación simultánea de varias fuerzas o incluso la modelización de ésta como una carga superficial variable. A juzgar por lo descrito en [11], es posible mantener constantes e iguales a 1 los coeficientes C2 y C3 (pues la variación del centro de presiones en y y z es reducida) y el coeficiente C1 como:

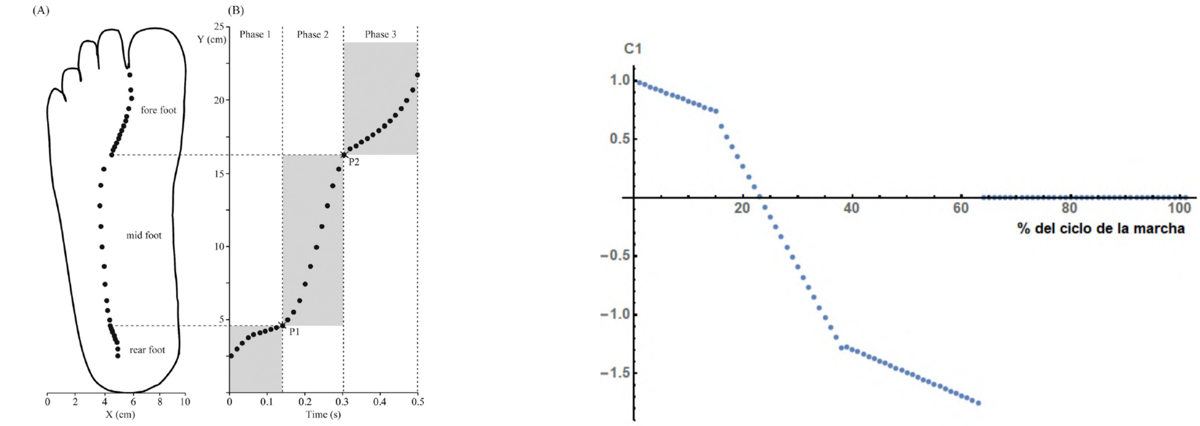
$$\mathrm{C_1} = \mathrm{C_{1,1}}\left(\mathrm{U(x)}\text{-}\mathrm{U(x\text{-}x_2)}\right)+\mathrm{C_{1,2}}\left(\mathrm{U(x\text{-}x_2)}\text{-}\mathrm{U(x\text{-}x_3)}\right)+\mathrm{C_{1,3}}\left(\mathrm{U(x\text{-}x_3)}\text{-}\mathrm{U(x\text{-}x_4)}\right) \quad (11)$$

donde U representa la función escalón unitario, x_i las coordenadas respecto del sistema de referencia local del pie de los puntos donde comienza cada una de las tres subfases en las que se divide el movimiento del centro de presiones a lo largo del ciclo (Fig. 2) y los coeficientes parciales $C_{1,1}$, $C_{1,2}$ y $C_{1,3}$ se definen de forma que, en función de los datos de entrada, se ajuste el resultado a la forma que se muestra en la Fig. 2.

Vea en la Fig. 2 que el coeficiente C_1 se extiende, como debe ser, desde el momento en que el talón toca el suelo hasta el final de la fase de balanceo o *swing*.

Una vez se dispone de los pares y fuerzas articulares para cada instante de la marcha en función de los datos cinemáticos en el mismo instante, es posible emplear esa información para, por ejemplo, controlar la acción de los actuadores de un exoesqueleto vestible que, mediante tirantes, produzca un porcentaje del par articular preciso y facilite así el desarrollo de la marcha. En aras de traducir ese porcentaje del par transmitido a la articulación a la fuerza precisa en el cable y, por tanto, al par preciso a la salida del motor, puede emplearse en primera aproximación un modelo geométrico sencillo como el de [12]. Un esquema simplificado para la cadera se muestra en la Fig. 3, adaptado de [12] para el caso de la cadera.

En nuestro caso, la Fig. 3 podría representar la articulación de la cadera, por ejemplo, siendo por tanto θ_k el ángulo γ en la Fig. 1. R_k es el radio de la articulación. Por supuesto, la Fig. 3 representa un caso en el que



(a) Evolución del centro de presiones de la fuerza de contacto con el suelo a lo largo de la marcha, [11].

(b) Representación del coeficiente C_1 a lo largo de la marcha.

Fig. 2. Centro de presiones de la fuerza de contacto y coeficiente C_1. Ejes *X* e *Y* en (a) según [11].

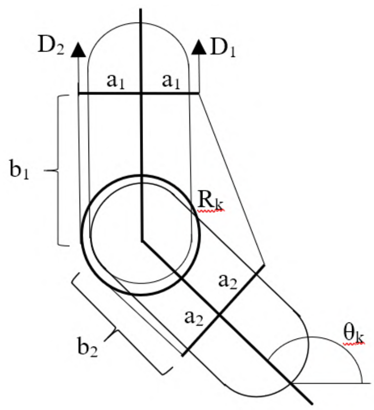


Fig. 3. Modelo geométrico simple de la cadera para el cálculo de la tracción en los cables a partir del par articular.

las distancias b_i y b_j allí mostradas entre los puntos de anclaje en torso y muslo respectivamente es distinta (en el caso de la cadera), así como el ancho de cada miembro a_i. Aquí multiplicaremos la longitud total de un miembro y otro por un factor t_i y t_j comprendido entre 0 y 1 para estudiar la influencia del punto de anclaje sobre el par demandado al motor en el caso particular de la flexión de cadera. Con todo, puede calcularse la extensión de ambos cables D_1 y D_2 como:

$$D_1 = \sqrt{a_i^2+b_i^2}*\cos\left(\varphi_i+\frac{\theta_k}{2}\right)+\sqrt{a_j^2+b_j^2}*\cos\left(\varphi_j+\frac{\theta_k}{2}\right)-b_i-b_j \tag{12}$$

$$D_2 = R_k\gamma \tag{13}$$

donde los ángulos φ_i se definen como:

$$\varphi_i = \mathrm{ArcTan}\left(\frac{a_i}{b_i}\right) \tag{14}$$

El par articular se relaciona con la tensión en el cable a través de la expresión (15), definidos los vectores **J** y **f** como se indica y siendo f_1 y f_2 las fuerzas de tracción en los cables. Se asume que existen dos cables para asistir los músculos agonistas y antagonistas.

$$\begin{gathered} n_{h,3} = \mathbf{J}\cdot\mathbf{f} \\ \mathbf{J} = \begin{pmatrix} \frac{\partial D_1}{\partial\gamma} & \frac{\partial D_2}{\partial\gamma} \end{pmatrix}^T \\ \mathbf{f} = \begin{pmatrix} f_1 \\ f_2 \end{pmatrix} \end{gathered} \tag{15}$$

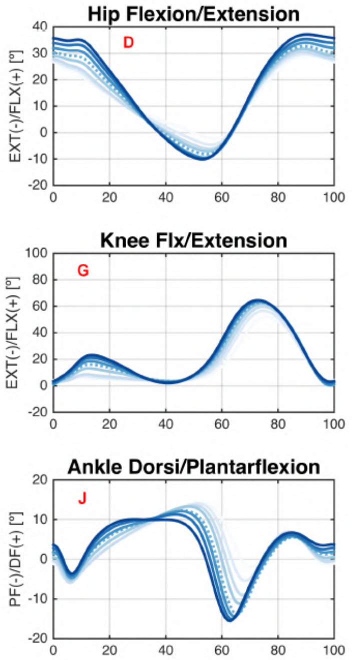
Una vez se dispone, por tanto, del par articular preciso y del porcentaje en cada fase de la marcha que desea asistirse, basta con despejar de (15) para obtener la fuerza precisa en cada cable (o el cable, si solo hubiera uno). Destacar que, allí donde el valor de $n_{h,3}$ sea negativo, el cable trabajaría a compresión y por tanto el resultado será nulo por no disponer, por lo general, de esa capacidad. Conocida la relación de transmisión de la reductora que acompañe al motor y el conjunto de poleas que transmiten el movimiento, es posible conocer el par instantáneo que ha de exigirse al motor a lo largo del ciclo de la marcha.

3. VALIDACIÓN DEL MODELO

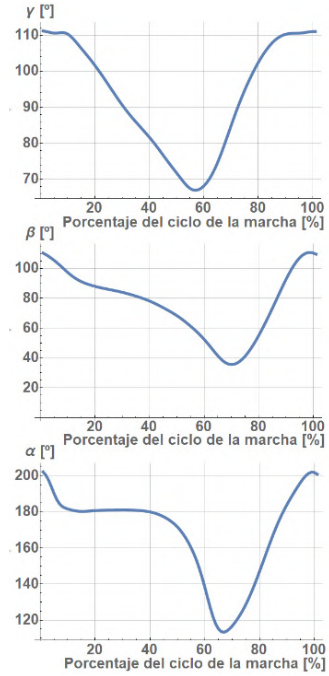
En aras de comprobar la veracidad del modelo planteado, es posible acceder a bases de datos, como la de [13], que recogen, para una serie de casos analizados a distintas velocidades de la marcha, los valores de pares articulares a lo largo de la marcha. Como [13] proporciona, además, los datos numéricos de los parámetros cinemáticos y de fuerza de contacto con el suelo para cada instante estudiado, por lo que la resolución a través del modelo planteado es inmediata. Así, basta introducir los parámetros cinemáticos (vea en la Fig. 4a los ángulos de flexión en tobillo, rodilla y cadera extraídos de [13]) para obtener los correspondientes valores de las reacciones en las articulaciones (fuerzas y pares articulares) en los tres ejes coordenados.

Los ángulos α, β y γ descritos en el apartado anterior y mostrados en la Fig. 1 pueden obtenerse, a partir de los ángulos de flexión de la Fig. 4a a través de (16), donde se han denominado A, B, y C a los ángulos de flexión en tobillo, rodilla y cadera respectivamente. Se muestran en la Fig. 4b.

$$\begin{gathered} \alpha = A-B+C+\pi \\ \beta = -B+C+\frac{\pi}{2} \\ \gamma = C+\frac{\pi}{2} \end{gathered} \tag{16}$$



(a) Ángulos de flexión articular de [13]



(b) Ángulos α, β y γ

Fig. 4. Representación gráfica de los ángulos de flexión articular en [13] y ángulos α, β y γ obtenidos a partir de ellos a través de (16).

Los datos de posición geométrica de los centros de gravedad de cada uno de los tres segmentos considerados de cada pierna pueden obtenerse con facilidad a partir de los datos de posición de las articulaciones que se indican en [13]. La fuerza de contacto con el suelo se incluye en [13] con su descomposición en cada uno de los tres ejes coordenados. En tanto que los datos ofrecidos por [13] corresponden a un análisis estadístico para un número de casos, se han adoptado como datos antropométricos aquellos que, para un individuo medio, se aceptan en [14], y que se muestran en la Tabla 1.

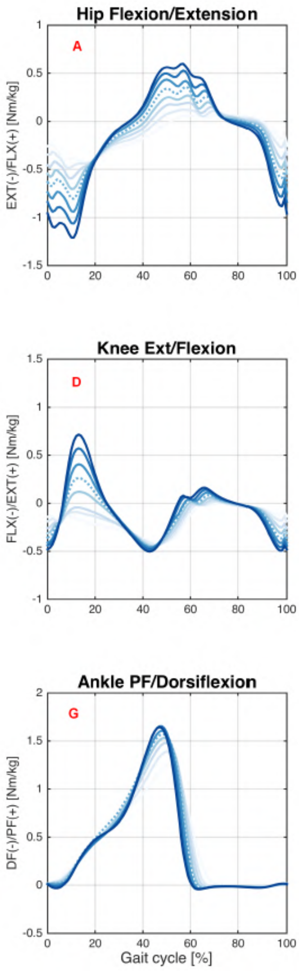
Introducidos los datos antropométricos en el modelo y definidos a partir de ellos los valores de los coeficientes C_i queda, junto con la cinemática de [13], completamente definido el problema para su resolución a través de análisis dinámico inverso, sin más que resolver el sistema de 18 ecuaciones con 18 incógnitas

Tabla 1. Datos antropométricos medios aceptados para el análisis comparativo.

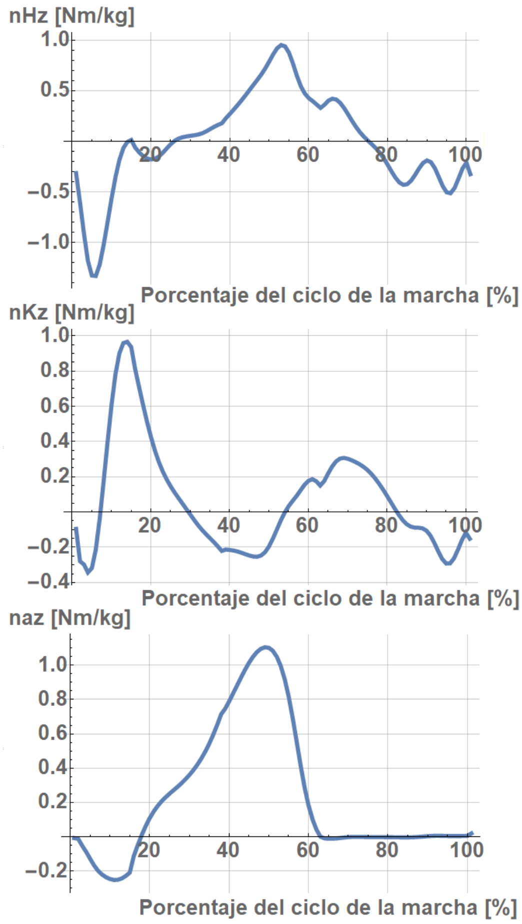
	Valores medios
Altura [m]	1.75
Masa [Kg]	75
Masa del muslo [Kg]	7.5000
Masa de la pierna [Kg]	3.4875
Masa del tórax [Kg]	16.2000
Masa del pie [Kg]	1.0875
Longitud del pie [m]	0.2660
Longitud de la pierna [m]	0.4305
Longitud del muslo [m]	0.42875
I_z del pie [Kg*m^2]	0.0035
I_z de la pierna [Kg*m^2]	0.0490
I_z del muslo [Kg*m^2]	0.1238

resultante. Véase en la Fig, 5 los resultados de pares articulares en tobillo, rodilla y cadera mostrados en [13] (a) y los obtenidos (b).

La Fig. 5 muestra como el modelo propuesto arroja resultados similares a los recogidos en [13], aunque sujetos a desviaciones provocadas, en parte, por la utilización de datos antropométricos medios, que no tienen por qué coincidir con los de los sujetos que arrojaron los resultados en la Figura 5a.



(a) Pares articulares según [13].



(b) Pares articulares obtenidos.

Fig. 5. Representación gráfica de los pares articulares normalizados con la masa del individuo medio en [14] (para varias velocidades de la marcha) y pares obtenidos con el modelo planteado, solo para la máxima velocidad de la marcha contemplada en [13].

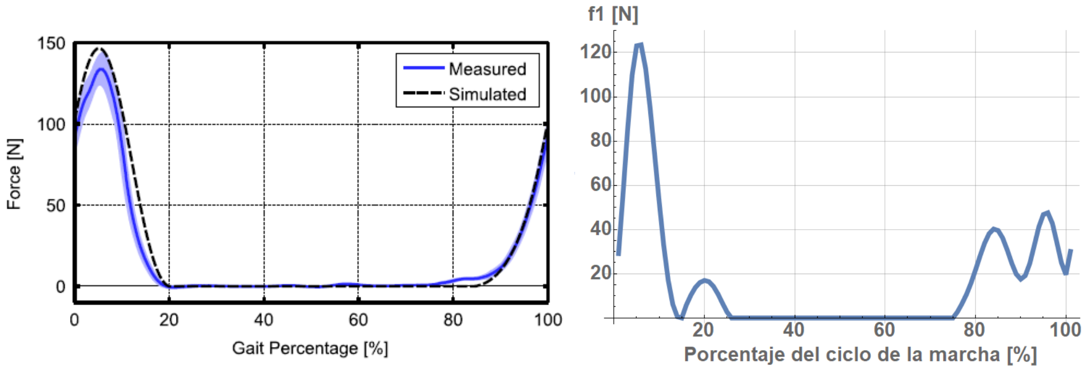
4. CÁLCULO DEL PAR MOTOR EN UN EXOESQUELETO VESTIBLE

Una vez son conocidos los pares articulares en cada instante de la marcha, resulta de especial interés, de cara a desarrollar un sistema de control para los actuadores de un exoesqueleto vestible para asistencia a la marcha, el cálculo de las fuerzas de tracción en los cables y del par demandado a los actuadores. En aras de comprobar qué valores se obtienen para un individuo medio de [14], basta aplicar las expresiones (12-15) para obtener una aproximación en el caso del modelo simplificado de la Fig. 3. Así, estableciendo que el motor ha de asistir, por ejemplo, el 30% del par articular y con un sistema de actuación con las características de la Tabla 2:

Tabla 2. Características generales del sistema de actuación de un exoesqueleto vestible.

	Valores medios
Par máximo en continuo [Nm]	4
Par máximo en continuo a la salida del motor [mNm]	95.6
Relación de transmisión [-]	1:33
Radio de la polea [m]	0.0126
t_1 [-]	0.5
t_2[-]	0.5
Radio de la articulación R_k [m]	0.15

se tienen las fuerzas de tracción de la Figura 6b.



(a) Tensión en el cable según [4]

(b) Tracción obtenida para el cable 1 de la Figura 3 con el modelo planteado y los datos de [13] para asistir el 30% del par

Fig. 6. Representación gráfica de la tracción en el cable según [4] y obtenida para los datos de [13].

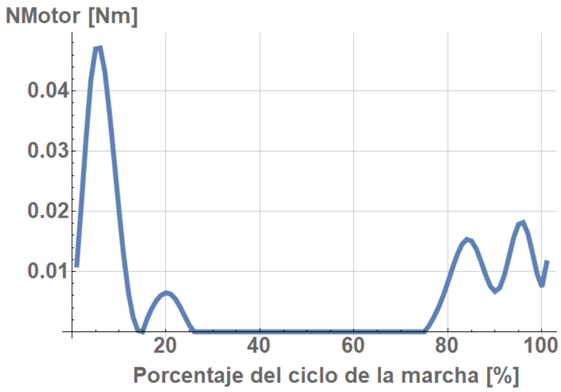


Fig. 7. Par resultante a la salida del motor.

Más que comparar los valores numéricos de la fuerza en los cables para el caso planteado en [4] y los obtenidos en la Figura 6b, se trata fundamentalmente de estudiar la similitud que existe entre ambos, a sabiendas de que los individuos y características de la marcha concretas en [4] no coinciden con los propuestos en [13] y [14], que son los empleados, a modo de ejemplo, para obtener los resultados del modelo propuesto. De igual forma, puede obtenerse el par instantáneo demandado al motor para ofrecer el 30% del par articular en la cadera, Fig. 7.

Se observa que existe una fase de la marcha, comprendida aproximadamente entre el 25 y el 75% en la que tanto la tracción en el cable 1 como el par demandado al motor por ese cable son nulos. Esta fase corresponde a aquella en la que el cable, de realizar trabajo, lo haría en compresión, algo que no es posible en los cables que de manera general se emplean en exoesqueletos vestibles. Por lo demás, es previsible un uso intenso de los actuadores en la fase de apoyo del talón y parte de la fase de apoyo del pie completo, así como en la fase final de balanceo. El par máximo se exige al término de la fase de apoyo del talón.

5. CONCLUSIONES

Se ha propuesto un modelo dinámico inverso para el análisis de las reacciones y pares articulares a lo largo de la marcha. El modelo admite seis grados de libertad para cada segmento corporal (tres de traslación y tres de rotación) y toma como datos de entrada las tres componentes de la fuerza de reacción con el suelo y los datos cinemáticos de la marcha (posiciones y ángulos de flexión articulares instantáneos). Como primer paso para el apoyo al diseño de un exoesqueleto vestible para la asistencia a la marcha, es imprescindible la capacidad de prever, para cada instante, el par que ha de requerirse al motor para cumplir con la actuación deseada en cada articulación y en cada momento de la marcha. El modelo aquí descrito, si bien se ha contrastado a través de una base de datos conocidos, ha de ser capaz, tras su implementación en el sistema de control pertinente, de tomar los datos de los sensores y controlar, de esta forma, a los actuadores. Como paso previo a la aplicación del modelo en un exoesqueleto vestible, se ha optado por introducir un modelo sencillo de actuación por cable, siendo ya capaces de emplear el modelo dinámico inverso para predecir el par demandado al motor y la fuerza de tracción en los cables a lo largo de la marcha. Se ha llegado a resultados cercanos a aquellos obtenidos de la bibliografía, tanto en lo que se refiere al propio estudio de la marcha (obtención de reacciones y pares articulares a partir de los datos cinemáticos) como al cálculo de las tracciones a que se someten los cables de una actuación por cable sencilla. Se espera poder implementar el modelo matemático descrito en un sistema de control adaptado al exoesqueleto en diseño.

AGRADECIMIENTOS

Los autores agradecen a la Consejería de Economía e Infraestructuras de la Junta de Extremadura que, a través del Fondo Europeo de Desarrollo Regional, haya financiado el proyecto IB18103 “Exoesqueleto híbrido vestible para el envejecimiento activo”, a partir del cual se ha desarrollado este trabajo.

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DESIGN OF AN INVERSE DYNAMICS MODEL OF A LOWER-LIMB ASSISTANCE EXOSUIT

Abstract – In the area of gait assistance, simple models based on cable-driven actuators with anchor points at the segments to be actuated stand out due to their high wearability, low weight and reduced cost. Force is transmitted from the actuators to the anchor points using cables, increasing user comfort by doing away with the rigid structures found in traditional exoskeletons. This paper deals with the inverse dynamics model to obtain joint torques needed during the gait cycle as a precursor to numerically dimension and design the components to develop an exosuit. A two-dimensional biomechanical model made up of seven segments was designed with three in each leg and one for the torso and torques in the hip, knee and ankle as the primary parameters. These torques are then distributed between the subject and the exosuit in different configurations. This model helps optimise the various characteristics of the exosuit such as the placement of anchor points, the exact number of actuators, cable arrangements among others. When finalising the force transmission system of the exosuit, it is possible to determine several variables such as: which joint(s) need the most external support and which ones are better left unactuated. The effect of the aforementioned design parameters on the force transmission, effectiveness of proposed designs in other literature can also be analysed.

Keywords – Exosuit, Cable Actuation, Inverse Dynamics Analysis, Human Gait.

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Force and Torque Characterization in the Actuation of a Walking-Assistance, Cable-Driven Exosuit

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Abstract: Soft exosuits stand out when it comes to the development of walking-assistance devices thanks to both their higher degree of wearability, lower weight, and price compared to the bulkier equivalent rigid exoskeletons. In cable-driven exosuits, the acting force is driven by cables from the actuation system to the anchor points; thus, the user's movement is not restricted by a rigid structure. In this paper, a 3D inverse dynamics model is proposed and integrated with a model for a cable-driven actuation to predict the required motor torque and traction force in cables for a walking-assistance exosuit during gait. Joint torques are to be shared between the user and the exosuit for different design configurations, focusing on both hip and ankle assistance. The model is expected to guide the design of the exosuit regarding aspects such as the location of the anchor points, the cable system design, and the actuation units. An inverse dynamics analysis is performed using gait kinematic data from a public dataset to predict the cable forces and position of the exosuit during gait. The obtained joint reactions and cable forces are compared with those in the literature, and prove the model to be accurate and ready to be implemented in an exosuit control scheme. The results obtained in this study are similar to those found in the literature regarding the walking study itself as well as the forces under which cables operate during gait and the cable position cycle.

Keywords: exosuit; wearable exoskeleton; cable-driven actuation

Citation: Rodríguez Jorge, D.; Bermejo García, J.; Jayakumar, A.; Lorente Moreno, R.; Agujetas Ortiz, R.; Romero Sánchez, F. Force and Torque Characterization in the Actuation of a Walking-Assistance, Cable-Driven Exosuit. *Sensors* **2022**, *22*, 4309. https://doi.org/10.3390/s22114309

Academic Editors: Nevio Luigi Tagliamonte, Diego Torricelli and Philipp Beckerle

Received: 20 April 2022
Accepted: 4 June 2022
Published: 6 June 2022

1. Introduction

The design of walking-assistance exosuits for humans can bring noticeable benefits to those who see their walking capacities reduced due to age or pathologies. Traditional exoskeletons are rigid, bulky, and generally heavy, and are gradually giving way to wearable exoskeletons or exosuits. Even though these are relatively new to the rehabilitation field, they offer a reliable and comfortable way to increase the user's moving capabilities without restricting movement or generating incompatibilities between the user and the exosuit's degrees of freedom [1]. The human skeleton itself becomes the support structure. Exosuits produce force transmission to assist the human gait, and are made mostly of textile elements, drastically reducing their weight and making them much more suitable, for instance, for elderly people who see their mobility reduced. Those who need walking assistance, whether due to age or other causes (spinal cord injury or stroke, among others) are expected to significantly increase in number in both Europe and every other area with elderly population growth. By the year 2080, for instance, the population over 65 is expected to grow by roughly 30% in Europe [2]. The use of wearable exoskeletons can help such persons to improve their life quality by increasing their independence. Of late, various solutions have been reached in order to increase user mobility, through both upper limb assistance [3] and lower limb assistance [4,5]. Tendon-driven exosuits, particularly those aimed at upper-limb assistance, have been the subject of study in several recent contributions, such as [6–8], which examines the design of a wearable exoskeleton for lower limb assistance intended to aid gait in the elderly. As a mandatory first step in the design and control of such systems, the calculation of joint torques throughout the gait cycle must be conducted. In [9], this fact is underlined and a model for joint torque prediction is developed in the context of attending to a series of gait phases with different foot support conditions.

In order to approach the problem from a mathematical perspective, two different points of view are especially interesting, namely, inverse dynamics models [10,11] which take kinematic data as input variables, and models where the problem is directly solved based on, for example, Lagrange Mechanics [12,13]. While approaching the problem via the Lagrange Equations is less demanding in computational terms than the application of dynamic equations in multibody systems [12], inverse dynamics is an excellent method to estimate reaction forces and torques at each joint during gait without solving any differential equations. Results obtained by the Newton–Euler equations via inverse models offer a good correlation with those from multibody analysis [14]. Here, an inverse dynamics model is aimed at, assuming the gait kinematic data and the ground contact force as known quantities. Prediction of the joint torques and forces during gait is the main goal in seeking to aid the design of an exosuit for gait assistance. A similar inverse dynamics model was presented in [10] for the whole body. Kinematic data such as position, acceleration, and segment angles can be measured using inertial measurement units and/or markers with a camera setup, as in [2], while ground contact forces can be measured via sensors embedded in the shoes, as presented in [15]. While there exist certain flexible exoskeleton proposals to assist the elderly, as in [16], soft exosuits with no support structure, as proposed in [17], provide the best case scenario for near-zero suit–human interaction.

Inverse dynamic models have been extensively used when designing exosuits for upper limb assistance, as in [18–20]. Inverse dynamics in the field of lower-limb assistive exosuits can be found in the recent literature, as in [2], although the results have rarely been studied with regard to their implication on the design of an exosuit. A general cable model for hip and ankle assistance during gait is presented in this paper, including the human body and cable stiffness effects. These two joints were chosen because they are the main power contributors during gait, as per [4,21]. A single-cable configuration is followed in this study, that is, torque can be transmitted in only one direction and the cable follows a certain trajectory along the human body via a set of pulleys or general cable guides. Multiple trajectory points along the path from the actuation system to the anchor points are considered in order to reliably predict the cable extension at both joints. Bowden cables could be used for the design of the exosuit, with the consequent non-negligible efficiency loss due to non-linear friction between the cable and the sleeve [22,23]. Because the model is a fully 3D approach to human motion inverse dynamics, it is possible to study the effects of the action of flexion–extension torque at a certain joint upon its abd/adduction and rotation torques, as well as to predict the cable's instant position regardless of the pulley configuration. Avoiding excessive undesired joint torques outside the sagittal plane may be a key design factor when approaching the positioning of the anchor points or the transmission system, among others; the model presented here can be used to parametrically quantify these torques as functions of said design variables. As the actuation depends on the results directly provided by the inverse dynamics model, the influence of gait and anthropometric data for a certain subject on the design or operation of the exosuit can be quantified as well. Here, the predicted torque at the motor shaft and the cable extension during gait are studied for certain gait conditions, namely, those of an elderly subject walking overground at a comfortable speed.

2. Materials and Methods

2.1. An Inverse Dynamics Model for Human Gait

In this section, the modeling of dynamic equations leading up to the derivation of joint torques and reaction forces is presented. Gait kinematic data, along with the ground contact forces, are input data. Applying Equations (1) and (2) with respect to a fixed inertial frame of reference will lead to the desired expressions. Such equations are applied as follows to each j-th leg segment and for each of the three global reference axes:

$$\sum \mathbf{F}_j = m_j \mathbf{a}_j \tag{1}$$

$$\sum \mathbf{n}_j = \mathbf{I}_j \dot{\boldsymbol{\omega}}_j + \boldsymbol{\omega}_j \times (\mathbf{I}_j \boldsymbol{\omega}_j) \tag{2}$$

In (1) and (2), $\mathbf{F}_j$ stands for the force on segment j, $\mathbf{a}_j$ for the linear acceleration of the center of mass of segment j, m_j for its mass, and $\mathbf{I}_j$ for its inertia tensor, while $\boldsymbol{\omega}_j$ is the angular velocity of segment j.

As the main goal is the deduction of torques and forces in the ankle, knee, and hip, a simple three-dimensional model for the leg is chosen. The full system consists of three segments plus the torso, all joined to one another, and each of them solitary to its own local reference frame. Joints allow rotational displacement around the three coordinate axes; thus, the linear reaction forces and the joint torques are the unknowns to be solved. Figure 1 shows the most essential elements in such a model, omitting for better clarity joint forces and torques other than those in the z-axis. Such local reference frames are defined by their corresponding Euler angles in the common Z-Y-X order.

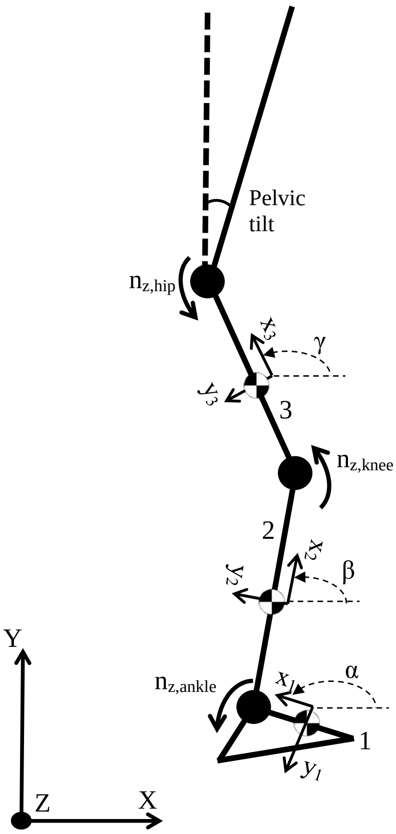


Figure 1. Simplified scheme for the considered lower-limb segments.

For each leg, Equations (1) and (2) are a system of eighteen differential equations: nine for force, nine for torque, and six in total for each leg segment. Said equations can be transformed into algebraic equations with the position data for each point (the centers of gravity of each leg segment and their joints), segment angles, and linear and angular accelerations as known quantities, which can be obtained by gait analysis in the laboratory. Equations (1) and (2), as particularized for each leg segment, are therefore as follows.

- Segment 1, foot:

$$F_{a,i} = m_1(a_{1,i} - g_i) - F_{r,i} \qquad i = x, y, z \tag{3}$$

$$n_{a,i} = I_{1,i}\dot{\omega}_{1,i} + (\boldsymbol{\omega}_1 \times (\mathbf{I}_1\boldsymbol{\omega}_1))|_i - \left(\mathbf{r}_{ga}^P \times \mathbf{F}_r\right)\Big|_i - \left(\mathbf{r}_{ga}^A \times \mathbf{F}_a\right)\Big|_i \quad (4)$$

In (4), $\dot{\omega}_{1i}$ represents the angular acceleration of the foot segment around its center of gravity with respect to the fixed i axis of the global reference frame, while θ_{ji} is the Euler angle i of segment j. The ground contact force is referred to as $F_{r,i}$ and the force and torque at joint j (a for ankle, k for knee, and h for hip) are $F_{j,i}$ and $n_{j,i}$, respectively. COM stands for the center of mass and vectors are described such that their subindex is their origin; r_{ga}^P, for instance, is a vector from the foot's COM to the application point of the ground contact force.

- Segment 2, shank or leg lower segment:

$$F_{k,i} = m_2(a_{2,i} - g_i) + F_{a,i} \qquad i = x, y, z \quad (5)$$

$$n_{k,i} = I_{2,i}\dot{\omega}_{2,i} + (\boldsymbol{\omega}_2 \times (\mathbf{I}_2\boldsymbol{\omega}_2))|_i - \left(\mathbf{r}_{gk}^k \times \mathbf{F}_k\right)\Big|_i + \left(\mathbf{r}_{gk}^A \times \mathbf{F}_a\right)\Big|_i + n_{a,i} \quad (6)$$

The force $F_{a,i}$ calculated in (3) was the joint force exerted by the rest of the body upon the foot segment. In (5) and (6), however, the same $F_{a,i}$ stands for the force exerted by the foot upon the shank segment; thus, the sign changes. The same applies to the moment, $n_{a,i}$ in (6).

- Segment 3, thigh or leg upper segment:

$$F_{h,i} = m_3(a_{3,i} - g_i) + F_{k,i} \qquad i = x, y, z \quad (7)$$

$$n_{h,i} = I_{3,i}\dot{\omega}_{3,i} + (\boldsymbol{\omega}_3 \times (\mathbf{I}_3\boldsymbol{\omega}_3))|_i - \left(\mathbf{r}_{gh}^h \times \mathbf{F}_r\right)\Big|_i + \left(\mathbf{r}_{gh}^k \times \mathbf{F}_k\right)\Big|_i + n_{k,i} \quad (8)$$

Combining all of the previous equations, all of them referring to the inertial frame shown in Figure 1, provides the system of equations to be solved.

2.2. Approach to the Design of a Lower-Limb Assistance Exosuit

When the joint torques and reaction forces are known for each time point during gait for the given kinematic data, it is possible to use said information to, for instance, control the action of the exosuit's actuators, which then use cables to produce a determined percentage of the total required joint torque, thus assisting the gait of the wearer. Next, a general geometric model is introduced to translate this percentage of the joint torque applied in the corresponding joint into the required forces in the cables, and consequently into the torque curve for the actuators. In this work, we particularize the general scheme to an exosuit that assists hip and ankle joints as an application example of the proposed model. A scheme for the hip and ankle joints is shown in Figure 2.

Generally, a more simplified geometric model is developed to obtain the relation between cable forces and joint applied torques, as in [18], where the problem is restricted to a 2D cable-joint interaction; although this simplifies the problem, it is missing out-of-plane information. For this multiarticular assistive exosuit, however, a precise geometric model is required in order to properly predict cable forces based on the desired instant torque as well as to provide a reliable model for the cables' extension. Therefore, a generalization of the method developed in [5] is proposed in this work, using the biomechanical model of the subject defined in the previous section in combination with an exosuit. In [5], a simplified 2D model for the hip actuation is presented, adding the human body's stiffness; however, the the cable's stiffness is neglected, and out-of-plane motion is ignored because the proposed exosuit is not expected to act under either ab- or adduction. However, cable misalignment might lead to undesired out-of-plane torques only approachable with a 3D cable model. Here, as both hip and ankle are actuated, the exosuit has two actuation units (one per joint) acting alternatively on both sides. Thus, the model considers two different subsystems for the actuation, one to assist the hip joint (Figure 2, blue) and the other to assist the ankle joint (Figure 2, green). The pulleys driving the cables are located at a certain distance perpendicular to the torso corresponding to the backpack where the exosuit actuation system is located. While the cable belonging to the hip subsystem goes straight to the anchor point, the one in the ankle sub-system follows a certain path defined by a number of trajectory points and the two final anchor points. The cable passes through such trajectory points at any given instant during gait, as in cable guides. The desired flexion/extension torque at any specific instant is known from the inverse analysis, while the existing relationship between such torque and the required cable force to achieve it is shown below:

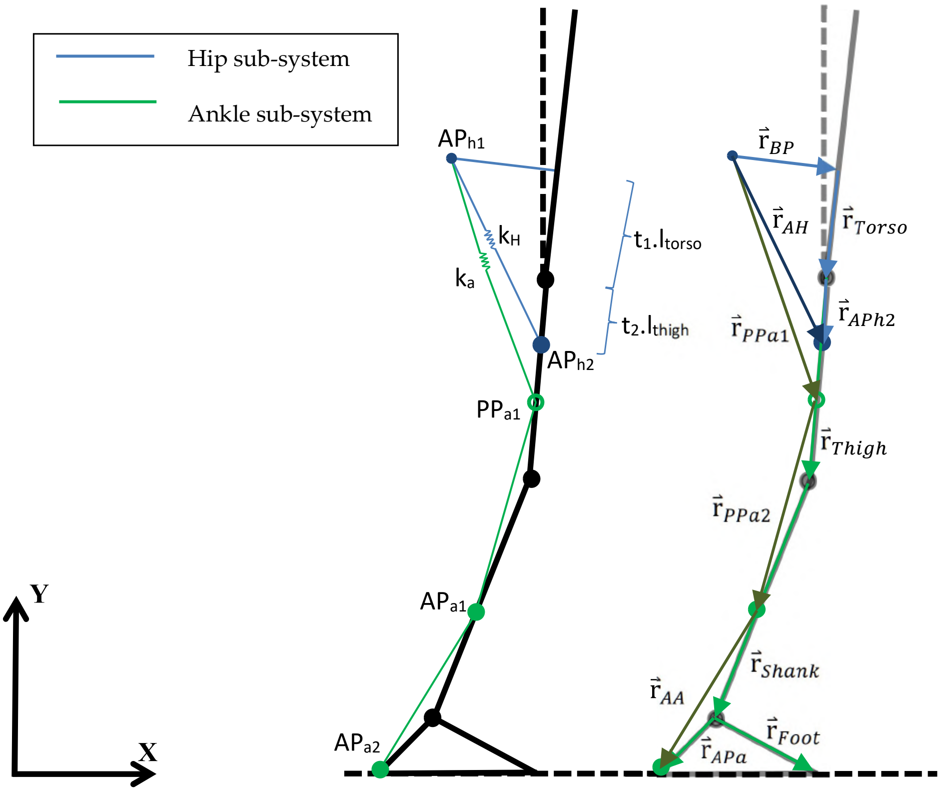


Figure 2. Geometric model for calculating the cable traction forces from the expected joint torques in both the hip and ankle sub-systems.

$$n_{mj,z} = -\mathbf{r}_{APj} \times \mathbf{f}|_z \quad (9)$$

In (9), $\mathbf{n}_{mj}$ stands for the motor torque applied to joint j and $\mathbf{f}$ for the cable force, while $\mathbf{r}_{APj}$ is a vector joining the joint j to the final anchor point.

The desired motor torque in joint j is generally expressed as a percentage of the corresponding biological joint torque whenever the consequent cable force is traction. The minus sign allows positive values to be obtained for the actuation range, where the desired torque is negative (clockwise). Constants k_i in Figure 2 represent the total equivalent stiffness of each sub-system, that is, the combined effect of the cable stiffness and that of the human body in each case. The latter can be approached via experimental methods, such as those described in [5], where the body responds to any force following a parabolic law. Because every exosuit defining vector shown in Figure 2 is strictly related to the body motion itself, their kinematics being completely defined, their coordinates and modulus are known along the gait cycle.

When the required joint torque is known and the desired percentage of additional support by the actuation system in each gait stage is defined, the necessary force in each cable can be directly obtained from (9). In cases where the obtained value for $n_{h3,z}$ is negative in certain periods, the cable is not able to actuate, as it cannot, in general, produce compression; the results will therefore be zero for said stages. If the reduction ratio of the gearbox installed with the actuators is known, along with the pulley array that transmits the movement, it is possible to obtain the instantaneous torque that must be demanded to the actuation system throughout gait. Furthermore, the total cable extension for each sub-system is known by considering both the inverse dynamics results and the cable force to predict the cable extension due to cable and body stiffness.

3. Results and Discussion

3.1. Inverse Dynamics Results

In order to check the model's reliability, databases as in [24] can be accessed and used to obtain the instantaneous values for joint torques during the walking cycle for a series of studied cases at different gait speeds. There, the authors provide an extensive set of kinematic results for a large number of subjects while walking overground or on a treadmill using camera-based tracking. Additionally, as in [24], with numerical data for the kinematic and ground contact force parameters for each time during gait the full solution can be reached via the proposed method. That is, simply by introducing the input parameters, the corresponding force and torque reaction values in the three axes at each joint can be obtained.

Information regarding the geometric position for the center of gravity of each considered leg segment for each leg can be easily calculated from the joint position data collected in [24]. Ground contact force is included in [24] for each of the three coordinate axes as well, along with the position of its center of pressure. As the data published in [24] correspond to a statistical analysis conducted for several cases, average anthropometric data according to [25] have been used with the proposed model. In [25], the average segment length and mass are normalized by the subject's height and total mass, as commonly used in applied software when testing several different subjects. In order to fully test the model's reliability on a target population, a group of ten subjects from [24] was selected, with ages ranging from 50 to 73 and both male and female, while walking over ground at comfortable speed. In [24], said subjects are referred to as subjects 27, 28, 29, 31, 33, 34, 35, 37, 41 and 42. Average anthropometric data as functions of each subject's height and mass are used, based on the common model described in [26], including segment length and mass, moments of inertia, etcetera; [26] provides a model for the lower limb specifically that is commonly used in numerical software for gait and human movement analysis. All necessary kinematic and ground contact force information is filtered using a fourth-order low-pass Butterworth filter using a cut-off frequency of 10 Hz, with sample rates of 150 Hz (kinematic) and 300 Hz (ground contact force and its center of pressure, COP), as indicated in [24]. Such a filter design is broadly used in the biomechanics field, including the authors of the dataset used in [27], where the authors provide another dataset focusing on running subjects instead of normal walking.

Using the anthropometric data as input for the model along with the kinematic variables in [24], the problem is already completely defined for its solution via the proposed dynamic inverse model. A system of eighteen equations with eighteen unknowns remains to be solved. Figure 3 shows the results for flexion–extension torques at the ankle, knee, and hip joints calculated in [24] and those obtained for the selected population, showing the average results and the standard deviation. Similar results to those included in [24] are obtained, although subject to deviations due mainly to the use of average anthropometric data in [25], such as segment moments of inertia or position of their centers of gravity. Anthropometric data are very relevant to the final solution, and information such as the exact position of the ground force center of pressure greatly impacts the result; the exact position of the COP during the contact phase is very determinant and a possible source of errors, showing in Figure 3 a stronger deviation in that phase. Moments in the upper joints are more sensitive to the solution of the lower segments, and might require different filtering in order to yield more accurate results, as stated in [28] wuth respect to the results of a large study in low-pass filtering focusing on inverse dynamics.

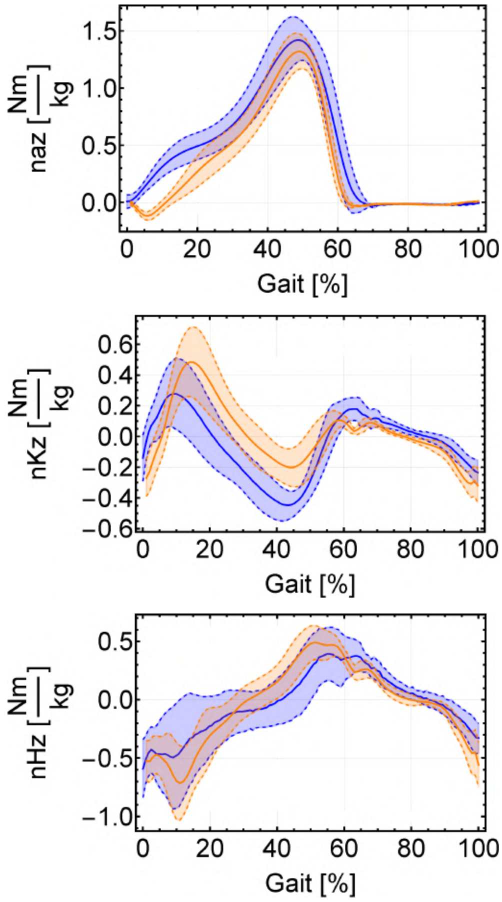


Figure 3. Representation of the normalized joint torques obtained via the proposed model for the selected population, according to anthropometric data in [25] (blue) and those in [24] for the "comfortable" gait speed (orange). Average values as solid lines. In dotted lines, ±standard deviation.

In [24], several attempts were performed for each subject, while only the average inverse kinematics results are provided. Because subjects were selected while walking over the ground at different speeds, comparing the average results with those of an specific attempt might be a considerable source of error. However, this way was chosen in order to analyze the problem from a realistic perspective; when designing an exosuit for walking assistance, the required actuation strongly depends on the walking conditions. Thus, a treadmill at constant speed was rejected as a means of studying variable phenomenon. In addition, the results are very different between subjects, as their height ranges from less than 1.5 m to more than 1.8.

3.2. Motor Torque in a Wearable Exosuit

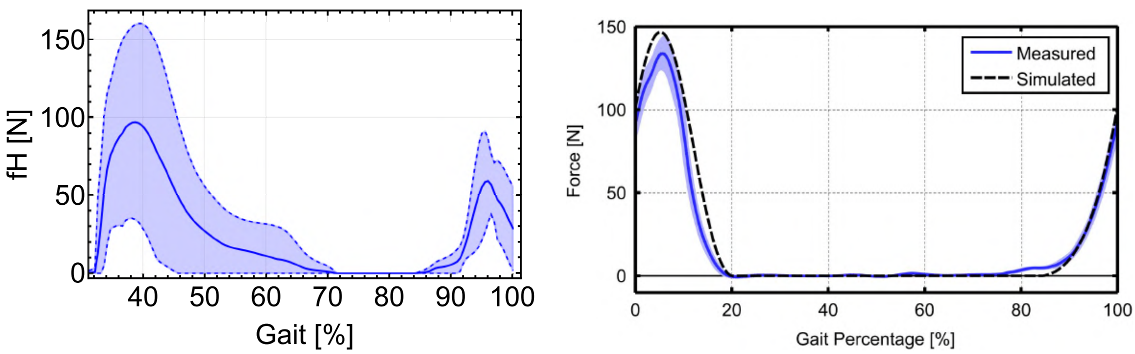
Knowing the joint torque patterns during gait is especially interesting regarding the development of a control scheme for the walking-assistance exosuit actuation system, the calculation of the forces applied to the tires, and the demanded motor torque. In order to estimate those forces and torques for the selected population using the data in [25] (Table 1), Equation (9) can be applied to reach an approximation for the case in Figure 2. Thus, a demanded joint torque of 30% of the total at the hip joint is established, as in [5], and 15% at the ankle joint, in order not to go beyond the motor maximum continuous torque. An actuation system characterized by the parameters in Table 1 can now be designed.

Table 1. General characteristics for the actuation system in a wearable exoskeleton.

	Hip Sub-System	Ankle Sub-System
Max. continuous torque [Nm]	4	4
Max. continuous torque at the motor shaft [mNm]	95.6	95.6
Transmission ratio [-]	1:33	1:79
Pulley radius [m]	0.0126	0.019
t1 [-]	0	0
t2 [-]	0.6	0.8
t3 [-]	-	0.3
Backpack length [m]	0.32	0.32
Trajectory points	0	2

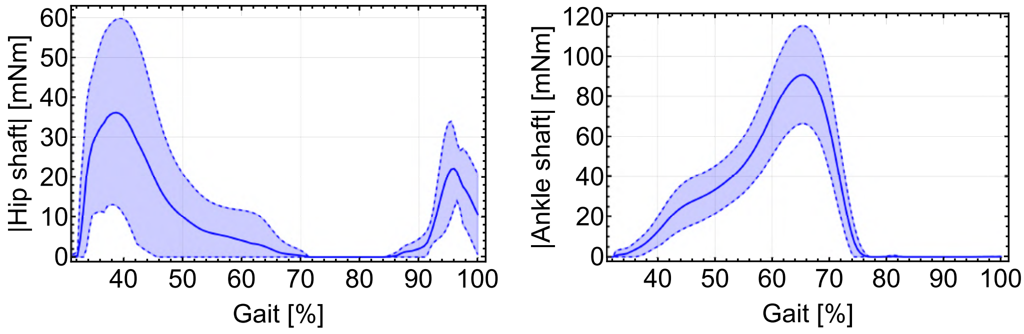
In Table 1, t_1 and t_2 represent, for the hip-subsystem, the distance from the contact point between the exosuit backpack and the back and the hip joint (t_1) and between the hip joint and the hip subsystem anchor point (t_2) as fractions of the torso and thigh segment lengths, respectively (as in Figure 2). They are similarly defined for the ankle subsystem. The path followed by the ankle cable is defined by two trajectory points located, respectively, at 80% of the thigh and 30% of the shank segment. These positions are chosen in order to locate the first trajectory point at a distant position from the final anchor point in the hip-subsystem, as well as to improve the scheme's visibility. How many trajectory points are included and where they are located impacts the cable extension function; these are therefore relevant factors in the design of the exosuit. In this case, the ankle anchor point is located at the same height as the heel camera-based marker except with the same y coordinate as the ankle to induce as little abduction moment as possible. Human body stiffness is neglected in the first place, although it is taken into account later on, and the cable stiffness is assigned a value of 6×10^5 N/m, following [5]. Now, the cable forces can be obtained and compared to those in [5] for the hip subsystem. To calculate the required cable force during gait, the biological hip moment in Figure 3 is filtered with a 10 Hz low-pass filter to avoid sudden changes in the moment being demanded of the engine, increasing user comfort [5].

Rather than comparing the exact values obtained in both cases for the cable force, Figure 4 is interesting in the way it reveals how the evolution of force varies in a similar trend for both. Individuals tested for each method had different biological characteristics, and therefore the results are not directly comparable. Moreover, the biological hip moment used in [5] was smoothed differently. Instantaneous torque demanded at the motor shaft can be predicted, which can assist 30% of the hip and 15% of the ankle joint torques during gait for both hip and ankle joints, as shown in Figure 5.



(**a**) (**b**)

Figure 4. (**a**) Cable force obtained using the proposed model for the data in [24] to assist 30% of the total torque. Average values as solid lines. In dotted lines, ±standard deviation. (**b**) Representation of the cable force of the hip subsystem according to [5].

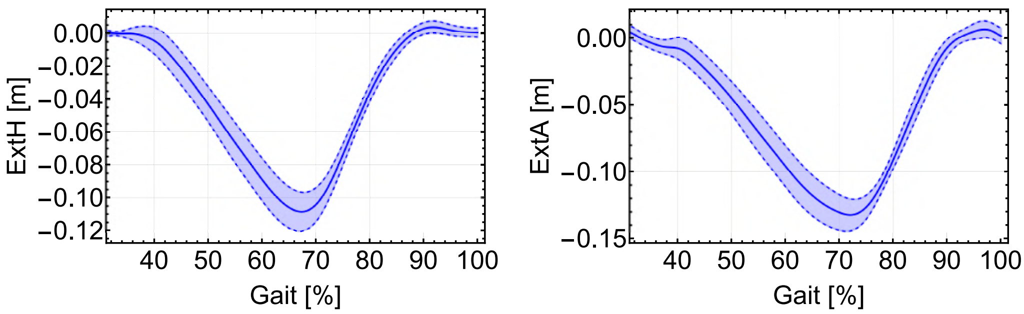


(**a**) (**b**)

Figure 5. (**a**) Torque at the motor shaft during the walking cycle for the hip subsystem and (**b**) for the ankle sub-system. Average values as solid lines. In dotted lines, ±standard deviation.

It is notable that in the case of the hip actuation there exists a walking phase, between 40% and 85% of the gait full cycle approximately, where both the force in the cable and the required actuation torque might be zero (Figures 4 and 5). In all cases, the gait cycle shown in the figures starts at 0% when the right heel first touches the ground and finishes at 100% when the same heel hits the ground at the start of the next cycle. During this phase, if the cable were to produce work it would be in the form of compression, which it is incapable of doing in general, and the same applies to the ankle. On the other hand, intense use of the actuators is predicted in the heel support phase and a portion of the full plantar support phase (up to around 65% of the gait cycle), as well as in the final stage of the swing phase (from around 65% of the gait cycle onwards) for the case of the hip subsystem. The maximum torque is required at the end of the heel support phase. Additionally, the peak in required torque at the ankle happens during the full foot support phase.

The total cable extension for each system is the first step for the motor position control. As the cable vectors are fully defined, it is possible to obtain the total cable extension with respect to the initial length (that at the beginning of the gait cycle). Figure 6 shows the extension of both hip and ankle cables without considering any cable or body stiffnesses. Increasing or decreasing the number and/or location of the trajectory points in the ankle subsystem highly modifies the instantaneous cable extension.



(**a**) (**b**)

Figure 6. (**a**) Cable extension during the walking cycle for the hip and (**b**) ankle subsystems. Average values as solid lines. In dotted lines, ±standard deviation.

Looking at the two graphs, a maximum decrease in cable length of approximately 12 cm and 15 cm, respectively, can be seen for each subsystems. Increasing the number of trajectory points for the ankle subsystem modifies the cable extension and is therefore relevant for implementing the position-based control scheme. The more cables in both systems deviate from being in-plane with the trajectory and anchor points, the higher a force needs to be transmitted by the cable to aid the same percentage of joint torque. Additional torques in x and y appear as well, which can be calculated when the required cable force is known, and can affect the user comfort.

Cable and human body stiffnesses have a great impact when controlling the actual cable position at the motor–pulley system. As an example, both kinds of stiffness data presented in [5] for the hip sub-system are used with the proposed model to obtain the instantaneous cable position at the pulley and the total cable plus human body elongation for subject 29 in [24]. An average cable stiffness of 6×10^6 N/m is considered for the cable, while a non-linear stiffness model is assumed for the human body, defined in [5] as:

$$F(x_s) = 8899.7x_s^2 + 99.546x_s \quad (10)$$

where x_s is defined in [5] as the total length of the equivalent spring measured in meters and F is the cable force at the hip sub-system measured in N. The spring stands for both the cable and human stiffness, the latter including the compression in thigh, waist, and shoulder (where the exosuit's backpack is in contact with the human body). Hip cable extension can be obtained by considering the stiffness effect, calculating only the elongation due to the instantaneous cable force, which is already a known quantity. Thus, Figure 7 shows the cable extension with and without the stiffness effects, the latter being the one shown in Figure 6a except with the zero located at the minimum length and the equivalent spring length x_s:

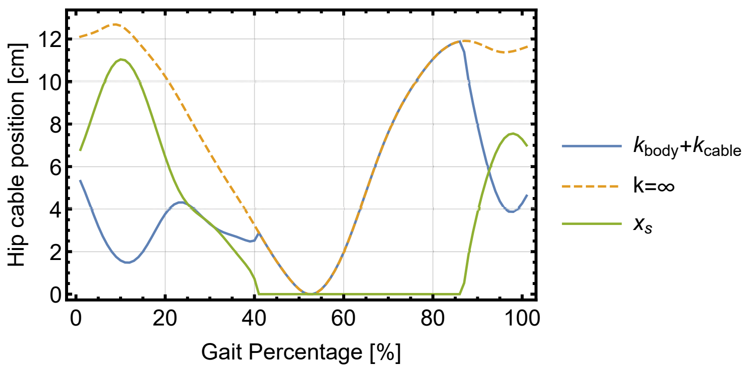


Figure 7. Cable position at the motor with and without stiffness along with x_s.

While the cable stiffness has a negligible effect due to its high value, it would probably be relevant if Bowden cables were used, as friction losses have a notable effect on their performance. However, the human body stiffness produces a displacement that reaches values of up to 11 cm in the studied case. Figure 8 shows the 3D model for the exosuit and the required cable force for 10%, 50%, and 95% of the gait cycle, showing the exosuit's backpack, the cables for both systems, and their corresponding trajectory and anchor points. All measurements correspond to the case solved in this section, without considering the body stiffness for either the hip or ankle joints.

Both systems act simultaneously at various stages during gait, although their peak values are reached at different times, as shown in Figure 4. The maximum value of the ankle cable force is noticeably higher than that of the hip, although this strongly depends on the exact position of the anchor point. The higher the distance from the ankle, the lower the force required to assist the same torque.

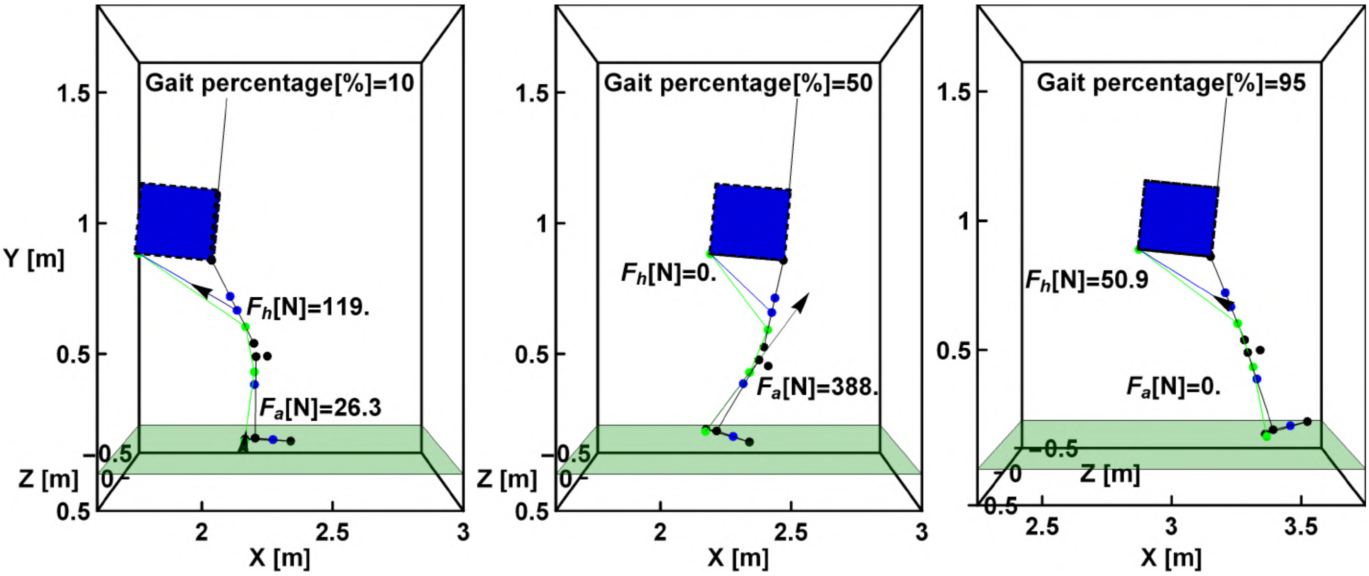


Figure 8. 3D model of the exosuit along with the lower segment and torso.

4. Conclusions

Here, an inverse dynamics model has been proposed for the analysis of joint reaction forces and torques during gait. Said model takes into consideration six degrees of freedom for each leg segment (three linear and three rotational) and takes the ground reaction force and the walking kinematic parameters of linear and angular position and acceleration as input variables. Flexion torques in hip, knee, and ankle were calculated and compared with those found in the literature as a preliminary step for the design of a walking-assistance exosuit. As the model is defined from the Euler angles, it is a full 3D inverse dynamics model and can be used as such as long as enough markers are used to determine each segment's angles, making possible an analysis of the impact of several anthropometric parameters on walking dynamics and their impact on the exosuit's design.

As a first step towards the design of a soft wearable exoskeleton for walking assistance, the capacity to fully predict the required joint torque to assist the user during gait is mandatory. The proposed model has been tested using bibliography data. Following installation in the exosuit along with the corresponding control scheme, it must be capable of receiving such information from the sensors and requesting the correct torque from the actuation system at each instance. As an implementation of a generally-designed exosuit, a 3D model for cable-driven actuation has been chosen to precisely predict the required motor torque and traction force in cables for the walking cycle. Outcomes close to those in the literature were obtained via the proposed inverse model along with a torque, force, and cable extension for both the hip and ankle actuation. The results obtained were acceptable regarding both the walking study itself (reaction forces and torques from the input ground force and kinematic data) and the forces under which cables can be expected to operate during gait. The described mathematical model is expected to be implemented in the control scheme for the currently under-development exosuit. It will be integrated using a musculoskeletal model to optimize coordination between the exosuit's actuation and the muscular system, allowing it to determine the best way to actuate when a certain muscle is handicapped.

Author Contributions: Conceptualization, D.R.J., A.J. and J.B.G.; methodology, D.R.J., A.J. and J.B.G.; software, D.R.J.; validation, D.R.J.; formal analysis, D.R.J.; investigation, D.R.J., A.J. and J.B.G.; resources, R.A.O. and F.R.S.; writing—original draft preparation, D.R.J.; writing—review and editing, R.A.O. and R.S.F.; visualization, R.A.O. and R.S.F.; supervision, R.A.O., R.S.F. and R.L.M.; project administration, R.A.O. and R.S.F.; funding acquisition, R.A.O. and F.R.S. All authors have read and agreed to the published version of the manuscript.

Funding: This research was partially supported by the Ministry of Science and Innovation – Spanish Agency of Research (MCIN/AEI/10.13039/501100011033) through project PID2019-107491RB-I00 and partially by Junta de Extremadura (Consejería de Economía, Ciencia y Agenda Digital) and European Regional Development Fund "A way of making Europe" through projects IB18103 and GR21069.

Data Availability Statement: The data that support the findings of this study are openly available in [24] at https://pubmed.ncbi.nlm.nih.gov/29707431/ (accessed on 18 April 2022).

Conflicts of Interest: The authors declare no conflict of interest.

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A synergy-based approach for the design of a lower-limb, cable-driven exosuit

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ABSTRACT

A novel, synergy-based design for lower-limb, gait assistance exosuits is proposed in this paper. The general design philosophy is followed by the mathematical model that leads to the exosuit's actuation system design from a certain set of postural synergies, which may reduce the number of total required actuators and, thus, the overall system weight and price. Cable extensions are chosen as the design variable upon which the principal component analysis is performed, given the high resulting cumulative variance that is obtained with few principal components. Expressions for the pulley radii are then derived and a general exosuit design is proposed, along with a novel transmission system that can combine any number of actuators to follow the gait movements from a synergy perspective. A basic testbench is also presented with such transmission system implemented and an empirical set of results are shown with a varying number of actuators, proving that cable extensions and synergy-based design provide an excellent solution when reducing the number of actuation units in a walking-assistance exosuit.

Nomenclature

$\bar{\mathbf{Q}}$ Mean of the system input variable matrix, where column j equals the mean of $\mathbf{Q_{All,j}}$.
$\Delta\theta_{ijk}$ Angular movement increase at the shaft of motor k, for joint j at a given time instant i.
Δl_{ijk} Extension at cable from motor k, for joint j at a given time instant i.
$\sqrt{\sigma^2}$ Diagonal matrix where element jj equals the standard deviation of $\mathbf{Q_{All,j}}$.
$\mathbf{e}$ System eigen-vector matrix.
$\mathbf{PC}$ Principal component matrix.
$\mathbf{Q}$ System input variable matrix.
σ Standard deviation.
θ_{ik} Angular displacement by motor k at instant i.
AP_{jq} Anchor point q at joint j.
C Pulley size constant.
C_k Vertical displacement constant for the final pulley at motor k.
i Subscript indicating the normalized time at that instant.
j Subscript indicating the joint.
k Subscript indicating the actuator.
l_{ijk} Cable extension at joint j, by motor k at instant i.
N Dimension of the measured statistical variables.
n Number of degrees of freedom.
n^* Number of chosen principal components/actuators.
PCA Principal component analysis.
PP_{ji} Trajectory point i for joint j.
r_{jk} Pulley radius in motor k for joint j.

1 Introduction

According to WHO (World Health Organization), more than 500 million people are affected by any kind of motion pathology worldwide [1]. This trend is not expected to be reversed in the close future, given the fact, for instance, that the European population over 65 may well grow by around 30% by the year 2080 [2]. Lately, numerous solutions have been proposed to improve users' mobility, among which exoskeletons and exosuits for any kind of assistance are to be found. The exosuit range of applications has notably grown during the last decade as a natural evolution for traditional exoskeletons, which offer a rigid solution with restricted degrees of freedom, while exosuits get rid of such inconveniences in exchange for a lower transmitted joint torque. Since exosuits dispense with rigid tutoring bars, their supporting role is assumed by the human body itself and cables behave as muscle-tendon actuation units, capable of transmitting forces to different body segments. The main advantages of these systems are the lower weight and the improved kinematic compatibility with the user, as opposed to exoskeletons [3]. Lacking tutoring bars not only decreases the weight of the device but also significantly improves its wearability. Thus, there is an increasing number of exosuits under development. For instance, there is a vast amount of recently developed assistive devices for the upper limb, including the arm itself and the hand [4–7], most of which are soft exoskeletons or exosuits, and also for the lower limb, such as in [8–11]. The transmitting force, however, becomes their most visible limitation: they will only be able to assist around 30% of each joint's total required torque. This barrier comes from the way power is transmitted in a cable-driven actuation system and the high contact forces caused by the human-cable interaction. Thus, exosuits' major benefit is providing additional support during everyday activities, such as walking, grabbing, and reaching objects or stabilizing gait under pathological conditions, or for the elderly. This paper proposes a new design philosophy for exosuits that assist the lower limbs during gait based on postural, kinematic synergies. Using statistical analysis to determine the best way to approach the design of a mechanical system itself is not new, and similar applications such as hand assistance can be found in the literature. In [12], five fingers are assisted with only two actuators, severely decreasing the weight and price of the device. In [13] a hand-assistance synergy-based glove is proposed using bowden cables. Synergies themselves are well known and have been commonly used for the past 30 years to decrease the dimensionality of human movement and to study the coupling between degrees of freedom for motor control [14–18]. Since weight and wearability are determining factors when it comes to assisting human gait in pathological and older users, it is our task to further expand the application field of synergy analysis and propose a low-weight, low-cost exosuit concept to assist hip, knee, and ankle during gait. That is, based on kinematic coordination inherent to the user's gait cycle, a novel design is proposed to drastically reduce the number of required actuators while still assisting both legs and multiple joints, thus reducing the price and weight of the device and increasing its wearability. It was certainly those benefits, lower weight, and price, as well as improved wearability, that inspired this work: extending the synergy-based approach to lower limb assistance, a field that benefits humankind on many levels. In [19], for instance, a practical application of lightweight exosuits for the lower limb is presented, enabling active rehabilitation on patients with various gait disorders. Another use that has been explored for lower limb exosuits is reducing the metabolic cost and muscle fatigue of the wearer. In [20], a light exosuit for hip assistance of around only 2.5 kg was developed and tested, showing how the metabolic consumption could decrease around 8%. However, it is also shown how this decrease has to take into account the increase in metabolic cost inherent to wearing an exosuit. Similar studies are presented in [21,22], all aiming at reducing the metabolic cost of walking, proving that whether it is for rehabilitation, assistance or simply to reduce the cost of walking in healthy subjects, weight and wearability take a key role in the design of lower-limb exosuits. Traditionally, a single actuator was used to actuate each degree of freedom, although new designs opting for a lower number of actuators, key elements regarding the total weight, have been proposed, as in [23]. The PCA study included here highlights which type of synergies are more beneficial when it comes to approaching the design of a walking-assistance exosuit and stresses how much it might reduce the problem's dimensionality. A synergy-based study on walking variables might yield interesting results: it has already proven very useful on arm and hand movement, and gait has a more cyclic nature than the latter. Furthermore, a general approach on how to translate postural synergies into actuation systems and mechanisms is presented, introducing a general transmission device that would allow for a general, synergy-based actuation. Results are tested using a developed 3D model for a sample exosuit and on a test bench. A parametric study on how the positioning of the anchor points may affect the joint assistance is also included.

2 Postural synergies

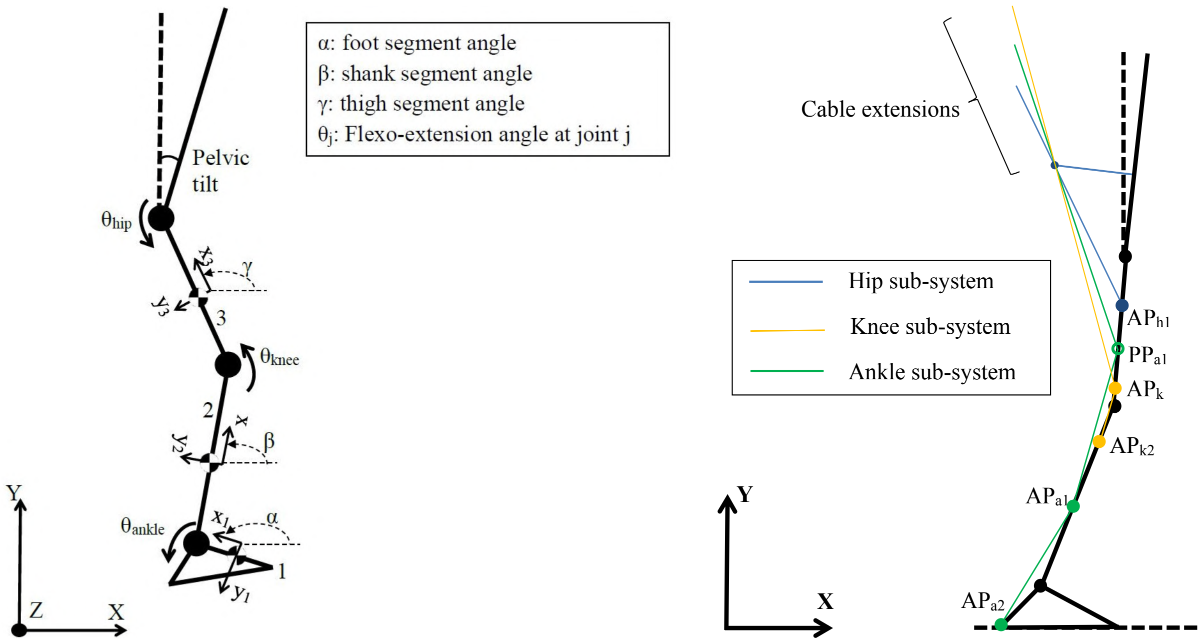
In general, a mechanical device designed to actuate n degrees of freedom that are completely independent of one another will require an equal number of actuators. Thus, finding relationships between such quantities is vital when it comes to simplifying the system to the point where it might be viable to construct and, presumably, market and sell. Regarding the design of exoskeletons and exosuits, there exists a trend of applying statistical analysis to the involved quantities, seeking for any kind of, again, trend that may lead to an overall system simplification. One of such methods is PCA or Principal Component Analysis. PCA is a multivariate statistical technique that generates, in general for any set of n statistical variables of dimension N, a set of n principal components (PCs) or scores that explain a decreasing amount of variance [17,24]. Said PCs or PC scores are linear combinations of the input variables obtained from the eigenvectors of the covariance matrix, as in 1. Based on the decreasing value of the VAF (Variance Accounted For), researchers can focus on only $n^* < n$ of said principal components to then reconstruct the original variables with results of increasing reliability the higher n^* is, returning of course the original variables as they were if $n^* = n$. A comprehensive guide on PCA can be found in [17,25], where some of the main purposes of PCA are highlighted: extracting the most relevant information from a given set of data, thus reducing its size (once reconstructed from the selected principal components). Such data will be related to the problem whose dimensionality is to be reduced: in this case, gait-related variables such as the joint or segment angles. The number of resulting principal components with high cumulative variance will yield the number of required actuators. Reducing that number will generally result in lower weight and price of the final exosuit, as well as increased wearability.

PCA has been extensively used in several fields, including walking [26] and postural balance [27]. In our case, if $\mathbf{Q}$ is a matrix containing the n statistical values of dimension N under study, that is, our input data and $\mathbf{e}$ are the eigenvectors of the covariance matrix of $\mathbf{Q}$, then, providing data have been standardized to zero mean and unit standard deviation, as is customary [25]:

$$\mathbf{Q}_{N\times n} = (\mathbf{PC}_{N\times n^*}\mathbf{e}_{n^*\times n})\sqrt{\boldsymbol{\sigma}^2}_{n\times n} + \bar{\mathbf{Q}}_{N\times n} \tag{1}$$

Now, a question that may arise is: which set of known quantities should be studied to aid the design of a walking assistance exosuit? Some research groups have focused on joint angles to approach their synergy-based designs for upper-limb exosuits and gloves, as in [13]. Those did yield very good results, enabling the researchers to propose actuation models with a low number of actuators, for a high number of DOFs. Another option comes in the form of studying segment angles. Fig. 1 shows the definition for both the segment and joint angles in a simple lower-limb model, as well as a simplified scheme of the proposed exosuit design, with a cable acting each joint via two anchor points.

In Fig. 1b a general exosuit design scheme to actuate hip, knee, and ankle using cables is shown. The actuation unit is located in a backpack or a supporting device located in the trunk and force is transmitted to the joints via cables. The design concept is common in lower-limb assistance exosuits, as those found in [9,10]. Variance accounted for in such segment angles proves especially interesting given its high value for the first principal component, that is, based on only one PC human gait can be approximately reconstructed, yielding results as in Fig.2. However, there might be other gait-related parameters that offer a better value of the cumulative variance corresponding to the first PCs and that might translate into interesting design proposals yielding better overall characteristics, such as lower prices, weight, or increased wearability of the final product.



(a) Segment and joint angles in the lower limb

(b) Exosuit's simplified design scheme

Fig. 1: Segment and joint angles in a planar model (left) and proposed exosuit model (right).

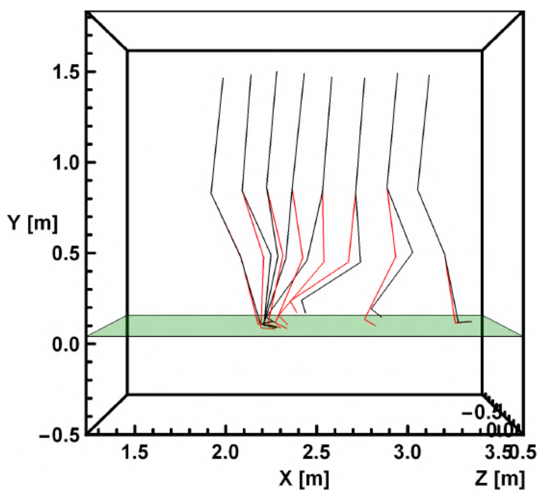


Fig. 2: Human gait reconstructed based on one PC (red) compared with measured results (black).

3 Method and Mechanisms

In Fig. 2, results based on kinematic data collected in [28] for subject 29 in overground, comfortable speed walking were used to reconstruct gait. It is clear that a design philosophy based on a number of reduced quantities, the so-called synergies, can yield very interesting, simplified results from both cost and efficiency standpoints. For instance, in [12] a robot hand with only two actuators is proposed to actuate no less than 17 DOF thanks to a tendon displacement-based PCA study carried out for several, distinct hand postures. A similar study aimed at lower limb assistance can also provide very interesting results, as will be proved here.

Even though elevation angles and their synergies have been extensively studied [29] and yield high values for the cumulative variance for a few principal components, total cable extension during gait for a cable-actuated exosuit will prove an even better option: data collected from a population of ten older adults (male and female) will be analyzed, yielding thought-provoking results regarding the cumulative variance achieved by applying the PCA to cable extensions instead of joint or segment angles, as well as a lower standard deviation for the studied population. In this paper, the authors propose a general design philosophy for such devices, having found that the total cable extension for the lower limb joints follows a similar pattern for the proposed exosuit. In general, an exosuit that acts on the ankle, knee, and hip joints for both legs during gait might need up to six actuators, one for each joint when assisting the flexion/extension rotations at the hip, knee and ankle for both legs (for a total of 6 DOFs). However, PCA analysis will prove an outstanding solution in order to drastically decrease that number. Now, given the kinematics of human gait and a certain exosuit design (with its corresponding anchor points for each joint subsystem, as in Fig. 1b), it is possible to obtain the instantaneous cable extension for each joint and actuator. If human body stiffness is neglected, this calculation is reduced to a mere geometric problem. Thus, now **Q** being the total cable extension for each joint, Eq. 1 can be re-written as:

$$Q_{ij} = \sum_{k=1}^{n^*} \left(PC_{ik} e_{kj} \sqrt{\sigma^2}_{jj} + \frac{\bar{Q}_{jj}}{n^*} \right) \tag{2}$$

where subindex i stands for the instant where the variable is measured and j for joint, the first being the rows of matrix **Q** and the second its columns. Now, since **Q** represents cable extension, it can be directly related to pulley angular displacement as:

$$\sum_{k=1}^{n^*} \Delta l_{ijk} = \sum_{k=1}^{n^*} \theta_{ik} r_{jk} = \sum_{k=1}^{n^*} \left(PC_{ik} e_{kj} \sqrt{\sigma^2}_{jj} + \frac{\bar{Q}_{jj}}{n^*} \right) \tag{3}$$

Of course, when $n^* = n = 6$, there will exist six actuators, each of which acting upon all six DOFs present in the system, each of which is therefore acted upon by six different actuators at the same time. Equation 3 shows how the cable extension produced by actuator k upon joint j at instant i equals the product of the angular displacement of the axis attached to actuator k by the pulley radius r corresponding to joint j. Each actuator has a set of n pulleys to act all n DOFs. The mechanical implementation of Eq. 3 is possible, for instance, by extending the design proposed in [12], which could effectively combine the effect of several actuators to obtain one linear output displacement thanks to a pulley mounted on a vertical guide. For n DOFs, a set of $n^* - 1$ devices as the one shown in Fig. 3 are required.

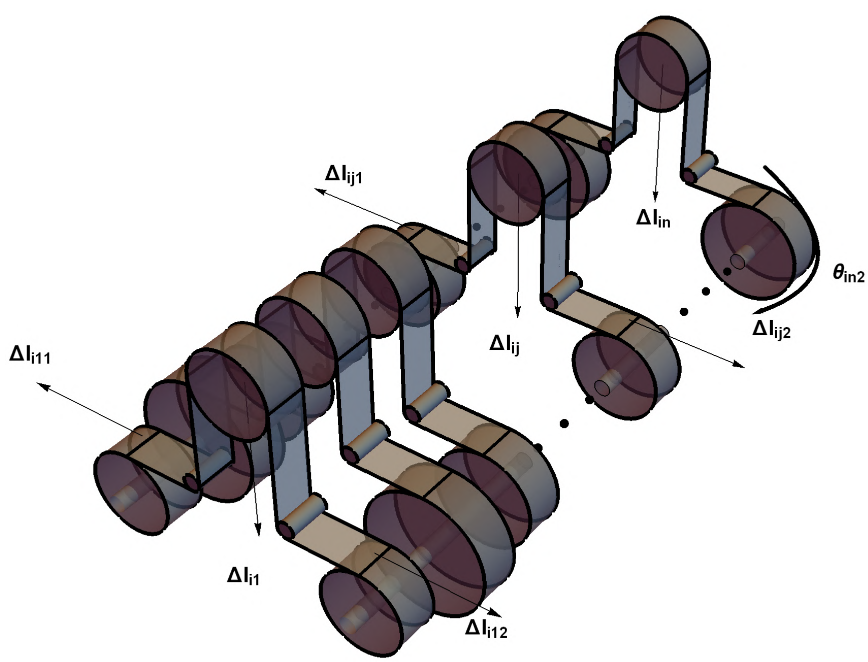


Fig. 3: General mechanism to act n DOFs with two actuators. Cable extensions and motor rotations are shown for an instant i, at a joint j, for a case where two actuators are selected, as Δl_{ij} and θ_{ijk} respectively.

If more than two actuators are necessary, then the system in Fig. 3 will be more complex. Figure 4 shows the system needed for one joint (DOF) actuated by three actuators:

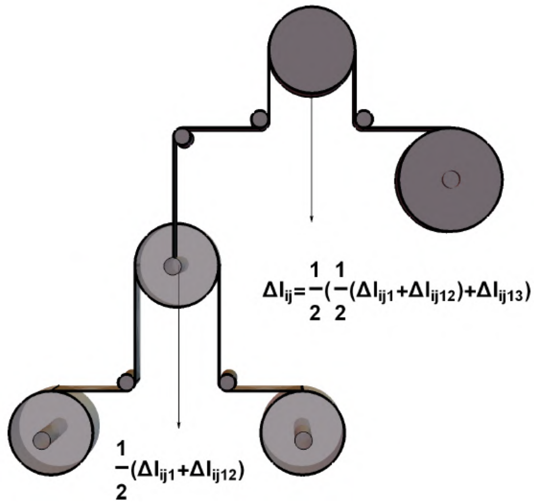


Fig. 4: Set of pulleys needed to act joint j with three actuators. Cable extensions are shown for an instant i, at a joint j, for a case where three actuators are selected, as Δl_{ij}.

In Fig. 4, two pulleys on mounted guides make it possible to act one single joint with three actuators. The system can be extended for a total of n^* actuators, being the vertical displacement of the final pulley the one that will translate as the total cable extension for such joint, defined as:

$$\Delta l_{ij} = \sum_{k=1}^{n^*} C_k \Delta l_{ijk}; \quad C_k = \begin{cases} \frac{1}{2^{n^*-1}} & \forall k \leq 2 \\ \frac{1}{2^{n^*-k+1}} & \forall k > 2 \end{cases} \tag{4}$$

Then combining Fig. 4, with as many pulley systems as the number of actuators n^* require, with Fig. 3, with as many stages as joints n are to be actuated, a system for a general synergy-based actuation is proposed. The main goal then is to reduce n^* as much as possible to simplify the system and reduce its weight and price, while still being able to follow human gait to a large extent. Combining Eq. 3 and Eq. 4 yields the angular displacement for axis k aimed at joint j at a given instant i:

$$\theta_{ijk} = \frac{PC_{ik} e_{kj}}{C_k r_{jk}} \sqrt{\sigma^2}_{jj} + \frac{\bar{Q}_{jj}}{n^* C_k r_{jk}} \tag{5}$$

Of course, a single shaft can only provide a certain angular displacement at any given time. That is:

$$\frac{\partial \Delta\theta_{ijk}}{\partial j} = 0 \longrightarrow r_{jk} = C \cdot e_{kj} \cdot \sqrt{\sigma^2}_{jj} \tag{6}$$

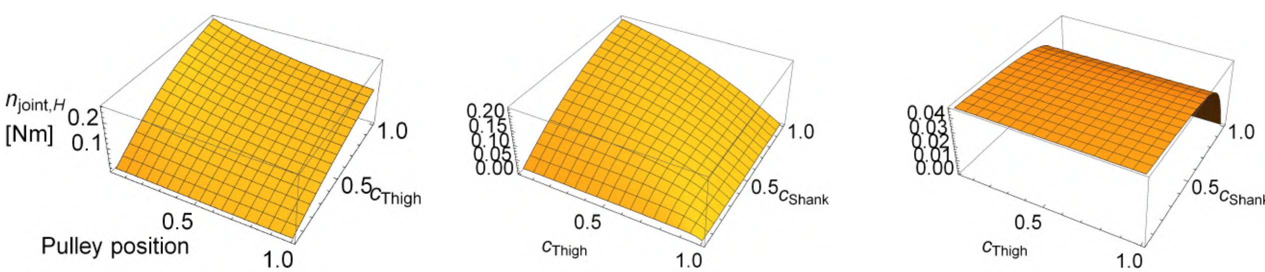
where C is any real number that yields as a result pulley diameters that must be enough to be manufactured and small enough to be comfortable. Reducing n^* will result in simpler, lighter, and cheaper transmission and actuation systems.

The model described above yields the pulley radii and the angular displacement required for every actuation unit and every actuated joint and predicts the impact that reducing the number of actuators will have on the reconstructed cable trajectories. In order to both have a visual look at the expected results and evaluate the proposed synergy-based approach using cable extensions, a study on a population of ten older adults, a possible target population for walking assistance exosuits, is performed. In order to study cases as close to the reality of our target population as possible, a public database including the gait parameters of several adults while walking overground and published in [28] is selected. This decision is motivated by the requirement of studying gait in normal conditions of overground, comfortable walking, rather than data collected on a treadmill, as well as having access to data from selected populations, in this case, older adults, both male and female. Said subjects are identified in [28] as subjects 27, 28, 29, 31, 33, 34, 35, 37, 41 and 42. They walk over the ground at a speed that each of them found comfortable, and they are older adults of ages ranging from 50 to 84 years. In [28], data for marker position, segment angles, and ground reaction forces are provided for the subjects. Since only basic anthropometric data of the subjects is available, the standard values for limb length and mass as functions of the subject's height and mass (which are known) published in [30] will be assumed. More mechanics-related parameters such as the segments' moments of inertia are obtained from the model developed in [31], commonly used in biomechanical models and software. The gait and body characteristics of the subjects are publicly available in [28].

Now in order to predict the cable extensions during gait that are expected given the design proposal shown in Fig. 1b, exact positions for the anchor and trajectory points (those through which cables pass before reaching the anchor points) are needed. In general, cables in Fig. 1b can follow any trajectory from the actuation unit to the anchor points, only by adding more trajectory points (those through which each cable is forced to pass). In this case, and exactly as shown in Fig.1b, only the ankle cable system will have one trajectory point, above the knee, before reaching the first anchor point at the shank. Notice that, for the hip cable, the backpack support structure itself acts as the first anchor point, as in Fig. 1b. No cable and body stiffness will be considered for the purpose of this paper. The exact position for the anchor points can be chosen to provide a joint torque close to the maximum achievable with the given configuration.

4 Results and discussion

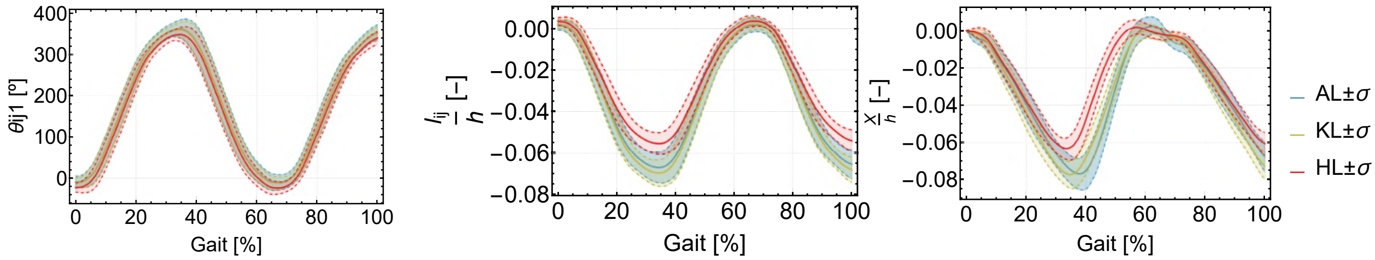
A first design choice has to be made prior to obtaining any results: the location of the anchor points. Figure 5 shows how the torque applied to each joint by the actuation unit depends on the position of both anchor points in the case of the hip and knee of one of the subjects of the studied population, more specifically, subject 29 (c_{Joint} has a value between 0 and 1 and represents the length of the segment at which the anchor point is located). The results obtained show a rather obvious conclusion: locating the anchor points at the joints themselves maximizes the torque produced in such joints since the moment arm is maximum.



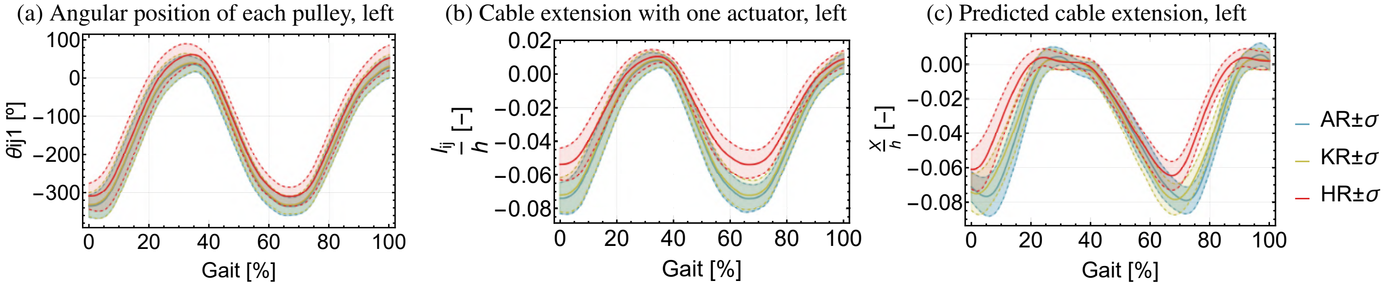
(a) Hip (b) Knee (c) Ankle

Fig. 5: Applied torque at each joint as a function of the anchor and trajectory points position for a unit cable force.

As expected, torque at the ankle joint does not depend on the position of the trajectory points, for only the anchor points' position makes the applied torque vary. As an example, anchor points are located at 60% of the thigh (downwards) for the hip, 80% of the thigh and 30% of the shank for the knee, and 30% of the shank for the ankle (the final anchor point, in this case, is located at the heel). This configuration is not the best from a torque perspective, but will serve as an example: maximizing the moment arm generally implies locating the anchor points at, or close to the joints themselves, which might not be possible or comfortable for the wearer, the main reason why we decided to illustrate our results with a more common configuration. An additional structure located at the heel could be added to locate the anchor point further away from the ankle and thus decrease the force needed in the cable to produce the same torque. Once the exosuit design is proposed and the gait data is provided, cable extensions are calculated and then taken as inputs for the PCA analysis that will ultimately result in the required number of actuators, pulley radii, and angular displacement at each motor's shaft. With the cable extension data standardized, PCA can be conducted to yield surprising results: only the first principal component accounts for around 94% of explained variance. This leads to a notable conclusion: a transmission system between both legs, such as a clutch, might be expendable, for only one actuator with six pulleys can replicate very accurately the human movement during gait for both legs. The pulley diameters, given by Eq. 6, might have opposite signs for either the left or right leg, meaning that cables would have to be rolled up in opposite directions. Notice that Eq. 6 gives the value of the radii for each pulley multiplied by a constant, as the system's eigenvalues, which allows the designer to choose a size that fits their specific requirements, attending parameters such as the cable section or the desired size for the actuation unit. Equation 5 yields the position control for the actuator. The six angular and linear cable displacement functions for this population are:



(a) Angular position of each pulley, left (b) Cable extension with one actuator, left (c) Predicted cable extension, left



(d) Angular position of each pulley, right (e) Cable extension with one actuator, right (f) Predicted cable extension, right

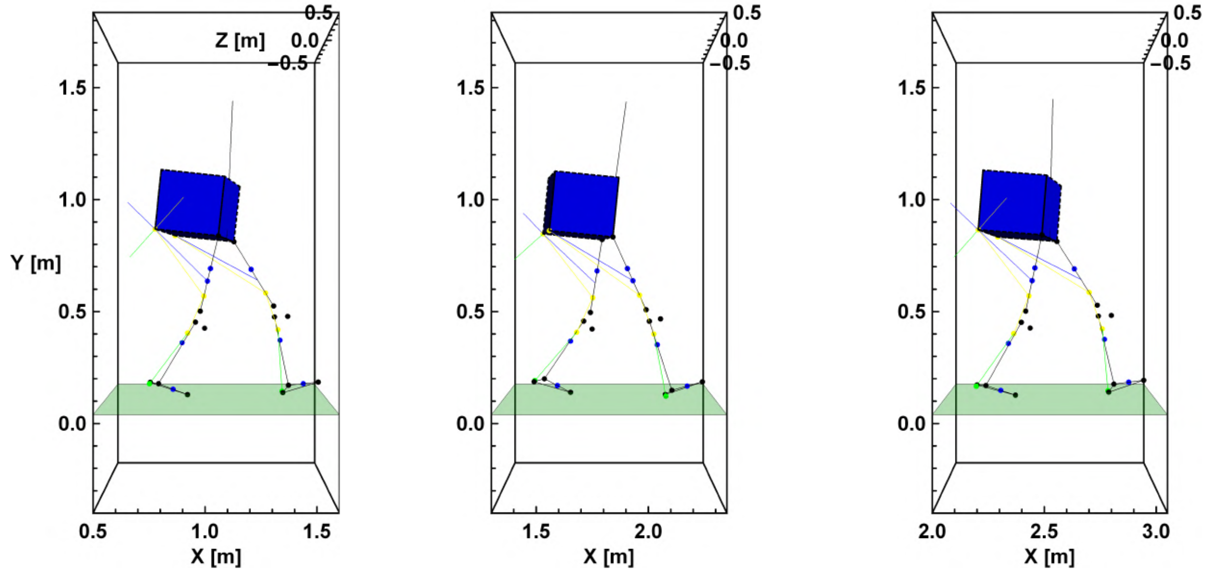
Fig. 6: Motor position control θ_{ij1} and cable extensions Δl_{ij}, along with the predicted cable extensions during gait X, for each leg (left and right) and joint (hip, knee and ankle). The average value is shown in a solid line, while dashed lines of the same color represent the standard deviation with both signs.

In Fig. 6, AR stands for joint "A" (ankle) and "R" (right) leg. Dashed lines represent the mean value plus and minus the corresponding standard deviation, while solid lines stand for the average values. The cable extension results have been normalized with the subject's height, so that they are comparable for all subjects. The decision of studying subjects of different heights and genders is to account for the whole target population that we focused on: elderly people who might need walking assistance. Notice that, even though each pulley has a different mean value for its angular displacement due to different starting values, the derivative is equal in all 6 of them, meaning that the use of a single actuator may be possible. The difference between their values represents the different amount of cable that is wrapped around its corresponding pulley at each time instant. As shown, the cable extensions obtained with only one actuator well resemble those predicted by the model for the chosen anchor point distribution. For the chosen pulley size, each of them has to spin at a noticeably high rotating speed, since the angular displacement in only around 40% of the gait cycle is over 250º for the right leg, as shown in Fig. 6. This can be solved, for instance, by increasing the pulley radii via coefficient C in Eq. 6. Cable extension is measured as the difference between the total cable length at a given instant and the instant when each leg's cycle starts, that is, when its initial ground contact happens. Now, to visualize the results of Fig. 6, a 3D model of the exosuit described above is shown in Fig. 7, when operating during the gait cycle at the time instants highlighted in Fig.6, that is, 0%, 33.5%, and 66% gait. Those instants correspond to the initial ground contact for the left leg, the initial ground contact for the right leg, and the second initial ground contact for the left leg, respectively.

Please notice that the gait cycle considered here starts when the left leg first touches the ground and ends when the right leg touches the ground for the second time so that the ground contact force information included in [28] can be considered for the solution. In Fig. 7, a human subject wearing the exosuit during gait is shown (subject 29 in [28], for illustrative purposes), including a backpack where the actuation system and the pulleys are to be located. The cable segments that stay above the pulleys represent the cable extensions, calculated as the total cable length minus the cable length at the start of each leg's cycle (initial ground contact). Thus, at 0% gait, the right leg shows high values of the extension, while they are zero for the left leg, and the opposite happens at 33.5% gait. At 66% of the gait cycle, the left leg touches the ground for the second time, yielding cable extension values similar to those at 0% gait, as highlighted by red lines in Fig. 6. Had the same time instant been considered as a reference for both legs, the offset between both sets of angular displacements in Fig. 6 would not exist.

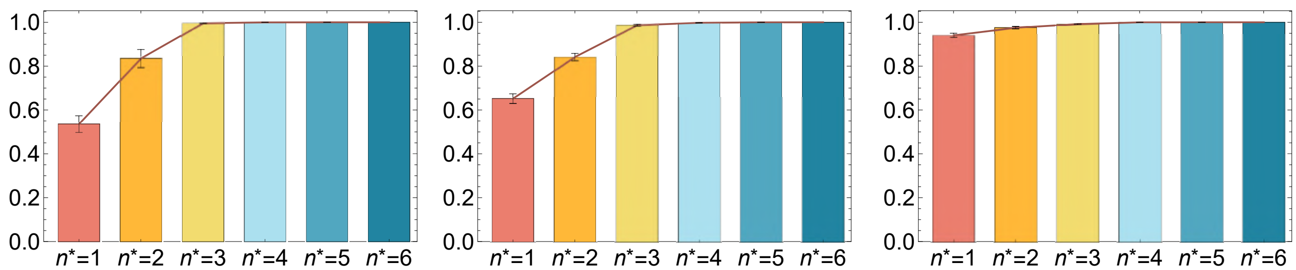
PCA yields the cumulative variance, a more precise way to estimate the correspondence between the obtained results and the predicted ones, rather than only comparing the expected cable extension to the one obtained with the selected number of PCs, as in Fig. 6. Figure 8 shows how different values of n^* are able to reconstruct the gait according to their respective cumulative variance for flexion-extension angles, segment angles, and cable extensions for the chosen population.

Notice that the VAF for the first PC ($n^* = 1$) is much higher for cable extensions than for any of the other two for the selected case. Moreover, the standard deviation for all PCs is also lower for the cable extension case when compared to the more traditional segment and joint angles.



(a) 0% Gait (b) 33.5% Gait (c) 66% Gait

Fig. 7: Cable extensions at 0%, 33.5%, and 66% of the gait cycle.



(a) Flexion-extension joint angles (b) Segment angles (c) Cable extension

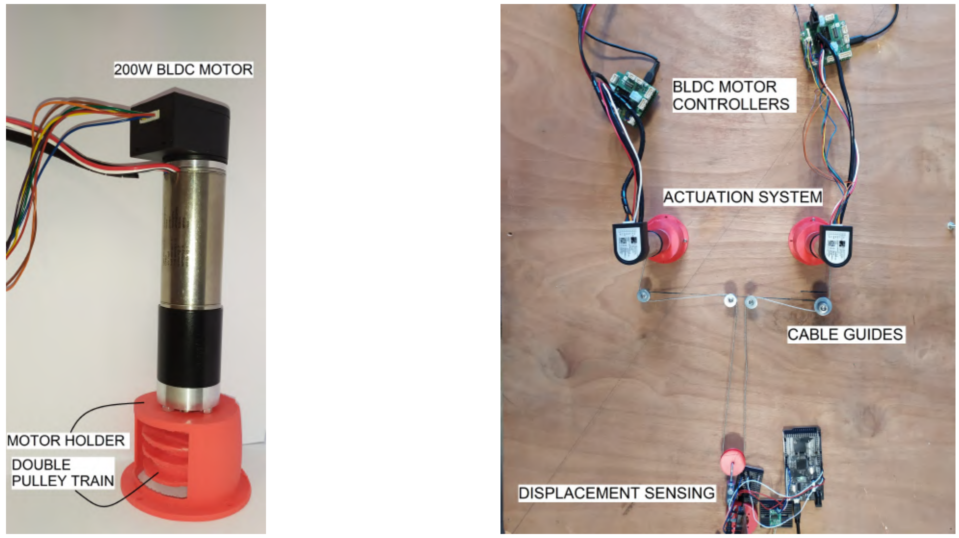
Fig. 8: Cumulative variance for various n^* values and gait variables for the target population. Average values are shown along with their corresponding standard deviation.

5 Validation

A testbench is designed and constructed to measure synergy-based cable actuation in comparison with full-measurement results with IMUs and a force plate. In order to test the results described above, a simpler version will suffice. Figure 9 shows a testbench designed to reconstruct the hip and ankle cable extensions for the same subject, on only the right leg and with one and two actuators simultaneously. The testbench is built with two 200W Maxon EC-4Pole actuators controlled by EPOS 4 50/8 controllers and with a 33:1 and a 79:1 reduction each. Pulleys and motor holders are 3D printed in PLA and the force is transmitted via a 1.5 mm, stainless steel cable with a 26 kg safe working load. Bowden cables could be used as well, although that would bring non-negligible friction losses with the sleeve [32, 33]. Cable extensions are measured using a linear displacement sensor based on a potentiometer, attached to the central pulleys thanks to steel cable guides. A spring with a maximum extension of 29.7 mm and a 1060 N/m spring rate joins the cable with the position sensors system.

The test rig can simulate cable extensions under scaled, normal gait conditions reconstructed based on either one or two synergies (actuators). Motors receive signals that follow the principal components obtained before, while the pulley trains have their diameters according to Eq. 6. If two motors actuate at the same time, results are expected to match 100% of the theoretical cable extensions, whereas if only the one corresponding to the first PC does, a slightly worse approximation is reached, as in Fig. 10.

Figure 10 shows how even with only one motor, the cable extension function is followed by the corresponding first PC without a significant difference when compared to the dual actuation results. Since only two joints are being actuated, two actuators acting simultaneously on both joints should perfectly replicate the measured cable extensions during the gait cycle. Both sets of results present an error due to imprecisions in the testbench but clearly show how the combination of two actuators reaches a closer solution than only one, but only to a small extent. This is of course due to the choice of cable extensions as a target variable for the PCA and the consequent design proposal of the transmission system, Fig. 3 and 4, and



(a) Actuation system. Motor, motor holder and double pulley

(b) Testbench overview

Fig. 9: Testbench for cable extension prediction with one and two simultaneous actuators.

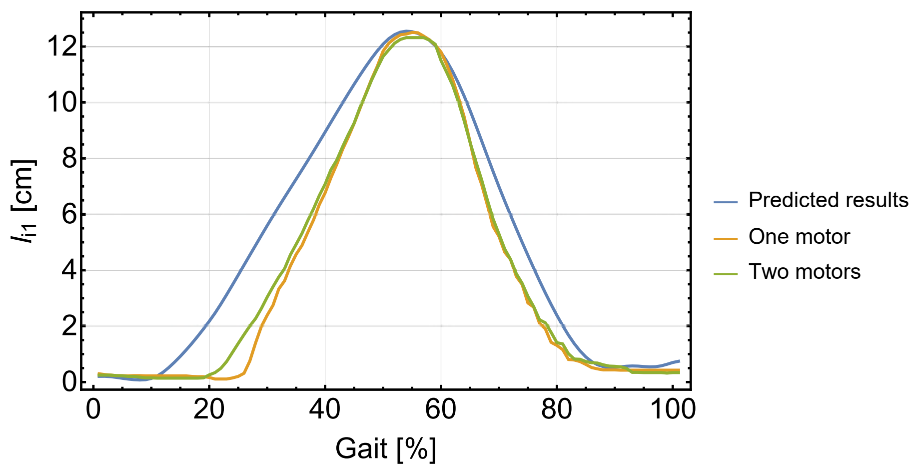


Fig. 10: Cable displacement results measured in the test bench for one motor (orange) and two motors (green) for the hip actuation system. Results predicted by the model in the considered gait cycle are shown in blue.

Eq. 6. Naturally, actuation might not be possible in every joint, at every time, for there are some limitations. For instance, depending on the exosuits design, a joint might not be suitable for actuation if the cable that actuates it is under compression, that is, the first derivative of its extension function is positive. It is the case in the design described in 1b. An extensive analysis on when it is possible to actuate in a soft-exosuit for leg rehabilitation is provided in [34].

Now, even though the synergy-based study proposed here suggests that it may be possible to actuate both legs during gait for human walking assistance, there are still concerns that should be approached in the future. A precise study on which actuation phases are possible, for instance, would have to consider many design choices, such as the maximum torque acceptable at the motor shafts or the power available at the actuation units: a compromise must be found to keep the weight and price of the overall system under control. During some stages of human gait, as Fig. 6 suggests, some cables will be under compression and, therefore, will be unable to transmit any forces. Different actuation systems to approach this circumstance in synergy-based assistance devices are found in the literature, as in [13] for hand assistance. Here, however, a valid synergy-based approach for the lower-limb assistance is proposed, aimed at guiding in the design of lighter, cheaper, and more wearable exosuits.

6 Conclusions

An innovative, synergy-based design philosophy was proposed for walking-assistance exosuits. A symbolic analysis of kinematic variables was conducted using PCA, which led to an expression for pulley radii and motor control schemes achieving around 94% of the total variance with only one actuator for six joints and a given population of elderly subjects walking comfortably over the ground. That was possible thanks to choosing cable extensions as the target variable, as has been done already for upper-limb assistive devices, instead of segment or joint angles. A general transmission system was conceptualized to act any number of joints with any number of simultaneous actuation units, along with an exosuit design based on cable-driven actuation, with anchor points located where they would provide the corresponding joint with higher torque. Reducing the number of required actuators in such a drastic manner opens the possibility of acting both legs with only one motor, once the actuation phases during gait are determined. Only having one actuator, however, might lead to high motor torque requirements in certain gait phases if acting all DOFs at the same time. Therefore, characterizing forces and torques in the chosen actuation phases may be relevant when designing synergy-based, gait assistance exosuits. Results were tested on a testbench designed to reconstruct the cable extension function for one joint (hip or ankle) in an actuation scheme built to actuate both joints simultaneously: removing one actuator, the one corresponding to the second and last PC, did not have a great impact on the result, as expected by the statistical model.

Acknowledgements

This research was funded by the Spanish Ministry of Science, Innovation, and Universities, through project PID2019-107491RB-I00 and the Junta de Extremadura through project IB18103, thanks to which the current research is possible.

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