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Design of a 3-DoF Joint System with Dynamic Servo-Adaptation in Orthotic Applications

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Abstract—**Most exoskeleton designs rely on structures and mechanical joints that do not guarantee the right match between the orthosis and the user. This paper proposes a virtual joint model based on three active degrees of freedom aimed to emulate a human joint. This joint is capable of performing a dynamic servo-adaptation in real-time to avoid misalignments and to provide a flexible adjustment to different users' sizes in order to avoid undesirable interaction forces.**

I. Introduction

One of the challenges in rehabilitation and assistive exoskeletons is achieving the reproduction of upper and lower limb movements, accurately and efficiently, to maximize the benefit of therapies. This means designing comfortable and safe robots, which final goal is to avoid pain, reduce recovery time and improve the functionality of the injured limb. In this sense, the main focus of research in this area has been oriented to control issues that deal with human-machine interaction. However, little advances have been made in the ergonomic design of exoskeleton structures that minimize the undesirable transmission of forces to the patient. Referring to limb joints, these interaction forces, between human and robot, can reach up to 250 N and torques up to 1.46 Nm [1]. Such forces and torques are produced mainly due to two reasons:

- Misalignment caused by the migration of the center of rotation of the biological joint during the therapeutic movement.
- Misalignment between the center of rotation of the orthotic joint and its biological counterparts, which can occur due to mismatches in the initial adjustment of the robot or during the movement.

These misalignments cause a significant alteration in the normal muscle activation pattern [2], which can produce fatigue, pain and even induce the user's rejection of the ongoing rehabilitation therapy.

Studies related to upper limbs design, [3][4] and [5] have demonstrated that the position of the rotation centers of these joints vary according to the movement and applied forces. Moreover, human anatomy is highly variable from one individual to another within a given population, having strong influence gender and age. Anthropomorphic measures of the arm and forearm vary considerably between patients. This fact determines that exoskeletons should be designed in such a way that they facilitate their adjustment in size, and furthermore, the position of the joint rotation center should be controllable. Failure to solve these problems produce a mismatch between the center of rotation of the human and the orthosis joints that together with the offset produced during the movements, can grow up to values of ± 10 cm [1].

The variability of joints location between individuals demands the design of systems that can be adjusted within a certain range. This requirement implies the need of taking anthropometric measurements of each patient and adjusting the exoskeleton in each therapy session, which most often constitutes a not simple process that takes long preparation times.

This paper proposes a new joint design, with three active Degrees of Freedom (DoF) aimed to emulate a 1DoF movement. This joint can perform a dynamic servo-adaptation in order to adjust itself to different user's sizes, compensate the migration of Instantaneous Center of Rotation (ICR) and avoid mismatches during the users' movements.

Section II describes the biomechanical modeling of upper and lower limbs joints, the solution developed for different rehabilitation devices to match the joints movements and the problems associated with them. Section III presents the joint system proposed and its application to the different upper/lower limbs joints. The model description and the equations of motion are developed in section IV and finally performance tests of the joint system for different types of misalignment are developed in section V.

II. Joints Anatomy v/s Mechanical Joints Design

Most exoskeleton designs are based on models of human joints that behave like a hinge, for those joints having 1DoF, or like a ball and socket joint for those having 3DoF. However, very few consider the complexity of biological joints, which have a combined movement of rotation-

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translation and one or more additional DoFs surpassing those considered in most mechanical models.

A. Upper Limb

In upper limbs, elbow joints design follows the hinge joint model (pronation-supination is associated with the forearm movement). An elbow joint constitutes a simple design, which minimizes energy consumption (by placing a single actuator), weight and sensorization needs furthermore simplifies control. However, its design presents some dynamic and kinematic constrains limiting its DoF. Most existing exoskeletons [6][7][8][9][10][11][12] and [13] have 1DoF for flexion-extension and a structure that fits one side of the arm, trying to match the axes of rotation, being all adjustable within a range of fixed measures. The exoskeleton ABLE [14] (1DoF), presents a structure below the elbow (which impedes its alignment to this joint), however this structure is conceived only for guidance, since the forearm and wrist are not subjected to the exoskeleton, thereby causing a relative motion between the arm and the device.

Other devices have tried to solve this limitation by adding more DoFs to the elbow joint, so that through redundancy they compensate the offset misalignment between axes. Thus, WREX [15] incorporates two passive DoF at the elbow joint, in different planes, allowing the user's elbow to naturally align with the exoskeleton.

The work in [16] goes one step further in the design of a system for the elbow joint with 4DoF, where two of them are actuated (flexion-extension and pronation-supination), while the other two are passive, and used to compensate the misalignments and to prevent that the forces generated by the orthosis are transferred to the arm. However, it can only compensate 70% of the produced interaction forces and 60% of the torques transmitted to the arm.

Referring to the shoulder, this articulation is considered as a ball and socket joint, and is usually modeled with 3DoF for flexion-extension (φ_{FE}), abduction-adduction (φ_{AA}), and internal-external rotation (φ_{IER}) movements [6][7][9][11][12] [14] and [15]. However, shoulder devices try to emulate the resulting motion and not the biological joints movements. Anatomically, the human shoulder has five joints working simultaneously, resulting in combined movements of rotation and translation. Some devices try to reproduce one such joint adding more DoFs. In [10] and [13], two active DoF for the sternoclavicular joint are added to reproduce the depression-elevation (φ_{DE}) and retraction-protraction (φ_{RP}) movements. ESA exoskeleton [16] uses 6DoF (two of them are passive) which do not intersect the rotation biological axes of the shoulder since the arm is wearable and is attached to the operator body. The problem with this design is that the device cannot exert enough force and torque for rehabilitation and assistance because it is designed to work in non-gravity environments.

B. Lower Limb

Referring to the lower limbs, the knee is considered as a hinge joint to execute the flexion-extension movement, while the hip constitutes a ball and socket joint to produce a wide movement. Robotic devices seek also to assist and rehabilitate the gait (lower limbs exoskeletons seek rehabilitation of several functions. Most of them are rehabilitation systems supported on a treadmill and consider both hip and knee joints as hinge joints (1DoF) [17][18][19] [20][21] and [22], simplifying models. The emulated resulting motion does not consider the true movement of the biological joints, causing problems associated with axes misalignments.

Studies of the knee have shown that its ICR is displaced with a mean value of 17 mm [23] and the orthosis slides along the legs, a run of about 20 mm [24] during the extension movement. Others have concluded that current knee orthosis do not provide efficient protection to the knee movement [25].

Referring to the hip joint, it has been shown that its ICR displaces up to 29 mm and there is an offset of up to 20 mm during sagittal plane movement [26]. Some researchers have developed joints with the aim of reducing misalignments, but they only compensate the displacement of the ICR and not the mismatch between the orthosis and the leg [27].

III. PROPOSED KINEMATIC MODEL

To avoid the misalignment of upper/lower limbs joints, the challenge is to find mechanical adaptations that satisfy joint kinematics requirements, trying to reproduce the true movement of the joint, avoiding the initial offset between orthosis and patient at the beginning of a therapy and compensating the mismatch with the biological joint during the rehabilitation movements.

In order to face these requirements, a dynamic servo-adaptation based on a three active DoF joint is proposed. The adequate control of the three actuators provides a variable center of rotation, variability that has three objectives; accommodate to different anthropometric arm/leg measures, avoid the mismatching produced during the execution of a movement and compensate the migration of the ICR produced by the angular joint movement.

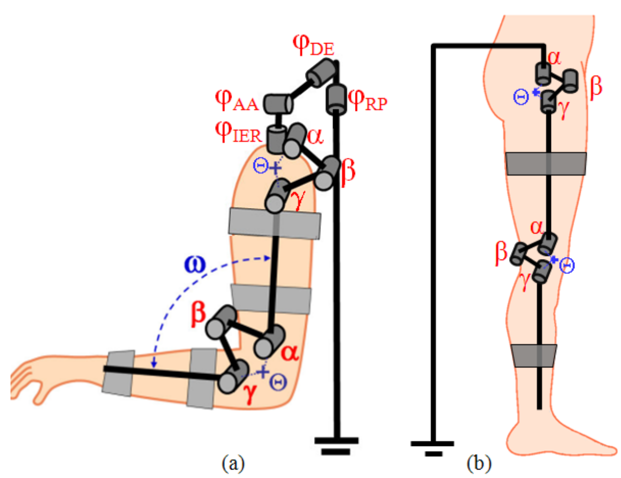


Fig. 1. Kinematic model of (a) upper limb. (b) lower limb.

This joint system can be applied to any of the joints that produce movements in the sagittal plane, both the upper and lower limbs, as shown in Fig.1, where α, β and γ are the joint angles of the 3DoF system actuators, **ω** is the sagittal plane angular movement and **Θ** is the ICR for the shoulder, elbow, hip and knee joints.

The use of three actuators instead of only one implies extra mass and inertia on human limbs, so such joint system cannot be used in assistive exoskeletons, where the devices should be portable, lightweight and autonomous. It is mainly aimed for its use as rehabilitation exoskeleton, where the weight structure is supported from its base and energy consumption is not a problem at all.

IV. Method

The system consists of two links (**1**, **4**) attached to upper/lower limbs that connect three actuators in serie (**A**, **B** and **C**) separated by other two symmetrical links of length **l** (**2**, **3**), Figure 2. Actuator **B** produces the joint rotation, while the other two (**A** and **C**) are necessary to compensate the displacement that is produced in the rotation plane.

The joint system should be fitted to the user at the beginning of a therapy and keep maintaining this configuration during its execution. Therapy starts from a rest position, where the distance between actuators **A** and **B** is **2l** (Fig. 2a). Then it is necessary to fit the orthosis to the specific anthropomorphic size of the user by reducing the separation in **2Δ** (Fig. 2b). For the execution of the therapy, joints **A**, **B** and **C** rotate respectively angles **α**, **β** and **γ**. These angles produce a symmetrical configuration of the system forming an isosceles triangle. In this case, angles **α** and **γ** are equal and they are calculated as a function of the desired length reduction, **2Δ**, as follows:

$$\left(2(l-\Delta)\right)^2 = 2l^2 - 2l^2\cos\beta \quad (1)$$

$$\boldsymbol{\beta} = \operatorname{acos}\left(\frac{2l^2-\left(2(l-\Delta)\right)^2}{2l^2}\right) \quad (2)$$

$$\boldsymbol{\alpha} = \boldsymbol{\gamma} = \frac{\pi-\beta}{2} \quad (3)$$

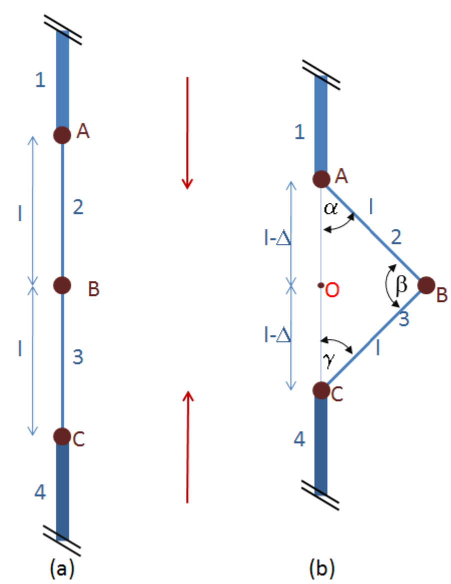


Fig. 2 (a) System joint at the beginning of a therapy. (b) Displacement **Δ** of links **1** and **4** to reduce the length of the chain.

From the initial set position (Fig.2b), the joint system is actuated to reach the desired angle **ω**, maintaining this initial setting, **2Δ**. Fig. 3 shows how the articulated system is attached to the patient arm and adapted to the elbow flexion-extension movement.

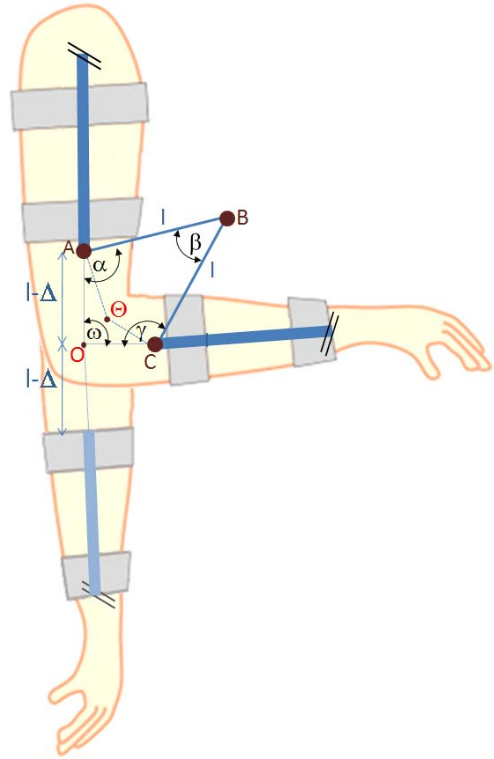
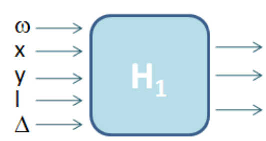


Fig. 3. Elbow joint system during the flexion-extension movement.

To perform a movement, the three actuators must rotate around the same center of rotation **Θ**, which in turn must match with the ICR of the biological joint. This requirement ensures the correct axes alignment and avoids mismatches between the orthosis and the user's limb during the therapy. This center of rotation **Θ** may shift, so that the actuators should constantly adapt to compensate this undesirable shift. Therefore, the actuators movements **α**, **β** and **γ** must be a function of the desired angle, **ω**, the geometry **l** of links **2-3**, the reduction length **Δ** and the position of the ICR **Θ(x,y)** with respect to its location at the starting of a therapy. Thus, it is necessary to find the transformation matrix $\mathbf{H_1}$:



The Parallelograms **ABCΘ**, **ABCO** and the triangles formed between them are used to find the expressions of **α**, **β** and **γ** (Fig. 4), and through trigonometry it is possible to determine:

$$n = \sqrt{x^2 + (l-\Delta-y)^2} \quad (4)$$

$$\varphi_1 = \tan^{-1}\frac{x}{x-\Delta-y} \quad (5)$$

$$\varphi_2 = \tan^{-1}\frac{x}{x-\Delta+y} \quad (6)$$

Where $\boldsymbol{n}$ is the distance between ICR **Θ** and actuator **A**, φ_1 is the angle formed by **Θ'**A**Θ** (**Θ'** is the projection of **Θ** in the axis **y**) and φ_2 is the angle formed by **Θ'CΘ**, all of them component of the triangles generated during the ICR migration.

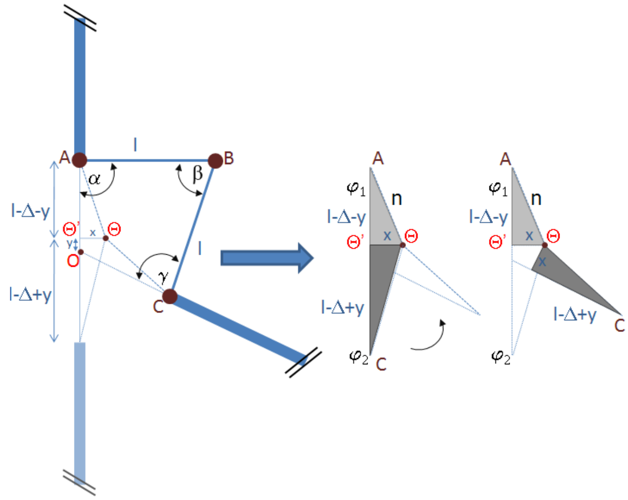


Fig. 4. Joint system movement for an ICR **Θ**.

Applying the law of sinus and cosinus, trigonometric identities and the symmetry of the model and using (4), (5) and (6), the equations of motion that govern the system can be obtained:

$$\boldsymbol{\alpha} = \pi - \sin^{-1}\left(\tfrac{n}{l}\cdot\sin\left(\tfrac{\omega}{2}+\varphi_1\right)\right) - \tfrac{\omega}{2} \quad (7)$$

$$\boldsymbol{\beta} = \sin^{-1}\left(\tfrac{n}{l}\sin\left(\tfrac{\omega}{2}+\varphi_1\right)\right) + \sin^{-1}\left(\tfrac{n}{l}\sin\left(\tfrac{\omega}{2}+\varphi_2\right)\right) \quad (8)$$

$$\boldsymbol{\gamma} = \pi - \sin^{-1}\left(\tfrac{n}{l}\cdot\sin\left(\tfrac{\omega}{2}+\varphi_2\right)\right) - \tfrac{\omega}{2} \quad (9)$$

Equations (7), (8) and (9), allow the system adapt itself to any ICR variation, maintain the initial adjustment of the orthosis and avoid any offset between device and user's limb.

V. Results

To evaluate the behavior of the designed joint, a modeling system was developed in LabVIEW, which interface includes the possibility to generate different ICR pathways, and control the variables **l**, **Δ** and **ω**. This interface allows obtaining the actuators motion graphics and generates an animation of the movement of the joint system.

The designed joint was tested under different variations of the ICR, considering the length of the links **1** and **4** of 20 cm, and that of links **2** and **3** of 10 cm and the initial adjustment **Δ** of 10 cm. Fig. 5 shows the joint system in its initial position and the initial adjustment to fit to the user. **A**, **B** and **C** are the actuators and Θ_i is the ICR for a flexion angle in limb configuration.

From this point the joint system can be adapted to any change of the ICR along a therapy. To test the joint system's behavior, three different pathways of the ICR are considered:

1. It is assumed that the biological joint has a Fixed Center of Rotation (FCR), hypothesis assumed by almost all exoskeletons.
2. Elbow ICR. The elbow joint ICR is modeled according to the data obtained from two upper extremities of an unembalmed cadaver with no pathological condition using radiopaque markers and roentgenograms [28]. Its pathway is shown in Fig. 6a.
3. Knee ICR. The knee joint ICR is modeled according to the data obtained from a mechanical model in which surface markers and an imaging system is used [23]. Its pathway is shown in Fig. 6b.

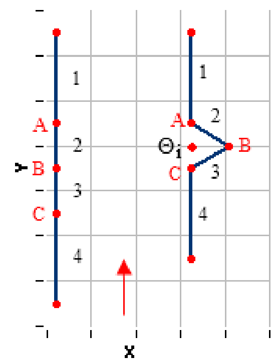


Fig. 5. Initial adjustment of the joint system.

These centers of rotation pathways were obtained for a variation of the elbow/knee, flexion-extension angle **ω** from 180º to 90º.

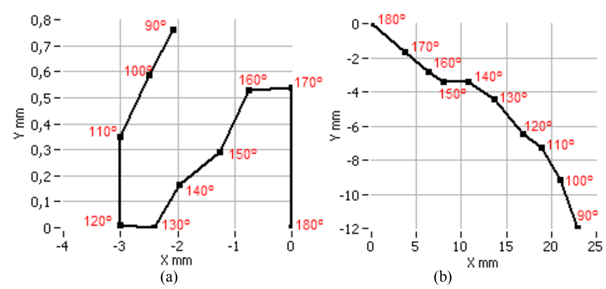


Fig. 6. ICR pathway of (a) Elbow joint. (b) Knee joint.

Applying the three different modeled pathways (Fig. 6) to the proposed joint model, and considering that the starting point at 0º corresponds to Θ_I, the joint system behavior for each of them is obtained.

Fig. 7 shows how the system adapts to any change in ICR for the three pathways and has different behaviors to maintain its alignment.

Fig. 8 shows the different responses of the three actuators to adapt to the ICR movement for each pathway. From the different cases, it can be seen that actuator B controls the flexion-extension movement of the joint system, while the other two seek to compensate the ICR shifts. In movements without ICR variation (red line), actuators A and C moves symmetrically as expected.

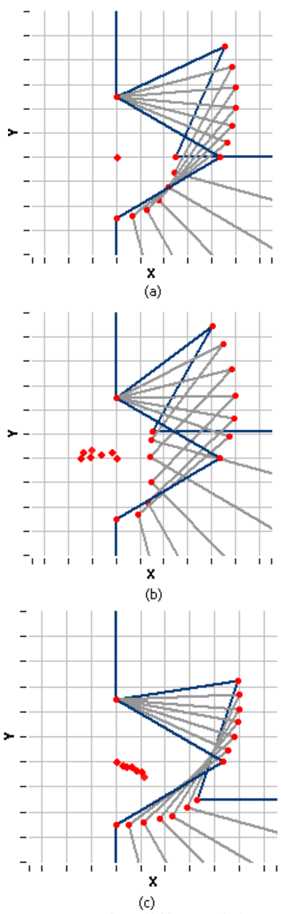


Fig. 7. Joint system response for different joint pathways (a) Fixed center of rotation (b) Elbow ICR (c) Knee ICR

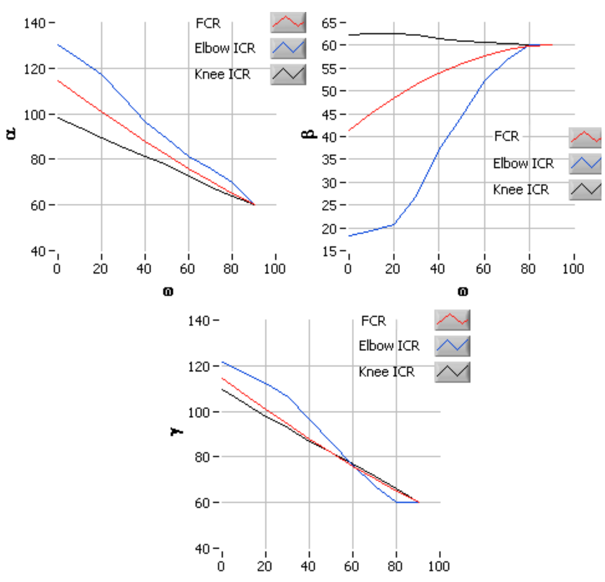


Fig. 8.Actuators response of the joint system for different ICR pathways.

VI. Discussion

The above results illustrate how the proposed joint system eliminates the offset that generates the initial adjustment of the orthosis with the patient and the mismatch that occurs during a therapeutic movement between the devices and the limb, thanks to its adaptability. In addition, it is able to track the ICR variation of the joint.

However, the determination of the joint ICR is not a simple task. It can be achieved via observational methods, geometric analysis, vision systems, electromagnetic motion tracking, etc., and from the obtained data compute its position dynamically through Euler angles or screw theory, among others.

Each method has assumptions and constraints that limit its effectiveness. The data acquisition systems most frequently used are based on markers (vision system and electromagnetic motion tracking), because they are more precise and accurate than the geometric analysis and observational method. However, several problems limit the reliability of these methods based on the data obtained from markers on the limb surface:

- First, the movement of soft tissues (muscles, skin) can generate a relative shift between markers and bones.
- Second, some assumptions (such as considering that human segments behave as rigid bodies) necessary to determine the center of the joints, are not accomplished.
- Third, the use of markers requires a suitable environment for signal acquisition and a considerable preparation time.

Motion analyses based on markers are extensivelly used to characterized the limb kinematic behavior like gait analysis, sports, etc., but the problems described above restrict their use in real therapeutic practice. Therefore, in this area it is necessary to propose practical solutions that allow the axes of the orthosis joint to track its anatomical counterpart.

The total misalignment between orthosis and limb is produced by three causes: ICR migration, initial offset and movement mismatch. However, ICR migration is much less relevant than offset and mismatch. Thus an exact tracking of the ICR during the movement is not necessary.

Therefore to know the position of the ICR and avoid the total misalignment a model is studied based on the measurement of interaction forces between the orthosis and joint.

To analyze these interaction forces, we are developing a force sensor from which a force model can be elaborated for its control in real time.

Next step will be the development of position and force based control strategies to be able to minimized the interaction forces and perform experiments to validate the joint system design.

VII. Conclusion

The purpose of this work is the replacement of 1DoF joint in the sagittal plane by a 3DoF joint system that allows avoiding the total misalignment between the orthosis and the user's limb. The proposed solution covers the whole anthropometric range of patients and reduces the time required to adjust the exoskeleton to the user's limb before starting a therapy.

By reducing misalignments, the interaction forces applied to the patient's are reduced as the joint system emulates the real behavior of human joints.

This study presents a theoretical model of a 3DoF joint system to solve axes misalignments. Further works will be addressed to the study of the interaction forces between orthosis and patient, to elaborate a force model that may be able to predict and minimize these forces and finally its implementation and validation.

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