Review Article
Interaction Control for Human-Exoskeletons

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In this work, a general concept of the human-exoskeleton compatibility and interaction control is addressed. Rehabilitation, as applied to humans with motor control disorder, involves repetitive gait training in relation to lower limb extremity and repetitive task training in relation to upper limb extremity. It is in this regard that exoskeletal systems must be kinematically compatible with those of the subject in order to guarantee that the subject is being trained properly. The incompatibility between the wearable robotic device and the wearer results in joint misalignment, thus introducing interaction forces during movement. This, therefore, leads to the introduction of the need for interaction control in wearable robotic devices. Human-exoskeleton joint alignment is an uphill task; hence, measures to actualize this in order to guarantee the safety and comfort of humans are necessary. These measures depend on the types of joints involved in the rehabilitation or assistive process. Hence, several upper and lower extremity exoskeletons with concepts relating to interaction forces reduction are reviewed. The significant distinction in the modelling strategy of lower and upper limb exoskeletons is highlighted. Limitations of certain exoskeletal systems which may not allow the application of interaction control are also discussed.

1. Background

Disabilities in the upper and lower limbs of humans may be age-related, accident-related, or pathology-related [1]. According to the world population prospects data, which were revised in 2019, 16% of the world population will be over the age of 65 in 2050. It was further stated that the number of individuals who are 80 years of age is expected to triple in 2050, i.e., from 143 million in 2019 to 426 million in 2050.

Stroke is known to be a major cause of disability and reduced life quality [2]. The physical effect of a stroke may include arm, leg, hand, and foot movement difficulties or disabilities. Stroke may happen to anyone at any time; however, aged people are at a higher risk. According to Garcia-Aracil et al. [3], about 75% of stroke sufferers survive 1 year after. This proportion is envisaged to increase in the coming years due to emerging technologies committed to providing additional therapeutic exercises through robotic interventions.

In recent years, the demand for physical therapy services across the globe has grown partly because of the ageing populations that we see today. The rate of survival from medical disorders such as stroke is on the increase due to the role of technology in the healthcare system.

In any of the disability-related cases highlighted above, the resultant effect could be linked to a nerve or muscular injury which may lead to improper motor functions and eventually paralyses. In a bid to provide a solution to people with these disabilities, the need for rehabilitative and assistive devices to support the mobility of such individuals is paramount.

Robotic rehabilitation and assistive technologies not only have the potential of changing older people's lives and

improving quality of life of disabled patients but also ease the stress on physiotherapists.

1.1. Robotic Rehabilitation.

Mobility disorder caused by SCI or related illnesses in people has been on the increase in recent years [4]. Patients with such disorders are often dependant on others in order to carry out their day-to-day activities. To help alleviate the difficulties these people go through, therapeutic intervention is often the sole recommendation. However, the effectiveness of a therapeutic intervention is dependent on its intensity and frequency.

Due to a lack of physiotherapy resources, a number of patients may not be able to perform the exercise required for their recovery, which, in turn, may prolong the overall period of their recovery. Robotic systems not only provide effective and repetitive gait training but also reduce the burden of physiotherapists while gathering quantitative data about a patient's performance in a bid to track the patients' progress or condition.

1.2. Assistive Robotics.

Disabilities or chronic conditions which may have resulted in the physical limitations of some patients often require assistive technologies in the form of exoskeletons and soft robotics. These assistive technologies do help enhance and prolong the patient's independence. Assistive robotics, in this case, is inclined towards the assistance of individuals with limited physical capabilities due to accident, disease, or illness and ageing people with reduced physical strength.

2. Introduction

Exoskeletal devices are characterized by their anthropomorphic nature. They are usually fitted with actuators at the human joint level and possess a level of the intelligence of humans in conjunction with robot mechanical strength [4]. The main purpose of these devices is to provide patients with limb disabilities a medium by which they can augment or restore a measure of their motor function so as to enable them to regain partial or complete control of their limbs [5]. To achieve this, it necessitates the design of rehabilitative and assistive (by assistive, the exoskeleton actuation mechanism does not enforce the movement of the user but only performs the duty of a supportive mechanism) protocols intended to be executed via certain control strategies. These protocols may be designed based on two approaches, which are motion intention detection and prespecified targeted task.

In the prespecified targeted task, the user is made to follow predefined or target trajectories usually collected from healthy individuals [6], those defined by the physiotherapist [7], or designed using central pattern generators to mimic the movement pattern of the subject [8]. Motion intention detection refers to utilising brain current motor activities extracted by EEG to predict the next voluntary motor task of humans [9]. Hence, it provides a basis for which the intent of completely paralyzed patients with reduced communication capabilities may be determined,

based on a range of different electrophysiological signals [10]. This may be achieved with the aid of a brain-computer interface (BCI). These approaches are all inclined towards achieving the following objectives or applications: gait rehabilitation (this application focuses on providing certain aids to patients having any form of mobility disorders so as help rehabilitate their musculoskeletal strength, motor control, and gait), human locomotion assistance (this type of application is aimed at assisting spinal cord injury (SCI) or paralysed patients with reduced or no motor/sensor function), and human strength augmentation (this application is targeted towards enhancing the performance of able-bodied persons). Control strategies needed for these applications to be performed effectively are centred generally on motion control, force control, and interaction control.

Motion control in humans typically encompasses the method by which humans control their movement when performing a number of different physical activities (these activities may include running, walking, standing, holding objects, and so on) [11]. To achieve this, the need for a working knowledge of the mechanical and control behaviour of the neuromusculoskeletal system is necessary. It is this known behaviour that is used for the modelling of rehabilitative and assistive devices in the form of orthoses and exoskeletons, which compensates for the failure of the human neuromusculoskeletal system. A diagrammatic description of this behaviour is given in Figure 1.

Considering the block diagram of a conceptual feedback control system scheme in Figure 1, the building blocks required involve the linkage system (body segments), actuators (muscles), sensors (proprioceptive and tactile sensors and visual and vestibular system), and the controller (Central Nervous System (CNS)).

The desired trajectory (electrophysiological signal) is compared with the actual trajectory of the limb, which, in turn, allows the CNS to send a neural signal to the muscles. The muscles then exert forces on the skeletal system which will start moving if it is not constrained by the environment. As seen in Figure 1, two inputs are considered. The desired trajectory input which is transferred to the actual trajectory in order to track the behaviour of the system, and the disturbance input which is introduced with an intention to deviate the system from its intended trajectory/position. These disturbance forces may be friction or environmental forces acting on the human body. τ represents the time-delays caused by transport and processing in the nervous system. This concept allows the modelling of the exoskeleton in conjunction with the movement pattern of humans. The desired trajectory is defined as the reference trajectory, while the actual trajectory represents the movement of the limbs irrespective of the disturbance encountered [1].

Based on the abovementioned description, motion control may be typically referred to as position control. In contrast to motion control, force control utilizes a slightly different method to accomplish its objective. In force control, the output fed to the controller is usually that of a force sensor. The desired trajectory often from a motor encoder used to effect control in position control is replaced by the desired force usually from a force/torque sensor. This type of
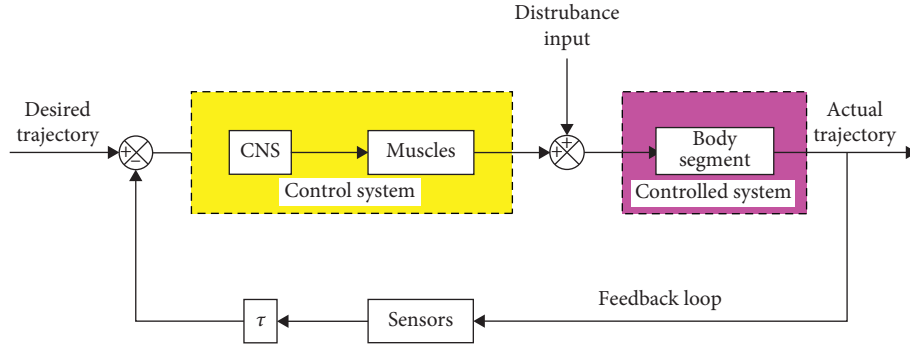
FIGURE 1: Conceptual scheme of the motion control of a musculoskeletal system [11].

control strategy is not as easy as it may seem because the objective of the control of robots could be seen as bioperational, which is ensuring that the desired trajectory is followed effectively while also ensuring the exact control of the force required is met. Force control may, therefore, be regarded as a hybrid force/position control system in that two references (desired trajectory and desired force) are required to compute the joint torques.

Exoskeletons are sometimes referred to as wearable robots which could be attributed to the fact that they are mostly worn by humans. Hence, controlling the position of the human-exoskeleton limb for a specific task involves the physical interaction of the exoskeleton with the human and also the environment. Controlling exoskeletal devices requires it to comply with the human forces based on the impedance that is generated due to their interactions. This impedance requires regulation; therefore, a typical position control method may not be able to handle the interaction tasks that exist between the robot, human, and its environment. This necessitates the introduction of a method that has the ability to modulate the impedance generated. Hence, interaction control becomes expedient.

3. Interaction Control

The successful control of the interaction forces between the robot and human or rather executing robot interaction tasks with the environment is determined by the robot's interactive behaviour. This behaviour may be referred to as the *feel*, which is a function of the dynamics of mechanical interaction [12]. Hence, interaction control may be defined as a control strategy inclined towards the realization of compliant behaviour during interaction. It does possess the ability to modulate and control the interaction dynamics of the human-robot system.

In rehabilitation and assistive robotics, direct physical interaction with humans is required. Hence, realizing its objective needs the proper execution of constrained motion tasks needed to be achieved by the human and, therefore, requires the use of compliant control systems which attempt to accommodate the interaction forces introduced by this physical interaction. Thus, this has led to the use of an interaction control design, which plays a very important role in the field of rehabilitation and assistive robotics. This is because of the cooperation needed between the human and

the exoskeleton. A description of this cooperation is given in Figure 2.

Mechanical interaction dynamics may be characterized by a phenomenon called "mechanical impedance" [12]. The objective of the interaction control design is to ensure or maintain a lower mechanical impedance level between the human and the robot so as to reduce interaction forces/torque. For example, axis misalignment which is one of the causes of interaction forces presents a high impedance at the interaction port, and this could be dangerous to the human and the exoskeleton itself. This is critical because interaction forces may affect controlled variables which may, in turn, cause the system to be unstable. Maintaining a low mechanical impedance at this interaction port necessitates the use of interaction control. Interaction forces in human-exoskeleton mainly occur in two ways. This may be the interaction torque generated at the interaction port usually at the joint level (human) with respect to the exoskeleton kinematics or at heel/foot contact (environment) with the ground.

3.1. Interaction Torque: Human-Exoskeleton. Exoskeletons are designed for two specific purposes. Firstly, it should be capable of transferring power to the limb of the wearer and, secondly, ensuring the safety of the wearer/user [13]. Their main function includes assisting able-bodied individuals to perform a rigorous task humanly impossible and also to provide assistive and rehabilitative measures to physically challenged and elderly people, so as to support their mobility and as such help them regain control of their limbs [14].

To achieve these purposes, the exoskeletal design must be tailored towards having the ability to adapt the wearers' kinematics to its kinematics. That is, the human-exoskeleton kinematic compatibility must be established in the design. This way, the wearer's safety may be guaranteed. However, the kinematic compliant exoskeletal design has remained a critical issue in the development of robotic rehabilitation exoskeleton [15]. This is due to the variation in the human and exoskeleton kinematics that results in joint misalignment. Consequently, at the interaction port between the physical device and the human limb, an uncontrollable force may appear [16]. These forces which are due to kinematic mismatch [17] are undesired and disturbing and may be very harmful to the human/wearer. Figure 3 presents an example
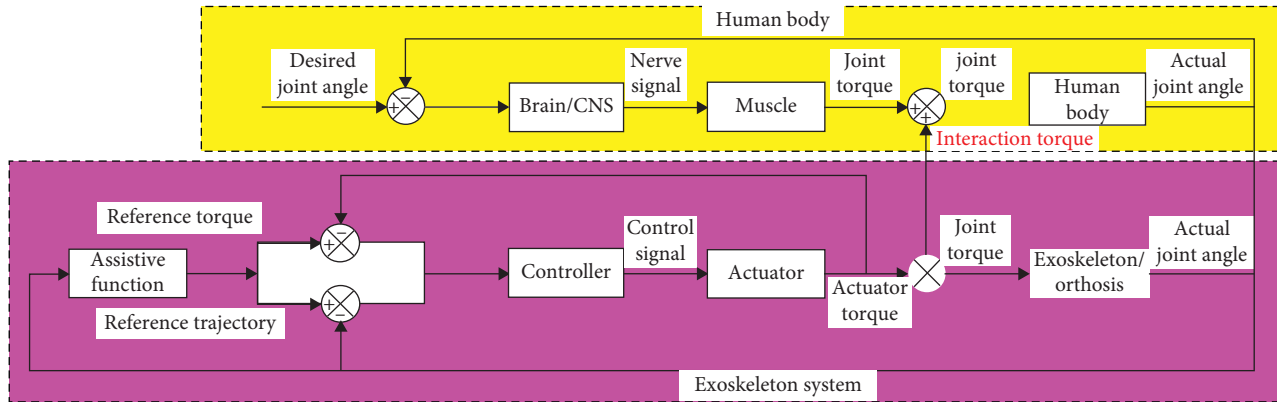
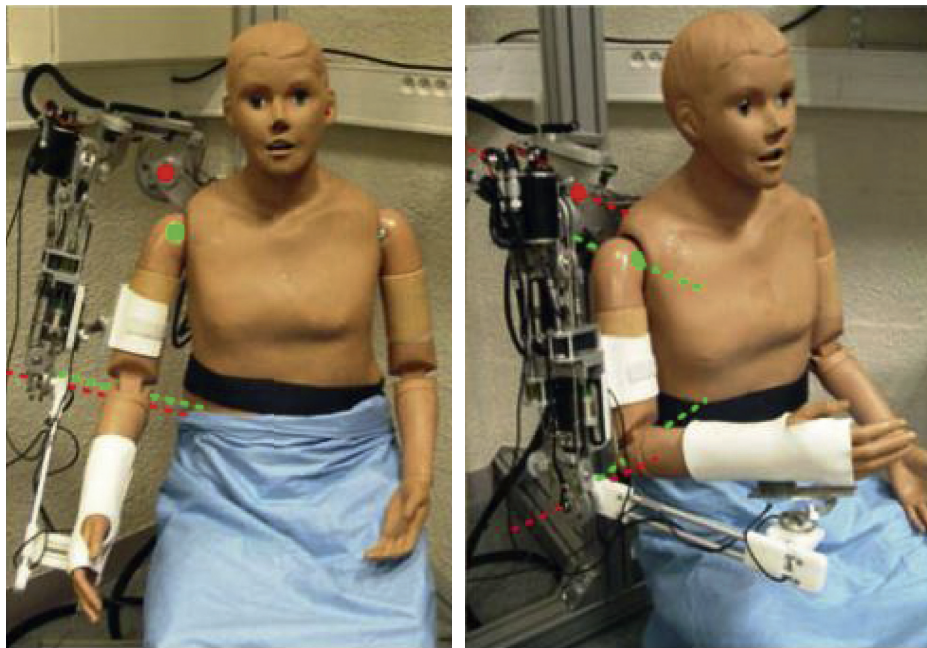FIGURE 2: Human-exoskeleton cooperation system [4].

FIGURE 3: Manikin-exoskeleton [18].

of a mismatch between an exoskeleton and a manikin, red dot (exoskeleton shoulder centre) and green dot (manikin shoulder centre).

With improper adjustment of the exoskeleton to the human, the resultant effect is the shifting of the exoskeleton along the limb of the human, while with an insufficient kinematic representation of the human, a restriction of the human joint axis over its range of motion may occur. Misalignment, in general, negates the compliance that is meant to exist between the exoskeleton and human tissue [19] and, thus, will hinder the smooth transfer of the desired driving force to the human or wearer.

Based on the abovementioned analogy, some of the research studies conducted over the past years in the area of exoskeleton have been channelled towards misalignment [20, 21]. In the literature, certain mechanical solutions have been provided to help alleviate the effects of misalignment in the robotic exoskeleton. These solutions may be categorized

into four based on the approach considered which are as follows:

(1) Self-alignment mechanism (automatic)
(2) Misalignment compensation joints (Addition of Passive Degree of Freedom (DoF))
(3) Manual Links regulation
(4) Compliant actuators

This will be elaborated upon in subsequent sections.

3.2. Interaction Torque: Human-Exoskeleton Environment.
Wearable robots do not walk in space; they walk on the ground and are, therefore, bound to interact with the environment. In addition to this, the upper limb (i.e., the hand) does perform movements that may require contact with an object. Performing these tasks requires robot compliance

with the human forces by exhibiting a form of impedance that must be modulated based on the application [22].

The robot impact (this could be either to the ground or an object) creates a pseudoimpedance at the lower limb joints, especially at the heel strike and toe-off [23]. Hence, measuring this impedance allows for the design of a control strategy that is based on the interaction forces generated in relation to the position or movement [24]. This may be referred to as interaction control. The main goal of interaction control in a human-exoskeleton system is to provide a form of regulatory measure with regard to the joint impedance of the human limb which may occur as a result of an impact or misalignment [25, 26]. A decrease or increase of the joint impedance may be beneficial to the wearer depending on the application (increased impedance may be beneficial in sports training/physical therapy, while reduced impedance may be useful in terms of assisting patients with a motor disorder). However, there remains a critical drawback in the implementation of interaction control, which is associated with the difficulties encountered in the exoskeleton design and construction in relation to the human interface [23]. Based on the human interface, it means that the safety, stability, comfort, and compliance (the user should be able to override its commands on the exoskeleton) are of critical importance and has to be met [27]. This drawback may be attenuated by making sure that the mechanical design of the exoskeleton is simpatico with the controller and meets its requirements [24]. The realization of a good mechanical design is a function of the kinematic compatibility of the human-exoskeleton.

The advancement of haptic technology in rehabilitation robotic devices has helped in the development of more effective ways in rehabilitating individuals with lost motor functions. Haptic technology, which may also be referred to as kinaesthetic communication or 3D touch, may be defined as any technology that can create an experience of touch by applying forces, vibrations, or motions to the user. Haptic devices may be categorized into end-effector devices and full-scale exoskeleton robots [28]. The latter may be divided into ungrounded and grounded exoskeleton [29]. Ungrounded robotic devices are devices that are not attached to an external frame of reference and are often used with grounded end-effector devices to target the entire arm and increase the range of movements and for rehabilitation. An example of this is the BRAVO system, as presented in [30]. Grounded exoskeletons may be referred to as robotic devices that have an external mechanism that allows them to be freestanding. These types of devices are highly effective in the rehabilitation of upper limb paralysis because they target the whole arm, which also includes the fingers. This will be expanded on in Section 5. By definition, end-effectors may be referred to as grounded haptic devices that allow interacting with the virtual or physical environment at the end of the device [31]. Beyond the advancement of neuro-rehabilitation of patients, haptic technology in exoskeleton presents a platform that allows the understanding of how humans interact with the world and their environment. It is also interesting to know that developed haptic-robotic devices are capable of providing a patient with motor deficiencies, with a particular type of force feedback that usually gives an indication of the objects that are either in a real or virtual environment.

Prior to the introduction of the measures taken to ensure adequate interaction control implementation on human-exoskeleton, it is important to understand the term "kinematic compatibility."

4. Kinematic Compatibility

Kinematic compatibility remains a critical principle in the design and manufacture of the human-exoskeleton device. This is because it helps avoid the introduction of interaction forces that may jeopardize the safety and comfort of the user of the said device [20]. In order to achieve complete kinematic compatibility in an anthropomorphic exoskeleton, a perfect alignment of the robotic and anatomical joints must have been established [32]. However, this is not an easy task to accomplish.

The notion behind kinematic compatibility lies on the sufficient kinematic representation of the exoskeleton and the ability to adjust the exoskeleton frame to fit a particular user. Failure to adhere to this principle results in axis misalignment. Axis misalignment may be better explained using the kinematic chain, as discussed in [9]. See Figure 4 for a representation of the human and exoskeleton links which form a closed kinematic chain.

In [9], the kinematic chain for the human and exoskeleton is described as closed chains. Hence, there exists a *twist* (a twist represents the velocity of a rigid body as an angular velocity around an axis and a linear velocity along this axis) as defined in screw theory (a screw is defined as a six-dimensional vector which is derived from a pair of three-dimensional vectors, e.g., forces/torques and linear/angular velocity. This phenomenon originates in the study of spatial rigid body movement. The components of the screw define the Plücker coordinates of a line in space and the magnitudes of the vector along the line and moment about this line) that gives a description of the movement final link with respect to the first. Considering both chains (human-exoskeleton) being rigid, the twists that occur in each chain are expected to be equal for any movement to occur (this means that angular velocity and the instantaneous axis of rotation (IAR) are the same for both chains). The twist may be defined based on Figure 4 as the instantaneous velocities of the body with respect to the reference body for both chains, respectively. However, this does not occur naturally. Mapping the IAR of the human to that of the exoskeleton is a rigorous task, and although movement still occurs with both IAR(s) not being equal, this tends to show the presence of hyperstatic (this means that the mechanical structure is statically indeterminate, i.e., a third twist occurs which allows movement to take place) forces. These forces result from an applied force, and as a consequence, deformation energy transferred through work becomes present in the system.

To alleviate the phenomenon discussed above, an alternative method, as presented in [18, 20, 33], is the inclusion of additional passive joints in the robot kinematic chain, thereby enabling isotacticity (possibility of complete control
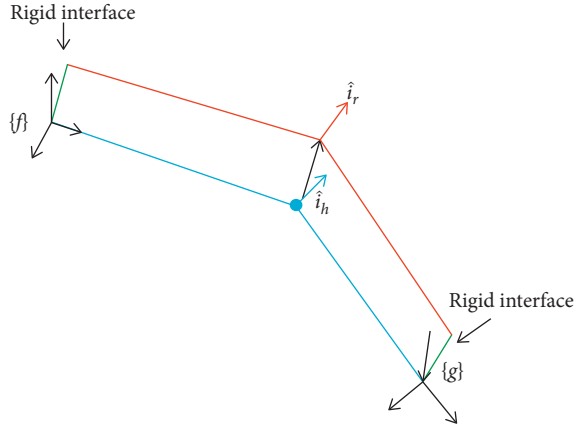
FIGURE 4: Human-exoskeleton kinematic chain representation [9]. g may be referred to as the body, while f is the reference body. $\widehat{i}_h$ is the IAR of the human, and $\widehat{i}_r$ is the IAR of the exoskeleton.

of the addressed degrees-of-freedom (DOFs) via the exchange of interaction forces that do not cause pain or discomfort to the user). Other methods will be discussed in subsequent sections and subsections.

5. Current Approaches

The actual generation of hyperstatic forces may be attributed to certain developments that may occur due to the way human articulation takes place or the use of flexible elements for the human-exoskeleton connection. These forces are known to be uncontrollable, and to ensure a successful rehabilitation or assistive protocol, the exoskeleton must smoothly interact with the wearer. Generally, the major aspects considered to ensure a smooth human-exoskeleton interaction may be divided into two [34], which are as follows:

(1) The mechanical and kinematic design structure

(2) The control strategy implementation

The actuation and motor control implementation in human-exoskeletons may be introduced in two forms, that is, either by supporting or imposing natural movement of the patients during training. With the introduction of impedance control measures and, of course, hybrid control measures, the safety of the user with regards to control of wearable robots may be guaranteed to a larger extent. A review of these control strategies with specifics on lower extremity exoskeleton may be found in [35].

However, a significant area that has been identified to be very important in the human-exoskeleton design is the kinematic setting of an exoskeleton. This kinematic setting, if not properly matched with those of the human, tends to generate undesired interaction forces during motion with or without actuation. The forces generated are usually due to misalignment between the robotic device and the human limbs which, often, may not be compensated by the device actuators.

It is a known fact that human actuation does not act on the sagittal plane of the body only; hence, neglecting the movement along the transverse and coronal plane may be a

deliberate attempt to allow misalignment. In addition, a small translation effect may result from the bones sliding [36]. During motion, an undesired or uncontrollable interaction force may develop due to this kinematic mismatch. It should be noted that these interaction forces may not be compensated by the actuator of the exoskeletal device and may cause a lot of discomfort on the user while jeopardizing his/her safety. Wearable robots may be classified with regard to the limbs they are meant to fit into. Hence, the wearable upper limbs and lower limbs exoskeletal device will be grouped accordingly.

In [37, 38], it was documented that the kinematic mismatch between the Lokomat leg orthosis and the patients is responsible for the injuries and discomfort they encountered while using the device. This kinematic variation may also somewhat change or modify the natural muscle activation pattern [39], which could negate the successful recovery of the user motor function. Due to the anomalies stated above, a number of second-generation rehabilitation exoskeletons are currently being developed in many laboratories [20].

The first step by researchers working on the Lokomat robotic orthosis was to analyze the difference in kinematic trajectories between walking on a treadmill and walking on the Lokomat. This study discovered that the amount of misalignment at the hip is larger than that at the knee when compared to the joint position in space [40]. Although this may differ in other orthotic/exoskeletal devices, it was established that joint centres are never perfectly aligned. The key factors to be considered when designing a rehabilitation robot are as follows:

(a) The device should be capable of training the total workspace of the human limb.

(b) The input applied to trigger the exoskeleton joint actuators should be in sequence (i.e., activated joint by joint). This is to ensure the exact type of movements in humans. Hence, the movement input should be rhythmic, periodic, and physiological. This is because the quality of induced movement is critical to the restoration of the motor functions in humans [41].

(c) It must be able to guarantee the safety of the wearer without causing any discomfort.

The effect of kinematic mismatch leads to a counterproductive result during physical therapy and may also lead to the wearer's injury. A review of methods which include actuation and motor control and kinematic design of human-exoskeleton devices, in particular, with a view to alleviating the effect of misalignment is presented below. Firstly, methods used in upper limb wearable devices will be presented and secondly, those of the lower limb wearable devices will be discussed accordingly.

5.1. Upper Limb Wearable Device. The arm has a lot of functions to help humans fulfil the activities of daily living. This involves performing a task that requires a high degree of mobility [42]. The resultant effect is a complex kinematic

structure which may need to be replicated by the wearable device. Based on this, researchers have, in recent times, begun to look into this possibility by proposing a different kinematic design. A number of researchers have adopted the sequence of rotation of the human arm described in Figure 5. However, a slight deviation from this conventional principle has been experimented and documented in the literature.

In a bid to incorporate the key factors stated above, a novel kinematic design paradigm aimed at improving the ergonomics in human-exoskeleton was developed and analyzed in [20]. The focus was on developing a 3-dimensional, 9-DoF human arm rehabilitation robotic device for a subject of 1.80 m. It should be noted that the aim was not to literally align the joint centres but to allow for sufficient workspace and ensure proper device functionality even when misalignment occurs. Experiments were conducted using the prototype built to conform to the mechanical design and tested with various subjects. The exoskeleton was said to be comfortable enough during motion. However, the weight of the exoskeleton was in question. To fulfil these key factors, Schiele and van der Helm [20] highlighted some key points that an exoskeletal device design must have:

(i) The kinematic structure of the exoskeleton might not necessarily be the exact replica of the human kinematics but rather provide a parallel facultative system over the joints of the human. This is to avoid the need for alignment and also to reduce the complexity of the robotic device.

(ii) It should be able to carry out movement in the Cartesian space more the human limb which it is in contact with.

(iii) The interacting interface of the human-exoskeleton must be rigid rather than flexible to avoid shifting.

(iv) A maximum of 6 DoF(s) should be allowed between consecutive attachments.

(v) Joints must incorporate mechanical end-stops to ensure the precise range of motion is adhered to.

The general objective of utilising multiple passive mechanisms for the connection of two kinematic chains which are similar in nature was studied in [16]. A constructive (graphical) method was proposed to enable the determination of all the possible segmentations of "*freed*" DoF(s) for several fixation mechanisms. This was to provide proper analysis and generality of the experimental solutions that have been proposed over time.

The actual goal is to design an exoskeleton fixation mechanism that could exhibit controllable forces' capabilities and be motionless when the wearer is in a still position. This is performed to fulfil global isostaticity (possibility of complete control of the addressed degrees-of-freedom (DOFs) via the exchange of interaction forces that do not cause pain or discomfort to the user) while avoiding hyperstaticity. Practical application of this method is carried out using an exoskeleton mechanism called "*ABLE*," which was reported in [44]. This is a DoF arm exoskeleton. The findings are seen to allow two fixations under normal
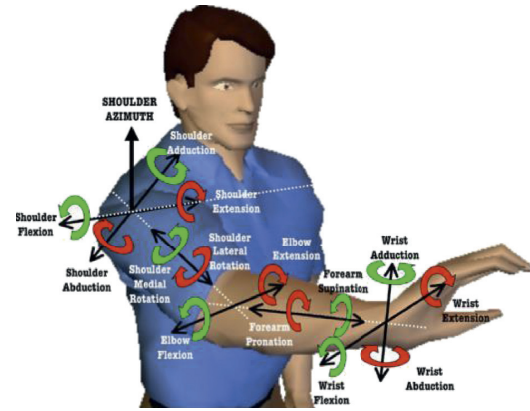

FIGURE 5: Conventional primary 7 DoF(s) of the human-exoskeleton arm [43].

operable conditions and also satisfy a necessary condition of singularity avoidance as stated in [45]. A similar approach was reported in [18].

Using the same principle (passive compensation joint) as in [16], Schiele proposed an analytical model that is suitable for the prediction and interpretation of constraint forces between the human and exoskeleton during interaction in [34]. However, for simplicity, the analysis is based on an explicit mechanical arm model with a single DoF at the elbow joint. The force transmission analysis is only on a plane, and overall analysis is dependant on the derived equations realized from the mechanical model. Aligning the IAR of both kinematic chains (human and exoskeleton) is critical for this design methodology while considering a rigid fixation at the upper arm and a soft (by soft fixation, it means that the element used has to present both viscous and elastic properties when experiencing deformation) fixation at the forearm. Model verification is based on experimental data, and since it is only one DoF, it lacks generality and somewhat does not consider off-plane forces. Experimental data is based on EXARM exoskeleton fitted to a subject of height 1.71 m and weight 63.0 kg.

The addition of supplementary passive DoF(s) may be introduced in two different ways. As previously explained, it may be introduced to connect the wearer limbs to the exoskeleton or rather may be introduced in the robotic device between two active joints [19]. Exploiting the capabilities of the latter, Stienen et al. [21], argued that increased number of passive links may require no joint alignment as in [20] but does not necessarily make it suitable for a large number of rehabilitation exercises and hence the proposition for a better option which involves the decoupling of the joint rotations from the joint translations. However, this is realized with a complex mechanical network and movement inertia. The use of a movable linkage system which allows for the mounting of an exoskeletal device on it provides the decoupling capabilities. However, this forms part of the mechanical complexity. In addition, this linkage system helps also to create self-aligning axes. Linkage systems have an influence on the wearer in aspects concerning the impedance force felt. This is as a result of the friction and inertia

in the linkage-exoskeleton system. Hence, a comparison of three types of linkage systems is made, and this includes the LG (Linear Guidance) mechanism, PH (Parallel Hexapod), and DP (Double Parallelpiped). The proposed DP linkage system is seen to be best suitable for the shoulder linkage system based on its passive mechanism functionality, slender design, and low space requirements. Implementation of the 3D DP linkage-exoskeleton system is carried on the Dampace and Limpace exoskeletons.

Designing a robotic device that may possess the ability of a high-performance system requires that certain factors are met [46]:

   (i) The weight and stiffness

   (ii) Operational workspace

   (iii) Desired joint torques

   (iv) Link design

   (v) Placement of the motor

   (vi) Selection of the cable

Using a collected database of different task categories (ADL: Activities of Daily Living) of the arm [47], the kinematic and dynamic analysis of a 7-DoF upper limb powered exoskeleton design is described in [46]. To ensure that the factors listed above are taken into consideration, three types of joint configurations were considered in conjunction with other requirements. The three parallel, noncollinear DoF(s) are found to be better than the proximally placed and the circumferentially placed single DoF configuration, although not at the expense of the weight. One of the requirements that are very crucial to the development of a 3-DoF spherical joint is the placement of singularities. This was achieved by placing the singularity at the edge of the workspace (i.e., hard to reach). The exoskeleton performance is measured by using a frequency that is higher than the command signal frequency of the human for its operation. This has been reported to be 10 Hz which is higher than the stated 2 Hz and 5 Hz in [48, 49].

ARMin II [50], another robot device developed for arm therapy, is designed to enable the user to perform different task categories such as ADL. Its workspace is chosen such that operation in the most functional range of the arm workspace is made possible. To overcome the challenge of robot adaptablity to different users, this robotic device is equipped with five adjustable segments, thanks to the four additional passive DoF implemented. It is a semiexoskeletal device (end-effector-based mechanism) (the contact to a user's limb is only at its utmost distal part [51]) with 7 DoF(s), of which 2 of them are coupled (translation and rotation) through the mechanical linkage. The device is meant to comply with the patient's natural movements of the arm; hence, there is an added DoF to allow for a vertical displacement of the exoskeleton arm elevation (centre of rotation) axis. The ARMin II is built to overcome the challenges of the shoulder fixed axis of rotation associated with ARMin. Originally, ARMin, the first prototype, is designed with 4 DoF(s) (2 active and 2 passive) [52, 53]. A pilot study is conducted to validate the mechanical design

functionality on patients. This test was able to establish the fact that though it could easily accommodate a range of different users in conjunction with their range of motion (RoM), the vertical shoulder rotation is not taken care of. Hence, a new version with an added DoF to compensate for the shoulder actuation in the vertical plane is proposed in [54]. The ARMin III is designed to provide a more ergonomic shoulder actuation, i.e., having a kinematic structure that closely imitates that of humans [55]. This was achieved using a simplified human shoulder model. It also incorporates added modules that provide actuation for the lower arm and wrist. Clinical tests on patients are currently being conducted.

For the design of an exoskeleton free from misalignment irrespective of the exoskeleton application, a method which is based on kinetostatic analysis of a coupled human-exoskeleton is proposed [19]. Validation of this concept is carried out via the design of a robotic link with 2 revolute joints to mimic the finger metacarpophalangeal. Just as in [20], analysis is based on the serial self-alignment mechanism (SAM) (this refers to exoskeletons that provide actuation for the human DoF(s), regardless of the human chain geometry). An elaborate analysis of the treatment of the misalignment problem is presented. However, it is only analytical and may need a physical design for its validation.

The design of an exoskeleton arm intended for shoulder rehabilitation is presented in [56]. The focus was on the development of a lightweight exoskeletal device with scapula motion. This presented the task of designing a distinct mechanical system in conjunction with the actuation system. Emphasis is placed on the utilization of a serially connected pin joint to replace the conventional ball and socket joint at the shoulder. This formulates a 3-DoF kinematic structure which differs from the ball and socket joint kinematics, but at the expense of a possible *gimbal lock*. To reduce the effect of singularities, the point at which singularity occurs on the workspace is moved to a likely unreachable point. Prototypes (MGA exoskeleton) were built to operate in the virtual reality and physical therapy mode. The control architecture which enables independent control of the scapula joint is also provided. The exoskeleton is still in the process of electronic integration and testing.

The *L-Exos* [57] is another wearable robotic exoskeleton right arm equipped with 5 DoF(s) (4 active and 1 passive). The kinematic design is based on 5 rotational joints connected serially, with three rotational axes arranged orthogonal to each other to represent the ball and socket joint of the shoulder. This is performed to allow a high degree of motion movement (about 70%) of the human shoulder joint when the exoskeleton is worn. The DoF at the wrist is passive and allows for supination and pronation movement. The actuation and control system for *L-Exos* and subsequent clinical trials in the three types of robotic therapy schemes (the reaching task, free-motion task, and the task of object manipulation) in virtual reality mode is reported in [58]. Results on patients who undertook these trials were found to provide evidence of the shoulder and back compensation strategies.

Pneu-WREX (Pneumatic-Wilmington Robotic Exoskeleton) is designed to compensate for the limitations in *T-WREX* (Training WREX [59]). This is because the *T-WREX* gravity balance has limited space to accommodate the full movement of the arm ROM, and it is incapable of applying a dynamic form (but fixed form) of assistive/resistive torque to the arm [60]. To achieve this compensation, the *Pneu-WREX* is equipped with pneumatic actuators, nonlinear force control, and passive counterbalancing. Its mechanical design incorporates the shoulder and elbows joint with a total of about 4 DoF(s). The elbow actuation is by a four-bar linkage system. The high force-to-weight ratio of pneumatic actuators and the possibility of active back drivability and position control is the main improvement of the *Pneu-WREX*. Its acceptability is strongly backed by a survey of 29 therapists. Another upper arm rehabilitation robot which uses pneumatic muscle actuators for its actuation mechanism is the RUPERT IV [61]. It is designed with a 5-DoF envisaged to assist the shoulder elevation, humeral external rotation, elbow extension/flexion, forearm supination, and wrist extension. RoM is limited using physical stops. PMA is compliant in nature and, therefore, provides a level of safety to the wearer.

The development of a novel exoskeleton for upper limb rehabilitation (*RehabExos*) is described in [62]. The *RehabExos* has the same kinematics and DoF [57] as the *L-EXOS* and it is built with the same objective to accomplish as the *L-EXOS* [58]. However, it has some modification that includes two links serially connected by a lockable passive prismatic pair at the forearm. This allows it to be fitted by users with different limb sizes to some degree. Just like the L-EXOS, the GH joint is not taken care of but has a different actuation system that ensures the exoskeleton's capability of controlling contact forces at points. Preliminary tests are used to validate the proposed design of the *RehabExos* and seen to be robust and reliable using a manikin.

In 2012, a robotic arm exoskeleton *ASSISTON-SE* [63] which extended the type of therapeutic exercise that may be given to a patient was developed. The *ASSISTON-SE* kinematic design is based on the serial connection of a serial RP kinematic chain, a parallel 3RRP mechanism, and a serial RR kinematic chain. The 3RRP mechanism is a generic 3-DoF self-aligning joint system designed to compensate for the GH mobilization and scapular stabilization exercise required originally by humans at the shoulder joint during rehabilitation. These kinematic and design considerations were highlighted in [64]. The exoskeletal device is symmetrical and may be applied to both left and right upper limbs. The first prototype of *ASSISTON-SE* is a simplified 4-DoF with an actuation system that uses direct-drive actuators which are back drivable. Implementation and testing on healthy patients are still on the way.

MEDARM (Motorized Exoskeleton Device for Advanced Rehabilitation of Motor function) [65] is an arm rehabilitation exoskeletons which prides itself as being the only robotic exoskeletal device with the ability to independently control all its 5 DoF(s) at the shoulder complex. Of which, this is its primary goal. It is designed with a total of 6 DoF(s). To complete the number is 1 DoF at the elbow. The shoulder complex comprises of the GH (glenohumeral) joint and the sternoclavicular joint. In the kinematic design, the GH joint is modelled as the spherical joint with 3 revolute joints having their axes intersecting at a single point, while the shoulder girdle mechanism is a 2-DoF to support the sternoclavicular joint movement (elevation/depression and protraction/retraction). These joint axes are arranged to avoid singularities during theor manoeuvre. Actuation of the joints is realized via electric motors, cables, and belt transmissions for efficient power-to-weight ratio. This, in turn, minimizes the inertia. With the aid of a robotics toolbox [66], a dynamic model of the exoskeleton and human limb was simulated. Results from this simulation were used in determining various parameters used in the prototype, as at the time of publication, clinical tests were not performed.

Wire-based robots are known to have a wider workspace than classical parallel robots [67]. Hence, they are robots made from a conscious effort to substitute actuators with wires to have an increased workspace. In 2003, a 3-DoF wire-based robot for rehabilitation called NeRebot (Neuro-REhabilitation roBOT) was proposed in [68]. Although there were signs of positive results as reported in [69], there were some limitations also recorded. These limitations are highlighted as follows:

(i) Heavy and cumbersome

(ii) Limited working space for the therapist

(iii) Operator-dependent

(iv) Hardware/software configuration does not support the real-time control system

Due to these limitations, the *MariBot* (MARIsa roBOT) [70] was designed. *MariBot*, a 5-DoF rehabilitation wire-based robot, is complex but has a much-widened working space (i.e., increases the possible number of movement) due to the added 2 DoF(s), and the eventual design does not require manual setup before rehabilitation exercises. Satisfactory clinical tests were reported in [71].

Most recently, an upper-body rehabilitation exoskeleton which is known as *Harmony* was presented in [72]. This exoskeletal device is built with 6 DoF(s) (1 DoF at the wrist and 5 DoF(s) at the shoulder). The kinematics is designed and made to conform to an anatomical shoulder joint. The anatomical shoulder joint, therefore, consists of a shoulder girdle mechanism and a ball and socket joint. In most cases, the shoulder joint mechanism is represented by a ball and socket joint. However, its usage may cause singularity to occur in the workspace. Hence, 3 revolute joints arranged perpendicular to each other at an acute oblique angle with axes intersecting at a single point to represent the ball and socket joint were used. The issue of misalignment was taken care of by making the shoulder girdle mechanism of the exoskeleton comply with that of the GH joint in humans. These are in the areas of the elevation of the shoulder mechanism: the distance between both sides of the revolute joint that allows for the axis of elevation and depression and the gap between the back and the shoulder mechanism. This system is powered by series elastic actuators previously

designed by Edsinger-Gonzales and Weber [73] and modified using a flat brushless DC motor and a harmonic drive reduction gear system, thus permitting the use of an impedance control strategy. The exoskeleton is seen to offer good kinematic compatibility with wide RoM and will provide significant improvement in the rehabilitation of the upper body.

The *exo-spine* [74] is another strategy to ensure the humans are able to move all the degrees of freedom his/her spine and shoulder girdle. Focus is on a full-body exoskeleton which is in relation to the rigid back parts carrying the power supply. These rigid parts may hinder the wearers' movement of the spine and shoulder. The *exo-spine* is proposed to be incorporated into a simplified full-body exoskeleton since its kinematic design and actuation system considerations are based on improving the upper limb movement pattern of the HAL Robot Suit *HAL-5* described in [75]. *Exo-spine* is equipped with a total of 5 vertebra and links connected in a chain-like form. This enables it to bend forward with a single DoF in spite of its numerous parts and its interconnections provide 3 DoF(s).

Upper limb robotic device may be classified into two types, namely,

(1) Exoskeletons robotic device

(2) End-effector-based robotic device

Figure 6 presents a pictorial description of the types of upper limb robotic devices.

There have been extensive development and evaluation of end-effector-based robotic devices. End-effector robots are characterized by their attachment to the subject's upper limb at its most distal part, i.e., the hand, and positioned in an oriented space. Examples of these end-effector robotic trainer devices are the *MIT-Manus* [76], *MIME* (Mirror Image Motion Enabler) [77], *ARM Guide* [78–80], *NeReBot* [68], *ACT3D* (Arm Coordination Training 3-D), and *REHABROB* [81]. Although extensive clinical research has been conducted on these rehabilitation devices for the evaluation of their effectiveness, it does pose a certain restriction to the application of therapeutic movements at the shoulder and elbow [20]. These effects were reported in [82, 83] to be a consequence of the lack of hand and wrist motor improvement. This is as a result of the kinematic redundancies in human limbs which negates the introduction of accurate joint trajectories for systems of this nature. However, it does have certain advantages such as a reduction of the kinematic structure and the problem of misalignment is not applicable to these devices.

5.2. Lower Limb Wearable Device. Wearable devices attached to the lower limbs often known as lower limb exoskeletons may be full-bodied, multijoint (i.e., hip-knee-ankle), or single joint rehabilitation devices. In any of these cases, the type of application they are designed for may be categorized into two:

(i) Human power augmentation

(ii) Human impairment movement [84]

Although the studies which correspond to the discovery of the extent of misalignment during training exercises for lower limbs exoskeleton have been conducted using the Lokomat [40], until recently, most work in this domain has been focused more on the upper limbs. Less research effort has been channelled towards lower limbs. Research studies on lower limbs are often centred on rehabilitative [85] and assistive [86] approaches in relation to the design of the exoskeleton control mechanism. Based on the periodic behaviour of the human walking movement, lower limbs present an added difficulty that is related to the speed of the human gait. This is as a result of the high angular velocities and acceleration produced at increased human gait speed which may affect the exoskeleton inertia interaction with joint misalignment [87].

Misalignment basically occurs at the joint axes, and the first step towards achieving human-exoskeleton compatibility is to allow for flexibility in the kinematic model of the exoskeleton while maintaining flexibility between the human and exoskeleton chains.

In the design of lower limbs kinematics structures, 6 DoF(s) are generally considered. This includes a 3 DoF(s) at the hip joint (circumduction, adduction/abduction, and flexion/extension), 1 DoF at the knee joint (dorsiflexion/plantarflexion), and 2 DoF(s) at the ankle joint (flexion/extension) [45]. See Figure 7 for the orientation of the human leg 6 DoF(s).

However, this number of DoF is not a direct replica of that of the natural human leg. Simplification of the model is needed, in order to reduce the complexity of designing a wearable exoskeleton. One feature that must not be compromised is that it must be kinematically compatible with that of the human leg. This will allow for a smooth transfer of the desired force to the subject during rehabilitation.

In order to ensure that the axis misalignment is taken into consideration, Jianfeng at al. [15] proposed a seemingly effective method for the structural design of the lower limb extremity. Using the LOPES as a case study, an analogy describing overconstrained and even-constrained effect of the closed human-exoskeleton kinematic chain is reported. The overconstrained effect is highlighted to be the main cause of misalignment in human-exoskeleton interaction during gait training. In its analysis, the adoption of a 2-DoF at the hip and 1-DoF at the knee to represent the exoskeleton structure, while considering prismatic joints for the hip and a four linkage bar for the knee, is stated. Necessary conditions which include the closed human-exoskeleton chain being a 3-DoF and having a single active knee joint and two active hip joints installed in a closed-loop are stated. The rationale behind this lies in the realization of the swing movements in the human lower extremity being made possible by the hip and knee joints.

Hence, in the kinematic modelling of the hip joint, its structure is designed to entail 3 revolute joints that will allow for 3 DoF(s) (flexion/extension, adduction/abduction, and internal/external rotation). In the same way, the knee joint is not modelled as a single revolute joint but rather as a four-bar knee joint. This is to ensure compliance with the small-angle adduction/abduction and internal/external rotation
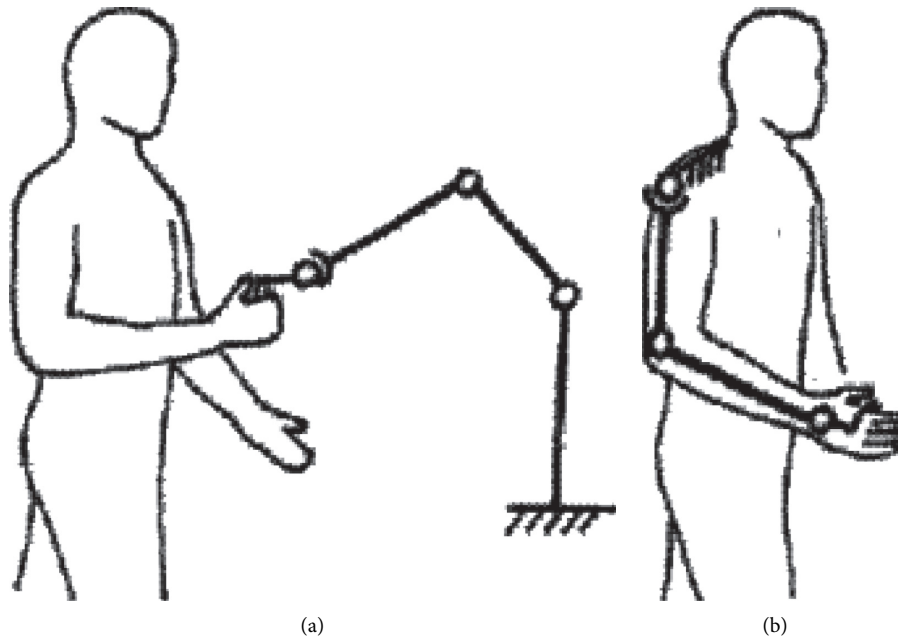
FIGURE 6: (a) End-effector-based robotic device and (b) exoskeleton robotic device [58].
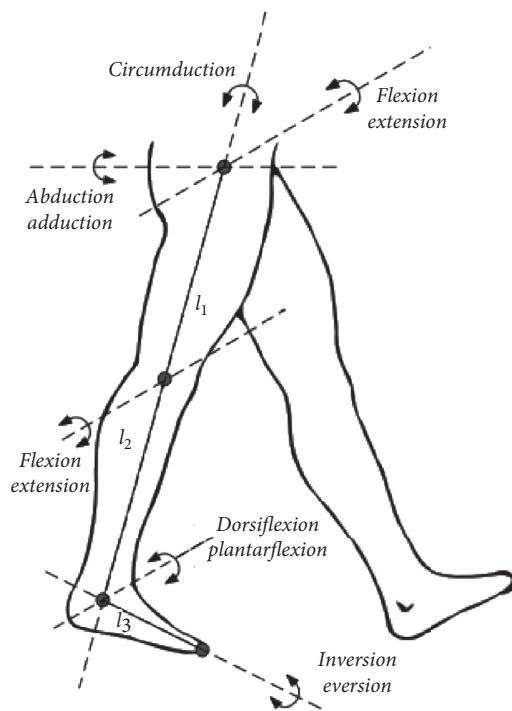

FIGURE 7: Human leg DoF orientation [45].

movement of the knee during human daily walking activities. For an even-constrained system, a total of 8 passive DOF(s) are considered to be added to the closed chain of the human-exoskeleton. This is grouped into 3 categories considering the basic and optimal exoskeleton structures. The optimal structure is achieved by making the subchains connecting the exoskeleton structure to the human lower limb shorter and simple. This is to provide a system that will

be more comfortable to the user and, hence, a replacement of several revolute joints with a universal, spherical and cylindrical joints to ensure a close workspace with that of humans. Bearing this in mind, further works may be considered and experimentation may be carried out to validate any deviation from this norm. This is because the utilization of interaction control in human-exoskeleton control may require mimicking the exact degrees of freedom at each joint. For reference purposes, reviews of several typical lower extremity exoskeletons are presented in [4].

At this juncture, it is important to note that a rigid exoskeleton presents the most challenge with regards to its alignment to the wearer. Rigid exoskeletons are usually fitted with motor-driven actuators which depend on sensory feedback and control for proper compliance between the device and which may also assist in handling unstructured or delicate objects [88].

However, soft robots have emerged to be a readily available alternative approach due to their compliance matching and back-drivability with compact and lightweight mechanical structures [89]. This has lead to their promising application in systems or devices that require interaction with soft materials, organisms, and/or replication of biological functionalities [90]; which is due to their elastic ability. Their application to robotic systems centres on the robot actuator design. This may have been demonstrated by the use of pneumatic artificial muscle (PAM) which may be applied to biocompatible devices as in [91], humanoid robots as in [92, 93], and complaint manipulators as in [94]. However, PAMs does have a drawback that is attributed to the limitation of its compliance with the overall robotic system by the rigid components involved. The use of soft bending actuators [95–97] was to address this limitation based on their ability to generate inherent bending motion without requiring any rigid components.

Another approach that has been recently conceived by Wyss Institute researchers is the use of soft clothing-like materials for the development of "exosuit" for lower extremity mobility. This exosuit is composed primarily of specially designed fabrics, which are presumed to be significantly lighter than an exoskeleton specifically because it has no rigid component. Due to its features, the exosuit is capable of providing minimal restrictions to the wearer's motions and, thus, avoids problems related to joint misalignment [98], thus minimizing the generation of interaction forces between the device and the wearer. Although the soft exosuits are developed specifically to enhance the functionality of lower extremities, measures are being put in place to develop prototypes that improve mobility of the upper extremities.

6. Conclusions

The kinematic structure is the first point of call when deriving the dynamic model for a human articulation system. Researchers have used various ways to model the human joint system depending on the understanding of the joint movement and most often reduce the complexity involved when considering all the possible DoF at a joint. Modelling the joints' DoF of the exoskeleton to represent the human articulation may be simplified to involve only the DoF of interest. In most cases, such as the elbow and knee, joints are modelled as a hinge joint, while the hip and shoulder joint may be modelled as a ball and socket joint. However, this does not represent the actual DoF of these joints. Hence, a purposeful consideration to minimize misalignment starts from the very moment when the kinematic design is being conceived. This may be seen as various researchers tend to design a kinematic structure that may be close to that of humans or rather increase the workspace of the exoskeleton kinematics to accommodate all possible degrees of freedom that a human joint exhibits. It should be noted that it is almost impossible for an exoskeleton joint to have the same DoF as that of a human and a close representation may be the farthest the researchers can go.

Having a close kinematic design to that of humans is just the first step to an ergonomic design. Certain other challenges still need to be overcome, such as singularities, macromisalignment, micromisalignment, and safety. Avoiding singularities has been approached by moving the point at which it occurs to an unreachable point in the workspace. This may only be possible using the Denavit–Hartenberg (DH) convention to design the kinematics. Macromisalignment may occur if the exoskeleton is not flexible enough to accommodate different sizes of limbs and if the interaction port is not rigid. However, in the literature, it is being said that the transmission of loads to human limbs may be applied through soft tissues. The problem relating to load application to human centres on the mode and load intensity and areas on the human body [45]. In addition, load application may require advanced control strategies [99]. It is necessary to make the exoskeletal device complaint, i.e., provide adequate and prompt response to the patients' effort. Micromisalignment may also occur because

it is not exactly possible to align the axis of rotation of the exoskeleton to that of the human. Most especially, the precautions taken to ensure safety have to be at the hardware and software level. Key factors considered are

- (i) The forces transmitted to the patient
- (ii) Excessive excursions of the upper limb segments
- (iii) Emergency stops protocols and what occurs after it has been issued
- (iv) Error detection of the sensors or operational software

Conflicts of Interest

The authors declare that they have no conflicts of interest.

Acknowledgments

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