

Chapter 2

Literature Review

A comprehensive literature review on rehabilitation robots is carried out to identify the key issues. The main design requirements and development complications are identified and the various approaches used in past robots are reviewed. It begins with a survey of existing human rehabilitation devices designed for use in human assistance and treatment. An overview of the kinematic and computational biomechanical models of the human limb is also provided. This is followed by a review of the state of the art of interaction control strategies, with primary focus on its application in rehabilitation robots. Finally, the reviewed materials are assimilated in a discussion that highlights issues in rehabilitation robots that require further development, and are hence the subject of investigation for this research.

2.1 Medical Needs and Existing Rehabilitation Devices

2.1.1 *Upper Limb Rehabilitation Robots*

Early research on rehabilitation robots for the human upper limb was based on end-effector robots. End-effector rehabilitation robots hold the patient's hand or forearm at one point and generate interaction forces at this sole interface as shown in Fig. 2.1a. The kinematic structure of these end-effector robots are based on industrial robots and the kinematics of the human limb are not considered in their design. This type of robot is simpler, easier to fabricate and can be used for patients with different arm lengths. However, determining the posture of the upper limb can be difficult with only one interface, especially if the interface is at the patient's hand. This is because the upper arm and forearm are unconstrained and are free to move about the pivots at the shoulder and hand. Controlling the torque at specific upper limb joints is also not possible, resulting in uncontrolled load transfer between upper limb joints. As a consequence, generating isolated movement at a single upper limb joint is difficult since movement of the end effector can cause a combination of movements at the wrist, elbow and shoulder joints. In addition, the range of motion that end-effector robots can generate for the upper limb tends to be limited therefore only a limited set of rehabilitation movements can be produced by

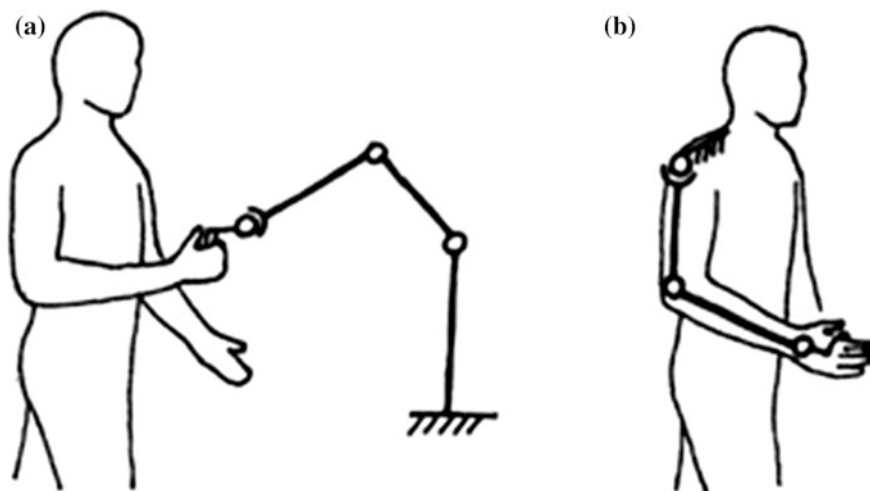


Fig. 2.1 Kinematics of **a** an end-effector robot and **b** an exoskeleton robot [10]

these robots. Examples of end-effector rehabilitation robots include the MIT-MANUS [1, 2], the MIME [3] and the GENTLE/s [4]. Extensive clinical testing has been done on these devices to evaluate their effectiveness as rehabilitative devices [5–9]. The results indicate reduced motor impairment of the upper limb for patients who received robotic therapy. The positive results justify research on the more sophisticated exoskeleton robots as rehabilitation devices.

Exoskeletons have a structure that resembles the human upper limb, having robot joint axes that match the upper limb joint axes as shown in Fig. 2.1b. Exoskeletons are designed to operate alongside the human upper limb, and therefore can be attached to the upper limb at multiple locations. Although this can make it more difficult for the robot to adapt to different arm lengths, multiple interfaces allow the exoskeleton to fully determine the upper limb posture and apply controlled torques to each upper limb joint independently. It is possible for exoskeletons to target specific muscles for training by generating a calculated combination of torques at certain joints. In addition, a larger range of motion is possible compared to end-effector robots which enable a wider variety of movements to be used in rehabilitation exercises.

The majority of past upper limb exoskeletons focus on movements for the 3-DOF spherical rotation of the shoulder joint and 1-DOF of the elbow. A lower number of exoskeletons have included movements for the 3-DOF wrist joint and even fewer have included movements for the 2-DOF translations of the shoulder joint. One exoskeleton studied during this literature review has also included 1-DOF for grasping movement of the hand. From this, it can be seen that the upper limb DOF that has larger influence on the hand's position have been the focus of upper limb exoskeletons. Rehabilitation of these movements is of the highest priority since they are the most important in controlling the position of the hand.

The ARMin III (Fig. 2.2a) [11], MGA [12] (Fig. 2.2b) and IntelliArm [13] exoskeletons have implemented an actuated DOF for shoulder elevation & depression. The MEDARM has included actuation for both shoulder elevation and depression, and retraction and protraction, allowing 5-DOF of actuated movement at the shoulder complex [14]. Other groups have opted to use passive DOF for these translation movements [13, 15, 16]. Passive DOF allows the joint to move freely but eliminates the ability to generate actuation forces at the joint.

There are a number of commercially available rehabilitation devices for the upper limb. One of the more sophisticated rehabilitation devices available are the Armeo products (Hocoma AG, Switzerland) [17]. These include the 7-DOF ArmeoPower active exoskeleton, ArmeoSpring passive exoskeleton and ArmeoBoom sling suspension system. The ArmeoPower is based on the ARMin III exoskeleton [11]. Examples of other commercial devices include the mPower arm brace (Myomo, Inc., Cambridge, MA) [18], a 1-DOF portable arm brace which uses electromyography (EMG) signals measured from the biceps and triceps muscles to generate assistive torques for elbow flexion and extension, and the Hand Mentor (Kinetic Muscles, Inc., Tempe AZ) [19], a 1-DOF wearable device for the rehabilitation of the wrist and fingers which provides force, position and EMG feedback and is actuated by an air muscle. The Robot Suit HAL-5 (Cyberdyne Inc., Japan) is a full-body exoskeleton for the disabled which uses measured EMG signals from the user to generate assistive torques. Examples of commercial end-effector rehabilitation robots include the InMotion robots (Interactive Motion Technologies, Inc., Boston, MA) [20], Biodex System 4 dynamometer (Biodex Medical Systems, Inc., New York) [21], HUMAC NORM (SCMi, Stoughton, MA) [22] and CON-TRES MJ (CMV AG, Switzerland) [23].

2.1.2 Ankle Rehabilitation Robots

Robotic devices had been developed for the rehabilitation of the human ankle. Although the main rehabilitation problem considered in this research is that of sprained ankle rehabilitation, devices used for gait rehabilitation for neurological disorders are also considered in this discussion for completeness. Ankle rehabilitation devices can be classified into two categories in terms of the mobility of the device during operation. These are wearable robots and robotic platforms with stationary bases. Wearable ankle robots typically take the form of a robotic orthosis or exoskeleton (Fig. 2.3) and are used to correct the user's gait pattern. Robotic platforms (Fig. 2.4) on the other hand, manipulate the user's foot using their end-effectors and are generally developed to facilitate the treatment of ankle sprains.

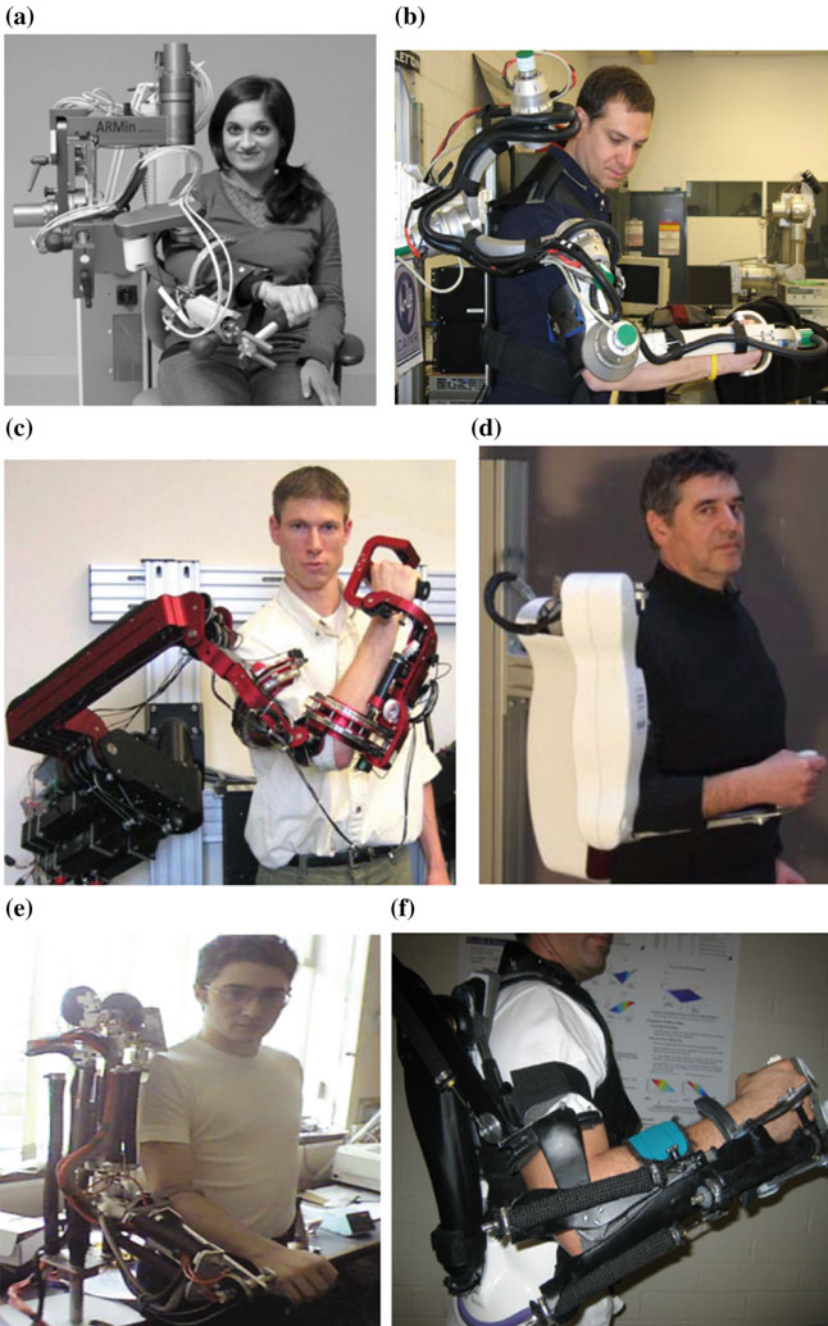


Fig. 2.2 Past upper limb exoskeletons. **a** ARMin III [11]. **b** MGA [12]. **c** CADEN-7 [24]. **d** ABLE [25]. **e** SRE [26, 27]. **f** RUPERT IV [28]

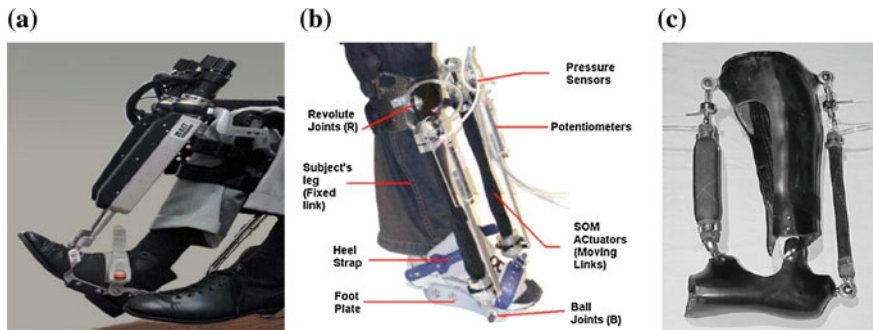


Fig. 2.3 Examples of wearable ankle rehabilitation robots. **a** The anklebot developed in [29]. **b** The robotic gait trainer developed in [30]. **c** The pneumatically powered ankle-foot orthosis developed in [31]. Images reproduced from [29–31]

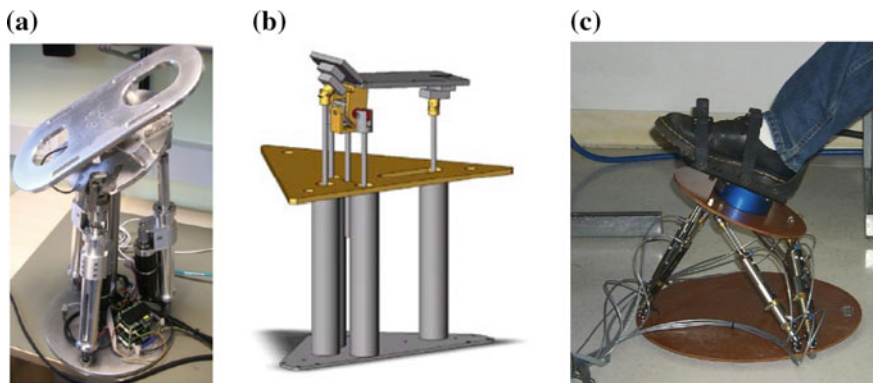


Fig. 2.4 Examples of platform-based ankle rehabilitation robots. **a** The ankle exerciser in [32]. **b** The reconfigurable ankle rehabilitation robot in [33]. **c** The Rutgers Ankle rehabilitation interface. Images reproduced from [32–34]

2.1.2.1 Wearable Ankle Rehabilitation Robots

To enable their use in gait rehabilitation, robotic devices developed for preventing foot drop must be wearable. It must also be controlled to limit the downward rotation of the foot during certain phase of gait. It is therefore not surprising that many robots used in this capacity take the form of actuated orthoses or exoskeletons. While some of these devices provide actuated motion in only 1-DOF to influence foot plantarflexion and dorsiflexion [35–38], others also include the possibility of controlled or passive inversion and eversion movements [9, 39, 40]. The internal-external rotation of the foot, however, is rarely controlled as it is assumed to be a negligible component of gait.

The actuators used in wearable ankle robots developed in the literature are typically of lower inertia to allow higher mobility due to the wearable nature of the device. The actuators are also chosen to be inherently backdrivable to ensure the safety of the user. An example of such actuators is the series elastic actuators used in [35, 38]. This family of actuators is constructed by placing an electric motor in series with a compliant elastic element. The compliant element therefore isolates the motor inertia from the actuator end point inertia and the force applied by the actuator can be regulated by controlling the deformation of the compliant element. These actuators are normally used with stiffness control and are in general utilised to influence the mechanical behaviour of the ankle joint rather than to provide large assistive moments.

Pneumatic muscle is another type of actuator commonly used in wearable ankle robots due to its high power-to-weight ratio and inherent compliance. It is typically used in systems with higher moment capacity, thus allowing these devices to provide a greater level of assistance during the user's gait. The disadvantage, however, is the requirement of a source of compressed air and the non-linear dynamics of the actuators. Both position control [30] and proportional myoelectric control [31, 36, 37] strategies had been applied on systems using pneumatic muscles. Position control is generally used to drive the length of the muscles to values which correspond to the desired foot configuration/orientation while proportional myoelectric control activates the pneumatic muscles according to the myoelectric signals measured from the user/patient's leg muscles.

Some notable features can also be identified in the wearable ankle robots considered in this review. The first is the incorporation of some element of intelligence into these devices. For example, adaptability was introduced in [35] to improve the performance of the AFO by adjusting the AFO stiffness to reduce the occurrences of drop foot gait, while the previous gait velocity is used in [38] to generate references for subsequent gait cycles. Additionally, knowledge of the general gait pattern had also been incorporated in higher level control schemes which coordinate the switching of AFO behaviour according to the current phase of gait.

Another important feature worth noting can be found in the mechanical designs of [9, 30, 40], where the AFOs were designed to be under-actuated when not attached to the user/patient. The advantage of this is that it will not be necessary to align the AFO's kinematic constraints to those of the human ankle, thus allowing the device to cater for a wider range of users and reducing set-up time. Furthermore, with an appropriate design, the device will be able to provide control or support in the important degree of freedom while at the same time acting passively in the remaining directions. This therefore helps to maintain natural movement of the ankle-foot structure and ensures that no unnecessary constraints are imposed on the user's ankle-foot complex.

2.1.2.2 Platform-Based Ankle Rehabilitation Robots

A range of platform-based devices had also been developed by researchers for the purpose of sprained ankle rehabilitation [41, 42]. They are therefore designed to carry out various ankle rehabilitation exercises such as motion therapy and muscle strength training. Motion therapy can be divided into passive, active-assist and active exercises, each requiring a different level of participation from the patient, ranging from no active effort in the passive exercises to full user-driven motion in active exercises. Strength training requires the robot to apply a resistive load to impede the user's movement to improve muscle strength.

One of the key differences between these platform-based devices and the wearable devices discussed previously is that the platform-based devices have a fixed base and thus cannot be used during gait training. Given the rather limited range of motion at the ankle-foot complex, parallel mechanisms are typically used for multiple DOF systems to reduce the size of the robot. With the exception of the Stewart platform-based device proposed in [43] which is capable of 6-DOF motion, most researchers have opted for designs which offer 2- or 3-DOF in rotational motion, where robot movements in the yaw direction (internal-external rotation) are typically constrained on 2-DOF devices. Most of the lower DOF devices also include a central strut in the robot's kinematic structure to provide the kinematic constraint required to restrict the movement of the end effector so that it is purely rotational [41, 44].

Different actuators had been used in platform-based ankle rehabilitation robots. The Stewart platform-based device in [43] and the reconfigurable ankle rehabilitation platform in [33] have utilised pneumatic cylinders to provide actuation, while electric motors were used in devices developed in [41, 45, 46]. A custom-designed electric actuator was proposed in [32, 48] to improve actuator backdrivability, whereby a cable-driven pulley system is used to convert the rotational motion of a DC motor to linear motion of the actuator rod.

A variety of control schemes had been implemented on these platform-based ankle rehabilitation robots. One approach involves the use of either pure force or pure position control for the execution of different exercises [43]. For instance, position control of the platform is typically used for passive range of motion ankle exercises where the user's foot is guided by the robot along the prescribed rehabilitation trajectory, or for isometric exercises where the orientation of the robot is kept constant while the user exerts a particular moment on the robot. Force control on the hand is used to maintain a desired level of interaction torque between the user and the robot during resistive or assistive exercises. Impedance/admittance control strategies had also been implemented, usually through a position-based approach whereby the robot's reference trajectory is modified based on the desired robot impedance and the measured interaction forces/moments [32, 33, 47, 48]. Such control schemes are also generally used with a computed torque/inverse dynamics based position controller to allow accurate tracking of the desired reference trajectories. While the basic interaction control schemes had been implemented on existing platform-based ankle rehabilitation robots, little emphasis had

been placed on the realisation of adaptive control in such devices to allow adjustment of the robot behaviour due to variation in the user's joint characteristics and capability.

2.1.3 Rehabilitation Robots for Masticatory System

There is no specific medical condition called “jaw motion disorder”. For the scope of this book, jaw motion disorder will be used to describe a deviation from ideal or normal function of the masticatory system, due to a single factor or combination of pathologies that affect mandibular movement. The causes of this dysfunctional movement can be from temporomandibular joint disorder (TMJD), which is an affliction of the TMJ, or paralysis or weakening of mandibular muscle.

TMJD is also known as myofascial pain dysfunction or Costen's syndrome and it is mostly caused by habitual behaviours that place excessive wear on the TMJ, trauma, disease or wear from aging [49, 50]. It is estimated that at any one time in the USA, 10 million people suffer from TMJD, making it a major health problem [51]. The symptoms and effects of TMJD are numerous and affect other areas in the craniofacial region as well as the mandible itself. Symptoms such as headaches, ear pain, dizziness, fullness of the ear (where the ear feels clogged) and tinnitus (ringing in the ear) can all be caused by TMJD. More intuitively, TMJD also causes muscle pain, pain within the TMJ and grinding, crunching or popping sounds [49]. Pain in the facial muscles and jaw joints can also be extended to the neck or shoulders. The most significant effect, however, is on the actual motion of the mandible. TMJD reduces the range of motion, causing the mandible to not fully open or to deviate to one side during opening. These unnatural movements feel awkward and affect the nature of occlusion [50].

Besides TMJD, the other significant contributing factor to jaw motion disorder is a loss in mandibular muscle function and this can be due to a weakened or paralysed muscle or group of muscles. The causes of detrimental mandibular muscle function are no different from those of other muscles within the body, and include muscular dystrophy (MD) [52], muscular atrophy [53] and stroke.

Most of the pathologies described in this section which contribute to jaw motion disorder can be treated with varying degrees of physical therapy. The therapy has to be tailored to a particular patient and should suit their specific needs. Passive motion therapy followed by active therapy can help strengthen and increase the movement range of muscle. However, existing aids are not sufficient at emulating all the motions that may be required for mastication and a more sophisticated device is required that accommodates the multiple DOFs of the mandible to restore mandibular function in rehabilitative processes.

2.2 Human Musculoskeletal Models

2.2.1 Movements of Upper Limb

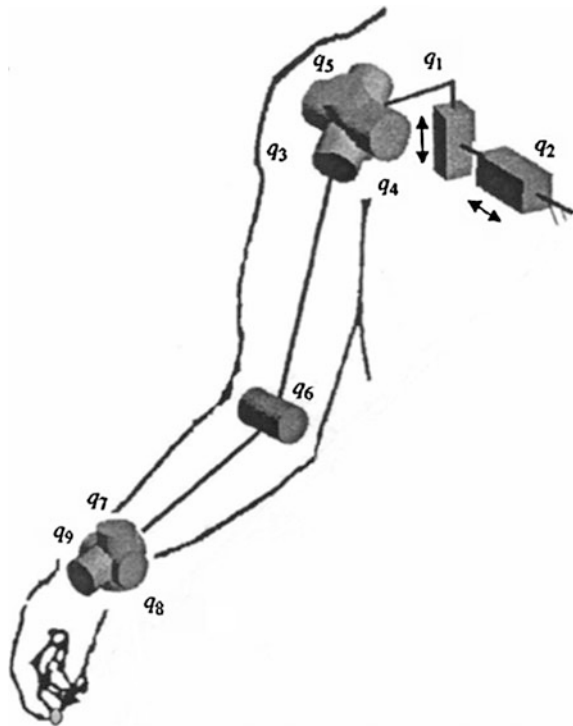
Operating alongside the human upper limb, exoskeletons need to be capable of producing movements similar to those of the upper limb. The upper limb effectively has a total of 9-DOF from the shoulder to the wrist with the finger joints excluded as shown by q_1 to q_9 in Fig. 2.5 [54, 55]. This 9-DOF gives the upper limb exceptionally high manoeuvrability and allows the hand to reach a very large workspace. The proximal joints of the upper limb are often considered a higher priority for rehabilitation as these joints have the largest influence on the hand's position and provide support for the rest of the limb.

The shoulder joint has 5-DOF, 3-rotational DOF which allow spherical rotation of the upper arm and 2-translational DOF which move the upper arm along the vertical axis and the anterior–posterior axis. The movements of each DOF are commonly described by a pair of terms, one for movement in the positive direction and one for the negative direction:

Shoulder flexion

Rotation of the upper arm about the shoulder ICOR (instantaneous centre of rotation) out of the plane of the torso so that it points forwards.

Fig. 2.5 9-DOF of the human upper limb [55]



<i>Shoulder extension</i>	Rotation of the upper arm about the shoulder ICOR out of the plane of the torso so that it points backwards.
<i>Shoulder abduction</i>	Rotation of the upper arm about the shoulder ICOR in the plane of the torso so that it is lifted upwards.
<i>Shoulder adduction</i>	Rotation of the upper arm about the shoulder ICOR in the plane of the torso so that it is dropped downwards.
<i>Shoulder medial rotation</i>	Axial rotation of the upper arm towards the torso.
<i>Shoulder lateral rotation</i>	Axial rotation of the upper arm away from the torso.
<i>Shoulder elevation</i>	Translation of the shoulder ICOR upwards.
<i>Shoulder depression</i>	Translation of the shoulder ICOR downwards.
<i>Shoulder protraction</i>	Translation of the shoulder ICOR forwards.
<i>Shoulder retraction</i>	Translation of the shoulder ICOR backwards.

An interesting phenomenon of the shoulder is that abduction of the upper arm above the horizontal plane will occur simultaneously with elevation as shown in Fig. 2.6 [56]. Without this elevation, abduction above the horizontal plane cannot be achieved.

The elbow joint has 1-rotational DOF which moves the forearm with the following movements:

<i>Elbow flexion</i>	Rotation of the forearm about the elbow joint so that the forearm is moved closer to the upper arm.
<i>Elbow extension</i>	Rotation of the forearm about the elbow joint so that the forearm is moved further from the upper arm.

The wrist joint has 3-rotational DOF allowing the hand to rotate spherically about the wrist joint. The movements are described as:

<i>Wrist flexion</i>	Rotation of the hand about the wrist joint towards the palm.
<i>Wrist extension</i>	Rotation of the hand about the wrist joint away from the palm.

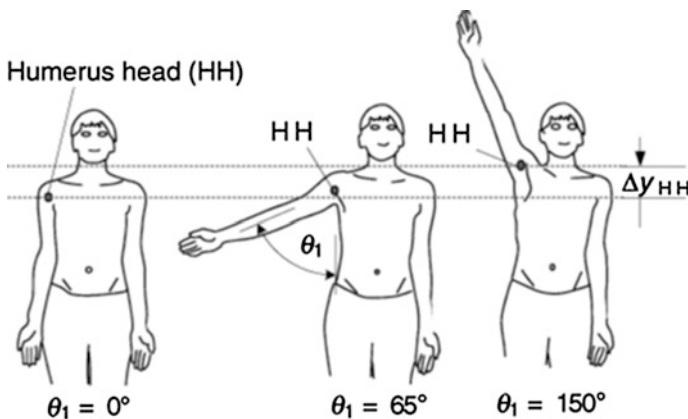


Fig. 2.6 Shoulder elevation during abduction of the upper arm [57]

<i>Wrist radial deviation</i>	Rotation of the hand about the wrist joint towards the thumb.
<i>Wrist ulnar deviation</i>	Rotation of the hand about the wrist joint away from the thumb.
<i>Forearm pronation</i>	Rotation of the hand about the axis of the forearm so that the palm faces backwards when the hand is pointing downwards.
<i>Forearm supination</i>	Rotation of the hand about the axis of the forearm so that the palm faces forwards when the hand is pointing downwards.

2.2.2 Model of Ankle Joint

Kinematics of the ankle-foot complex had been extensively studied in the literature. The simplest representation of ankle-foot motion is that of a hinge joint perpendicular to the sagittal plane. This description considers the entire foot as a rigid body that can rotate about the shank in the plantarflexion and dorsiflexion directions. This is a gross oversimplification of the ankle-foot motion as movements in other DOF are ignored. Additionally, early studies had found from examination of the talus bone surface geometry that the axis of rotation of the talus will vary with its orientation [58, 59]. The actual kinematics of the foot is therefore very complex as it is governed by the articulating surfaces between the different foot bones, as well as constraints imposed by ligaments, tendons and soft tissues. This was highlighted in various studies which investigated the movement patterns of foot bones in terms of 6-DOF motion in either in vitro or in vivo scenarios [60–63]. The general findings of these works were that the axes of rotations of the ankle and subtalar joints do vary rather considerably between different foot orientations and different individuals/specimens. Additionally, translational motions of the joint centres were also recorded, although it was found that these movements are typically within the range of one to two centimetres.

Information of ankle kinematics is essential in applications such as gait analysis, diagnosis of normal ankle-foot function and design of implants for total ankle replacement. However, the complex motion observed at the ankle makes it difficult to describe the complete ankle kinematics concisely with a mathematical model. Models of varying levels of complexity had been established for different applications. As discussed above, the simplest model used is that of a single hinge joint model (Fig. 2.7a). Furthermore, ankle-foot motion had been described as purely rotational using an effective spherical joint (Fig. 2.7b) [64], while the biaxial model which considers the foot motion to be equivalent to rotations about two hinge/revolute joints in series was also widely adopted in literature (Fig. 2.7c)

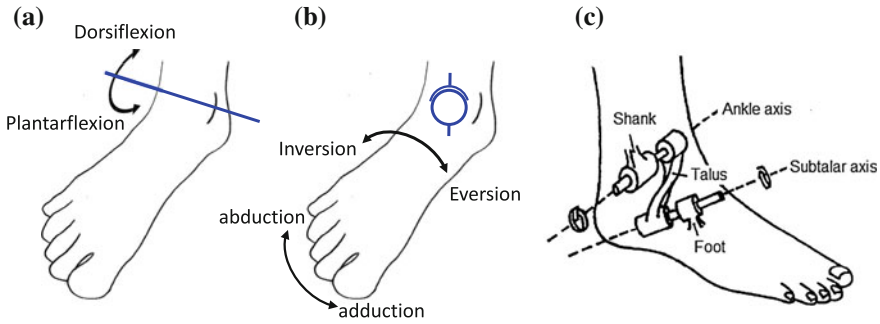


Fig. 2.7 Kinematic models used to describe ankle motion. **a** Hinge joint model. **b** Spherical joint model. **c** Biaxial joint model. Adapted from [65]

[64–69]. Additionally, recent studies had modelled the ankle-foot kinematics using four-bar linkages and spatial parallel mechanisms [63, 70].

Since large inter-subject variability is observed in ankle kinematics, a user-specific description of ankle kinematics should be used to adapt the robot behaviour to suit the current user. Most of the studies in literature that considers subject-specific ankle kinematics in full 6-DOF have utilised either motion tracking systems or medical imaging techniques [71, 72]. As these methods typically require offline processing, they are not suited for use in real-time systems. A simpler kinematic model with reduced degree of freedom which is amenable to online parameter identification is therefore more appropriate for this research. Additionally, since this representation of the ankle kinematics will also be incorporated in the dynamic model of the ankle-foot structure, the use of a straightforward model will reduce the computational complexity of the system, thus making its simulation more tractable. For the above reasons, the biaxial ankle model appears to be sensible choice and its parameter estimation will be further discussed.

Parameter identification for a biaxial kinematic model was investigated by van den Bogert in an in vivo manner using visual markers placed on the subject's foot [68]. The biaxial model considered has 12 parameters and these are determined through minimisation of the discrepancies between marker positions obtained from the assumed model and from measurements using the Levenberg–Marquardt algorithm. The resulting ankle and subtalar joint orientations using this method were found to be similar to corresponding values obtained from in vitro anatomical studies of the ankle. Good fit of the model in terms of the marker positions was also reported, with relatively small rigid body errors. Lewis et al. had also investigated the parameter identification of the biaxial ankle model on both biaxial mechanical linkage and cadaveric foot specimens [71]. The optimisation algorithm used is largely similar to that described by van den Bogert except that the ankle and subtalar joint displacements were estimated through optimisation using the Gauss–Newton algorithm. It was reported that the parameter identification of the biaxial mechanical linkages shows results that are largely consistent with the actual

kinematic parameters of the structure. Considerable discrepancies, however, were observed between the ankle and subtalar joint orientations computed from the optimisation algorithm and the average helical axes obtained from successive measurements of the foot bone orientations. This had therefore led them to conclude that the biaxial ankle model with fixed revolute joints can only give a limited representation of the actual ankle-foot kinematics, and that an alternative model, perhaps one with configuration-dependent joint axes orientations, be explored. It should be noted that both the previous works discussed above on the identification of biaxial ankle kinematic model parameters were completed using offline optimisation techniques.

Studies in the biomechanical characteristics of the ankle go beyond the understanding the kinematic behaviour of the ankle. It seeks also to identify how the human ankle will react under certain loading conditions, as well as the loading distribution among different anatomical structures of the ankle-foot complex such as foot bones, ligaments, tendons and other soft tissues.

One of the core components of a computational ankle model is a description of the ankle-foot kinematics as it determines how the foot bones will move relative to one another, thus ultimately influencing the length of ligaments and muscle-tendon units, as well as deformation of other soft tissues. While the use of three-dimensional contact constraints [73–75] can lead to more realistic results, it can be computationally intensive and therefore limit the speed of simulations. In this aspect, the biaxial ankle kinematic model described previously appears to be able to provide a good balance between simplicity and the ability to provide a reasonably description of the ankle-foot motion.

Another important modelling decision is found in the treatment of bones and soft tissues. Some models treat the bones as rigid bodies and ignore effects caused by deformation of soft tissues [67, 73], while others apply finite element analysis on the bones and soft tissue in order to obtain the stress distribution across the articulating bone surfaces [74, 75]. Clearly, use of finite element analysis will improve the accuracy of the model at the expense of increased computational complexity.

Effects of ligaments on the ankle-foot biomechanics had also been considered in some models. Typically, they are treated as tension-only elastic elements whose lengths are dependent on the configurations of foot bones [73–75]. Some models, however, include the influence of ligaments on passive joint stiffness as a lumped effect, and describe it through application of non-linear resistive moment-displacement functions at the ankle and subtalar joints [76, 77]. Properties of muscles and tendons are also commonly included in computational models which require consideration of active muscular contractions [67, 76, 77], and these models typically employ a Hill-based muscle model and are often used for gait analysis. Models which involve explicit modelling of the ligaments and muscle-tendon units generally require the acquisition of bone geometry and ligament/tendon attachment locations by means of medical imaging, and this can add to the complexity of the model. However, as forces and strains along the ligaments/tendons can be extracted from such models, the added complexity can be justified for applications requiring greater insights into the loading on these anatomical elements.

2.2.3 *Model of Masticatory System*

The purpose of the human masticatory system is to perform the initial breakdown of food via chewing and prepare it for swallowing. It includes the bones and soft structures, such as muscles, ligaments and tendons of the face and mouth, that are involved in mastication. There are multiple complex mechanisms involved in this process, including the secretion of saliva, manipulation of the chewed food into a bolus with the tongue and the muscular control of the mandible that produces chewing motions, which is the focus of this and following parts. The main problem is that if the temporomandibular joint (TMJ) or mandibular muscles are weakened, the patient is unable to properly chew and process food. Current solutions include modifying the texture or consistency of the food for easier processing, or bypassing the masticatory system by tubular feeding through the oesophagus or abdomen. However, there are also efforts that are attempting to reduce the effects of physical symptoms through therapy and physical rehabilitation of the mandibular muscles. Similar to gait or upper limb rehabilitation, these exercises consist of repetitive exercises that are conducted with the assistance of a therapist, and can therefore also benefit from the advantages robotics and exoskeletons provided in those applications.

This section introduces the components of the masticatory system directly involved in the formation of chewing motions, including the skeletal structure, primary mandibular muscles and the temporomandibular joint. These components form a unique complex structure that is not replicated elsewhere in the human body and allows the mandible to be manipulated in three-dimensional space. A review of the available literature regarding masticatory robotics and the development of jaw exoskeletons for rehabilitation is also presented, which lead to the motivations behind this particular case study.

2.3 Control of Rehabilitation Robots

2.3.1 *Motion/Force Control Strategies*

The main goal of interaction control is to establish a certain relationship between force and motion, and this relationship is typically expressed as either a mechanical impedance or admittance. To realise these relationships, both force and motion of the robot have to be obtained from sensors and acted upon accordingly through application of suitable control laws. However, the most tightly controlled loop in a rehabilitation robot typically deals with only one of the two interaction variables, and these control loops are considered as low-level controllers in this review. These lower level control loops of the interaction controllers are generally implemented using conventional position (or force) control to ensure that the desired motion (or force) is applied to the robot. An outer loop is then applied to alter this desired

motion (or force) depending on the measured force (or motion) so that the overall behaviour of the robot resembles that of a mechanical system exhibiting the desired impedance or admittance.

2.3.1.1 Inner Loop Position/Velocity Control

In addition to the commonly used proportional-integral-derivative (PID) controller, another popular strategy used in the implementation of a position-controlled inner loop is the computed torque control [78]. This method is an established method for position tracking of robotic manipulators and operates by linearizing the robot dynamics through application of feedback terms which aim to cancel the non-linear terms in the robot dynamic equations. An additional proportional derivative (PD) term acting on the position error is also applied to facilitate tracking of the reference position. The computed torque control scheme therefore requires a good knowledge of the robot dynamics as well as the ability to measure actuator velocities. In applications where the robot velocity is low, the velocity-dependent terms can be neglected and gravity compensation alone can be used to reduce the computational complexity of this approach [79].

Variants of the computed torque control laws had been used in interaction control of rehabilitation robots [80, 81]. In robots with inner position control loops, the observed interaction forces are used to compute reference accelerations according to the desired impedance relationship. These reference accelerations are then fed to the inner motion control loop to realise the prescribed interaction behaviour.

2.3.1.2 Inner Loop Force/Torque Control

Inner force or torque control loops can also be used to provide the required interaction behaviour. In this alternative approach, the motion of the robot is used to generate the force/torque reference. Similar to the case of motion control, the simplest force controller can be obtained through the use of PID-type controllers. More advanced control strategies such as disturbance observers [82, 83] had also been used to reject disturbances stemming from frictional forces and modelled dynamics. It should be noted that computed torque control used in robot motion control also ultimately requires some form of actuator level force/torque control. This is because it operates on the assumption that the desired torque is accurately delivered by the actuators.

Naturally, actuator force control can be carried out with the feedback of actuator forces. The main challenge associated with the implementation of control laws requiring force feedback, however, is system stability. Since compliant force sensors are typically required to measure the actuator force, it contributes to additional position feedback [84]. As a result, large sensor stiffness will lead to a large, effective position feedback gain, thus creating severely under-damped systems

which could become unstable when higher order modelled dynamics are taken into account. Force sensors that are too soft on the other hand will result in inaccurate position measurements due to additional unmonitored force sensor deformations.

Researchers have proposed that the passivity of the controlled system must be preserved if stability were to be maintained during interaction with arbitrary passive environments [85, 86]. This imposes an upper limit on the force feedback gain that can be used depending on how the actuator mass is distributed when the actuator's first resonance is considered in the actuator model. Since the main contribution of increasing the force feedback gain is shown to be a reduction in the apparent actuator inertia, the above limitation also restricts the extent to which this inertia can be reduced. Recent work, however, had proposed the use of environmental information to relax the passivity criterion to permit performance improvements of the actuator [87, 88]. The authors have imposed bounds on the expected human arm impedance and utilised it to numerically compute force feedback gains that satisfy the robust stability criterion based on the small gain theorem.

An alternative strategy in the regulation of actuator force involves the use of a force sensor-less control scheme. Instead of measuring the actuator force/torque through force/torque sensors, this method uses a disturbance observer based approach [89] to estimate the reaction torque/force from current and motion variables. This was shown in [90] to reduce the oscillations found in the resulting force response. Such a control strategy, however, requires the measurement of actuator velocity and a good knowledge of model parameters such as actuator inertia, damping and friction. Torque control was also achieved in [91] through the use of a position disturbance observer in the control of a rotary series elastic actuator, which consisted of a highly geared motor coupled in series with a torsional spring. In this approach, torque control is realised by accurately controlling the deformation within the torsional spring.

While a considerable amount of research had been made in force/torque control, manipulator force control is still mainly achieved by independent control of individual actuators, where the torque/force of each actuator is regulated in its own feedback loop. It is therefore worthwhile to investigate whether force control performance can be improved when the robot actuators are treated collectively as a multi-input, multi-output system.

2.3.2 EMG Signals Based Control

Muscle tissue can be divided into two main types based on their fundamental structure: striated muscle and smooth muscle [92]. Striated muscle is involved with conscious movement and its basic structure consists of thick and thin filaments that slide against each other to produce movement [93]. Skeletal muscle is a type of striated muscle that is usually attached to a bone on each end by tendons. Upon voluntary activation, the muscles contract and shorten pulling on the tendons, and the resulting tension causes the bones to move relative to each other.

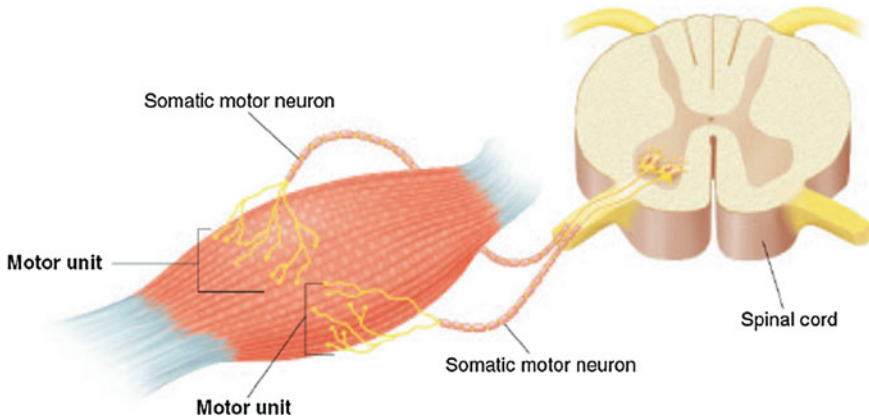


Fig. 2.8 The motor unit [94]

Each skeletal muscle cell is essentially a muscle fibre and two stages of bundling of muscle fibres result in the formation of skeletal muscle tissue. Somatic motor neurons externally stimulate the muscle cells to cause contractions. The cell body of a motor neuron is located in the grey matter of the spinal cord and its axon (main branch) splits into a number of collateral branches (motor neuron fibres) before being distributed about the muscle fibres as nerve fibre branches. Each branch innervates or activates a single muscle fibre, and the somatic motor neuron, together with all the muscle fibres it innervates, is collectively known as a motor unit, as shown in Fig. 2.8. For more details of muscle structure and function, see [94].

The control signals of skeletal muscle are called action potentials and these are electrical impulses that originate in the brain. The signals propagate through the central and peripheral nervous systems and are transmitted down the axons of motor neurons where they reach a specialised synapse called the neuromuscular junction, where the action potentials cross the boundary from muscle motor neuron to muscle fibre, and stimulate contraction [95]. This is illustrated in Fig. 2.9. Note that the muscle fibres of a single motor unit are not necessarily adjacent to each other and are instead interspersed throughout the muscle.

To take advantage of the benefits, the EMG signal can provide to interfacing, the term “neuromuscular interface” (NI) has been coined to describe a new type of self-contained module with a specific role. This role is to obtain and process EMG signals into a predicted joint torque or position of the joint or limb of interest. The NI includes all the hardware and software components that would be required to perform this procedure. The part an NI plays in the grander scheme of a complete control system and its more detailed constituents is shown in Fig. 2.10. The input to the interface are the EMG signals of the target joint, and the output is the user-intended torque or position of the associated limb, which can then be used by a controller or similar component to operate an exoskeleton. The focus on the development of a dedicated module to serve as the interface means that the NI can

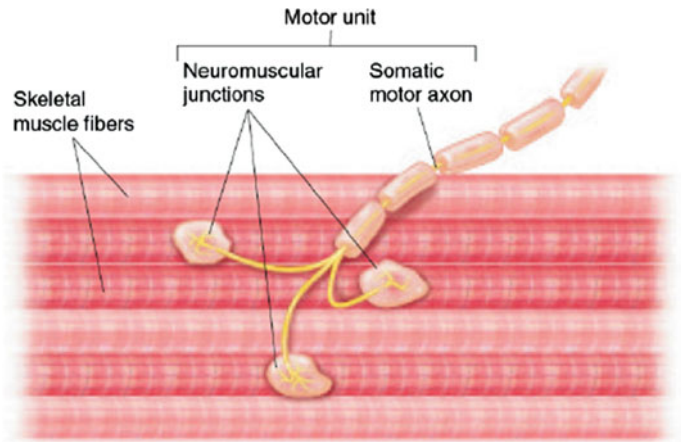


Fig. 2.9 The neuromuscular junction [94]

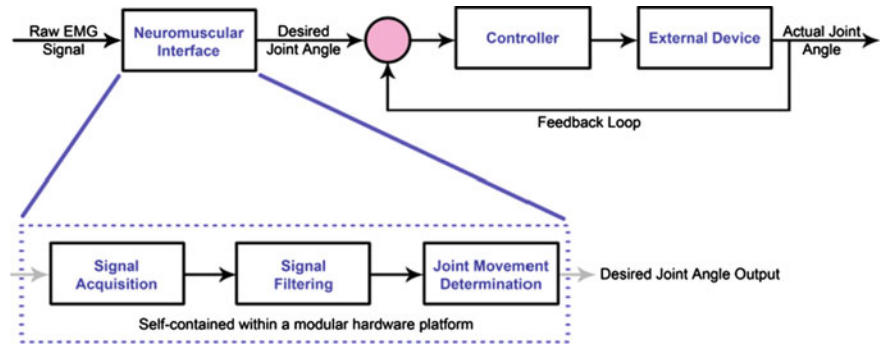


Fig. 2.10 The concept of a neuromuscular interface

be included in an unlimited number of applications both within and outside the health industry. The concentration on such a specific module for interfacing is unprecedented, with interfacing methods commonly a secondary consideration as a component of an overall system that includes controller, actuator and sensor design, and implementation. The output of the NI will only need to be post-processed further depending on the purpose, and the input signal should be able to drive exoskeletons, prostheses, mobility devices, communication tools, industrial equipment and any other relevant applications.

The realisation of an NI depends on and requires the development of several key areas of importance. Challenges in areas such as the EMG signal filtering process, conversion and interpretation of the signals into an equivalent joint torque or position and the specialised hardware development of the interface; all need to be addressed before an NI can become robust, reliable and practical enough for

real-world applications. As mentioned previously, one of the main limitations of current exoskeleton devices is the lack of an adequate interface for information exchange between the user and the device. This is the current-limiting technology, which will continue to hold back exoskeleton development despite any advances in power supply, actuator or sensor technologies.

Most of the interfacing methods and approaches presented and discussed use commercial EMG or biosignal acquisition systems that have been specifically designed for clinical or research purposes. The conversion of an interface into a re-usable module includes the specialised hardware necessary to acquire EMG signals and the non-existence of such hardware is a key problem that is preventing EMG interfaces from becoming practical. Custom hardware interfaces are required to maximise performance while reducing unnecessary bulk and reviews of examples of wearable hardware systems can be found in [96, 97]. These hardware systems are targeted more towards patient or individualised monitoring applications, rather than interfacing and control, but they share similar design considerations, such as durability, portability, power consumption requirements, cost and aesthetics.

The realisation of EMG interfaces on hardware is an important consideration because the transfer from a laboratory environment to a real-world embedded system will inevitably have lower quality hardware components (to save cost) and have to account for additional sources of noise and disturbances. There is also an emphasis on reducing power consumption and volume to ensure that electrodes and corresponding processing circuitry and systems are as unobtrusive to the user as possible [98]. Progress in this area has been relatively lacklustre with only a few minor developments. This is partly because interfacing methods have their own fundamental problems and are barely on the cusp of being robust or practical enough to make it beyond the research or laboratory stage.

2.3.3 Interaction Controllers for Rehabilitation Robots

2.3.3.1 Basic Interaction Controllers

One of the most basic forms of interaction controllers can be found in simple impedance control, which essentially applies the torque command in an open loop manner without any force feedback. This torque command, however, is determined based on the desired impedance relationship and the discrepancies between the desired and actual robot motion. Owing to the lack of force feedback, this interaction control approach has poorer disturbance rejection but does not suffer from the stability issues discussed previously. It is therefore suitable for the use with devices with low inherent inertia and low friction. Force feedback control can also

be used in impedance control schemes to allow reduction of the apparent robot inertia and improve the force tracking ability of the robot. However, the force feedback gain and hence the performance improvement are again limited due to stability constraints. Natural admittance control can be used to regulate the end point admittance of a robotic manipulator. It does so by using both force and velocity feedback in the same control loop. It was proposed that the mechanical admittance used in this approach be selected in such a way that the apparent end point mass of the controlled system is identical to that of the actual physical system to maintain passivity. Stiffness and damping characteristics, however, can be chosen as desired. Additionally, the velocity feedback gain is chosen to be large so that effects of disturbance forces such as friction can be reduced.

2.3.3.2 Higher Level Interaction Control

In addition to the basic interaction control strategies described above, higher level interaction control schemes had also been investigated in rehabilitation robots, with many such schemes focusing on improving the safety and incorporating adaptability in the rehabilitation robots. These higher level controllers are also generally designed with a particular type of rehabilitation exercise in mind.

Safety and adaptability in rehabilitation robots are somewhat related. For instance, different patients will have different joint or limb kinematics. It is therefore unreasonable to have the robot strictly enforce one set of rehabilitation trajectories for all patients as it may result in application of large forces and thus lead to discomfort/injuries. In fact, impedance control in itself can be viewed as having a built-in adaptive mechanism as it permits positional deviation from a virtual reference when external forces are encountered. Some higher level interaction controllers extend on this and provide greater freedom to the user to dictate the actual path taken in rehabilitation. However, the extent of this freedom must also be bounded to ensure that the required exercises are still being carried out.

In order to achieve adaptability of this nature, some controllers for lower limb rehabilitation define a particular region or tunnel around the reference trajectory within which the interaction forces between the robot and user are minimised [99, 100]. This is typically achieved through feed forward compensation of the robot inertial and gravitational forces. It is also possible to reduce the time dependency nature of the reference trajectory by identifying the reference point using a nearest neighbour approach [99]. Various strategies for the adaptation of rehabilitation trajectory had also been considered in [101] for a position-controlled gait rehabilitation robot. Some of these strategies were aimed at reducing the active patient torques through modification of the reference gait trajectory while another utilises impedance control to allow deviation from the reference trajectory. The recorded deviation due to impedance control is then incorporated into the reference trajectory of the next gait cycle.

An alternative approach taken in [102] to provide adaptability in upper limb rehabilitation is to avoid the prescription of reference trajectories in the Cartesian space, and instead define the virtual trajectory in terms of Euclidean distance to the desired end point. In other words assistance is given to the user through impedance control when the distance between the current position and the end position exceeds that desired for the particular time instant. In [103], a moving potential field is used to define the level of forces applied to move the user's limb to the current reference position. This potential field, however, is selected in such a way that it will not impede the user's movement should the current arm position be closer to the target destination compared to the current reference. This means that the controller is designed so that it would not penalise users when they are performing better than required.

One other way to improve safety is to use a smaller manipulator impedance to allow larger deviations from the reference trajectory. An obvious shortcoming associated with this approach is that certain positions in the limb or joint range of motion will not be reached as insufficient forces are available for guidance. A method to overcome this problem is to apply a reference force on top of the force command generated from impedance control. This will provide adequate forces to move the affected limb or joint while also allowing for greater flexibility in terms of the limb or joint position. In [104], this reference force is generated from a series of radial basis functions whose weights are adaptively tuned to compensate the inertial and gravitational forces of the robot and user.

Another aspect of adaptability can be described as the ability of the robot to cater for the physical capability of the patients. Various researchers have proposed that robots used in neuromotor training should encourage the patient to actively participate in the rehabilitation exercises by providing assistance or intervention only when it is needed [103, 105, 106]. It was also observed that given the opportunity, the user will decrease their effort and rely on the robot's assistance to complete the rehabilitation exercises [107]. Robots used in rehabilitation must therefore also be able to adjust the task difficulty or level of assistance it provides to the user according to some performance indicator. A common approach is to reduce the level of assistance over time. This can be done by reducing the assistive forces by decreasing the impedance or feed forward force parameters as in [102, 104]. Clearly, a mechanism must also be put in place to halt the decay in assistance should the performance of the patient deteriorate, and this is typically accomplished via the addition of a term which increases the reference force or impedance parameter based on variables derived from motion error. A fuzzy inference system has also been used in [80] to vary the robot behaviour between that of a minimal interaction force-controlled and impedance-controlled robot depending on the position-tracking errors. This means that when the user is moving as required, the robot will merely actuate to support its own gravitational and inertial forces. However, as the user fails to follow the required motion trajectories, the robot will provide assistance according to the prescribed impedance relationship.

2.4 Discussion

Recent technological advances have enabled the development of feasible exoskeleton robots. Modelling software has allowed exoskeletons to be tested in simulations before they are fabricated, allowing rapid prototype development. Biomechanics modelling allows the exoskeleton to mimic the dynamics of the human limb. Sensor technologies, control strategies and computing power have advanced to the extent where they are no longer major obstacles. However, actuator and power supply technologies still have limitations.

One of the fundamental limitations in past upper limb exoskeletons is the inability of the shoulder mechanism to achieve the entire human shoulder workspace with adequate performance. This is due to the singular configurations present in the 3R spherical wrist mechanism that are used in the shoulder complex of the exoskeletons. Operating at a singular configuration results in the loss of 1-DOF in the 3R mechanism and therefore the exoskeleton becomes unable to achieve the same movements as that of the 3-DOF human shoulder. Even operating near a singular configuration reduces the ability of the 3R mechanism to rotate about the affected DOF and requires the 3R mechanism to move at undesirably high velocities. Several recent upper limb exoskeletons have considered this problem and designed the exoskeleton so that the singular configurations occur at uncommon shoulder postures at the edge of the shoulder workspace. Although this change partially improves the performance of the exoskeleton, the singular configurations are still present in the shoulder workspace and operating near these configurations causes poor performance in the 3R mechanism.

Among the 9-DOF in the human upper limb, the 3-DOF spherical movement of the shoulder has the largest range of motion and has the most influence on the rest of the upper limb since it is the most proximal joint in the limb. Therefore, recovery of the shoulder joint is often more urgent than the other joints in the upper limb. The shoulder joint is also highly complex and is the most powerful joint in the upper limb making it the most difficult joint to provide rehabilitation for. Thus, designing an exoskeleton that is capable of implementing all movements of a normal human shoulder is highly challenging, but such capabilities can provide significant improvements to existing shoulder rehabilitation methods.

Clinical results are available for some of the early end-effector type robots which provide strong evidence that robotic rehabilitation has a beneficial effect on motor function. However, comparing clinical data is rather difficult as different groups use different devices, control strategies, intervention strategies and assessment criteria. There are many patient-specific parameters that can affect the outcomes of the treatment which may also need to be taken into consideration. There are currently insufficient guidelines and tools used in clinical evaluations of robotic rehabilitation, and to some degree in conventional rehabilitation, which is limiting the amount of quality data that can be acquired. Many assessment methods, such as the assessment of posture, are based on subjective impressions which make it difficult to justify the effectiveness of rehabilitation treatments. Future research will need to

focus on developing and refining these guidelines and tools to ensure researchers to get as much reliable data as possible out of clinical evaluations. With better data, the effects of variations in the rehabilitation treatment and the patient's condition on motor and functional recovery can be better understood. This will enable the development of more effective rehabilitation exoskeletons and intervention strategies. Exoskeleton technologies have the potential to initiate new areas of research as well as support existing research work. New approaches to rehabilitation treatment and patient assessment may be discovered and a better understanding of the human neuromuscular system can be achieved.

One promising approach for patient treatment is the application of task-based exercises in rehabilitation. There is evidence that suggests task-based rehabilitation specifically designed to deal with lost abilities produce better results than resistance-strengthening exercises. However, realistic task-based exercises are difficult to achieve with manual rehabilitation methods. Exoskeletons have the ability to accurately control multiple joints at the same time, enabling them to produce more realistic task-based exercises for the patient. In addition, studies have found that rehabilitation is more effective when the patient exerts voluntary effort in intensive and frequent exercises, much like recreational exercises. Incorporating rehabilitation exercises into virtual games can make rehabilitation more enjoyable thus motivating the patient to put in effort and encouraging more exercise. In addition, the use of virtual reality enables more realistic task-based exercises to be performed. The concept of using virtual games to provide therapy exercises has already been applied in a number of exoskeletons. The next step is to design games based on rehabilitation principles and allow the games to be adjusted to better match the patient's level of motor deficiency.

It can be seen from the above review that rehabilitation robots had already been proposed in the literature, with wearable devices mainly aimed at gait rehabilitation and platform-based devices focusing more on treatment of ankle sprains. However, it should be noted that the Stewart platform-based ankle rehabilitation robot had also been applied in the area of stroke rehabilitation, thus indicating that it is worthwhile to develop a rehabilitation robot which can be potentially extended to cater for treatment of both ankle sprains and neurological disorders.

One major shortcoming in existing platform-based ankle rehabilitation devices with 2- to 3-DOF is that the rotation of the robot end effector is typically constrained about a point on the robot rather than allowing the user's lower limb to govern the end-effector motion as in the designs proposed. The consequence of this is that the motion of the user's foot will not be limited to movements between the shank and the foot during operation of the robot. Under such conditions, measurements of the robot end-effector orientation may no longer be the true ankle joint displacements, thus limiting the repeatability of the actual ankle-foot motion while also compromising the ability of the robot to act as a reliable evaluation/measurement tool. This issue is therefore addressed during the mechanical design of the ankle rehabilitation device developed in this research.

Additionally, even though existing platform-based ankle rehabilitation devices are already capable of basic interaction control and can perform various

rehabilitation exercises, not much emphasis was placed on the adaptability of these devices. As the kinematics and impedance characteristics of the ankle can vary considerably between individuals, the controller for rehabilitation robots should ideally be able to detect these variations and adjust for it accordingly. An example of this is the reduction of robot impedance in regions of large stiffness to prevent exertion of excessive forces. It is therefore the intention of this research to incorporate adaptability into an ankle rehabilitation robot through online parameter estimation. A suitable interaction control scheme can then be developed to capitalise on the additional information available to improve the safety of the device. Furthermore, the assistance adaptation schemes available in robots designed for motor training were also considered in this research so that the developed device will be able not only to accommodate variations in the users' joint characteristics but also to adapt its behaviour to ensure that the level of assistance provided is based on the user's capability to carry out the required exercises. While the aim of this research is to create a system which is primarily targeted at rehabilitation of sprained ankles, development of an assistance adaptation scheme will also facilitate future extension of the developed system to cover neuromotor rehabilitation.

It is worth noting that many of the assistance adaptation schemes vary the assistive effort either directly or indirectly based on observation on the position-tracking errors, and the adaptation rules are typically formulated in ways which do not place much consideration on the possibility of constrained motion in the robot's task space. This is perhaps due to the predominant application of these algorithms in upper limb rehabilitation where the subject's arm can normally move within the workspace of interest in a constraint-free manner. This is, however, unlikely to be the case for ankle-foot movements due to the existence of coupled rotations which imposes constraints in the three-dimensional rotational space. Assistance adaptation rules which are more suitable for constrained motion are therefore investigated in this work.

It can be seen from the above discussion on ankle models that numerous computational ankle models had been developed to study foot pathology and biomechanics. However, to the best of the author's knowledge, none of these models were applied in the controller development of ankle rehabilitation devices. In addition to its use in controller simulation and in providing information on the configuration-dependent ankle characteristics such as ankle stiffness which can be used for parameter adaptation and stability analysis of the interaction controller, a suitable computational ankle model can also be used to approximate the forces along different ligaments or muscle-tendon units as well as reaction forces and moments encountered at the ankle and subtalar joints. It can therefore also serve as a tool to evaluate the performance of a controller or the effectiveness of a particular rehabilitation programme. It can be seen that a computational ankle model which provides all the functionalities above will greatly facilitate the overall goal of this research in the development of an adaptive ankle rehabilitation robot. Such a model is therefore developed in this research to facilitate both the design and the implementation of the adaptive control scheme.

Lastly, given the considerable variation in the ankle kinematics and the need to incorporate adaptability into the developed system, user-specific ankle kinematic parameters should ideally be available to facilitate adjustment of the controller parameters. It can be seen that while identification of the biaxial ankle kinematic model had been explored in the literature, such identification was carried out in an offline manner. However, due to the real-time requirements of this application, an online parameter identification algorithm is required. Consequently, the development of such an algorithm is also addressed in this research. Owing to the importance of computational tractability, it is proposed that a biaxial ankle model be used to describe the ankle kinematics in the identification algorithm. However, as it is commonly found in literature that orientations of the ankle and subtalar joint axes change with foot configuration, the conventional biaxial ankle model with constant axes orientations is also extended in this research to allow variation of these parameters with foot displacement so that a better fit between the model and measured foot orientations can be obtained.

2.5 Summary

This chapter presented a review of existing works relevant to this research. Numerous issues in the development of robots for rehabilitation have been identified and there is much room for improvement. Of particular concern is the inability of the robots to achieve the full range of motion of the joint's spherical movement with adequate performance. The different types of human rehabilitation devices developed in the literature were considered, with particular focus on their mechanical design, actuation methods and control schemes. Subsequently, studies relating to human joint kinematics and computational modelling were also examined. The state of the art of control strategies for rehabilitation robots was reviewed.

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