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Improving Effectiveness of Robot-Assisted Ankle Rehabilitation via Biomechanical Assessment and Interaction Control

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Co-Supervisor: Dr. T. Claire Davies

Abstract

The human ankle joint plays a significant role in maintaining body balance during ambulation, but it is particularly susceptible to musculoskeletal and neurological disorders. A general rehabilitation program for ankle injuries requires intensive efforts from therapists and patients over prolonged sessions. Robot-assisted rehabilitation solutions, as therapeutic adjuncts to facilitate clinical practice, have been actively researched in the past few decades and provide an overdue transformation of rehabilitation from labour-intensive operations to technology-assisted operations.

Various rehabilitation devices have therefore been developed for the treatment of ankle injuries to reduce the physical workload of therapists and supplement the resources required to facilitate a comprehensive rehabilitation regime so that adequate therapy can be delivered to patients. However, the effectiveness of existing ankle rehabilitation robots is limited by a variety of shortcomings including kinematic incompatibility (misaligned centre of rotation of the robot and the ankle joint), non-compliant actuation, less than three rotational degrees of freedom (DOFs), and the lack of real-time ankle assessment and adaptive interaction training schemes.

This research aims to improve the effectiveness of robot-assisted therapy for the treatment of musculoskeletal and neurological ankle injuries. The fundamental technology is the development of an intrinsically-compliant ankle robot with real-time biomechanical assessment and interaction control. The current device is named the Ankle Assessment and Rehabilitation Robot (AARR).

The AARR has a bio-inspired design, with the functions of both assessment and rehabilitation, devised after a systematic review of a variety of ankle rehabilitation devices. Mechanically, this novel robot is designed with flexibility in generating varying training ranges of motion (ROMs), and employs four intrinsically-compliant Festo fluidic muscles (FFMs) that mimic skeletal muscles to actuate three rotational DOFs. Functionally, the AARR incorporates sensor-based and model-based ankle assessment techniques to facilitate the robotic control for enhanced safety and rehabilitation efficacy.

The ankle assessment protocol for this robot aims to extract biomechanical information of the ankle joint from sensors and computational models. Ankle biomechanical assessment via sensors is implemented by three magnetic rotary encoders for measuring ankle position and a six-axis load cell for measuring patient-robot interaction. Two computational models are developed for estimating ankle ligament kinematics and passive joint torque. They are
distinguished by different definitions of their rotation axes, where the torque model moves about three perpendicular rotation axes (named the PRA-Model) while the rotation axes of the ligament kinematics model are not perpendicular (named the Non-PRA-Model). The proposed ankle assessment techniques are demonstrated to be valid and reliable in extracting ankle biomechanical information during the robotic training through comparisons with published data and experimental validation.

The trajectory tracking of the AARR is implemented by controlling the individual FFM length in joint space, or a cascade controller with position feedback in task space (the outer loop), and force feedback in joint space (the inner loop). With position controllers in the low level, two adaptive interaction training schemes are proposed to enhance patient engagement and rehabilitation efficacy. One scheme is implemented through the predefined trajectory that is adaptive to the movement intention of the patient. The other scheme employs a high-level admittance controller whose performance is adaptively tuned according to real-time patient-robot interaction.

Experiments were conducted on a sprained ankle to evaluate the proposed control strategies when implemented on the AARR, with all normalised root mean square deviation (NRMSD) values of the trajectory tracking at less than 5.4%. To conclude, the AARR has the potential for clinical applications of ankle assessment and rehabilitation, and both interaction training schemes are safe and effective for patients by considering the movement intention and real-time patient-robot interaction.
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Global Acronyms, Abbreviations and Mathematical Notations

Acronyms and Abbreviations

AARR  Ankle Assessment and Rehabilitation Robot
FFM  Festo Fluidic Muscle
MAD  Mean Average Deviation
NRMSD  Normalised Root Mean Square Deviation
PAM  Pneumatic Artificial Muscle
PID  Proportional-Integral-Derivative
RMSD  Root Mean Square Deviation
SD  Standard Deviation

Mathematical Notations

\( \theta_d \)  Desired Robot Position
\( \theta_m \)  Measured Robot Position
\( L_d \)  Desired Muscle Length
\( L_m \)  Measured Muscle Length
\( S_d \)  Desired Muscle Strain
\( S_m \)  Measured Muscle Strain
\( T_r \)  Robot Torque by Actuators
\( T_a \)  Ankle Torque or Patient-Robot Interaction Torque

Note:

1) The rotation axes of ankle dorsiflexion/plantarflexion (DP), inversion/eversion (IE),
and adduction/abduction (AA) are defined as axis-X, Y and Z, respectively. Therefore,
the positions for ankle DP, IE and AA are respectively denoted as \( X, Y \) and \( Z \); the
torques for ankle DP, IE and AA as \( T_x, T_y \) and \( T_z \); the forces along ankle DP, IE and
AA as \( F_x, F_y \) and \( F_z \).

2) Other local acronyms, abbreviations and mathematical notations are defined in each
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Please indicate the chapter/section/pages of this thesis that are extracted from a co-authored work and give the title and publication details or details of submission of the co-authored work. Chapters 4 and 8 are extracted from the below submitted work (Under review).

M. Zhang, W. Meng, T.C. Davies, and S.Q. Xie, "Design and intelligent control of an Intrinsically-Compliant Ankle Assessment and Rehabilitation Robot", IEEE/ASME Transactions on Mechatronics.

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Chapter 1. Introduction

This thesis comprises five published journal articles, two under-review journal manuscripts, and two independently-written chapters of introduction and conclusion. To make the completed thesis functions as an integrated whole, connecting texts have been provided at the beginning of each chapter. This chapter presents a general overview of various ankle injuries and the physiotherapy, and the research issues encountered in the course of developing ankle rehabilitation robots, as well as the thesis structure.

1.1. Ankle Complex

The ankle-foot complex contributes significantly to the function of the lower extremity, and is fundamental to balance, support, and propulsion [1,]. However, it is particularly susceptible to musculoskeletal or neurological injuries. Ankle sprains are very common and mainly involve the overstretching or tearing of tendons and ligaments. There are over 23,000 cases estimated to occur in the United States (US) every day [3]. In New Zealand (NZ), 230,288 new claims and 278,657 active claims related to musculoskeletal ankle injuries were made to the Accident Compensation Corporation from July of 2009 to June of 2014 [4]. Neurological injuries may also cause the ankle to operate inefficiently – a typical example is drop foot following stroke. Based on a report from the American Heart Association, approximately 795,000 people experience stroke in the US each year [5]. There are an estimated 60,000 stroke survivors living in NZ, and many of them have mobility impairments [6].

The ankle-foot complex mainly consists of tibia, fibula, talus and calcaneus [2], as shown in Figure 1.1(a). The tibia and fibula are considered as a unit to simplify the motions of the ankle-foot complex. The ankle joint is the articulation between the tibia-fibula unit and the talus [2]. The subtalar joint, also known as the talocalcaneal joint, locates between the talus and the calcaneus [2]. The term “ankle” used in this study encompasses both the ankle and subtalar joint, is primarily rotational, and is often described by rotations on three mutually perpendicular anatomical planes. The plane which distinguishes the left and right sides of the body is termed the sagittal plane, where ankle plantarflexion and dorsiflexion exist and occur about the X-axis. Ankle inversion and eversion exist and occur about the Y-axis in the frontal plane that divides the body into front and back halves. The transverse plane is perpendicular to the sagittal and the front planes, and divides the body into top and bottom portions, where ankle adduction and
abduction exist and occur about the Z-axis. Rotational motion of the foot in the sagittal plane is termed plantarflexion when the toes are pushed further away from the head and dorsiflexion in the opposing direction. Inversion is used to describe the rotation of the foot on the frontal plane where the inner or medial side of the foot is raised upwards, with eversion being its complementary motion. Adduction is used to describe rotational motion on the transverse plane which moves the toes towards the centre of the body while movement in the contrary direction is termed abduction. The above mentioned ankle motions are also presented in Figure 1.1(b). In this research, as Wu et al. [7] recommended, ankle dorsiflexion, adduction and inversion are defined as positive while plantarflexion, abduction and eversion as negative.

![Figure 1.1](image_url)

**Figure 1.1**: (a) Anatomy of the human ankle foot complex. (b) Rotational motions of the human ankle-foot complex as a three-DOF revolute joint, X-axis in red, Y-axis in blue and Z-axis in pink.

### 1.2. Ankle Rehabilitation Devices

A general rehabilitation program for musculoskeletal ankle injuries involves an acute phase treatment by immobilization and a sub-acute phase with passive/active range-of-motion (ROM) and muscle strengthening exercises. Stroke neuro-rehabilitation usually includes repetitive, intensive, and task-specific movement training to promote motor recovery [8]. However, rehabilitation exercises for both musculoskeletal and neurological ankle disabilities require cooperative and intensive efforts from therapists and patients over prolonged sessions [9].

Robot-assisted ankle rehabilitation solutions, as therapeutic adjuncts to facilitate clinical practice, have been actively researched during the past few decades. There are two types of ankle rehabilitation devices. In one group are the wearable exoskeletons aimed at ankle rehabilitation during gait training, such as the MIT Anklebot developed by Roy et al. [10] and the bio-inspired soft ankle robotic device developed by Park et al. [11]. The other group consists of platform-based robots that conduct only ankle training [12-16]. Platform-based ankle robots
can be further divided into single-degree of freedom (DOF) and multi-DOF devices. The single-DOF device is mostly actuated by a rotating motor [16], while multi-DOF robots are usually based on parallel platforms actuated by linear actuators.

Further, and most significantly, parallel ankle rehabilitation robots can be classified in terms of the mechanical structure, although they possess the same advantages as parallel manipulators. Specifically, robot systems actuated from below [14, 15] require varying positions of the ankle joint and synergic movement of the lower limb from the patient during the training due to the fact that the rotation centre of the robot does not coincide with the ankle joint. In contrast, the patient can keep his/her ankle stationary and fully relaxed on devices actuated from above [12, 13]. From this point of view, the parallel manipulator wherein the rotation centre of the robot coincides with the ankle joint is more suitable for ankle rehabilitation. To distinguish the structural differences, I name the actuated-from-below parallel ankle robot B-PAR, while the actuated-from-above parallel ankle robot is called the A-PAR. To make the description clear, a hierarchy chart is presented in Figure 1.2 with a variety of ankle rehabilitation robots, and examples are provided in Figure 1.3. This research will focus on the development of a novel A-PAR that is considered to be the ankle rehabilitation robot with the most potential.

Figure 1.2: Hierarchy chart of existing ankle rehabilitation robots.

Figure 1.3: Existing ankle rehabilitation robots. (a) The MIT Anklebot [16], (b) an intelligent single-DOF ankle stretching device [10], (c) the B-PAR [14], and (d) the A-PAR [12].
1.3. Control Strategies of Rehabilitation Training

Passive and active training both play significant roles in the rehabilitation of injured ankles. Passive exercise is usually needed when the patient can hardly move his/her foot, and can be accomplished by a position control scheme that can guide the injured ankle to move along the predefined trajectory. Following passive exercises, active training is mostly conducted to make the patient fully regain his/her ankle range of motion (ROM) and strength, and can be implemented using an interaction controller based on impedance or admittance techniques.

A variety of robot-assisted ankle training strategies have been implemented and evaluated through clinical trials [17]. A typical example of a single-DOF device is the intelligent ankle stretching device developed by Zhang et al. [16] for use on neurologically-impaired patients. Clinical evidence has demonstrated that the combined passive stretching and active movement is effective for children with cerebral palsy [18] and stroke patients [19]. For parallel ankle rehabilitation robots, some advanced control strategies have been implemented to conduct passive and active training. Two typical B-PARs exist. One is the Rutgers Ankle based on a six-DOF Stewart-Gough platform [15], and the other one is a parallel manipulator that is actuated by three customised cable-driven linear electric actuators [14]. Both robots have been programmed to conduct interaction training [20, 21]. Deutsch et al. [20] allowed the patient to use the robot to interact with virtual-reality simulations, while Saglia et al. [21] adopted an admittance controller combined with a position control scheme for patient-active exercises. Their potential for active training has been demonstrated on either stroke patients or healthy subjects.

A-PARs, considered to be suitable for ankle rehabilitation, are still in their early stage in terms of robotic design and control scheme. Only two devices have been constructed for ankle rehabilitation, with one developed by Tsoi et al. [13] and the other by Jamwal et al. [12]. The former is a three-DOF parallel ankle rehabilitation robot actuated by four brushed DC motor-driven linear actuators, and the interaction control has been implemented by allowing for the adaptation of robot stiffness. The latter, also a parallel manipulator, employs four pneumatic artificial muscles (PAMs) for actuation. When an adaptive fuzzy-logic controller was implemented on the robot during testing of a healthy subject it led to satisfactory trajectory tracking in the presence of unpredicted patient-robot interaction, and the adaptation scheme was adopted for the length control of the PAMs rather than the patient-robot interaction. This
information is summarised in Figure 1.4 where the control schemes of a future A-PAR are also defined.

![Diagram of A-PARs with potential training modes and their controllers](image)

**Figure 1.4:** Future development of the A-PAR in terms of training strategy.

### 1.4. Research Motivations and Objectives

There are three main motivations in adopting robot-assisted rehabilitation techniques for the treatment of ankle injuries. First, the use of robot-assisted ankle rehabilitation techniques can reduce the physical workload of therapists. Secondly, the biomechanical information of the injured ankle can be accurately measured and estimated during robotic training, which then allows for the objective evaluation of the ankle disability. Lastly, patient engagement can be encouraged by implementing advanced interaction training schemes or virtual-reality training environment on the robot.

While robot-assisted rehabilitation techniques show potential for the treatment of ankle injuries, their effectiveness for clinical applications is impeded due to a range of limitations. To identify the research gap, a systematic review [17] was conducted focusing on the effectiveness of robot-assisted therapy for ankle rehabilitation. These limitations include kinematic incompatibility that requires varying ankle positions during the robotic training, stiff actuation, rehabilitation exercises with less than three DOFs, and the lack of real-time biomechanical assessment and adaptive interaction training schemes. Another review was conducted to identify the optimal real-time technique for three-dimensional ankle assessment during the robotic training [22].

To enhance the effectiveness of robot-assisted rehabilitation as an adjunct to traditional physiotherapy interventions for ankle-impaired subjects, this research develops an intrinsically-compliant ankle rehabilitation robot with the integration of real-time biomechanical assessment to facilitate the implementation of advanced interaction training schemes. From the viewpoint of functionality, this device is named the ankle assessment and rehabilitation robot (AARR). Three main objectives of this research are: 1) to develop an intrinsically-compliant AARR; 2) to develop and validate sensor-based and model-based ankle assessment techniques for providing biomechanical measurements of the injured ankle; and 3) to implement advanced
interaction control schemes to allow for adaptive robotic training with enhanced safety and rehabilitation efficacy. To summarise, this AARR will be able to conduct adaptive passive and active training according to real-time ankle assessment, which requires the robot to guide the motion of the injured ankle, and simultaneously comply with real-time patient-robot interaction and movement intention for enhanced safety.

In general, the clinical relevance of the developed AARR mainly include:

- Reducing the workload of both patients and therapists;
- Allowing objective evaluation from sensors built in the robot and computational models;
- Engaging patients in rehabilitation exercises in a highly motivating manner;
- Enhanced training comfort and practicality to encourage patients using the robot more.

1.5. Thesis Outline

This thesis details the work carried out in this research to meet the above objectives, as shown in Figure 1.5. Specifically, Chapter 4 implements the first objective with the development of the AARR that was proposed based on the reviews in Chapters 2 and 3. The second objective is achieved in Chapters 5, 6 and 7. The sensor-based assessment technique was developed in Chapter 5, involving a rotary sensor to measure real-time ankle orientation and a six-axis load cell for measuring human-robot interaction torque. Chapters 6 and 7 developed model-based assessment techniques (two computational ankle models) for estimating ligament kinematics and passive torque of the ankle joint. These two models are driven by real-time ankle orientation that can be measured by three magnetic rotary sensors built in the AARR. The last major objective of this research is the implementation of advanced interaction training strategies on the AARR in Chapters 8 and 9.

In general, this research develops an intrinsically-compliant AARR with the integration of real-time ankle assessment and the implementation of advanced interaction control strategies. The materials of this thesis have been either published or submitted for possible publication (Appendix F). The work is organised as follows:

Chapter 1 provides an introduction to the mechanical design and control strategies of existing ankle rehabilitation robots. The research objectives are detailed and the thesis outline is presented.
Chapter 2 provides a systematic literature review regarding the effectiveness of robot-assisted ankle therapy. This review analyses and compares 29 studies with a total of 164 patients and 24 healthy subjects involved. While most rehabilitation devices and control strategies have been demonstrated to be suitable and effective for individuals with a variety of ankle disabilities, the most effective ankle rehabilitation robot and control algorithm cannot be determined due to the lack of universal evaluation criteria. However, a compliant multi-DOF ankle rehabilitation device is encouraged. It should be able to conduct both passive and active training through adaptive interaction control algorithms with the consideration of real-time biomechanical ankle assessment. This chapter contains materials that have been published in [17] as:


Chapter 3 provides a critical literature review of studies that have investigated ankle assessment techniques in conjunction with robot-assisted therapy. This review includes 76 publications, of which 29 studies with 465 subjects introduced techniques that have the potential to be integrated with real-time monitoring systems. It was found that most quantitative techniques are reliable in measuring ankle kinematics and dynamics, however, they are mostly only available in the sagittal plane and few studies have investigated ankle assessment in a three-dimensional space where actual ankle motions occur. This chapter contains materials that have been published in [22] as:

Chapter 4 presents the development of a novel intrinsically-compliant AARR that was proposed based on the findings from two reviews in Chapters 2 and 3. This robot is actuated in parallel by four Festo fluidic muscles (FFMs) for three rotational DOFs that are for ankle dorsiflexion/plantarflexion, inversion/eversion and adduction/abduction, respectively. This chapter contains materials that have been submitted for part of the possible publication as:


Chapter 5 develops an assessment technique via sensors for measuring ankle orientation and torque/stiffness for use in robot-assisted therapy. Validity and reliability was experimentally demonstrated on nine healthy subjects, which provides the basis for its application on the AARR. This chapter contains materials that have been published in [23] as:


Chapter 6 presents the development and validation of the Non-PRA-Model whose rotation axes are not perpendicular. This model was proposed to estimate kinematics of ankle ligaments and validated by comparing modelling results with published data. This kinematic information can be used for the optimisation of the robotic training trajectory, and also provides the basis for its further development in predicting passive ankle torque. This chapter contains materials that have been published in [24] as:


Chapter 7 presents the development and validation of the PRA-Model whose rotation axes are perpendicular. This model was proposed to estimate passive ankle torque for a specific subject on the basis of the Non-PRA-Model (Chapter 6). The PRA-Model was structured according to the AARR design, and simulates the actual foot movement during the robotic training. Its effectiveness in estimating passive ankle torque was validated by comparing it with published data and experimentally on nine subjects. The biomechanical assessment of the ankle joint could
be used to facilitate the implementation of advanced interaction schemes with enhanced training safety and efficacy. This chapter contains materials that have been published as [25] in IEEE Transactions on Biomedical Engineering as:


Chapter 8 proposes an interaction training strategy on the AARR via an adaptive predefined trajectory. The proposed interaction training is implemented by an intelligent algorithm as a high-level controller for trajectory adaptation and a cascade low-level position controller for trajectory tracking. The adaptation of the predefined trajectory is determined based on the movement intention of the patient, which makes the training safe and effective by encouraging patient engagement. By this interaction training scheme, the AARR can deliver adaptive passive and active robotic training, which has been experimentally validated with a sprained ankle. This chapter contains materials that have been submitted for part of the possible publication as:


Chapter 9 proposes another interaction training strategy on the AARR according to real-time patient-robot interaction torque. This interaction training scheme is implemented via an admittance law as a high-level controller for trajectory adaptation and an FFM-length-based low-level position controller for trajectory tracking. The adaptation of the parameters of the admittance controller is determined based on real-time patient-robot interaction torque, joint position and passive torque. Using this interaction training scheme, the AARR can also deliver adaptive passive and active robotic training, which has been experimentally validated with a sprained ankle. This chapter contains materials that have been submitted for possible publication as:

Chapter 10 summarises the main conclusions of this research and provides recommendations for its further development.

1.6. Chapter Summary

This chapter has highlighted the main motivations of this research through an overview of ankle injuries, anatomy, existing rehabilitation robots, and control strategies. The main incentives for the use of AARR for the treatment of ankle injuries include: 1) reducing physical workload of therapists; 2) extracting biomechanical measurements to allow for the objective assessment of the ankle disability level and facilitation of advanced interaction control schemes; 3) encouraging patient engagement to improve the rehabilitation efficacy; and 4) improving training comfort and practicality to encourage patients using the robot more.

The AARR was proposed based on the findings from two literature reviews (Chapters 2 and 3) that investigate existing ankle rehabilitation devices and robot-assisted assessment techniques, respectively. The proposed AARR, considered to be an ankle rehabilitation platform with great potential, is characterised by intrinsic compliance, a compatible robot structure that allows the ankle to be stationary during the robotic training, three DOFs, and real-time biomechanical assessment to allow for safe and effective interaction training. Three major research objectives include: 1) developing the AARR; 2) developing assessment techniques via sensors and computational models to extract biomechanical information of the human ankle; and 3) implementing interaction control strategies to allow for adaptive robotic training.
Chapter 2. A Review of Robot-Assisted Ankle Rehabilitation Techniques

This chapter presents a systematic review investigating the effectiveness of existing robot-assisted rehabilitation techniques for the treatment of musculoskeletal or neurological ankle injuries. This review was conducted to identify the research gap to further the development of existing ankle rehabilitation devices in terms of mechanical design and control strategy. While the most effective robot-assisted ankle rehabilitation technique cannot be determined, some features that can contribute to enhanced training safety and efficacy have been identified. The materials presented in this chapter has been published as [17] in Journal of NeuroEngineering and Rehabilitation.

This published review goes as follows. Thirteen electronic databases of articles published from January, 1980 to June, 2012 were searched using keywords ‘ankle*’, ‘robot*’, ‘rehabilitat*’ or ‘treat*’, and a free search in Google Scholar based on effects of ankle rehabilitation robots was also conducted. References listed in relevant publications were further screened. Twenty-nine studies met the inclusion criteria and a total of 164 patients and 24 healthy subjects participated in these trials. Ankle performance and gait function were the main outcome measures used to assess the therapeutic effects of robot-assisted ankle rehabilitation. The protocols and therapy treatments were varied, which made comparison among different studies difficult or impossible. Few comparative trials were conducted among different devices or control strategies. Moreover, the majority of study designs met levels of evidence that were no higher than American Academy for Cerebral Palsy and Developmental Medicine (AACPDM) level IV. Only one study used a Randomized Control Trial (RCT) approach with the evidence level being II. The review concluded that all selected studies showed improvements in terms of ankle performance or gait function after a period of robot-assisted rehabilitation training. However, the most effective robot-assisted ankle rehabilitation technique cannot be determined due to the lack of universal evaluation criteria. Future research into the effects of robot-assisted ankle rehabilitation should be carried out based on universal evaluation criteria, which could determine the most effective method of intervention. It is also essential to conduct trials to analyse the differences among different devices or control strategies.
Chapter 2. A Review of Robot-Assisted Ankle Rehabilitation Techniques

2.1. Introduction

The human ankle joint is a very complex bony structure in the human skeleton and plays a significant role in maintaining body balance during ambulation [2]. In fact, the ankle is the most common site of sprain injuries in the human body, with over 23,000 cases estimated to occur per day in the United States [3] and about 100,000 emergency department presentations per year in Australia [26]. In New Zealand, more than 82,000 new claims and 17,200 ongoing claims related to ankle injuries were made to the Accident Compensation Corporation (ACC) in the 2000/01 year, costing an estimated 31.8 million New Zealand dollars and making ankle related claims the fourth biggest cost to ACC. Additionally, neurological injuries like stroke, traumatic brain and spinal cord injuries are also leading causes for ankle disabilities. In the United States, at least 750,000 incident and recurrent strokes occurred with the prevalence rate being about 200 to 300 patients per 100,000 inhabitants in 1995 [28]. However, the biggest effect on patients with ankle disabilities and their family members is usually a result of long-term impairment, limitation of activities and reduced participation.

Traditionally ankle injuries are rehabilitated via physiotherapy and however evidence suggests that without sufficient rehabilitation: 44% of people will have future problems [29, 30]; ambulation is markedly compromised; re-injury prevalence is high; and approximately 38% of people will have recurrent activity limitations affecting their function [31]. Furthermore, during a rehabilitation treatment, cooperative and intensive efforts of therapists and patients are required over prolonged sessions [9]. Robotics technology can provide an overdue transformation of rehabilitation clinics from labour-intensive operations to technology-assisted operations as well as a rich stream of data that can facilitate patient diagnosis, customisation of the therapy, and maintenance of patient records (at the clinic and at home) [32]. Thus, robotic devices have been developed for human ankle rehabilitation by some research groups [33-35]. Currently, there are mainly two kinds of robot-assisted ankle rehabilitation devices: those that are wearable devices mainly aiming at improving ankle performance during gait and those that are platform-based devices focusing solely on improvement of ankle performance [36-38].

However, little is known about the effects of robot-assisted therapy on ankle recovery from disabilities and the intervention most effective for a specific case. The purpose of this systematic review was to provide a comprehensive investigation and examination of published evidence on the effectiveness of robot-assisted therapy used to help human ankles recover from musculoskeletal or neurological injuries.
Chapter 2. A Review of Robot-Assisted Ankle Rehabilitation Techniques

2.2. Methods

2.2.1 Search Strategy

The literature search was limited to English-language articles (including journal articles, extended abstracts, and conference proceedings) published between January 1980 and June 2012 in the following electronic databases recommended by a librarian in The University of Auckland: PubMed, EMBASE (Excerpta Medical database), MEDLINE (OvidSP), CDSR (Cochrane database of systematic reviews), Web of Science, Scopus, Compendex, IEEE Xplore, ScienceDirect, Wiley Online Library, Digital Dissertations, Academic Search Premier, SpringerLink. The electronic search terms were “‘Ankle*’ AND ‘Robot*’ AND (‘rehabilitat*’ or ‘treat*’).” A free search in Google Scholar was also conducted and valuable references listed in relevant publications were screened, which made the search as systematic and complete as possible.

Three reviewers are Mingming Zhang (MZ), T. Claire Davies (TCD) and Shane Xie (SX). The two primary reviewers (MZ, TCD) were a PhD student with expertise in mechatronics system and a senior lecturer with expertise in assistive technologies of rehabilitation, respectively. The third reviewer (SX) was a professor with expertise in medical robotics, assistive and rehabilitation robots. Two of reviewers (TCD, SX) hold doctorates in their respective fields. The initial search was conducted in thirteen electronic databases by one primary reviewer (MZ) and the total number of articles identified was 686. To start with, one primary reviewer (MZ) assessed all the titles for eligibility using the screening criteria described below. MZ independently assessed all abstracts after the first round screening. Abstracts considered as meeting the inclusion criteria by MZ were automatically included in the full review. Otherwise, they were excluded. Studies one primary reviewer (MZ) was not sure whether to include or not were discussed by two primary reviewers. Discrepancies between the two primary reviewers were resolved through participation from the third reviewer and discussion among these three reviewers. A search in Google Scholar was conducted by one primary reviewer (MZ) based on therapeutic effects of ankle rehabilitation robots and the references of the included papers were also screened by MZ for any additional studies. These studies newly selected were also reviewed.

Twenty-nine studies were included in the final review. Data extraction was then undertaken. The review concentrated on evidence based therapeutic effects of robot-assisted ankle rehabilitation, design types, levels of evidence and quality levels of the articles. The schematic overview of selection process with search results is shown in Figure 2.1.
2.2.2 Inclusion and Exclusion Criteria

Robotics was defined as an application of electronic and computerised control systems to mechanical devices to perform human functions [39]. The American Heritage Dictionary defined a robot as a mechanical device that sometimes resembled a human and was capable of performing a variety of often complex human tasks on command or by being programmed in advance, or a machine or device that operated automatically or by remote control [40].

All trials assessing the clinical outcomes of robot-assisted ankle rehabilitation training were included. These included participants who sustained any grade of ankle disabilities caused by musculoskeletal or neurological injuries. Both male and female participants from athletic and non-athletic populations were included to allow the generalisation of results to different populations. Papers involving platform-based ankle rehabilitation robots or wearable ankle rehabilitation robots were included.
However, studies were excluded if participants underwent ankle surgeries or wore ankle prostheses. Animal based trials assessing humans and healthy subjects based trials assessing patients were also excluded. Studies focusing on the whole lower limb but not related to ankle recovery were excluded as well. Only English articles published in peer-reviewed journals or published as conference papers or abstracts were included.

2.2.3 Organization of Evidence

The data extraction form used for this study was the critical review form for quantitative studies developed by the Occupational Therapy Evidence-Based Practice Group at McMaster University, Hamilton, Ontario, Canada provided guidelines for the reviewer to summarise information about study purpose, background literature, design category, sample size, outcome measures, treatment interventions, results and conclusions [41]. Levels of evidence for the selected studies were assessed according to guidelines from AACPDM [43] and were evaluated by the two primary reviewers.

2.3. Results

The search results are summarised in Figure 2.1. Sixty-three abstracts appeared to meet the inclusion criteria, and the associated full articles were obtained through downloading from electronic databases. Moreover, ten papers that appeared to meet the inclusion criteria were obtained from Google Scholar. Sixteen papers were excluded because they attempted to assess the effects of ankle rehabilitation robots by recruiting healthy subjects [10, 44-58]. A further two papers were excluded because they validated the feasibility of robotic devices based on only simulations [59, 60]. Eighteen papers solely discussing the design of ankle rehabilitation robots without any application were also excluded [13, 33-35, 61-74]. Four papers descriptively reviewed a variety of rehabilitation robots of the lower extremity [36-38, 75] and were excluded. A conference paper [76] with its main content included in a journal article [35] was excluded; two studies [77] and [18] involved the same patients and the former was excluded. Two studies [78] and [79] which focused on a novel approach of walking therapy were excluded. Another two studies [80, 81] concerning the lower limb recovery but no ankle were excluded. Eventually, 29 papers that still met the inclusion criteria were selected for this systematic review [16, 18, 20, 82-107]. These studies were carried out to evaluate the therapeutic effects of ankle rehabilitation robots through participation from related patients.
2.3.1 Study Characteristics

A total of 29 original papers, with data from 164 patients and 24 healthy participants used as control participants in different studies, met the inclusion criteria after the eventual discussion among three reviewers. 16 of these studies focused on evaluating the therapeutic effects of platform-based ankle rehabilitation devices, as shown in Table 2.1. Another 13 papers discussed the therapeutic effects of wearable ankle rehabilitation robots (three studies involved the Anklebot, one was for in-bed ankle rehabilitation robot, six studies focused on ankle-foot orthoses (AFOs) and three papers involved a robotic-assisted locomotor training device partly focused on ankle rehabilitation), as shown in Table 2.2. The 164 patients spanned the ages seven to 81 while the age of 30 patients in six papers [20, 84, 87, 95, 106, 108] was not stated. The majority of participants sustained ankle disabilities mainly caused by musculoskeletal injuries and stroke. Additionally, traumatic or non-traumatic, brain, spinal cord injuries, Cauda Equine syndrome and Guillain-Barre syndrome were also key factors for ankle disabilities.

Studies meeting the inclusion criteria were quantitative. Study designs included one RCT with only 18 participants, one single case design, nine before-after designs, six single case series and six case study designs [41]. The remaining six studies adopted combined designs. Specifically, two studies [88, 99] adopted case control design and single case design, two studies [16, 103] applied case control design and before-after design, Ward et al. [105] adopted a combination of single case series and before-after design, Deutsch et al. [100] applied both single case design and before-after design. Deutsch et al. [101] presented three case reports. RCTs’ essential feature is a set of clients/subjects are identified and then randomly allocated to two or more different treatment groups; for single case designs, changes in accessibility using several types of technology are compared with baseline data from the same individual using intervention sequences such as ABA, ABAB, ABACA, or ABCD; single case series designs involve more than one subject/client but evaluate pre- and post-treatment for that individual; before-after designs allow the evaluator to collect information about the initial status of a group of clients in terms of the outcomes of interest and then collect information again after treatment is received; in case control designs characteristic or situation of interest is compared with a control group of people who are similar in age, gender and background; case study designs involved task completion exercises without control group to provide descriptive information about the relationship between a particular treatment and an outcome of interest [43].
Table 2.1: Reviewed studies of platform-based ankle rehabilitation robots.

<table>
<thead>
<tr>
<th>Studies</th>
<th>Design</th>
<th>Subjects</th>
<th>Characteristics</th>
<th>Age (years)</th>
<th>Intervention</th>
<th>Measures</th>
<th>Outcomes</th>
<th>Assumptions</th>
</tr>
</thead>
<tbody>
<tr>
<td>M. Girone, 2000 [98]</td>
<td>Level V, Case Study</td>
<td>N=4</td>
<td>2 patients exhibited hypermobility secondary to chronic ankle instability and the other 2 presented with hypomobility as the sequelae of fractures</td>
<td>26-81</td>
<td>Rutgers Ankle prototype</td>
<td>Displacement and torque</td>
<td>The displacement of the uninvolved leg was comparable to normal ROM at the ankle with five degrees of dorsiflexion to 45 degrees of plantarflexion and that of the involved limb reflects a loss of ROM of –10 degrees of dorsiflexion and 28 degrees of plantarflexion; The maximum torque generated by the uninvolved limb was much larger (4 ft·lbs. for dorsiflexion and 8 ft·lbs. for plantarflexion) than that generated by the involved limb (0.5 ft·lbs. for dorsiflexion and 4 ft·lbs. for plantarflexion)</td>
<td>Increase in ROM and ankle torque can result in improvements in ankle performance and gait</td>
</tr>
<tr>
<td>J. E. Deutsch, 2001 [101]</td>
<td>Level IV, Single Case Series</td>
<td>N=3</td>
<td>Musculoskeletal ankle injuries</td>
<td>14-56</td>
<td>Rutgers ankle system with a 3-D piloting of an airplane</td>
<td>ROM, torque generation capacity and ankle mechanical work</td>
<td>Task accuracy improved to 100% for Case 1; a fivefold increase in ankle power output for Case 2 and a three-fold increase for Case 3; both Case 2 and Case 3 reached 100% task accuracy</td>
<td>Improved task accuracy means improved ankle performance and gait</td>
</tr>
<tr>
<td>J. E. Deutsch, 2001[100]</td>
<td>Level IV, Before-After, Single Case</td>
<td>N=1</td>
<td>A left cerebral vascular accident</td>
<td>69</td>
<td>Rutgers ankle system with a 3-D piloting of an airplane</td>
<td>Ankle and foot mobility, force generation, coordination and the ability to walk and climb stairs</td>
<td>Laboratory functional improvements correlate with activities of daily life</td>
<td></td>
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<tr>
<td>Author(s)</td>
<td>Series Type</td>
<td>N</td>
<td>Condition Description</td>
<td>Method</td>
<td>Outcome</td>
<td>Notes</td>
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<tr>
<td>R. F. Boian, 2002</td>
<td>Level IV, Single Case Series</td>
<td>N=3</td>
<td>3 patients with post-stroke</td>
<td>The Rutgers Ankle with two video games</td>
<td>Increase in power generation for all motions and walking endurance increase for one patient</td>
<td>Increase in power generation and walking endurance means improved ankle performance and gait</td>
<td></td>
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<tr>
<td>R. F. Boian, 2003</td>
<td>Level IV, Single Case Series</td>
<td>N=3</td>
<td>2 patients had normal sensation and the third had a decrease with 8/12 on the FM lower extremity sensory score</td>
<td>The second version of virtual reality-based ankle rehabilitation system</td>
<td>Muscle strength</td>
<td>Increase in ankle muscle strength means improved ankle performance</td>
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<tr>
<td>J. E. Deutsch, 2004</td>
<td>Level IV, Single Case Series</td>
<td>N=6</td>
<td>Post-stroke</td>
<td>A robotic device (the Rutgers Ankle was the input to the virtual environment)</td>
<td>Gait and elevation speed</td>
<td>Gait speed increased 11% (p=.08) and elevation time decreased 14% (p=.05); gait endurance increased 11%; gait and elevation speed improved from 0 to 44% and 3 to 33% respectively</td>
<td>Improved elevation speed means improvements in ankle performance and gait</td>
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<tr>
<td>R. W. Selles, 2005</td>
<td>Level IV, Single Case Series</td>
<td>N=10</td>
<td>spasticity and/or contracture after stroke</td>
<td>A feedback-controlled and programmed stretching device</td>
<td>Significant improvements were found in the passive ROM, maximum voluntary contraction, ankle stiffness, and comfortable walking speed</td>
<td>Improved ROM, muscle strength, joint stiffness, joint viscous damping, reflex excitability, walking speed and subjective experiences means improved ankle performance and gait and all these correlate with activities of daily life</td>
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</tr>
<tr>
<td>D. Cioi, 2011</td>
<td>Level IV, Single Case (ABA)</td>
<td>N=1</td>
<td>A child with mild ataxic cerebral palsy (CP)</td>
<td>Rutgers Ankle</td>
<td>Strength, motor control, gait function, overall function and quality of life improved obviously</td>
<td>Laboratory functional improvements correlate with activities of daily life</td>
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<tr>
<td>Study</td>
<td>Level</td>
<td>Design</td>
<td>N</td>
<td>Participants</td>
<td>Measures</td>
<td>Results</td>
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<tr>
<td>G. C. Burdea, 2012</td>
<td>V</td>
<td>Case Study</td>
<td>3</td>
<td>3 male children with CP</td>
<td>Rutgers Ankle Impairment, function, quality of Life and game performance</td>
<td>Strength, motor control, gait function, quality of life and game performance improved obviously</td>
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<tr>
<td>L-Q. Zhang, 2002 [16]</td>
<td>IV</td>
<td>Before-After, Case Control</td>
<td>9</td>
<td>5 healthy subjects and 4 chronic stroke patients with ankle contracture and/or spasticity</td>
<td>All subjects (36.8 ±12.8), 4 stroke patients (53.2±7.9)</td>
<td>A custom-designed joint stretching device ROM, joint stiffness, viscous damping and reflex excitability</td>
<td>The passive and active ROM of the ankle joint increased; joint stiffness and viscosity were reduced; reductions in reflex excitability were also observed</td>
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<tr>
<td>J. E. Deutsch, 2007 [20]</td>
<td>IV</td>
<td>Before-After (Group performance)</td>
<td>6</td>
<td>Post-stroke</td>
<td>Rutgers Ankle prototype robot with VR</td>
<td>Accuracy of ankle movement, exercise duration, training efficiency, mechanical power of ankle and number of repetitions ROM and pressure distribution</td>
<td>All measures improved in the first three weeks and did not decrease during the transition</td>
<td></td>
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<tr>
<td>K. Homma, 2007 [88]</td>
<td>IV</td>
<td>Case Control, Single Case</td>
<td>5</td>
<td>4 healthy subjects and a male with hemiplegia</td>
<td>A passive exercise device for ankle dorsiflexion and plantarflexion</td>
<td>These improvements were within the margin of the measuring error</td>
<td>Improved ROM means improved ankle performance</td>
<td></td>
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<tr>
<td>A. Mirelman, 2008 [92]</td>
<td>II</td>
<td>RCT</td>
<td>18</td>
<td>Chronic hemiparesis after stroke</td>
<td>VR Group: (61.8±9.94, 41–75); Robotic Group: (61±8.32, 45–71)</td>
<td>Velocity and distance walked</td>
<td>Greater changes in velocity and distance walked were demonstrated for the group trained with the robotic device coupled with virtual reality than Improved velocity and distance walked mean improved ankle performance and gait</td>
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<tr>
<td>Author</td>
<td>Level</td>
<td>Study Design</td>
<td>N</td>
<td>Population</td>
<td>Intervention/Equipment</td>
<td>Outcomes</td>
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<td>P. Cordo, 2009 [86]</td>
<td>Level IV</td>
<td>Before-After</td>
<td>11</td>
<td>Patients with post-stroke and severe motor disability of the lower extremity</td>
<td>AMES treatment device for ankles</td>
<td>Strength, joint position and motor function increased in most ankles; joint position improved in all ankles; motor function improved significantly</td>
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<tr>
<td>Y.-N. Wu, 2011 [18]</td>
<td>Level IV</td>
<td>Before-After</td>
<td>12</td>
<td>Children with CP 5-15 and mean age is 8 years 6 months</td>
<td>A portable rehabilitation robot with computer game</td>
<td>Improvements in dorsiflexion PROM (P=.002), AROM (P=.02), and dorsiflexor muscle strength (P=.001); spasticity of the ankle musculature was reduced (P=.01); selective motor control improved (P=.005); functionally, participants improved balance (P=.0025) and increased walking distance within 6 minutes (P=.025)</td>
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<tr>
<td>G. Waldman, 2011 [104]</td>
<td>Level IV</td>
<td>Before-After</td>
<td>8</td>
<td>Stroke survivors 50.4±8.9</td>
<td>A portable ankle rehabilitation robot</td>
<td>Active dorsiflexion range and dorsiflexor muscle strength improved (p=0.001 and 0.01, respectively) as well as the average MAS, STREAM, and Berg Balance mean improved ankle performance and gait</td>
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</table>
Table 2.2: Reviewed studies of wearable ankle rehabilitation robots.

<table>
<thead>
<tr>
<th>Studies</th>
<th>Design</th>
<th>Subjects</th>
<th>Characteristics</th>
<th>Age (years)</th>
<th>Intervention</th>
<th>Measures</th>
<th>Outcomes</th>
<th>Assumptions</th>
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<tr>
<td>single subject research designs (SSRD)</td>
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<tr>
<td>J. Furusho,</td>
<td>Level V,</td>
<td>N=1</td>
<td>A man (case: right ankle flaccid paralysis; height: 157cm; weight: 44kg)</td>
<td>59</td>
<td>An AFO with MR brake</td>
<td>Ankle angle, reaction force and a bending moment</td>
<td>In swing phase, the subject can maintain the dorsal flexion and prevent the drop foot; the subject can contact ground at heel; at contact ground, GRF doesn't lack smoothness; maximal value of a bending moment with control is larger than one without control; walking cycle is shorter than one without control</td>
<td>Preventing drop foot in swing phase and slap foot at heel strike can result in gait improvement</td>
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<tr>
<td>2007 [89]</td>
<td>Case Study</td>
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<tr>
<td>S. Tanida,</td>
<td>Level V,</td>
<td>N=1</td>
<td>A patient of the Guillain-Barre syndrome (183cm and 83.1kg)</td>
<td>34</td>
<td>I-AFO</td>
<td>Ankle joint angle and reaction force</td>
<td>The foot clearance in the swing phase was kept effectively by preventing the drop foot and the initial contact occurred in the primary stance phase normally</td>
<td>Preventing drop foot effectively in swing phase means good ankle joint control and performance</td>
</tr>
<tr>
<td>2009 [97]</td>
<td>Case Study</td>
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<tr>
<td>Y. Ren,</td>
<td>Level V,</td>
<td>N=4</td>
<td>Acute post-stroke</td>
<td>Not stated</td>
<td>A wearable robot for in-bed acute stroke rehabilitation</td>
<td>Passive and active biomechanical properties</td>
<td>Changes of passive and active biomechanical properties can be detected</td>
<td>These changes contribute to ankle performance and gait</td>
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<tr>
<td>2011 [87]</td>
<td>Case Study</td>
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<tr>
<td>L. W. Forrester,</td>
<td>Level IV,</td>
<td>N=8</td>
<td>Chronic stroke</td>
<td>62.4±10.4</td>
<td>A visually guided, impedance controlled, ankle robotic intervention</td>
<td>Ankle ROM, strength, motor control, and overground gait function</td>
<td>Increased target success, faster and smoother movements, walking velocity whereas durations of paretic single support increased and double support decreased</td>
<td>Improved target success, movement and walking velocity contribute to ankle performance and they correlate with activities of daily life</td>
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<tr>
<td>2011 [85]</td>
<td>Single Case Series</td>
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<tr>
<td>K. McGehrin,</td>
<td>Level V,</td>
<td>N=2</td>
<td>Sub-acute stroke</td>
<td>Not stated</td>
<td>A single session of Ankletbot</td>
<td>Ankle motor control</td>
<td>Increased targeting accuracy, faster speed and smoother movements.</td>
<td>Improved target success, movement and</td>
</tr>
<tr>
<td>2012 [84]</td>
<td>Case Study</td>
<td></td>
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<tr>
<td>Group Research Designs (CRD)</td>
<td>Level IV,</td>
<td>N=5</td>
<td>2 drop-foot subjects and 3 normal participants</td>
<td>AAFO</td>
<td>Occurrence of slap foot and swing phase ankle kinematics</td>
<td>The occurrence of slap foot was reduced and swing phase ankle kinematics more closely resembled normal compared to zero and constant control schemes</td>
<td>Decreased slap foot means improved ankle performance and gait</td>
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<tr>
<td>J. A. Blaya, 2004 [82]</td>
<td>Before-After</td>
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<tr>
<td>M. M. Mirbagheri, 2005 [106]</td>
<td>Level IV, Before-After</td>
<td>N=5</td>
<td>Incomplete SCI</td>
<td>Not stated</td>
<td>Robotic-Assisted Locomotor Training</td>
<td>Reflex stiffness, ROM, peak-velocity, peak-acceleration</td>
<td>Decreased ankle stiffness and increased ankle movement mean improvements in ankle performance and gait</td>
<td></td>
</tr>
<tr>
<td>G. S. Sawicki, 2006 [90]</td>
<td>Level IV, Before-After</td>
<td>N=5</td>
<td>Chronic incomplete SCI</td>
<td>44.6±13.4</td>
<td>PAFO</td>
<td>Push-off kinematics and muscle activation amplitude</td>
<td>Improvement in push-off kinematics means improved gait function</td>
<td></td>
</tr>
<tr>
<td>J. Ward, 2010 [105]</td>
<td>Level IV, Before-After, Single Case Series</td>
<td>N=3</td>
<td>stroke syndrome</td>
<td>60, 48 and 48</td>
<td>PAFO</td>
<td>Robot Assisted Gait</td>
<td>Laboratory functional improvement in six-minute walk correlates with activities of daily life</td>
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<tr>
<td>L. F. Chin, 2010 [93]</td>
<td>Level IV, Before-After</td>
<td>N=23</td>
<td>Both inpatients and outpatients with mobility problems secondary to an acquired brain injury</td>
<td>51±13, 26-68</td>
<td>A robotic-assisted locomotor training device</td>
<td>Functional independence measure (FIM), the Rivermead Motor Assessment (RMA) gross function subscale and Motricity Index (MI)</td>
<td>Laboratory functional improvement correlates with activities of daily life</td>
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<tr>
<td>Study</td>
<td>Design</td>
<td>Participants</td>
<td>Interventions</td>
<td>Outcomes</td>
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<td>A. Roy, 2011 [103]</td>
<td>Before-After Case Control</td>
<td>N=14 7 chronic stroke who had residual hemiparetic deficits and an equal number of age- and sex-matched nondisabled control subjects</td>
<td>A single session of Impedance-controlled ankle robot (Anklebot)</td>
<td>Increased targeting accuracy (21.6 ± 8.0 to 31.4 ± 4.8, p = 0.05), higher angular speeds (mean: 4.7 ± 1.5 degrees/s to 6.5 ± 2.6 degrees/s, p &lt; 0.01, peak: 42.8 ± 9.0 degrees/s to 45.6 ± 9.4 degrees/s, p = 0.03), and smoother movements (normalized jerk: 654.1 ± 103.3 s⁻² to 537.6 ± 86.7 s⁻², p &lt; 0.005, number of speed peaks: 27.1 ± 5.8 to 23.7 ± 4.1, p &lt; 0.01) while nondisabled subjects did not make significant gains except in the number of successful passages (32.3 ±7.5 to 36.5 ± 6.4, p=0.006)</td>
<td>Improved target accuracy, movement and angular speed mean improvements in ankle performance and gait</td>
<td></td>
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<tr>
<td>k. A. Shorter, 2011 [99]</td>
<td>Level IV, Case Control, Single Case</td>
<td>N=4 3 nondisabled male volunteer subjects and 1 male volunteer subject with a diagnosis of CES</td>
<td>A novel PPAFO</td>
<td>Data from nondisabled walkers demonstrated functionality and data from an impaired walker demonstrated the ability to provide functional plantar flexor assistance</td>
<td>Providing functional assistance contributes to ankle rehabilitation</td>
<td></td>
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<tr>
<td>M. M. Mirbagheri, 2011 [91]</td>
<td>Level IV, Before-After</td>
<td>N=10 Incomplete SCI</td>
<td>Robotic-Assisted Locomotor Training</td>
<td>Reflex stiffness and intrinsic stiffness was respectively reduced up to 65% and 60% after LOKOMAT training; MVCs were increased up to 93% in ankle extensors and 180% in ankle flexors following 4-week training</td>
<td>Decreased ankle stiffness and increased ankle movement mean improvements in ankle performance and gait</td>
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2.3.2 Outcome Measures

None of the studies involved outcomes classified using the health dimensions of the World Health Organization’s International Classification of Functioning, Disability, and Health. Most studies tested outcomes in terms of either ankle performance including ankle strength, ankle ROM and ankle motor control [16, 18, 20, 83, 84, 87-90, 95, 97, 98, 101, 103, 104, 106-108] or gait functionality [82, 92, 93, 96, 99, 105]. Five studies assessed both ankle performance and gait functionality to verify the effects of robot-assisted ankle rehabilitation devices [85, 86, 94, 100, 102]. One study [88] also tried to assess the device’s effectiveness through pressure distribution on the footplate, but whether pressure distribution could be used as an indicator of ankle recovery is not clear. Two studies [94, 98] also considered satisfaction level of participants as the evaluation criteria.

It is difficult to know how functional improvements in laboratories correlate with activities of daily life. One study [110] showed that for people with stroke, the six-minute walk test was correlated to StepWatch monitor outputs over three days. However half of the variability in usual walking performance was not explained by clinical walking tests, and the study concluded that activity monitoring should also be included in functional assessment.

Several studies have shown that gait performance is affected by ankle muscle strength (in stroke [111] and cerebral palsy [112]) and ankle joint position [113]. One study [114] concluded that the isokinetic torques of the paretic ankle plantar flexors had moderate to high correlations with gait and stair-climbing speeds. Another study [115] revealed that the dorsiflexor strength was the most important factor for gait velocity and dynamic spasticity was the most important determinant for gait spatial symmetry. It also showed that adequate ankle control during gait was important for normal gait pattern. To some extent, however, a functional recovery of gait can be thought of as an indicator of ankle joint functional recovery.

2.3.3 Methodological Quality

The level of evidence was based on the AACPDM guidelines. 28 studies were conducted with evidence no higher than level IV. Only one study was designed through RCT with only 18 participants and thus the evidence level was II [43].
Chapter 2. A Review of Robot-Assisted Ankle Rehabilitation Techniques

2.3.4 Research Results

The studies were grouped into two general areas based on the type of devices used for robot-assisted ankle rehabilitation. One was platform-based ankle rehabilitation robot, and the other was wearable ankle rehabilitation robot.

2.4. Discussion

The goal of this systematic review was to identify research describing the therapeutic effects of ankle rehabilitation robots. In total, 29 studies were selected to investigate the effectiveness of robot-assisted ankle rehabilitation on individuals with any grade of ankle disability.

2.4.1 Platform-Based Ankle Rehabilitation Robots

The platform-based ankle rehabilitation device has a fixed platform and thus is usually not used during the gait training [58]. Parallel mechanisms are typically used for multi-DOF systems to reduce the size of robots. With the exception of the Stewart platform based device which is capable of six DOFs, most researchers have opted for designs which offer two or three DOFs for ankle rehabilitation exercises.

A range of platform-based devices have been developed for the purpose of ankle rehabilitation. They are usually 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 patients, ranging from no active effort in passive exercises to full users driven motion in active exercises. Strength training on the other hand requires robots to apply a resistive load to impede the movement of users for muscle strength training.

The Rutgers Ankle Rehabilitation System was adopted in nine studies [20, 83, 92, 95, 96, 98, 100, 101, 107]. All these studies except one [92] focused on the development and effects of this ankle rehabilitation system. Mirelman et al. [92] used an RCT design and concluded that a robot with virtual reality had better outcomes compared with a robot alone whose improvements were modest and did not transfer to significant functional changes on the gait of individuals after stroke. Two studies [96, 100] adopted Rutgers Ankle with virtual reality to conduct post-stroke rehabilitation and assessed the effects based on different criteria. One was gait and elevation speed, and the other was ankle and foot mobility, force generation, coordination, and the ability to walk and climb stairs. The results indicated that lower extremity rehabilitation of a post-stroke
individual was promising [100] and could transfer to improve gait and elevation speed [96]. Boian et al. [107] further verified the conclusion of the research [100] through a two-month study in which three chronic post-stroke individuals underwent lower extremity rehabilitation. The second version of the Rutgers Ankle robot included virtual reality-based ankle rehabilitation with task-level haptic effects to enhance patients’ experience and alleviate boredom [95]. Results through a single case series design with three participants indicated that strength capabilities for some ankle muscles were improved and haptic effects did not interfere with patients’ ability to use the platform. Cioi et al. [83] proposed an updated Rutgers Ankle robot to allow access by youths with cerebral palsy. It was concluded that patient function and quality of life improved in terms of increased ankle strength and motor control after 36 rehabilitation sessions. Burdea et al. [102] further verified the conclusion through tests on three children with cerebral palsy by wider evaluation criteria. The telerehabilitation system with virtual reality consisting of Rutgers Ankle and a local computer connected with a remote one over the internet was evaluated on six post-stroke patients [20]. Performance in terms of accuracy of ankle movement, exercise duration and training efficiency, ankle mechanical power and number of repetitions did not decrease during the transition from the third week to the fourth week. Two studies [98, 101] involved orthopaedic rehabilitation using the Rutgers Ankle haptic interface. In [98], a proof-of-concept patient trial found that this device can be used for ankle rehabilitation in patients with hyper and hypomobile ankles by comparison between the healthy ankle and the injured ankle. Furthermore, Deutsch et al. [101] presented three case reports about rehabilitation of musculoskeletal injuries using the Rutgers Ankle haptic interface. The results showed improvements in ROM, torque generation capacity and ankle mechanical work over six rehabilitation sessions. However, the evidence level of these papers involving Rutgers Ankle was relatively low.

All in all, ankle rehabilitation using virtual reality-based Rutgers Ankle as compared to the Rutgers Ankle Robot alone was encouraging based on the testing of 45 participants (a child with cerebral palsy and 44 with post stroke or varying musculoskeletal ankle injuries). The main effect is likely due to the alleviation of users’ boredom.

Rehabilitation through ankle stretching was conducted in five studies [16, 18, 87, 94, 104]. An intelligent stretching device was developed in [16] to treat the spastic/contractured ankles of neurologically impaired patients. This device stretched the ankle safely to extreme dorsiflexion and plantarflexion position where spasticity and contracture were significant until a specified peak resistance torque was reached and then the ankle was held at the extreme position for a
period of time to let stress relaxation occur before it moved to the other extreme position. This made the treatment more effective than existing methods in terms of active and passive ROM, joint stiffness, viscous damping and reflex excitability in the sample of spastic patients. Selles et al. [94] further verified the therapeutic effects of the intelligent stretching device through a single case series design and found improvements with more outcome measures including additional muscle strength, walking speed and subjective experience of the subjects. Waldman et al. [104] developed a portable robot used for ankle rehabilitation after stroke. Each training session in this trial included passive stretching under intelligent control and biofeedback active movement training through motivating games with the robot providing assistance or resistance as needed. After 18 training sessions, eight subjects showed improvements in active dorsiflexion range, dorsiflexion muscle strength, MAS, STREAM and Berg Balance. These improvements were still observed six weeks after the study was completed. Such a device with similar training was also used for rehabilitation of lower extremity impairments in children with cerebral palsy, and results showed improvements in joint biomechanical properties, motor control performance, and functional capability in balance and mobility [18]. Although numerous articles have shown significant improvements of ankle performance or gait functionality, the mode by which this is achieved is unknown. Other types of training have not been compared to game based robotic assistance or resistance during passive stretching under intelligent control and biofeedback active movement training. Homma and Usuba [88] developed an ankle dorsiflexion and plantarflexion exercise device with passive mechanical joint and its effects were evaluated by pressure distribution on the footplate. Whether pressure distribution can be used as an indicator of recovery was not clear and the relation with biological data should be further investigated.

A novel therapeutic approach (Assisted movement with enhanced sensation (AMES)) was proposed in [86]. The effects of AMES as a treatment for hemiplegia was assessed through strength and joint positioning tests as well as motor function. For 11 subjects with severe lower extremity motor disability, improvements on most functional tests were sustained for six months. This strategy appeared safe and effective in chronic stroke patients. However, further studies with high level evidence should be conducted to verify its therapeutic effects.

2.4.2 Wearable Ankle Rehabilitation Robots

Wearable robots known as exoskeleton robots or as powered orthoses are being developed in contrast to platform-based rehabilitation robots [116]. In this review wearable ankle rehabilitation robots mainly referred to wearable Anklebot and AFOs. The AFO is a single-joint orthosis designed to assist and support movements of the ankle joint. It plays an important role
Chapter 2. A Review of Robot-Assisted Ankle Rehabilitation Techniques

during human walking. The first AFO was made in the late 1960s [38]. Some original robotic orthoses have been developed around the world and some trials with patients have been conducted for assessing their effects.

Three studies [84, 85, 103] proposed a visually guided, impedance controlled, seated Anklebot intervention for ankle rehabilitation and corresponding trial designs were conducted for assessing its effects on the paretic ankle. This control approach allowed subjects to reach targets unassisted while automatically tracking their performance; however, if subjects failed to move their ankles to reach a target in time, the robot provided assistive ankle torques [103]. In [84], two sub-acute stroke survivors performed ankle targeting movements in plantarflexion/dorsiflexion and inversion/eversion ranges with robotic assistance-as-needed and improved their ankle motor control in terms of targeting accuracy, faster speed and smoother movements. These short-term improvements were accompanied by changes in electroencephalogram power and coherence, which was possibly useful for the development of more effective Anklebot training that may translate to gains in gait function. Roy et al. [103] evaluated short-term ankle motor performance in chronic hemiparetic stroke through a double-arm pilot study with 560 movement repetitions training only in plantarflexion/dorsiflexion ranges. Statistically significant gains were achieved as indexed by increased targeting accuracy, higher angular speed, smoother movements and number of speed peaks. Forrester et al. [85] adopted a similar training protocol as [103] but with a different trial design and the purpose focused on effects on hemiparetic gait after a stroke. Results showed promise for the use of a modular impedance controlled Anklebot in the treatment of post-stroke hemiparesis, and seated Anklebot training could reduce ankle impairment and improve gait function. However, whether seated Anklebot training with a task based video game and impedance controller has better therapeutic effects compared with other ankle rehabilitation robots has not been shown.

Ren et al. [87] proposed a wearable robot for in-bed acute stroke rehabilitation and this device also applied passive stretching and active movement training through playing a game. Four patients participating in this trial were satisfied with this device and positive changes of active and passive biomechanical properties were detected.

An active ankle-foot orthosis (AAFO) was developed in [82] where the impedance of the orthotic joint was modulated throughout the walking cycle to treat drop-foot gait. A before-after trial design with two drop-foot participants and three control subjects showed that a variable-impedance orthosis might have certain clinical benefits for the treatment of drop-foot gait compared to conventional AFO having zero or constant stiffness joint behaviours. University of
Michigan Powered Ankle-foot orthoses’ (PAFOs’) effects were assessed through five patients with chronic incomplete spinal cord injury [90]. It has been shown that mechanical assistance from PAFOs improved ankle push-off kinematics without substantially reducing muscle activation during walking. Robotic plantarflexion assistance could be used during gait rehabilitation without promoting patients passivity. Two studies [89, 97] assessed an intelligent ankle-foot orthosis (IAFO). In [89], an experiment carried out using the IAFO with developed Magneto-rheological (MR) brake and control algorithm demonstrated that drop foot in swing phase and slap foot at heel strike was prevented in control participants, which was further confirmed by a patient with Guillain-Barre Syndrome in [97]. In [105], the PAFO utilized robotic tendon technology with a single DOF resulting in ankle rotation in the sagittal plane. All participants in that study showed some positive changes in their key gait variables and these improvements were more dramatic while harnessed and using a treadmill. A portable powered ankle-foot orthosis (PPAFO) was proposed in [99] to provide untethered assistance during gait. The PPAFO provided torque assistance for both plantarflexion and dorsiflexion by way of a bidirectional pneumatic rotary actuator. Healthy volunteers and a participant with a diagnosis of cauda equine syndrome participated in this research. Data from healthy walkers demonstrated functionality, and data from an impaired walker demonstrated the ability to provide functional plantarflexion assistance. Significant evidence exists to support the use of AFOs and however the therapeutic differences among them should be further investigated.

Two studies [91, 106] assessed the effects of robotic-assisted locomotor training on spasticity and volitional control of the spastic ankle in persons with incomplete Spinal Cord Injury (SCI). The results showed this training was effective in reducing spasticity and improving volitional control for patients with SCI. Another study [93] collected data from both inpatients and outpatient patients with mobility problems secondary to an acquired brain injury, before and after robotic-assisted locomotor training from September 2008 to May 2009. It showed significant improvement in ankle dorsiflexion.

2.4.3 Control Strategies

Several categories of strategies have been reported, including, assistive, challenge-based, haptic simulation, and coaching for robotic movement training following neurological injuries [117]. There were several studies that have examined the effects of various control strategies for robot-assisted ankle rehabilitation by unimpaired subjects. Sixteen studies [10, 44-58] were referred for more information. All selected studies with clinical evidence showed positive therapeutic effects for patients with ankle disability. Studies [83, 95, 96, 100, 101, 107] took challenge-
based robotic control algorithms by providing resistance to the injured ankle during exercise and haptic simulation strategies by haptic interfaces for interacting with virtual reality. In [83], the Rutgers Ankle was upgraded in hardware and software, creating a more powerful reaction force and more accurate direct kinematics, adding inverse kinematics for passive training. Three recent studies [84, 85, 103] proposed the assistance-as-needed control by a mechanism applied based on the error between the target location and the proximity of the subject’s ankle, as well as the robotic torsional stiffness and damping [10]. While many studies have demonstrated that training with different robotic control strategies reduces motor impairment as assessed with various outcome measures, only a few studies have found differential benefits of particular robotic control strategies with respect to other control strategies. However, which control strategy is more effective for a certain ankle disability has not been clearly addressed yet and should be further investigated.

2.4.4 Safety and Reliability

Not much attention has been paid to the technologies of human robot symbiosis to date because almost all robots have been designed and constructed on the assumption that the robots are physically separated from humans [118]. In particular, safety and reliability are the underlying evaluation criteria for mechanical design, actuation, and control architectures [119]. Among all selected studies, ten studies [16, 82, 83, 87, 88, 91, 94, 95, 99, 106] involved the safety issue when ankle rehabilitation was conducted on patients. To ensure user safety during operation, Shorter et al. [99] used an equipment within the manufacturer’s published specifications. In [95], the interaction between the platform stiffness and the vibrations imposed safety limitations on the system. ASME device mentioned in [86] was non-invasive and the relatively small amplitudes of tendon vibration and movement make it safe to use. In [88], the given ROM was smaller than the subject’s actual ROM and the speed was set slow enough to avoid induction of spasticity for safety reasons. Appropriate procedure of emergency stop should be further examined since sudden stop may induce injury of foot. Cioi et al. [83] proposed a snowboard foot binding on the top platform of the robot to allow safe and easy attachment to the patient’s foot. Two studies [91, 106] involved locomotor training ensured subject safety by an accessible panic switch and monitored by therapists. Three other studies [16, 87, 94] applied velocity control for the purpose of safety. Specifically, stretching velocity slowed gradually down with increasing resistance torque or at the joint extreme positions. Zhang et al. [16] also mentioned the safety screws used as mechanical stop to restrict the motor ROM and a digital signal processor controller for position limit. However, few studies were conducted in terms of safety...
Chapter 2. A Review of Robot-Assisted Ankle Rehabilitation Techniques

and reliability assessment. There are only two studies [94, 98] whose outcome measures contained the subjective evaluation of the subjects. Subjects in [94] showed very positive subjective evaluation in terms of the comfort of stretching. Participants in [98] responded favorably to the use of the device and stated that they would enjoy having this device complementing their current rehabilitation programs.

2.4.5 Optimal Ankle Therapy

There is no reason to believe that a “one-size-fits-all” optimal treatment exists [32]. In other words, therapy should be tailored to each patient’s needs and abilities. Robot-assisted therapy can be delivered in a variety of ways to reduce motor impairment and enhance functional motor outcomes. For instance, goal-directed therapeutic games can be designed to address motor impairments including poor coordination, impaired motor speed or accuracy, and diminished strength, as well as addressing cognitive or perceptual impairments [18, 20, 83, 85, 92, 95, 96, 98, 100, 101, 107]. Depending on the participant’s impairment, robotic-assisted treatment can provide passive, active-assistive, active, and active-resistive exercises [32]. As with [32], an optimal therapy could be tailored to each stroke patient through a novel performance-based impedance control algorithm. Further investigation is needed to assess its therapeutic effects.

The question of what is the most appropriate robot-assisted ankle rehabilitation is not evident. Riener et al. [120] demonstrated the form of therapy may be more important than its intensity: muscle strengthening offers no advantage over movement training. Experienced rehabilitation therapists advocated “active-assist-exercise” or “assistance-as-needed”, which refers to the principle of helping the patient perform a movement with the minimal amount of manual assistance possible [121]. Moreover, Siegler et al. [1] showed neither ankle joint nor the subtalar joints were acting as ideal hinge joints with a fixed axis of rotation and motion of the foot-shank complex in any direction is the result of rotations at both the ankle and the subtalar joints. In detail, the contribution of the ankle joint to dorsiflexion/plantarflexion of the foot-shank complex is larger than that of the subtalar joint, the contribution of the subtalar joint to inversion/eversion is larger than that of the ankle joint, and the ankle and the subtalar joints have an approximately equal contribution to adduction/abduction rotation movements of the foot-shank complex. Therefore, robotic movement assistance-as-needed given in different rotation directions should base on different joints/axes, which can be defined as “effective movement assistance-as-needed”.

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2.4.6 Limitations of the Search Strategy

An attempt was made to ensure that all studies related with robot-assisted ankle rehabilitation with any grade of ankle disability were reviewed. It is assumed in this review that all studies used different patients, but because some studies were conducted at the same place and at the same time, I cannot be certain whether only unrelated study populations were used. However, other research may exist in which robot or ankle was not identified as a key term within the article. For instance, some articles about ankle rehabilitation robots were probably described in terms of devices and lower extremity/lower limb. Only articles after 1980 were included in this study as robot-assisted ankle rehabilitation was very limited before then. This review included published conference papers and abstracts as well as full peer-reviewed papers but did not include abstracts written in languages other than English or unpublished data. Some studies may therefore have been excluded on this basis, leading to potential incomplete search.

2.5. Conclusion of the Review

Even though a range of robot-assisted ankle rehabilitation devices and control strategies are available for individuals with ankle disability, the most effective ankle rehabilitation device and control algorithm is still vague. This is due to a lack of universal evaluation criteria with effective outcome measures. Although using RCTs to assess effects of robots on ankle outcomes is expensive and time-consuming, so too are the robots designed to assist.

In terms of control strategies, providing too much assistance has negative consequences [117] and therefore “effective movement assistance-as-needed” control strategy is probably encouraging for ankle rehabilitation. Specifically, that means to assist the participant only as much as needed according to real-time ankle performance or systematically reduce its assistance as recovery progresses. This will be better if combined with virtual reality designed to be in dynamic accordance with ankle performance. In other words, the virtual-reality system should be updated online automatically as ankle rehabilitation progresses. However, to achieve dynamic and real-time ankle rehabilitation level is extremely important for realizing “effective movement assistance-as-needed” control, which could be realised through dynamically evaluating ankle stiffness or based on patients’ task accuracy. The technological challenge in integrating it into a virtual-reality system is significant as well.

Few studies have undergone rigorous experimental assessment with high-level evidence. Therefore, higher level trials like RCTs should be conducted aiming at assessing the therapeutic effects of robot-assisted ankle rehabilitation devices and control strategies. These trials should
base on rigorous comparison with each other, and with simpler, non-robotic conventional therapy in terms of devices and control strategies. It is also necessary to set up universal evaluation criteria that should contain systematic and comprehensive outcomes to evaluate devices and control strategies, including the assessment of end-user comfort, safety and training performance. These evaluation criteria may improve the consistency of results and facilitate comparisons among ankle rehabilitation robots.

2.6. Chapter Summary

This chapter introduces a variety of ankle rehabilitation devices in terms of mechanical design, actuation system, control strategy, and treatment efficacy. Limitations of existing ankle rehabilitation robots can be summarised as: incompatible robot design that requires varying ankle positions during the robotic training, stiff actuation, the lack of three-dimensional robot motion, and real-time ankle assessment. While the majority of existing rehabilitation robots have been demonstrated to be effective for the treatment of a variety of ankle injuries, the most effective robot design and control strategy cannot be determined due to the lack of universal evaluation criteria. However, this chapter provides guidelines in improving existing ankle rehabilitation robots, which leads to the conclusion that an optimal ankle rehabilitation robot should have soft actuation, compatible mechanical structure that allows the ankle to be stationary during the training, three DOFs for three-dimensional exercises, the function of real-time biomechanical assessment, and the implementation of adaptive interaction training.
Chapter 3. A Review of Robot-Assisted Ankle Assessment Techniques

This chapter provides a comprehensive review of studies that investigated ankle assessment techniques to better understand those that can be used in the real-time monitoring of rehabilitation progress for implementation in conjunction with robot-assisted therapy. This review was conducted to identify the suitable technique for real-time three-dimensional assessment during the robotic ankle training. Some quantitative techniques have been identified suitable for comprehensive ankle assessment during the robotic training. The materials presented in this chapter has been published as [22] in Journal of Rehabilitation Research and Development.

This published review goes as follows. Seventy-six publications published between January, 1980 and August, 2013 were selected from eight databases. They are divided into two main categories with 16 qualitative and 60 quantitative studies. These quantitative methods involve 13 goniometer studies, 18 dynamometer studies, and 29 studies regarding innovative techniques. A total of 465 subjects participated in the 29 quantitative studies of innovative measurement techniques that may potentially be integrated in a real-time monitoring device. Results show that qualitative techniques are not suitable for real-time ankle monitoring during robot-assisted therapy, while quantitative methods show great potential. While the majority of quantitative techniques are reliable in measuring ankle kinematics and dynamics, they are mostly available for only the sagittal plane. Limited studies determine kinematics and dynamics in all three planes (the sagittal plane, transverse plane and frontal plane) where motions of the ankle joint and the subtalar joint actually occur.

3.1. Introduction

Ankle injuries are very common either in sports or daily life [2, 122-125]. In New Zealand, about 100,000 claims related to ankle sprains were made to the Accident Compensation Corporation in 2000/2001 at a cost of an estimated 31.8 million New Zealand dollars. From 2002 to 2006, a total of 82,971 ankle sprains were identified in the NEISS database and an estimated 2.15 ankle sprains occurred per 1000 person-years in the United States [126]. Neurological injuries such as stroke and spinal cord injuries also cause various ankle problems.
Chapter 3. A Review of Robot-Assisted Ankle Assessment Techniques

[16, 127]. Ankle injuries cause complications such as oedema, disuse atrophy and arthrosis unless treated properly [128]. Additional symptoms usually include chronic pain, reduced range of motion (ROM), weak strength and increased joint stiffness as well as severe functional limitations [2, 129].

Clinicians often use a qualitative assessment method to assess ankle impairment based on a pre-defined scoring system. When quantitative methods are undertaken, these most commonly include the use of a goniometer or dynamometer. The goniometer is a tool to assess ankle ROM [130, 131] and the dynamometer is usually used for assessing ankle strength [132, 133]. Other devices have been developed for measuring ankle stiffness, for monitoring the progress of a rehabilitation programme, or for tracking the changes in joint stiffness [134, 135]. These measurement tools can guide clinicians in determining the most effective intervention.

There have also been significant advances in robotic rehabilitation in an effort to reduce the strain on the clinician. Various robot-assisted ankle rehabilitation devices have been developed in recent years [12, 13, 35, 67, 104, 136]. They usually lack the function of real-time ankle assessment that should be included in a robot-assisted ankle rehabilitation program to allow the robot to adjust the control strategy for a specific rehabilitation stage. Having a better understanding of the clinical tools that are most effective in providing intervention and how they might provide quantitative inputs for robot-assisted ankle rehabilitation is necessary to develop a robotic rehabilitation device that engages all users.

This review seeks to critically compare various published studies in terms of the development, application, reliability and validity of existing ankle measurement devices and techniques. It will provide a better understanding of the requirements for a real-time assessment strategy implementable within a robot-assisted rehabilitation program that can be used throughout the rehabilitation process.

### 3.2. Methods

#### 3.2.1 Search Strategy

Only English-language articles published from January 1980 to August 2013 were searched in the following six databases: Scopus, Web of Science, ScienceDirect, Academic Search Premier, Embase and MEDLINE (OvidSP). The search terms were “Ankle* AND “Performance OR Function OR Disability* OR Disorder* OR Injur* OR Spastic* OR Stabilit* OR Stiff* OR Torque OR Moment OR Strength OR Kine* OR Dynamic* OR Dorsiflexion” AND “Evaluat*
Chapter 3. A Review of Robot-Assisted Ankle Assessment Techniques

OR Assess* OR Measur* OR Examinat*. Additional search in Google Scholar and SpringerLink was further conducted for latest studies as an important supplement. Valuable references listed in relevant publications were also screened.

A total of 411 articles were identified initially. The first two rounds of screenings were conducted based on titles and abstracts, respectively. Studies considered to meet the predefined inclusion criteria were included in the final analysis and the others were excluded. Discussion amongst authors resulted when inclusion of certain papers was questionable. Figure 3.1 describes the selection process.

Figure 3.1: Flow diagram of selection process for final review.

3.2.2 Inclusion and Exclusion Criteria

This study aims to review existing ankle assessment techniques that can provide necessary information to allow for an evaluation of improvement during ankle exercises that are implemented using robot-assisted rehabilitation. The review attempts to better understand all methods of evaluation including qualitative and quantitative assessment of ankle recovery level. Papers involving ankle performance or functional qualitative assessment methods such as the
Foot Function Index (FFI) and Foot and Ankle Disability Index (FADI) were included. All quantitative studies assessing ankle performance or function (including ankle disability level, kinematics and dynamics) were included. All papers had to include trials involving either normal ankle or injured ankle. Trials assessing animal ankle performance or function were excluded due to significant differences between the animal ankle and the human ankle. Studies involving management or identification of ankle injuries and those related to emergencies were excluded, as were invasive ankle measurement techniques. Observation based physiological assessment techniques were excluded due to the unreliable accuracy \cite{137}. Image-based methods were also excluded because they cannot be used to evaluate functional improvement in ankle injury in combination with robot-assisted therapy. Image based techniques that examine kinematics in-vivo such as magnetic resonance imaging, computed tomography and X-ray tend to be expensive and not implementable in a typical robotic system, though they can be used for identification of ankle injury \cite{138, 139}. This review is not seeking to comment on the ability to detect ligament tears (it is assumed that the correct identification of ankle injury has already occurred), rather to examine the functional improvement before, during and after rehabilitation interventions. The data extraction applied the similar way as another review conducted by Zhang et al. \cite{17}.

3.3. Results

After excluding studies involving invasive measurement techniques \cite{140-142}, animal-based methods \cite{143, 144}, image-based methods \cite{138, 139, 145-147}, diagnosis of ankle injuries \cite{123, 148-151} and management of ankle injuries \cite{124, 152-156}, there were a total of 76 publications identified for further analysis. These were divided into two main categories with 16 qualitative studies \cite{157-172} and 61 quantitative studies. These 61 studies were further grouped into 13 studies \cite{130, 131, 173-183} using goniometers to measure ankle ROM, 18 studies involving dynamometers to measure ankle strength (4 studies \cite{133, 184-186} about hand-held dynamometers and 14 studies \cite{132, 187-199} about isokinetic dynamometers) and 29 studies \cite{134, 135, 200-226} with innovative ankle measurement techniques developed to measure various ankle parameters including ankle ROM, strength, torque and stiffness that may be used for real-time assessment of patient improvement.

Assessment techniques requiring specialist training were included in 16 qualitative studies. An additional 31 quantitative studies involving goniometers or dynamometers were also found, and these mainly measure either ROM or strength; both parameters easily measured by a robot.
Additional studies that provide information about parameters that can potentially be implemented in robot-assisted training were the main focus of this chapter and included 29 quantitative studies. A total of 465 subjects participated in these 29 quantitative studies, of which 19 studies were conducted on less than 20 participants, see Table 3.1. These participants contain both healthy volunteers and patients with diverse ankle injuries.
Table 3.1: Reviewed studies of quantitative ankle measurement techniques.

<table>
<thead>
<tr>
<th>Studies</th>
<th>Subjects</th>
<th>Characteristics</th>
<th>Age (years)</th>
<th>Methods</th>
<th>Measures</th>
<th>Reliability</th>
</tr>
</thead>
<tbody>
<tr>
<td>[213]</td>
<td>N=1</td>
<td>A healthy subject</td>
<td>No stated</td>
<td>Instrumented shoes</td>
<td>Ankle moment</td>
<td>The root mean square (RMS) difference of ground reaction force (GRF) was 19.1 N and for the centre of pressure (CoP) the RMS difference was 17.9</td>
</tr>
<tr>
<td>[216]</td>
<td>N=1</td>
<td>A healthy subject</td>
<td>No stated</td>
<td>Instrumented shoes</td>
<td>Foot and ankle dynamics</td>
<td>The RMS difference of the GRF was ((0.012 \pm 0.001)) N/N; the CoP estimation RMS difference was ((5.1\pm0.7)) mm; the RMS difference of the heel position estimates was ((18 \pm 6)) mm; the ankle moment RMS difference was ((0.004 \pm 0.001)) Nm/N and the RMS difference of the estimated power was ((0.02 \pm 0.005)) W/N</td>
</tr>
<tr>
<td>[221]</td>
<td>N=22</td>
<td>12 patients with ankle osteoarthritis and 10 healthy subjects</td>
<td>12 patients (age: 58±13 years and 10 healthy subjects (age: 61±13 years)</td>
<td>An ambulatory system consisting of plantar pressure insole and inertial sensors</td>
<td>Ankle force, moment and power</td>
<td>High repeatability (CMC&gt;0.7)</td>
</tr>
<tr>
<td>[205]</td>
<td>N=31</td>
<td>10 unimpaired physiotherapy students and 21 subjects who had suffered stroke</td>
<td>Impaired group: (mean age 75.4 (SD = 8) years); healthy group: (mean age 24.3 (SD = 3.9) years)</td>
<td>Lidcombe Template</td>
<td>The magnitude and direction of force applied to dorsiflex the foot</td>
<td>Highly reliable for both groups (r&gt;0.92)</td>
</tr>
<tr>
<td>[203]</td>
<td>N=15</td>
<td>Five members of staff, five people who had suffered a CVA and five head-injured adults from Lidcombe Hospital</td>
<td>Not stated</td>
<td>Lidcombe Template</td>
<td>Angle and force in ankle dorsiflexion</td>
<td>ICC for the combined group data was 0.97 and the percentage intertester agreement was 77%</td>
</tr>
<tr>
<td>[209]</td>
<td>N=29</td>
<td>17 subjects (repeatability) and 12 physical therapy graduate students with no</td>
<td>53 ±14 for repeatability and not stated for validity</td>
<td>The IAROM device</td>
<td>Ankle dorsiflexion ROM</td>
<td>The average ICC value was 0.92 and a mean correlation value is 0.96</td>
</tr>
<tr>
<td>Reference</td>
<td>N=</td>
<td>Participants</td>
<td>Device/Measurements</td>
<td>Notes</td>
<td></td>
<td></td>
</tr>
<tr>
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</tr>
<tr>
<td>[222]</td>
<td>29</td>
<td>12 participants (6 male, 6 female; height, 1.7 ± 0.1 m; body mass, 72 ± 12 kg); intertester reliability: 17 participants (7 male, 10 female; height, 1.7 ± 0.1 m; body mass, 88 ± 21 kg)</td>
<td>The IAROM device</td>
<td>Validity testing: ICC values ranges from 0.95 to 0.98 and reliability testing: ICC values ranges from 0.90 to 0.95; the ICCs for ankle joint dorsiflexion stiffness were 0.71 and 0.85 for the knee in an extended and flexed position</td>
<td></td>
<td></td>
</tr>
<tr>
<td>[211]</td>
<td>11</td>
<td>Six children with cerebral palsy (CP) and five healthy children</td>
<td>MSE</td>
<td>Not stated</td>
<td></td>
<td></td>
</tr>
<tr>
<td>[220]</td>
<td>22</td>
<td>12 children with CP who had ankle spasticity and five healthy children and five healthy adults</td>
<td>MSE</td>
<td>High reproducibility with ICC = 0.82, Pearson r = 0.81 and p = 0.002</td>
<td></td>
<td></td>
</tr>
<tr>
<td>[215]</td>
<td>4</td>
<td>Two male and two female with height (cm) 147.3-177.8, weight (kg) 45-74 and passive ROM (deg) 64-94</td>
<td>Anklebot</td>
<td>Errors (deg) in dorsiflexion and plantarflexion are 0.75 and 0.89, respectively and correlation coefficient is 99.61</td>
<td></td>
<td></td>
</tr>
<tr>
<td>[217]</td>
<td>20</td>
<td>10 healthy males and 10 healthy females, height (males: 177.16 ± 6.01 cm, females: 167.15 ± 5.98 cm), weight (males: 80.79 ± 13.53 kg, females: 71.10 ± 11.87 kg)</td>
<td>A unique medial/lateral swaying cradle device</td>
<td>The trial-to-trial reliability ICC coefficient was 0.96 with a standard errors of measurement (SEM) of 2.05 Nm/rad, and the day-to-day reliability ICC coefficient was 0.93 with an SEM of 3.00 Nm/rad</td>
<td></td>
<td></td>
</tr>
<tr>
<td>[134]</td>
<td>2</td>
<td>Stroke patients</td>
<td>A manual ankle assessment device</td>
<td>No stated</td>
<td></td>
<td></td>
</tr>
<tr>
<td>[219]</td>
<td>10</td>
<td>Male subjects with hemiplegia</td>
<td>A manual ankle assessment device</td>
<td>High reliability with ICC values over 0.97</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Reference</td>
<td>N</td>
<td>Group Description</td>
<td>Age Range</td>
<td>Equipment</td>
<td>Measurement</td>
<td>Results</td>
</tr>
<tr>
<td>-----------</td>
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</tr>
<tr>
<td>[218]</td>
<td>46</td>
<td>Gender-matched healthy subjects</td>
<td>20 or older</td>
<td>An Intelligent stretching device</td>
<td>Ankle stiffness</td>
<td>The reliability ICC coefficient of ankle stiffness between-day for both examiners was 0.77 for the right ankle and 0.76 for the left ankle with a 0.05 SEM for ankle stiffness for the right side and 0.04 for the left side</td>
</tr>
<tr>
<td>[225]</td>
<td>83</td>
<td>46 (16 female) healthy volunteers, 14 spinal cord injured (SCI) (1 female) and 23 multiple sclerosis (MS) participants (14 females)</td>
<td>46 healthy subjects (32 ± 7 years), 14 SCI patients (48 ± 12 years) and 23 MS participants (53.3 ± 12 years)</td>
<td>A portable Neurokinetics RA1 Rigidity Analyzer</td>
<td>Ankle joint stiffness</td>
<td>High intra-rater reliability (ICC: 0.60–0.89, SCI; 0.63–0.67, controls), inter-rater reliability (ICC: 0.70–0.73, SCI; 0.61–0.77, controls)</td>
</tr>
<tr>
<td>[206]</td>
<td>15</td>
<td>Eight healthy males and seven healthy females</td>
<td>20-68</td>
<td>Ankle stiffness measuring apparatus</td>
<td>Ankle stiffness</td>
<td>The coefficient of variation is 5 %</td>
</tr>
<tr>
<td>[212]</td>
<td>18</td>
<td>Healthy subjects with half females and half males</td>
<td>21-31</td>
<td>The apparatus consists of a force platform and a motorised footplate</td>
<td>Intrinsic ankle stiffness during quiet standing</td>
<td>Not stated</td>
</tr>
<tr>
<td>[210]</td>
<td>4</td>
<td>Two healthy males and two healthy females</td>
<td>29-70</td>
<td>A computational method with MATLAB</td>
<td>Ankle postural stiffness</td>
<td>High coefficient values of determination</td>
</tr>
<tr>
<td>[135]</td>
<td>11</td>
<td>Eight males and three females</td>
<td>Not stated</td>
<td>A parallel-cascade system identification method</td>
<td>Ankle dynamic stiffness and separation of intrinsic and reflex components</td>
<td>High intrasubject reliability ( r&gt;0.8 ) but a high intersubject variability</td>
</tr>
<tr>
<td>[207]</td>
<td>1</td>
<td>A normal subject</td>
<td>Not stated</td>
<td>A bilateral electro-hydraulic actuator system</td>
<td>Dynamic ankle joint stiffness during standing</td>
<td>Not stated</td>
</tr>
<tr>
<td>[201]</td>
<td>15</td>
<td>Healthy subjects</td>
<td>23-39 years</td>
<td>A system identification technique</td>
<td>Ankle dynamics</td>
<td>Intrasubject reliability was as good as or better than most clinical measures and intersubject variability was somewhat larger</td>
</tr>
<tr>
<td>[202]</td>
<td>3</td>
<td>Patients (two males and one female) recovering from fractures of the lower leg</td>
<td>28, 28, 50</td>
<td>A system identification technique</td>
<td>Ankle dynamics</td>
<td>Ankle dynamics measures were qualitatively similar to those of normals: stiffness was low in a region near the mid-range and increased</td>
</tr>
<tr>
<td>Ref.</td>
<td>N=</td>
<td>Description</td>
<td>Subject Details</td>
<td>Measurement/method</td>
<td>Note</td>
<td></td>
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<tr>
<td>200</td>
<td>10</td>
<td>Healthy female subjects</td>
<td>26-46</td>
<td>An ankle torque measurement system</td>
<td>Passive ankle mechanical stiffness</td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td>High reproducibility</td>
<td></td>
</tr>
<tr>
<td>214</td>
<td>1</td>
<td>An experienced snowboarder with 185 cm tall and a mass of 74.8 kg</td>
<td>Not stated</td>
<td>An “elbow-type” ISL</td>
<td>Measure dynamic ankle motion in a field environment</td>
<td></td>
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<tr>
<td></td>
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<td></td>
<td></td>
<td></td>
<td>Root-mean-squared errors were 0.59 deg for orientation and 1.00 mm for position; maximum measurement deviations were 0.05 deg in orientation and 0.10 mm in position</td>
<td></td>
</tr>
<tr>
<td>223</td>
<td>12</td>
<td>Six cadaveric specimens and six female subjects (height = 1.60 ± 0.04 m, body mass = 54.8 ± 5.8 kg)</td>
<td>25.8 ± 2.7</td>
<td>A mechanical jig</td>
<td>Ankle supination and pronation torque</td>
<td></td>
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<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td>Good internal consistency of trials was obtained with typical error of 0.3 and 0.1 in standing and sitting position, respectively</td>
<td></td>
</tr>
<tr>
<td>204</td>
<td>10</td>
<td>Five male and five female subjects</td>
<td>73-92</td>
<td>Footplate apparatus</td>
<td>Voluntary isometric strength and evoked isometric twitch properties, M-wave amplitude and passive tension</td>
<td></td>
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<tr>
<td></td>
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<td></td>
<td></td>
<td>Mean reliability coefficient of all measurements on the dorsiflexion and plantarflexor muscle groups was 0.91 ± 0.05</td>
<td></td>
</tr>
<tr>
<td>226</td>
<td>4</td>
<td>Healthy male subjects (height: 170.3 ± 5.2 cm, weight: 61.5 ± 15.4 kg)</td>
<td>30 ± 11.5</td>
<td>Estimation of muscle length parameters based on measurement data</td>
<td>Passive ankle joint moment versus ankle angle, and ankle muscle length</td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td>Predicted data are consistent with measured data</td>
<td></td>
</tr>
<tr>
<td>208</td>
<td>21</td>
<td>A healthy volunteer (male, 75kg, 185cm) for assessing the experimental set-up and 20 healthy volunteers (168.6 ± 8.8 cm and 71.0 ± 11.3 kg) for investigating ankle muscular functionality</td>
<td>First group: 28; second group: 50.4 ± 14.6</td>
<td>A measurement device for obtaining the kinematic characterisation and isometric loading of ankle under different working conditions</td>
<td>Ankle kinematic and dynamic characterisation</td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td>High linearity and overall accuracy are found in the desired torque ranges and inaccuracies are found in kinematic measurements</td>
<td></td>
</tr>
<tr>
<td>224</td>
<td>26</td>
<td>13 subjects (eight female, five male) with history of a single lateral ankle sprain and 13 healthy subjects (nine male, four female)</td>
<td>21.46 ± 1.17</td>
<td>A fulcrum device</td>
<td>Dynamic inversion speed</td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td>High reliability was found for time to maximum inversion (ICC= 0.81) and mean inversion speed (ICC= 0.79)</td>
<td></td>
</tr>
</tbody>
</table>
3.4. Discussion

3.4.1 Qualitative Ankle Assessment Techniques

With a view to understanding the clinical functional scales and how an assessment of improvement is conducted (and to better understand the accuracy requirements for measurement by robot-assisted techniques), the following describe the qualitative measurement techniques.

Scoring systems: The traditional method of describing ankle injuries was to group the assessment results: good, fair and poor [227]. In recent years, more accurate scoring systems have been developed for ankle performance or function assessment. SooHoo et al. [162] demonstrated that the FFI (a self-administered index consisting of 23 items) was a reasonable measure to monitor ankle status by examining its level of correlation to the Medical Outcomes Study Short Form-36 on 73 patients. Karlsson and Peterson [157] presented a scoring scale based on the subjective assessment of the patient’s symptoms and level of function and the evidence on 148 patients demonstrated that this system could be used to evaluate ankle function before and after treatment of ankle joint. More sophisticated, Kaikkonen et al. [160] proposed a performance test protocol and scoring scale for functional evaluation of ankle injuries based on both subjective patient feedback and clinical ankle examinations including the measurement of ROM, laxity of ankle joint and muscle strength. This method showed excellent reproducibility and the total score correlated obviously with isokinetic ankle strength, subjective opinion of recovery and subjective functional assessment on 148 patients. All these systems involve subjective assessment either from physiotherapists or patients.

Niki et al. [166] proposed new scales with improved expressions for Japanese people based on the clinical rating systems established by the American Orthopaedic Foot and Ankle Society and the Japanese Orthopaedic Association's foot rating scale (JOA scale). Reliability analysis on 65 clinicians and 610 patients was further conducted in [167] and the results showed that this newly established standard rating scales and JOA scale are highly reliable, to some extent which demonstrated its population-specific characteristic.

In contrast, two studies did not show positive effects for a certain group of people. Ansari et al. [158] found that reliability of Modified Tardieu Scale in the assessment of post-stroke ankle plantar flexor spasticity was not high, and Campanini et al. [159] demonstrated that the treatment of ankle spasticity for cerebral vascular accident (CVA) patients could not rely on the Ashworth score completely.
Other studies investigated and compared various qualitative ankle assessment systems. Haywood et al. [161] summarised seven disease-specific assessment methods of ankle performance: Ankle Joint Functional Assessment Tool, Clinical Trauma Severity Score, Composite Inversion Injury Scale, Kaikkonen Functional Scale, Karlsson Ankle Function Score, Olerud and Molander Ankle Score and the Point System and concluded that any measure should be used with caution until appropriate evidence is provided. However, further investigation into the effectiveness of functional outcome scores specific to patients was necessary, which was also supported by Farrugia et al. [165].

**FADI and FAAM**: Although FADI and FAAM belong to the category of scoring systems, they were discussed independently since a systematic review [168] identified them as the most appropriate outcome instruments to quantify functional limitations in patients with chronic ankle instability (CAI).

Hale and Hertel [169] advocated the use of the FADI and FADI Sport self-report instruments in clinical care and research applications in young adults with CAI, and these instruments appeared to be reliable in assessing functional limitations. Further, Wikstrom et al. [172] found that self-assessed disability got from these systems was greater in subjects with CAI than uninjured groups, which showed their patient-specific characteristic. More advanced, Foot and Ankle Ability Measure (FAAM) as the later version of FADI showed satisfactory validity and reliability on groups with various ankle injuries [163]. Martin et al. [170] developed FAAM for measuring region-specific and non-disease-specific function of the foot and ankle and concluded that it was a reliable and valid measure of physical function for individuals with a wide array of musculoskeletal foot and ankle disorders. Further, Garcia et al. [171] concluded that the FAAM may be used to detect self-reported functional deficits related to CAI and Cosby and Hertel [164] verified its reliability and validity on a male basketball player suffering from inversion ankle sprain. As for the region-specific attribute, direct comparison among different populations is quite necessary although some versions have proved to be reliable. For example, Mazaheri et al. [163] conducted tests on 93 Persian patients and demonstrated that the FAAM was a reliable and valid measure quantifying physical function. Therefore, it can be summarised that FADI and FAAM are usually reliable but larger sample sizes with a greater diversity of populations and various ankle injuries should be investigated in future research.

Taking all studies into consideration, it can be concluded that these scoring systems are usually region-specific and disease-specific and thus a universally designed method with convincing validity and reliability is essential. Further, these currently available systems usually require
participants to conduct a series of functional activities and answer subjective questionnaires, and then rate function on a pre-specified scale based on ankle performance. This assessment, which usually lasts a few minutes or longer, makes it difficult for the realization of real-time evaluation when used in robot-assisted therapy though some of the techniques and questions may be valuable in engaging the user in understanding the therapy being undertaken especially during a robot-assisted program.

3.4.2 Quantitative Ankle Assessment Techniques

To better be able to quantify functional improvement using robot-assisted therapy, quantitative measures are required. These should incorporate all aspects of ankle control including ankle kinematics and dynamics such as ROM, muscle strength and joint stiffness. This section discusses the tools (standard techniques and innovative techniques) for measurement of these aspects of functional control.

Standard Techniques

**Ankle ROM**: The ankle ROM is an important functional parameter that relates to the efficiency of gait. An accurate measurement of ankle ROM in all three planes is important to better understanding functional improvement. For this reason, ankle ROM is frequently assessed clinically. This section discusses devices that can be used for determining ankle ROM and provides important information about requirements in the development of an assessment method implementable with a robot to assist ankle therapy.

Goniometers have been considered as the standard method for the measurement of ankle ROM, especially for ankle dorsiflexion [130, 131, 173-178, 180-183]. One study [179] showed its poor interrater reliability in ankle dorsiflexion and measures using goniometry have been shown to be tester dependent [222]. However, it has been reported that the validity and repeatability can be problematic due to goniometer alignment, as well as potential variations in location and magnitude of forces applied to the foot [209]. To address these limitations, the Iowa ankle ROM (IAROM) device [209, 222] allows angular measurements at pre-determined force levels using a digital inclinometer and a hand-held force gauge to measure ankle dorsiflexion ROM in a clinically friendly and cost-effective way. The clinical test on 29 participants proved its repeatability and validity but the application combined with robot-assisted therapy is restricted due to the manual operation of a hand–held force gauge.
Chapter 3. A Review of Robot-Assisted Ankle Assessment Techniques

**Ankle Strength**: Ankle strength is the amount of force ankle muscles can generate. The ability to generate force is necessary for all types of movement.

A hand dynamometer is usually used to measure grip strength, pinch strength, and to perform muscle fatigue studies. Some studies [133, 184-186] have demonstrated that hand-held dynamometers can be reliably used for measurement of ankle strength. Different from the normal application, there is an innovative hand-held dynamometer using a torque wrench and a goniometer that can measure static ankle angle and moment reliably and precisely [186]. Unfortunately, for the same reasons as the IAROM device, the manual operation of hand-held dynamometer impedes its adoption when combined with robot-assisted therapy.

The isokinetic dynamometer is widely used during varying phases of rehabilitation. It is a precision-based and electrical instrument that measures the performance of various muscle groups in the body. Isokinetic dynamometers including the Biodex dynamometer, the Cybex Norm and the Kin Com II dynamometer have been demonstrated as the gold standard for measurement of ankle strength [132, 187-191, 193-199]. For example, Araujo et al. [198] assessed passive ankle stiffness with the isokinetic dynamometer and demonstrated satisfactory validity and reliability by tests on 15 subjects. However, Aydog et al. [192] concluded that the validity of the measurement of ankle torque in inversion and eversion using a manufactured prototype ankle attachment device on the Kin Com II dynamometer was questionable. Although isokinetic dynamometers can be used for measuring ankle ROM, strength and stiffness, the main limitation is that ankle measurement and assessment is solely available in the sagittal plane.

**Innovative Techniques for Ankle Torque and Stiffness**

The ankle stiffness is an important mechanical parameter that indicates the moment required for rotation and the resistance to an external perturbation [228, 229]. Devices and techniques specifically to assess ankle stiffness have been developed. The working principles of most of these systems are similar. The system normally includes an actuator to generate rotation torque, a potentiometer to measure the angular displacement, and a torquemeter for measuring ankle moment. During tests, patients are usually required to move their ankles with different speeds and the ankle stiffness will be calculated as the derivative of torque over angular displacement. Several of these systems have been evaluated in clinical settings and the results showed that these types of devices can be a useful tool in the clinical assessment of ankle stiffness.

The Lidcombe template consisting of a spring balance and a perspex sheet ruled with parallel lines was adopted in [203, 205] to measure the angle and force in ankle dorsiflexion, showing
high reliability between testers. Lorentzen et al. [225] investigated the accuracy and reliability of a portable Neurokinetics RA1 Ridgidity analyser used for measuring ankle stiffness on 83 participants. Results showed that it could potentially be a useful diagnostic tool for measuring ankle stiffness although it was originally developed to test elbow rigidity. These devices require manual operation from physiotherapists and thus their applications when combined with robot-assisted therapy are restricted.

Direct ankle assessment techniques using the potentiometer and torquemeter are easier to apply during robot-assisted therapy. Two studies [134, 219] constructed a manual device to measure ankle joint ROM and stiffness on stroke patients and showed promising for clinical application, of which [219] presented an improved design based on the device in [24]. The manual spasticity evaluator (MSE) was used for the quantitative evaluation of ankle spasticity and stiffness in two studies [211, 220] that conducted clinical trials on six children with cerebral palsy, five typically developed children and five typically developed adults. The results showed that ankle spasticity assessment could be more accurately done using MSE. These two devices also need manual drive when measuring ankle stiffness but this limitation can be overcome easily using a motor or other actuators.

Ankle stiffness can also be determined by measuring torque and joint angles in a more sophisticated way. A bilateral electro-hydraulic actuator system with a rotary potentiometer and torque transducers was used in [207] to measure dynamic ankle joint stiffness during upright human stance. Sung et al. [218] used an intelligent stretching device for ankle stiffness measurement. They applied system identification techniques to characterize dynamic joint properties including joint stiffness, viscous damping and foot inertia properties during small-amplitude perturbations. Forty-six gender-matched healthy subjects participated in the trial and results showed this method was reproducible and consistent in ankle dorsiflexion and plantar flexion measurements. Chesworth and Vandervoort [200] demonstrated that the proposed ankle torque measurement unit consisting of a potentiometer and a strain gauge could be a useful tool in the clinical assessment of passive ankle stiffness.

Different from single-plane assessment, Giacomozzi et al. [208] developed a device for ankle kinematic and dynamics characteristics in three planes. This device measured the three-dimensional movement of the foot with respect to the shank and evaluated torques around the three articular axes based on the measured position and moment information from transducers. There are also some studies focusing on ankle stiffness in quiet standing or postural control. Loram and Lakie [206] used an inverted pendulum with a position transducer and a torque cell
Chapter 3. A Review of Robot-Assisted Ankle Assessment Techniques

for ankle stiffness measurement in quiet standing. Casadio et al. [212] used a device consisting of a motorised footplate mounted on a force platform for the direct of intrinsic ankle stiffness in quiet standing. Ji et al. [210] proposed a computational method to evaluate postural stiffness through ground reaction forces.

From a biomechanical perspective, quantitative assessment of ankle muscles and ligaments based on measured joint kinematics and dynamics is also necessary. Ankle stiffness is mainly determined by grouping all muscles and ligaments surrounding the joint in which the passive component is the result of their viscoelastic properties [230]. Unfortunately, there have only been a few studies, in which the muscular-skeletal properties of the ankle complex are considered. Naito et al. [226] applied a musculoskeletal structure with a Hill-type muscle model for calculating individual muscle length based on the data from the device used in [134, 219] and the results from four healthy subjects suggested its success. A kinematics based ankle model with major muscles and ligaments can be a promising approach to study passive ankle torque or stiffness and comparisons with traditional methods in terms of accuracy should be conducted.

These ankle stiffness measurement methods applied different measuring techniques and thus the applicable scopes varied. The majority of these studies were able to assess ankle stiffness only in the sagittal plane while a unique medial/lateral swaying cradle device in [217] was developed to measure inversion/eversion ankle stiffness with a satisfactory validity and reliability. More directly related to robot-assisted therapy, the Anklebot showed its potential to estimate ankle stiffness in three planes although only tests in dorsiflexion/plantarflexion have been conducted [215]. Another two limitations of these studies are the small sample size and the lack of reaction forces acting on the device.

In summary, ankle stiffness assessment techniques mainly consist of direct measurement using the potentiometer and torquemeter and an inverse dynamics based method to determine the kinematics using reaction forces as the inputs. Some studies [186, 198] used the hand-held dynamometer to estimate ankle stiffness. Table 3.2 is presented to analyse their prospect when used in robot-assisted therapy. Hand-held dynamometer based method (HD-BM) is subject to manual operation, to some extent which affects the measurement accuracy. Potentiometer and torquemeter based method (PT-BM) cannot be used in parallel robots due to the use of the torquemeter that is usually installed between the power producer and the load. In other words, three rotary potentiometers for the three-dimensional ankle assessment are required as well as three torquemeters. By contrast, the inverse dynamics based method (ID-BM) is promising when
combined with robot-assisted therapy. The three-dimensional ankle assessment using a six-axis load cell will be available but the validity and reliability need to be analysed prior to use.

Table 3.2: Prospect analysis of HD-BM, PT-BM and ID-BM when used in robot-assisted therapy.

<table>
<thead>
<tr>
<th>Methods</th>
<th>Advantages</th>
<th>Disadvantages</th>
<th>Prospect for use in robot-assisted therapy</th>
</tr>
</thead>
<tbody>
<tr>
<td>HD-BM</td>
<td>Simple</td>
<td>Manual operation</td>
<td>Poor</td>
</tr>
<tr>
<td>PT-BM</td>
<td>Reliable</td>
<td>Restricted by robot structure and mostly used in single plane</td>
<td>Reasonable</td>
</tr>
<tr>
<td>ID-BM</td>
<td>Three-dimensional assessment</td>
<td>Validity and reliability are not clear</td>
<td>Good</td>
</tr>
</tbody>
</table>


**Other Ankle Kinematics and Dynamics**

Although ankle ROM, strength, torque or stiffness has been measured quantitatively based on various devices and techniques, there are still some devices developed that can measure certain parameters not common in clinical application. For example, Knight and Weimar [224] developed a fulcrum device to measure dynamic inversion speed and the data from 26 participants showed high reliability for assessing maximum inversion and mean inversion speed.

In addition to these devices commonly used in laboratory setting, three studies [213, 216, 221] developed ambulatory measurement systems of foot and ankle kinematics and dynamics in the sagittal plane, of which two studies [213, 216] assessed instrumented shoes on healthy subjects and results showed good correspondence between the proposed system and the reference, and another study [221] also showed good reliability but with a different ambulatory ankle dynamics measurement system on 12 patients and ten healthy subjects. Additionally, a new instrumented spatial linkage (ISL) was used in [214] to measure dynamic ankle joint motion and data from an experienced snowboarder demonstrated its utility in a field environment. These ambulatory measuring techniques showed their potential in certain activities.

Identification methods are commonly used to estimate ankle mechanics (passive and active stiffness) based on measured information. Kearney et al. [201] used a system identification method to estimate ankle mechanics based on measured information from a potentiometer and a torquemeter. Tests on 15 young adults showed it had a good intrasubject variability and a somewhat larger intersubject variability. Mirbagheri et al. [135] described a parallel-cascade system identification method that had similar variability with [201] when used to determine the intrinsic and reflex contributions to dynamic ankle stiffness. Further, Weiss et al. [202] tried to
monitor the mechanical consequences of soft tissue injury based on joint dynamics and tests on three patients demonstrated its promise in clinical application.

To summarise, while Rome [231] concluded that ankle joint dorsiflexion assessment were controversial in terms of measurement accuracies due to different study designs, most quantitative ankle measuring techniques have proved to be reliable for a certain individual or group. But they were usually available for only ankle dorsiflexion and plantarflexion under passive motion in terms of kinematics and dynamics. Studies involving direct comparison in terms of reliability and validity among different devices and techniques are also lacking. Studies with less than 10 participants should be further validated with a larger sample size in future.

3.4.3 Ideal Measurement Device for Use with Robot-Assisted Therapy

An ideal system to evaluate functional improvement using robots would include the three-dimensional assessment in terms of kinematics and dynamics.

In general, goniometers and dynamometers have been used commercially and can be considered as standard tools for measuring ankle joint ROM and muscle strength, but are not used to measure other attributes of functional improvement after ankle injury. Both ROM and strength can readily be measured using robotic techniques. Dynamometers and goniometers should be used as the gold standard with which to compare when testing reliability and reproducibility of robot measurement techniques. For torque or stiffness measurements, a combination of a six-axis load cell and a parallel robot shows great potential for the three-dimensional ankle assessment during robot-assisted therapy. Further, a kinematics-based ankle model with major muscles and ligaments looks promising when used to study the passive components. A combination of the three-dimensional measurement and the model-based method allows for differentiation between the active and passive components.

Kinematic and kinetic parameters of measurement are diverse with no consensus as to device, technique, clinician expertise, or even whether to test passive or active motion. Studies involving direct comparison in terms of reliability and validity among different devices and techniques are important to better understand how these can predict function in the future. Further research should also focus on analysing real-time ankle muscle and ligaments parameters in all three planes based on measured ankle kinematics and dynamics information. A robot assessment technique can be developed that is consistent in all aspects of kinematic and kinetic measurement allowing for consistency for both before and after intervention and additionally among different patients.
3.4.4 Limitations of the Search Strategy

An attempt was made to include all studies related with ankle measuring techniques. It is assumed that these selected studies used different participants. Other publications may exist when foot, lower extremity or lower limb was identified as a key term instead of ankle. However, these may lead to potential incomplete search as well as some constrains like published dates and languages.

3.5. Conclusion of the Review

While most qualitative ankle assessment systems have been shown to be reliable, they are usually region-specific and disease-specific and thus a universally designed method with convincing validity and reliability is essential. Further, the assessment items usually involve functional ankle tests, questionnaire answers, as well as clinical ankle examination, which make the generation of immediate evaluation results (real-time monitoring from a robotic perspective) difficult.

Most quantitative ankle assessment techniques are reliable in measuring ankle kinematics and dynamics, but are usually available for only the sagittal plane. Limited studies determine kinematics and dynamics in all three planes where motions of the ankle joint and the subtalar joint actually occur [2]. Once these kinematics and dynamics are better understood, online modelling may allow for alteration of interventions or control strategies during robot-assisted therapy based on real-time ankle characteristics. In addition, these innovative ankle assessment devices were usually evaluated with no more than 30 participants and should be further validated with a larger sample size. Direct comparison among different devices and techniques for a specific ankle parameter should also be conducted to determine what could be the ideal effective ankle assessment tools in clinical environment.

3.6. Chapter Summary

This chapter presents a variety of ankle assessment techniques including both qualitative and quantitative methods, and discusses their potentials for use in robot-assisted therapy. It was found that qualitative ankle assessment methods are not suitable for real-time monitoring in robot-assisted therapy while quantitative methods using sensors show great potential for this application. While most quantitative techniques have been demonstrated to be reliable in measuring ankle kinematics and dynamics, they are mostly available for only the sagittal plane. Limited studies conducted assessment in all three planes where actual ankle motions exist. By
investigating the functionality and installation of quantitative ankle assessment techniques, it was identified that angular rotary sensors are suitable for the position measurement of the end effector when physical rotation axes of the robot exist, a six-axis load cell is ideal for the measurement of real-time patient-robot interaction (forces and torques). It was also found that ankle assessment via computational models can be used for estimating ankle ligament kinematics and joint torque. These information provides guidelines in implementing real-time ankle assessment during the robotic training.
Chapter 4. Development of an Intrinsically-Compliant AARR

This chapter develops an intrinsically-compliant ankle assessment and rehabilitation robot (AARR) that was devised based on two critical reviews (Chapters 2 and 3). The robot kinematics and dynamics are also presented, as well as its construction. The AARR is characterised by compliant actuation, compatible robot structure that allows the ankle to be stationary during the robotic training, three rotational degrees of freedom (DOFs) for three-dimensional exercises, real-time biomechanical assessment, and adaptive interaction training schemes. The function of real-time ankle assessment is also integrated using three magnetic rotary sensors measuring the position of the end effector and a six-axis load cell measuring patient-robot interaction.

This chapter begins with an overview of the design requirement of an ankle rehabilitation robot. Following it, the conceptual design of the AARR, kinematics and dynamics are successively presented. The AARR was finally constructed, including the robot structure and the control box. Partial materials presented in this chapter have been submitted for possible publication in IEEE/ASME Transactions on Mechatronics.

4.1. Design Requirements

4.1.1 Ankle ROM and Torque

Ranges of motion (ROMs) of ankle dorsiflexion/plantarflexion (DP), inversion/eversion (IE), and adduction/abduction (AA) was suggested by Siegler et al. [1] and would be considered as references in designing the workspace of the AARR, as summarised in Table 4.1. It can be seen that the ROMs in different directions are quite different, with around -45° to 25° in the sagittal plane, -15° to 20° in the front plane, and -20° to 25° in the transverse plane. However, since the robot is designed to cater for both the left and right ankles, different motion limits in ankle IE and AA will be inverted in the robot coordinate frame when a foot from the different side of the body is placed on the robot. The limits of the required robot workspace on the frontal and transverse planes are therefore symmetric.

Ankle torque has to be also considered on the design of the AARR, play a significant role in the control of the robot, especially for active training strategies. The maximum passive ankle torques obtained from 32 fresh human lower legs are summarised in Table 4.1 [232]. Results
from a study [201] with 15 subjects have shown that the required torque to drive an ankle from maximum plantarflexion to maximum dorsiflexion can reach up to 71.7Nm. This study also evaluated the torques produced by maximum voluntary contraction of the subjects and the corresponding values for dorsiflexion and plantarflexion are 54.4 Nm and 126.0 Nm, respectively. Unfortunately, maximum torque for ankle adduction/abduction is not available from the above studies. The AARR to be developed in this research was therefore designed by assuming that the maximum adduction and abduction rotation torques are similar in magnitude to the inversion and eversion torques. In general, the torque requirements of the AARR are set at 120 Nm for ankle dorsiflexion/plantarflexion, and 60 Nm for ankle inversion/eversion and adduction/abduction to meet the general use of human users.

<table>
<thead>
<tr>
<th>Type of motion</th>
<th>Ankle motions</th>
<th>Maximum PAT (Nm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Dorsiflexion</td>
<td>20.3° to 29.8°</td>
<td>24.68° ± 3.25°</td>
</tr>
<tr>
<td>Plantarflexion</td>
<td>37.6° to 45.75°</td>
<td>40.92° ± 4.32°</td>
</tr>
<tr>
<td>Inversion</td>
<td>14.5° to 22°</td>
<td>16.29° ± 3.88°</td>
</tr>
<tr>
<td>Eversion</td>
<td>10° to 17°</td>
<td>15.87° ± 4.45°</td>
</tr>
<tr>
<td>Adduction</td>
<td>22° to 36°</td>
<td>29.83° ± 7.56°</td>
</tr>
<tr>
<td>Abduction</td>
<td>15.4° to 25.9°</td>
<td>22.03° ± 5.99°</td>
</tr>
</tbody>
</table>

PAT: Passive ankle torque; SD: Standard deviation; NA: Not available.

### 4.1.2 Robot Flexibility

An optimal robot for ankle rehabilitation should be able to deliver a subject-specific training with suitable robotic ROM and torque capacity for a given patient. As can be seen from the systematic review of the effectiveness of existing ankle rehabilitation robots in Chapter 2, the robotic training can be passive, active-assist and active ROM exercises, as well as muscle strengthening schemes. Passive ROM exercises involve the robot guiding the patient’s foot through the predefined training trajectory when the patient’s foot remains relaxed. A relaxed ankle means that the participant does not actively engage with the robot. Active-assist ROM exercises on the other hand require the robot to cooperate with the patient to perform the predefined motion, providing certain assistance based on real-time ankle assessment. Active ROM exercises are conducted completely depending on the patient’s intention and ankle disability level, with the robot providing minimal interaction forces/torques. For muscle strength training exercises, the robot should be able provide certain resistance to the foot according to
the joint position and capacity to challenge the patient over time. Take all into consideration, an optimal ankle rehabilitation robot should be adjustable in providing certain robotic ROM and driven torque to meet a variety of training purposes. It should be noted here that the training mode is categorized on the basis of whether it is conducted under passive or active conditions. It is passive training if without any active contribution from the participant, active training if without any assistance from the robot, or training in a passive-active mode where the robot provides certain assistance but not enough for the training.

4.2. Conceptual Design

As what has been discussed in Chapter 1, an actuated-from-above parallel ankle robot (A-PAR) can be considered as the most suitable mechanism for ankle rehabilitation. The major benefit of such a structure design is that the robotic training can be delivered without requiring synergic movement of the lower limb of the patient due to compatible rotation centres of the robot and the ankle joint. The A-PAR mechanism, considered as a general parallel platform, has several advantages over their serial counterparts. First, the parallel robot has high positioning accuracy since errors in the actuated joints no longer accumulate as in the case of serial robots. Secondly, the robotic end effector is supported by multiple actuators, which allows for a higher load capacity of the mechanism. Lastly, as actuators of a parallel robot is located at the fixed platform instead of the moving links, the total load driven by the robot is also minimized. Take all into consideration, an A-PAR mechanism will be adopted in this research for ankle rehabilitation.

The proposed A-PAR for ankle therapy is expected to provide three rotational DOFs of ankle training for a variety of rehabilitation applications, and thus actuated in parallel by four Festo fluidic muscles (FFMs) that are intrinsically-compliant and light weight with superb force/weight ratio. The reason why the proposed ankle robot is redundantly actuated is that the FFM can only pull and cannot push, which means that n+1 actuators are required to achieve n-DOF motion [233]. Moreover, it was found that few ankle rehabilitation devices can conduct real-time ankle assessment to facilitate the robot control [17]. A variety of assessment techniques have been also reviewed by Zhang et al. [22] for use in robot-assisted ankle therapy. To incorporate the function of real-time ankle assessment into the proposed A-PAR, in this research, three magnetic rotary encoders are employed for the position measurement of the robotic end effector and a six-axis load cell for measuring patient-robot interaction. From the viewpoint of functionality, the proposed A-PAR is named the AARR.
A preliminary concept design of the AARR is presented in Figure 4.1. Specifically, the robot consists of an upper platform and a lower platform that are both fixed with respect to the ground. These two platforms are connected with the sliding rail that is further connected with the support leg. Out from the lower platform, there is a three-link serial manipulator whose end is a moving platform. To allow for the measurement of patient-robot interaction, a six-axis load cell is mounted between the end effector/footplate and the moving platform. The AARR can also be adjusted for use on different patients, as shown in Figure 4.2.

Figure 4.1: The concept design of the proposed AARR.
4.3. Robot Kinematics

The inverse kinematics of the proposed AARR is easy to be implemented and provides a unique solution of the length of the FFM for a given end effector pose. The kinematic geometry of the proposed AARR is presented in Figure 4.4, where the upper platform is fixed while the lower platform is a moving one that is rigidly attached on a three-DOF serial manipulator. The fixed coordinate frame for the fixed platform is denoted as $O_f$ and the moving one as $O_m$. Connection points on the fixed and moving platforms are denoted as $P_i^f$ and $P_i^m$ respectively. Their position vectors $P_i^f$ and $P_i^m$ are defined in (4.1) as well as the position vector $O_f O_m$. The position vectors $L_i^f$ of the actuators are described in (4.2), where the rotation transformation matrix of the end effector with respect to the fixed platform using a fixed axis rotation sequence of its orientation $\theta_x, \theta_y, \theta_z$ is denoted as $R_{m}^{f}$ in (4.3), and $R_{12}, R_{13}, R_{22},$ and $R_{23}$ are defined in (4.4).

\[
\begin{align*}
\mathbf{p}_i^f &= [x_i^f \ y_i^f \ 0]^T \\
\mathbf{p}_i^m &= [x_i^m \ y_i^m \ 0]^T \\
\mathbf{o} &= \mathbf{O}_f \mathbf{O}_m = [0 \ 0 \ -H]^T
\end{align*}
\]
Chapter 4. Development of an Intrinsically-Compliant AARR

Figure 4.3: Kinematic geometry of the AARR. (The red, blue and pink lines are denoted as axis-X, Y and Z, respectively, representing the rotation axes of ankle DP, IE and AA).

\[ L_i^f = O + R_m^f p_i^m - p_i^f \] (4.2)

\[ R_m^f = \begin{bmatrix}
\cos \theta_z \cos \theta_y & R_{12} & R_{13} \\
\sin \theta_z \cos \theta_y & R_{22} & R_{23} \\
-sin \theta_y & \cos \theta_y \sin \theta_x & \cos \theta_y \cos \theta_x \\
\end{bmatrix} \] (4.3)

\[
\begin{aligned}
R_{12} &= -\sin \theta_z \cos \theta_x + \cos \theta_z \sin \theta_y \sin \theta_x \\
R_{13} &= \sin \theta_z \sin \theta_x + \cos \theta_z \sin \theta_y \cos \theta_x \\
R_{22} &= \cos \theta_z \cos \theta_x + \sin \theta_z \sin \theta_y \sin \theta_x \\
R_{23} &= -\cos \theta_z \sin \theta_x + \sin \theta_z \sin \theta_y \cos \theta_x \\
\end{aligned} \] (4.4)

The Jacobian matrix is further derived in (4.5), where \( l_i^f \) represents the length of the \( i \)th actuator [9]. This matrix relates the link velocities and the twist of the end effector, and is extensively used for kinematics and dynamic analysis.

\[ J_i = \left( p_i^f \times \frac{L_i^f}{l_i^f} \right)^T \] (4.5)

Note that for any type of Euler angles, the angular velocity is not equal to the rate of change of the Euler angles. The angular velocity relates to the rate of change of the Euler angles as in (4.6), where \( E(\theta_x, \theta_y, \theta_z) \) is defined in (4.7) and its inverse matrix in (4.8) [234]. For late use, (4.6) can be also expressed in (4.9) and its differentiation in (4.10).
\[
\omega = E(\theta_x, \theta_y, \theta_z)[\dot{\theta}_x \quad \dot{\theta}_y \quad \dot{\theta}_z]^T \tag{4.6}
\]

\[
E(\theta_x, \theta_y, \theta_z) = \begin{bmatrix}
\cos \theta_y \cos \theta_z & -\sin \theta_z & 0 \\
\cos \theta_y \sin \theta_z & \cos \theta_z & 0 \\
-\sin \theta_y & 0 & 1
\end{bmatrix} \tag{4.7}
\]

\[
E^{-1}(\theta_x, \theta_y, \theta_z) = \frac{1}{\cos \theta_y} \begin{bmatrix}
\cos \theta_z & \sin \theta_z & 0 \\
-\cos \theta_y \sin \theta_z & \cos \theta_y \cos \theta_z & 0 \\
\sin \theta_y \cos \theta_z & \sin \theta_y \sin \theta_z & \cos \theta_y
\end{bmatrix} \tag{4.8}
\]

\[
\omega = E(\theta)\dot{\theta} \tag{4.9}
\]

\[
\dot{\omega} = E(\theta)\ddot{\theta} + \dot{E}(\theta)\dot{\theta} \tag{4.10}
\]

The kinematic configuration of the AARR used for experiments in this research is summarised in Table 4.2, by which the robotic ROM respectively for ankle dorsiflexion, plantarflexion, inversion, eversion, adduction and abduction can reach 36°. By considering normal ankle ROM presented in Table 4.1, it is found that the AARR provides enough workspace for the training of ankle IE and AA, not for DP. Fortunately, the AARR can generate varying robotic ROMs by changing the connection location of the FFMs. For example, the robotic ROM can reach up to 45° for both ankle dorsiflexion and plantarflexion if the absolute value of the Y coordinates of the lower connection points of the FFMs is changed to 40 mm. Other configurations regarding the adjustment of the posture of the AARR were set to make the robot suitable for a subject with the height of 170 cm.

<table>
<thead>
<tr>
<th>Robot kinematic configuration</th>
<th>Absolute values of coordinates (mm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Distance between the UP and the LP</td>
<td>X</td>
</tr>
<tr>
<td>Upper connection points of the FFMs</td>
<td>202.5</td>
</tr>
<tr>
<td>Lower connection points of the FFMs</td>
<td>65</td>
</tr>
</tbody>
</table>

UP: Upper platform; LP: Lower Platform; The upper connection points of the FFMs are located at the UP and the lower connection points of the FFMs are located at the moving platform.

### 4.4. Robot Dynamics

The dynamics of the AARR can be described by (4.11) in task space, where \( I(\theta) \) is the inertial matrix of the robot, \( G(\theta) \) represents the gravity term, \( C_f(\theta, \dot{\theta})\dot{\theta} \) is the friction term, \( d(\theta, \dot{\theta}, \ddot{\theta}, t) \)
is all other forces acting at the joints, $T_r$ is the torque generated by the robot, and $T_a$ is the ankle torque from the patient or the patient-robot interaction torque.

\begin{equation}
I(\theta)\ddot{\omega} + \omega \times I(\theta)\omega + C_f(\theta, \dot{\theta})\dot{\theta} + G(\theta) + d(\theta, \dot{\theta}, \ddot{\theta}, t) = T_r + T_a \tag{4.11}
\end{equation}

Substitute (4.9) and (4.10) into (4.11), as in (4.12), (4.13) can be obtained, where $M(\theta) = I(\theta)E(\theta)$, $C(\theta, \dot{\theta}) = I(\theta)E(\theta) + \ddot{\omega}I(\theta)E(\theta)$, $\ddot{\omega}$ is the skew symmetric matrix of $\omega$. The friction term $C_f(\theta, \dot{\theta})\dot{\theta}$ and other force interventions $d(\theta, \dot{\theta}, \ddot{\theta}, t)$ are considered very small and ignored in this research, thus the dynamic equation can be simplified into (4.14).

\begin{equation}
I(\theta)(E(\theta)\ddot{\theta} + E(\theta)) + \ddot{\omega}I(\theta)E(\theta)\theta + C_f(\theta, \dot{\theta})\dot{\theta} + G(\theta) + d(\theta, \dot{\theta}, \ddot{\theta}, t) \tag{4.12}
\end{equation}

\begin{align*}
= & I(\theta)E(\theta)\ddot{\theta} + (I(\theta)E(\theta) + \ddot{\omega}I(\theta)E(\theta))\dot{\theta} + C_f(\theta, \dot{\theta})\dot{\theta} + G(\theta) \\
+ & d(\theta, \dot{\theta}, \ddot{\theta}, t) = T_r + T_a \\
M(\theta)\ddot{\theta} + C(\theta, \dot{\theta})\dot{\theta} + C_f(\theta, \dot{\theta})\dot{\theta} + G(\theta) &= T_r + T_a \tag{4.13}
\end{align*}

\begin{equation}
M(\theta)\ddot{\theta} + C(\theta, \dot{\theta})\dot{\theta} + G(\theta) = T_r + T_a \tag{4.14}
\end{equation}

### 4.4.1 Ankle Force and Torque

Ankle forces and torques denoted as frame $a$ can be computed by transforming the forces and torques measured from a six-axis load cell denoted as frame $slc$ using a force/torque transformation [234], where $a$ and $slc$ represent the ‘ankle’ and the ‘six-axis load cell’, respectively. Specifically, ankle forces and torques can be calculated using (4.15) and (4.16), where $aR$ is a $3 \times 3$ rotation matrix from frame $a$ to frame $slc$, $P_{slc}^a \times aR$ is a $3 \times 3$ skew matrix from frame $a$ to frame $slc$.

\begin{align*}
\begin{bmatrix}
P_{a}^{R}_{slc}^3 \times 1 \\
T_{a}^{R}_{slc}^3 \times 1
\end{bmatrix} &= \begin{bmatrix}
P_{slc}^a \times aR \\
0_{3 \times 3}
\end{bmatrix} \begin{bmatrix}
P_{slc}^a \times aR \\
T_{slc}^a_{slc}^3 \times 1
\end{bmatrix} \tag{4.15}
\end{align*}

\begin{equation}
P_{slc}^a = \begin{bmatrix}
0 & -p_z & p_y \\
p_z & 0 & -p_x \\
-p_y & p_x & 0
\end{bmatrix} \tag{4.16}
\end{equation}

### 4.4.2 Inertial Property of the Moving Unit

The knowledge of the inertial property of the moving unit is crucial to the motion control of the AARR. As shown in Figure 4.1, the moving unit consists of the three-link manipulator, the six-
Chapter 4. Development of an Intrinsically-Compliant AARR

axis load cell and the end effector, of which the three-link serial manipulator are for ankle DP, IE and AA, respectively. These inertial parameters of the moving unit mainly include the mass, centre of mass (COM) and inertia tensor for each link. They were obtained from the Creo model presented in Figure 4.1 and summarised in Table 4.3. The global inertial tensor and gravitational effect of the moving unit can be calculated using (4.17) and (4.18), respectively, where \( I \) is the global inertial tensor, \( I_i \) is the inertial tensor of the \( i \)th link, \( R_i \) is the rotation matrix that relates the \( i \)th link coordinate frame to the global frame, \( O_i^{com} \) represents the coordinate of the COM of the \( i \)th link, and \( m_i \) is the mass of the \( i \)th link. The global coordinate system is located at the rotation centre of the end effector with the axis-X pointing to the right, the axis-Y pointing forwards and the axis-Z pointing upwards with respect to patients.

Table 4.3: Inertial parameters of the three-link moving platform.

<table>
<thead>
<tr>
<th>No.</th>
<th>Link</th>
<th>Mass (kg)</th>
<th>COM (mm)</th>
<th>IT at CS(_i) (Tonne mm(^2))</th>
</tr>
</thead>
</table>
| 1   | 1.61 | (0,0,-122)|          | \[
| 74.8 | 0  | 0        | 0  | 41.5 | 0        | 0  | 33.7 |
| 2   | 1.01 | (0,-133,-88)|         | \[
| 40.7 | 0  | 0        | 0  | 13.7 | -6.1     | 0  | 27.3 |
| 3   | 1.76 | (0,10.5,-119)|      | \[
| 29.4 | 0  | 0        | 0  | 27   | 1.8      | 0  | 5    |

COM: Centre of mass; IT: Inertia tensor; CS\(_i\): The coordinate system used in the Creo model of the \( i \)th link.

\[
I(\theta_x, \theta_y, \theta_z) = \sum_{i=1}^{3} I_i = R_i I_i R_i^T \tag{4.17}
\]

\[
G(\theta_x, \theta_y, \theta_z) = \sum_{i=1}^{3} G_i = R_i O_i^{com} m_i \tag{4.18}
\]

To derive the global inertia and gravitational effect of the moving unit, the Denavit-Hartenberg transformation matrix \( A_i^{i-1} \) has to be obtained as in (4.19) to relate the \( i \)th coordinate system to the \((i-1)\)th one. Since only rotations exist for the moving unit, \( A_i^{i-1} \) can be simplified into \( R_i^{i-1} \) as expressed in (4.20). However, four parameters \( \alpha_i, a_i, d_i, \) and \( \theta_i \) have to be identified before deriving the rotation matrix \( R_i^{i-1} \), which requires \((n+1)\) coordinate systems to be defined for an \( n \)-DOF manipulator [234]. Although the origins of the coordinate systems of link 0, link 1, link 2, link 3 and the global frame intersect at the centre of rotation of the end effector,
they are placed separately in Figure 4.4 to facilitate the description of defined coordinate frames. The specific values of the four parameters are presented in Table 4.4.

\[
A_i^{-1} = \begin{bmatrix}
\cos \theta_i & -\cos \alpha_i \sin \theta_i & \sin \alpha_i \sin \theta_i & a_i \cos \theta_i \\
\sin \theta_i & \cos \alpha_i \cos \theta_i & -\sin \alpha_i \cos \theta_i & a_i \sin \theta_i \\
0 & \sin \alpha_i & \cos \alpha_i & d_i \\
0 & 0 & 0 & 1
\end{bmatrix}
\]  
(4.19)

\[
R_i^{-1} = \begin{bmatrix}
\cos \theta_i & -\cos \alpha_i \sin \theta_i & \sin \alpha_i \sin \theta_i \\
\sin \theta_i & \cos \alpha_i \cos \theta_i & -\sin \alpha_i \cos \theta_i \\
0 & \sin \alpha_i & \cos \alpha_i
\end{bmatrix}
\]  
(4.20)

Figure 4.4: Coordinate systems of the moving unit of the AARR. (The red, blue and pink dotted lines represent axis-X, Y and Z, respectively).

Table 4.4: Four parameters used to relate different coordinate systems.

<table>
<thead>
<tr>
<th>Coordinate system</th>
<th>$\alpha_i$</th>
<th>$\alpha_i$</th>
<th>$d_i$</th>
<th>$\theta_i$</th>
</tr>
</thead>
<tbody>
<tr>
<td>Global</td>
<td>$\pi/2$</td>
<td>0</td>
<td>0</td>
<td>$\pi/2$</td>
</tr>
<tr>
<td>1</td>
<td>$\pi/2$</td>
<td>0</td>
<td>0</td>
<td>$\pi/2 + \theta_x$</td>
</tr>
<tr>
<td>2</td>
<td>$\pi/2$</td>
<td>0</td>
<td>0</td>
<td>$\pi/2 + \theta_y$</td>
</tr>
<tr>
<td>3</td>
<td>0</td>
<td>0</td>
<td>0</td>
<td>$\theta_z$</td>
</tr>
</tbody>
</table>

4.4.3 Force Distribution

Cable-driven robots may lose controllability if certain cables are not in tension during the robotic operation [235]. The AARR employs four FFMs that work similar to cables along an actuator, and thus the determination of feasible force distribution of this AARR is required for a given robot driven torque $T_r$. In this study, two potential force distribution methods are compared and analysed.
Pott et al. [236] proposed a closed-form force distribution scheme that is shown in (4.21), where
\[ A = J^T, \quad A^+ = A^T (A A^T)^{-1}, \quad T_r \] represents the robot torque in task space generated by four FFMs, 
\( f \) is the mean feasible force distribution and can be defined according to specific applications. 
The distributed forces are continuous along trajectories and differentiable at most of the points, 
and the computational efforts are also strictly bounded and small even for large numbers of wires.

\[ F = f - A^+ (T_r + Af) \]  

(4.21)

The other method is an analytic-iterative force distribution technique proposed by Taghirad et al. [237]. This method is formulated into an optimisation problem (4.22), where \( B = I - A^+ A, \) and then handled by the Karush–Kuhn–Tucker theorem to draw an analytic-iterative solution.

\[
\begin{align*}
\{ \min f(y) &= (F_o + By)^T(F_o + By) \\
\text{Subject to} \quad F_{\text{min}} - (F_o + By) &\leq 0 
\end{align*}
\]  

(4.22)

Since this approach adopted an optimisation search algorithm, the absolute sum of link forces can be smaller than that of the closed-form method, which means that the robot implemented using the analytic-iterative method can achieve a predefined motion with less energy consumption.

The closed-form method and the analytic-iterative one are evaluated by simulation in MATLAB. 
Both methods are applied to the developed AARR to achieve the same robot driven torque. To 
compare the force distribution performance of these two methods, three cases with different 
angular trajectories and torques are analysed for each method. Case 1: the angular trajectory in 
task space is defined as: \( \theta_x = 0.3 \sin(2\pi ft), \theta_y = 0, \theta_z = 0, \) and the defined interaction 
torque is defined as: \( M_x = 10, M_y = 0, M_z = 0; \) Case 2: the angular trajectory in task space is 
declared as: \( \theta_x = 0.3 \sin(2\pi ft), \theta_y = 0.2 \sin(2\pi ft), \theta_z = 0, \) and the defined interaction 
torque is defined as: \( M_x = 10, M_y = 0, M_z = 0; \) Case 3: the angular trajectory in task space is 
declared as: \( \theta_x = 0.3 \sin(2\pi ft), \theta_y = 0.2 \sin(2\pi ft), \theta_z = 0, \) and the defined interaction 
torque is defined as: \( M_x = f(t), M_y = 0, M_z = 0. \)
Figure 4.5: Force distribution results of (a) the closed-form method and (b) the analytic-iterative method. For each figure, the three plots from top to bottom represent the results of Case 1, 2 and 3, respectively.

The calculated force results of each link are presented in Figure 4.5. It is found that both methods can achieve satisfied force distribution results with appropriate link force levels for a given interaction torque along a predefined angular trajectory. However, these two methods present significantly different force distributions due to their inherent characteristics. For the closed-form method, the force results largely depend on the predefined mean force parameter \( f \) (set at 70 N in the simulation). With the increase of the required torque in task space, the force distributed to each link increased significantly and even result in large negative force for link 3 and 4 (Case 3), which is undesired for cable-driven robots. For the analytic-iterative approach, however, the link forces are generated based on the predefined minimal force \( f_{\text{min}} \) (set at 10 N in the simulation). Thus, all the forces are positive and every cable can be kept in tension, which is very important for cable-driven manipulators. Meanwhile, as the analytic-iterative approach is developed with an optimisation search algorithm, the absolute sum of link forces is much smaller than that of closed-form results, which means the robot controlled by analytic-iterative method is able to reach the required movement with much less energy consumption. On the other hand, the closed-form approach has less computation burden and numerical efforts compared to analytic-iterative approach. The calculation time of the closed-form method for 3000 samples is 0.1926s, 0.1905s, and 0.1875s for Case 1, 2 and 3, respectively, while the counterparts of the analytic-iterative method are 1.1647s, 1.3198s, and 1.3274s. Hence, the closed-form method is more suitable for robots with strict time bound. It should be noted in the
Chapter 4. Development of an Intrinsically-Compliant AARR

analytic-iterative method that the time for each sample is less than 0.0005s indicating that this scheme can meet the real-time calculation requirement for the control of AARR (the control loop operates with a period of 0.02s).

Three cases were compared and analysed with different angular positions and drive torques. These include (a) closed-form method, (b) analytic-iterative method. Figure 4.5 shows the results where both methods calculate continuous force distribution. These two schemes also meet the real-time requirement in the control system of the AARR, although the analytic-iterative method has a higher computation burden due to a maximum of 16 iterations. In the simulation of Figure 4.5(b), the computation time for 3000 samples in Cases 1, 2, and 3 is 1.1647s, 1.3198s, and 1.3274s respectively. The time for each sample is less than 0.0005s indicating that the analytic-iterative scheme can meet the real-time requirement (because the control loop operates with a period of 0.02s). However, these two methods have significant differences that may affect the robotic training safety. For the closed-form method, the calculated force largely depends on the predefined mean force determined by the desired joint torque. As the desired torque increases, the force distributed to each actuator increases accordingly and can result in negative values, potentially reducing robot controllability (Figure 4.5(a)). For the analytic-iterative method, the actuator forces are generated based on a predefined minimal force, so that all distributed forces are positive with their sum optimized to a minimum. Therefore, the analytic-iterative force distribution scheme was adopted

4.4.4 FFM Modelling

During the past a few decades, there has been an increase use of PAMs in the industrial and medical areas due to their advantages such as low power to weight ratio, high strength and light weight [238] and the demand for a simple but physically meaningful model for control purposes is rising [239]. Based on the summary by Tri et al. [239], available models are mostly approximations and no model has achieved a satisfactory contraction force prediction, which was also stated by Inoue [240] “it is nearly impossible to create a precise mathematic model”.

A recent survey conducted by Kelasidi et al. [241] summarised geometrical models of PAMs and demonstrated that Chou and Hannaford [242] model and Tondu and Lopez [243] model have been widely used for control applications. Chou and Hannaford [242] method is a static model based on trigonometric analysis but with assumptions like cylindrical shape and frictionless contact. Tondu and Lopez [243] added a dry-friction term to the contraction force to improve the modelling response, however, which is limited to the assumption that hysteresis
is caused by only the thread-on-thread dry friction between cords. Davis and Caldwell [244] developed a mathematical model to capture hysteresis due to friction between braid strands during muscle operation and verified it experimentally. However, these models give rise to complex problems in terms of control, since some parameters are difficult to be quantified. Some more complex models have been also developed recently. Andrikopoulos et al. [245] developed a dynamic model of PAMs based on a switch system approach. Xing et al. [246] proposed an Echo State Network as a basis to implement PMAs’ modelling and control, and demonstrated that the proposed procedure has better dynamic performance and strong robustness over other classical approaches.

The above-mentioned models were developed based on McKibben PAMs. In addition to the conventional advantages (light-weight and intrinsically-compliant behaviour) of McKibben PAMs, FFMs can generate a larger contraction force with the same size and contraction length, which makes them increasingly popular in robotic devices. FFMs follow well with the McKibben principle, and consists of a flexible cylindrical isotropic rubber tube with two fixed connection flanges at its ends. In between the rubber tube is a shell of non-contractive fibres. When the muscle is inflated with compressed air, it widens and a contraction force arises as well as a contraction movement in the longitudinal direction. The muscle can only generate contraction forces with its maximum at the beginning of the contraction and decreases as the contraction increases. Similar to McKibben PAMs, the FFM is highly nonlinear and its modelling is required for control purposes. Kerscher et al. [247] developed a dynamic model of fluidic muscles based on Tondu and Lopez [243] model and their own experience, but this method predicts less force in lower pressure range (<=2.5bar). Hosovsky and Balara [248] modelled Festo MAS-20 using an adaptive neuro-fuzzy inference system with data points extracted from the characteristic supplied by manufacturer. Szepe [249] conducted 340 measurements on a testbed using the Festo DMSP-20-400 type artificial muscle and modelled its static contraction force using an exponential function with six unknown parameters with 6.44N root-mean-square error (RMSE) in the whole operating range. The benefit of using this function is that only six unknown parameters result in fast and reliable parameter search. Furthermore, its inverse function is capable to express the pressure required to maintain a given contraction with a given force. This property of the function is necessary in an open-looped controlled system or generally, when it is not possible to take direct measurements for all three parameters but one of them can be expressed from the others. With reduced number of parameters, another approximation of the generated force of the FFM (Festo DMSP-20-400) was obtained by Sarosi [250] with the hysteresis considered.
Robot-assisted ankle rehabilitation exercises are considered to be slow and in quasi-static environment. Therefore, the function approximation proposed by Sarosi [250] is adopted in this research to derive the FFM contraction force based on its pressure and contraction strain, as shown in (4.23). These five parameters were experimentally obtained by changing the muscle length when the contraction force and pressure were recorded. Specifically, for the inflation process of the DMSP-20-400-RM-RM $a=232.89$, $b=-38.32$, $c=-904.01$, $d=294.86$ and $e=-289.06$ while for deflation $a=272.70$, $b=-32.58$, $c=-908.24$, $d=298.83$ and $e=-262.85$. The FFM strain is expressed as the ratio of the contraction length to the initial length. To facilitate its use in this research, a modified expression (4.24) is derived to calculate the desired FFM pressure based on the measured force and the desired strain. The specific relation among these three variables (force, pressure and strain) is also plot in Figure 4.6 for certain discrete pressure values. With the hysteresis loop neglected, the corresponding relation for continuous pressure is shown in Figure 4.7.

\[
F(p,k) = (p + a)e^{bk} + cpk + dp + e
\]  (4.23)

\[
p(k,F) = \frac{F - ae^{bk} - e}{e^{bk} + ck + d}
\]  (4.24)

![Figure 4.6](image-url): Static model of the FFM (DMSP-20-400N-RM-RM) for certain separate pressure values. (FFM Strain = 100*FFM contraction length/400).
4.5. AARR Construction

The physical picture of the developed AARR is presented in Figure 4.8 together with a three-dimensional Creo model in Figure 4.8(a) to show the robotic motion, and a photo of its use on a patient in Figure 4.8(c). The bill of materials needed for this AARR can be found in Appendix D. The AARR, as discussed earlier in this chapter, has three rotational DOFs that are actuated in parallel by four FFMs (Festo DMSP-20-400N) that are numbered in Figure 4.8(a). These three DOFs are respectively for ankle DP, IE and AA. This robotic design makes the AARR have bio-inspired design by mimicking the configuration and actuation of the ankle joint by natural muscles.

The control box of the AARR is shown in Figure 4.8(b) with all hardware included. To actuate the AARR, four proportional pressure regulators (Festo VPPM-6L-L-1-G18-0L6H) are used to control the pressure of the FFMs. The sensing components include three magnetic rotary encoders (ams AS5048A) for measuring angular positions of the robotic end effector, four single-axis load cells (Futek LCM 300) together with four amplifiers (Futek CSG110) for measuring contraction forces of FFMs, and a six-axis load cell (SRI M3715C) for measuring robot-patient interaction torque.

These electronic components including the actuation system and the sensing unit communicate with the embedded controller (NI Compact RIO-9022) through three independent modules (NI 9401, NI 9205 and NI 9263) for digital input/output (DI/O), analog input (AI) and analog output (AO), respectively. Additionally, the six-axis load cell communicates with the controller through the RS232 port using a customised interface box (SRI M8125). The connection layout of the control box is clearly presented in Appendix E with the physical picture in Figure 4.9(a).
and the connection block diagram in Figure 4.9 (b). To ensure the training safety of the patient, an emergency stop button is also set to prevent any further movement of the FFMs.

Figure 4.8: An intrinsically-compliant AARR. (a) The three-dimensional model of the AARR, where the red, blue and pink dotted lines represent axis-X, Y and Z, respectively; (b) The control box of the AARR; (b1) The interface of the control box; and (c) The AARR in use on a subject with ankle sprain. In (a), three rotational DOFs are for ankle DP, IE and AA, respectively. In (b1), connectors 1, 2, 3 and 4 are for a six-axis load cell, four pressure regulators, three magnetic rotary encoders and four single-axis load cells, respectively.

Figure 4.9: The control box of the AARR. (a) The physical picture of the inside of the control box; (b) The connection block diagram.
4.6. Chapter Summary

This chapter presents the development of an intrinsically-compliant AARR. Specifically, it involves the robot kinematics and dynamics, the FFM modelling, the force distribution that distributes a given robot torque in the task space into individual FFM force of the joint space, the calculation of ankle force and torque, and the construction of the AARR. These information provides the basis for the implementation of advanced interaction training schemes that will be presented in Chapters 8 and 9.

In general, the AARR devised from two literature reviews (Chapters 2 and 3) has advantages over other rehabilitation devices, including intrinsically-compliant actuation system, compatible robot structure that allows the ankle to be stationary during the robotic training, three DOFs for three-dimensional rehabilitation exercises, and the integration of real-time ankle assessment. By using the AARR, robot-assisted ankle rehabilitation can be more comprehensive, safe and effective.
Chapter 5. Ankle Biomechanical Assessment via Sensors

From previous chapters, real-time ankle assessment is essential and can be conducted using the ankle assessment and rehabilitation robot (AARR) through built-in sensors (three rotary sensors and a six-axis load cell). This chapter aims to validate the effectiveness of the proposed sensor-based assessment technique in measuring ankle position and interaction torque, providing experimental basis for its use on the AARR. A manual single-degree of freedom (DOF) device was developed with a rotary sensor for measuring ankle orientation and a six-axis load cell for measuring ankle torque/stiffness. It should be noted that this validation work was conducted before the AARR being constructed. It is assumed that this sensor-based technique is valid and reliable for three-dimensional ankle assessment if its effectiveness on a single-DOF device is demonstrated. The materials presented in this chapter has been published as [23] in Journal of Biomechanics.

This published paper goes as follows. The measurement of ankle orientation and stiffness can provide insight into improvements and allow for effective monitoring during a rehabilitation program. Existing assessment techniques have a variety of limitations. Dynamometer-based methods rely on manual manipulation. The use of torque meters is usually for single-DOF devices. This study proposes a novel ankle assessment technique that can be used for multi-DOF devices working in both manual and automatic modes using the position sensor and the multi-axis load cell. As a preliminary evaluation, an assessment device for ankle dorsiflexion/plantarflexion (DP) was constructed. Nine subjects participated to evaluate the effectiveness of the assessment device in determining ankle orientation and stiffness. The measured ankle orientation was consistent with those from the NDI Polaris optical tracking system. The measured ankle torque and stiffness compared well with published data. The test-retest reliability was high with intraclass correlation coefficient (ICC$_{2,1}$) values greater than 0.846 and standard error of measurement (SEM) less than 1.38.

5.1. Introduction

Ankle assessment in terms of range of motion (ROM) and joint stiffness on subjects with a variety of pathologies can assess and improve outcomes while monitoring a rehabilitation program [2]. The information allows the physiotherapist to understand ankle function and provides the basis for advanced robotic controls such as adaptive training. However, most ankle
rehabilitation robots do not conduct real-time assessment during training [17]. A number of ankle assessment devices for the use of robot-assisted therapy that have been developed over the past few decades were reviewed by Zhang et al. [22]. Hand-held methods based on dynamometer are effective in measuring ankle strength [185], but due to the requirement for manual operation they are impractical in robot-assisted ankle therapy. Additionally, there are torque meters for measuring joint torque that have shown potential in clinical trials on neurologically impaired patients [219, 220]. Torquemeter-based methods are subject to special mechanical structures that are usually single DOF devices. Since torquemeters have to be mounted between the power producer and the load, this technique is not suitable for parallel ankle robots [12, 251]. By contrast, an inverse dynamics-based ankle assessment method shows advantages for more extensive applications since the force plate or multi-axis load cell can be easily mounted on the robotic end effector [22]. This study proposed a novel assessment technique in measuring ankle orientation and stiffness by the use of angular potentiometers and six-axis load cells.

5.2. Methods

5.2.1 Participants

Nine healthy subjects volunteered to participate in this study, as summarised in Table 5.1. The study was approved by the University of Auckland, Human Participants Ethics Committee (9348) and consents were obtained from all participants (Appendix B).

<table>
<thead>
<tr>
<th>Participants</th>
<th>Gender</th>
<th>Age (years)</th>
<th>Height (cm)</th>
<th>Weight (kg)</th>
</tr>
</thead>
<tbody>
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<td>1</td>
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<td>23</td>
<td>173</td>
<td>80</td>
</tr>
<tr>
<td>2</td>
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</tr>
<tr>
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<td></td>
<td>25</td>
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<td>75</td>
</tr>
<tr>
<td>7</td>
<td>Female</td>
<td>25</td>
<td>165</td>
<td>61</td>
</tr>
<tr>
<td>8</td>
<td></td>
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</tr>
<tr>
<td>9</td>
<td></td>
<td>27</td>
<td>157</td>
<td>44</td>
</tr>
</tbody>
</table>

<table>
<thead>
<tr>
<th>Mean ± SD</th>
<th>25.50±1.64</th>
<th>178.67±8.16</th>
<th>79.83±13.82</th>
</tr>
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<tbody>
<tr>
<td>7</td>
<td>25</td>
<td>165</td>
<td>61</td>
</tr>
<tr>
<td>8</td>
<td>Female</td>
<td>30</td>
<td>160</td>
</tr>
<tr>
<td>9</td>
<td></td>
<td>27</td>
<td>157</td>
</tr>
</tbody>
</table>

| Mean ± SD   | 27.33±2.52 | 160.67±4.04 | 51.33±8.74  |

SD: Standard deviation.
5.2.2 Instrumentation

This ankle assessment device consists of a handle, a moving platform, a footplate rigidly fixed with an ankle orthosis and the base, as shown in Figure 5.1. Its sensing system consists of an angular potentiometer and a six-axis load cell. The angulation was based on an angular potentiometer with the neutral position (NP) considered as 0°. The ankle torque was measured using the six-axis load cell mounted beneath the footplate. This device was driven to work in a low-velocity environment (≤ 8°/s) to mimic clinical ankle rehabilitation exercises. However, two assumptions exist in calculating ankle torque: 1) no friction is introduced by the ankle orthosis and 2) the effects caused by different rotation speeds are negligible.

![Diagram of single-DOF ankle assessment device](image)

**Figure 5.1**: A single-DOF ankle assessment device.

To evaluate the measurement of ankle ROM, the Polaris tracking system was used. Four markers on the tibial crest and four on the foot shown in Figure 5.1 were used to generate a rigid-body representation to determine ankle position relative to the tibia. Its schematic diagram is shown in Figure 5.2, where $H$ is the distance from the tibial crest markers to the ankle joint, $h$ is the vertical distance from the ankle joint to the footplate, $L$ is the distance from the foot markers to the central line and $D$ represents the distance of two sets of markers. Solid lines represent the neutral position while dotted lines describe the position at a certain moment. The rotation angle, denoted as $\alpha$, equals zero in NP and can be calculated based on (5.1) and (5.2). Ankle forces and torques can be calculated using (4.15) and (4.16).
5.2.3 Experimental protocol

Subjects were instructed to sit on a height-adjustable chair with the hip and knee joints in 90° of flexion. The ankle-foot complex was fixed onto the footplate by an ankle orthosis. The handle of the device was driven by the experimenter starting from the NP, through maximum dorsiflexion, back to NP and maximum plantarflexion, and finally back to NP for a cycle. The experimenter practised for smooth operations prior to the tests. In actual experiments, -10° in ankle plantarflexion was chosen as the starting point to minimize the effects caused by the rotation velocity. Participants were verbally encouraged to relax their feet to minimize the effects by active contributions. Extremes of ankle DP depend on the sensation of the participants. The rotation was reversed when the tested feet attained a position closer to their manoeuvrability limits. For test-retest reliability analysis, the measurement was conducted five cycles with a three-minute interval on each subject.

5.2.4 Statistics

Six measures were analysed for test-retest reliability: maximum position (MP) in dorsiflexion/plantarflexion (MPD/P), peak passive torque (PPT) in dorsiflexion/plantarflexion (PPTD/P)
and passive stiffness (PS) in 20°/30° dorsiflexion (PS-20°/30°D). ICC\textsubscript{2,1} was analysed for test-retest reliability. Absolute reliability was determined by calculating the SEM and smallest real difference (SRD) with 95% confidence interval, as shown in (5.3) and (5.4) [252], where standard deviation (SD) is the mean SD of all measurements.

\[
SEM = SD\sqrt{1 - ICC} \quad (5.3)
\]

\[
SRD = SEM \ast 1.96 \ast \sqrt{2} \quad (5.4)
\]

### 5.3. Results

Results from the potentiometer showed close comparison with those from the NDI Polaris system, in Figure 5.3. However, comparison in ankle plantarflexion was not conducted since markers fixed on the foot were blocked away from the camera by the side aluminium extrusion. Figure 5.4 presents the angle-torque relationship in ankle DP on a participant. The stiffness was calculated as the slope of the fitted curve at two points, 20° and 30° in dorsiflexion (20°D and 30°D). The curve trends in Figure 5.4 compare well with the data by Sung et al. [218] with PS-30°D being 0.3315 (Nm/°) and PS-20°D being 0.2317 (Nm/°). Figure 5.5 plots the distribution of six measured parameters over five cycles, and the data for each participant can be found in Appendix A. Table 5.2 summarises the statistical results, where all ICC\textsubscript{2,1} values are greater than 0.846 and SEM values are less than 1.38.

![Figure 5.3](image.png)

**Figure 5.3:** Measurements of ankle position from the potentiometer and the Polaris system.
Chapter 5. Ankle Biomechanical Assessment via Sensors

5.4. Discussion

The comparison with the NDI Polaris system demonstrated that this device can accurately measure ankle position. This is based on the premise that no relative motion exists between the
footplate and the foot. The benefit of this method of measuring ankle joint position is that it is not limited to the use of angular potentiometers. It can also be alternative such as an inclinometer when a physical axis does not exist. For the measurement of ankle torque and stiffness, the data shown in Figure 5.4 are qualitatively consistent with published results [218, 220]. The measured ankle torque and stiffness was not quantitatively compared with published data due to subject-specific characteristics. A commercial or well-validated measurement device of ankle torque can be used to allow the subject-specific comparison. The test-retest reliability was considered high according to Munro's correlation description [253] with all ICC\(_{2,1}\) values greater than 0.846 and SEM less than 1.38.

### Table 5.2: Statistical results of the test-retest reliability.

<table>
<thead>
<tr>
<th>Indices</th>
<th>ICC(_{2,1})</th>
<th>Mean SDs</th>
<th>SEM</th>
<th>SEM%</th>
<th>SRD</th>
<th>SRD%</th>
</tr>
</thead>
<tbody>
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<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>MPD</td>
<td>0.846</td>
<td>3.01</td>
<td>1.18</td>
<td>3.02</td>
<td>3.27</td>
<td>8.38</td>
</tr>
<tr>
<td>MPP</td>
<td>0.958</td>
<td>6.71</td>
<td>1.38</td>
<td>2.77</td>
<td>3.81</td>
<td>7.64</td>
</tr>
<tr>
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<td></td>
<td></td>
<td></td>
<td></td>
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</tr>
<tr>
<td>PPTD</td>
<td>0.949</td>
<td>2.70</td>
<td>0.61</td>
<td>7.10</td>
<td>1.69</td>
<td>19.68</td>
</tr>
<tr>
<td>PPTP</td>
<td>0.858</td>
<td>1.61</td>
<td>0.61</td>
<td>6.44</td>
<td>1.68</td>
<td>17.75</td>
</tr>
<tr>
<td><strong>PS</strong></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>PS-20(^\circ)D</td>
<td>0.863</td>
<td>0.0669</td>
<td>0.0248</td>
<td>13.21</td>
<td>0.0686</td>
<td>36.54</td>
</tr>
<tr>
<td>PS-30(^\circ)D</td>
<td>0.865</td>
<td>0.1299</td>
<td>0.0477</td>
<td>16.95</td>
<td>0.1323</td>
<td>47.01</td>
</tr>
</tbody>
</table>

Note: SEM% and SRD% are calculated to facilitate the comparability with other studies or protocols as follows: SEM%=SEM/mean of the means of all trials, and SRD%=SRD/mean of the means of all trials.

The main advantage of the proposed ankle assessment technique is the use of a six-axis load cell, which allows applications in combination with various robotic devices. Hand-held dynamometers rely on manual operation and are not suitable for robotic applications. The proposed ankle assessment technique is compared with torque meter-based techniques regarding the issues of installation when integrated with robotic devices. Torque meter-based techniques are usually suitable for single-DOF ankle robotic devices. When multiple DOFs are introduced for ankle rehabilitation robots [12, 251], this proposed technique has advantages since the six-axis load cell can be easily mounted beneath the robotic end effector. The installation of the six-axis load cell for multi-DOF robotic devices is same as that shown in Figure 5.1.

### 5.5. Conclusion of the Paper

Although the use of a six-axis load cell allows for three-dimensional measurement, the performance of the proposed ankle assessment technique was only evaluated in ankle DP, which
can be considered as one limitation of this study. However, results from nine participants suggest that this ankle assessment technique is both valid and reliable even in a manual assessment environment. It has potential to be extensively used on a variety of robotic devices, especially on parallel robots with multiple DOFs for three-dimensional ankle assessment. Future studies will investigate the effectiveness of this technique when used for three-dimensional ankle assessment.

**5.6. Chapter Summary**

This chapter presents the evaluation of the proposed sensor-based assessment technique for measuring ankle position and ankle torque/stiffness. Experimental results from nine healthy subjects demonstrated its validity and reliability. For validity, the measured ankle orientation using the rotary sensor was consistent with those from the NDI Polaris optical tracking system, and the measured ankle torque/stiffness using the six-axis load cell compared well with published data. The test-retest reliability was high with ICC$_{2,1}$ values greater than 0.846 and SEM less than 1.38. To summarise, this sensor-based assessment technique is valid and reliable in measuring real-time ankle position and human-robot interaction torque, which supports its use on the AARR for real-time ankle assessment.
Chapter 6. Non-PRA-Model for Estimating Kinematics of Ankle Ligaments

This chapter develops and validates the Non-PRA-Model whose three rotation axes are not perpendicular (Chapter 1), and is driven by three rotary sensors built in the ankle assessment and rehabilitation robot (AARR) and outputs ligament kinematics of the ankle joint. The assessment of ankle ligament kinematics is crucial in understanding injury mechanisms and optimising training trajectories. This model also serves as the basis of the PRA-Model for estimating passive ankle torque (Chapter 9). It should be noted that model-based techniques can be used for assessment at the level of muscles and ligaments in-vivo. The materials presented in this chapter have been published as [24] in Computer Methods in Biomechanics and Biomedical Engineering.

This published paper goes as follows. Accurate assessment of ankle ligament kinematics is crucial in understanding injury mechanisms and can help improve the treatment of an injured ankle especially when used in conjunction with robot-assisted therapy. A number of computational models have been developed and validated for assessing the kinematics of ankle ligaments. However, few of them can do real-time assessment to allow for input into robotic rehabilitation programmes. A computational model was proposed and validated to quantify the kinematics of ankle ligaments as the foot moves in real-time. This model consists of three bone segments with three rotational degrees of freedom (DOFs) and 12 ankle ligaments. This model uses inputs for three position variables that can be measured from sensors in many ankle robotic devices that detect postures within the foot ankle environment and outputs the kinematics of ankle ligaments. Validation of this model in terms of ligament length and strain was conducted by comparing with published data from cadaver anatomy and magnetic resonance imaging (MRI). Model-based ligament lengths and strains compare well with those from published studies, although are sensitive to ligament attachment positions. This ankle computational model has the potential to be used in robot-assisted therapy for real-time assessment of ligament kinematics. The results provide information regarding the quantification of kinematics associated with ankle ligaments related to the disability level and can be used for optimizing the robotic training trajectory.
6.1. Introduction

Rehabilitation robots have been developed for the treatment of ankle sprains and include both wearable exoskeletons and platform-based devices [17]. These robots are designed to follow a reference trajectory which is a predefined motion path that corresponds to limb trajectories typically encountered during activities of daily living. These trajectories can be altered during the operation of the robot through application of impedance or admittance control strategies to maintain a certain relationship between the human-robot interaction and the motion tracking error [14]. However, little work has been done in optimizing robot-assisted rehabilitation trajectories for a specific injury based on real-time assessment of ankle ligament kinematics. This work seeks to identify kinematics of the ankle ligament which will then be used to analyse how different motion trajectories influence tension in ankle ligaments, also could contribute to the understanding of ankle function and injury mechanisms [2]. Also, the assessment of the kinematics of ankle ligaments contributes to the quantification of ligament related disability level, although a three-grade evaluation protocol has been generally accepted based on how many ligaments get injured [2].

A variety of assessment techniques were developed for measuring ankle range of motion (ROM) and stiffness over the past few decades and clinical evidences indicated their potential for use in robot-assisted therapy [22]. Kobayashi et al. [219] developed a manual device by using a potentiometer and a torquemeter for the measurement of ankle ROM and stiffness. Peng et al. [220] applied a similar technique to evaluate ankle spasticity on patients with neurological injuries. However, these techniques cannot be used for ankle assessment at the level of ligaments. Commonly-used techniques for ligament assessment consist of cadaver-based sectioning studies, invasive techniques and image-based methods [22]. Two studies [254, 255] examined ankle ligament strains on cadavers by sectioning. Beynnon et al. [256] used an in-vivo but invasive strain gauge to study anterior cruciate ligament strain behaviour during rehabilitation exercises. Asla et al. [257] adopted a combination of dual-orthogonal fluoroscopic and MRI technique for measuring lengths of anterior talofibular ligament (ATaFL) and calcaneofibular ligament (CaFL), but only in static foot positions. The main limitations of the studies so far are invasiveness and high-computational cost, which make them impossible for robot-assisted ankle therapy that requires real-time assessment.

Computational models can be used for ankle assessment at the level of ligaments. Two typical approaches are finite element analysis-based methods and multibody kinematics/dynamics-
based methods. However, multibody modelling can execute much faster than the continuum-based finite element analysis [258]. Multibody modelling methods are more suitable for real-time ligament assessment due to their highly efficient algorithms, although both have advantages for specific applications. Lindner et al. [259] developed an ankle model with lateral ligaments to simulate ankle sprains. Wei et al. [260] established an ankle model for determining dynamic ligament strains during external foot rotation. A more recent model by Wei et al. [261] utilized three DOFs to estimate ligament strains and joint moments for a supination sprain injury. These multibody modelling methods are provided real-time inputs from the motion analysis system allowing continuous updating, but at a significant cost.

The aim of this study was to develop and validate a computational ankle model to assess ligament kinematics in real-time for supporting robot-assisted therapy. The model provides real-time ankle ligament kinematic data, which can be used for determining and improving rehabilitation programmes. The hypothesis was that the lower limb is stationary during robot-assisted ankle therapy, which allows the fusion of the tibia and the fibula in the developed model. This hypothesis was made based on experimental protocols of ankle rehabilitation devices [12, 13, 94] in which the lower limb was fixed on a leg holder for robotic training.

### 6.2. Methods

The proposed computational ankle model consists of three bone segments with three rotational DOFs and 12 ligaments. It considers three independent variables as inputs, and they are three angular positions respectively for ankle dorsiflexion/plantarflexion (DP), inversion/eversion (IE) and adduction/abduction (AA). From this model individual ligament kinematics are output. This model was initially developed in OpenSim [262] by identifying each ligament based on the lower extremity model by Delp et al. [263]. Model expansion regarding the definition of three rotation axes and analysis were conducted in MATLAB. Figure 6.1 presents the proposed model in terms of bone structure, rotational DOFs, ligament modelling and input variables. The model consists of three bone segments (tibia/fibula, talus and calcaneus/toes) painted in different colours, three rotational axes \((A_{dp}, A_{ie}, \text{and } A_{aa})\) respectively for ankle DP, IE and AA, and 12 ankle ligaments. Three independent position variables are denoted as \(\theta_{dp}, \theta_{ie} \text{ and } \theta_{aa}\), respectively. The transformation from robotic measurements to joint positions is not focused in this study as well as its further application for evaluating dynamics of ankle joint and ligaments.
6.2.1 Model Formulation

Joint Kinematics

The definition of the global coordinate system of the ankle joint presented in Chapter 4 is adopted for kinematic analysis. For the specific calculation, the coordinate system used by Delp et al. [263] in the lower extremity model is adopted to facilitate data extraction. The lower extremity model includes DP at the ankle joint and IE at the subtalar joint. The ankle AA between the tibia and talus was introduced in this computational model, to account for a three-dimensional foot motion [2]. As shown in Figure 6.1, this model includes three segments with three rotation axes. $A_{dp}$ was defined at an angle of 80° and 84° against the sagittal plane and midline of the foot [264]. $A_{ie}$ was defined an inclination of 38° against the horizontal plane and 21° medial from the midline of the foot based on data from six fresh-frozen cadaveric lower leg specimens [265]. $A_{aa}$ vertically runs along the tibial axis. The orientations of three rotation axes were redefined in the form of unit vectors in (6.1).

$$\begin{bmatrix} A_{dp} \\ A_{ie} \\ A_{aa} \end{bmatrix} = \begin{bmatrix} 0.1045 & 0.1387 & -0.9848 \\ 0 & -1 & 0 \\ -0.7018 & -0.6157 & 0.3584 \end{bmatrix}$$ (6.1)
Three coordinate frames respectively denoted as tibia/fibula, talus and calcaneus were defined to describe foot motions. In Figure 6.1, the black, yellow and red coordinate frames represent the tibia/fibula frame $xyz_{tib}$, the talus frame $xyz_{tal}$ and the calcaneus frame $xyz_{cal}$, respectively, of which $xyz_{tib}$ is fixed. The fine line, the heavy line and dotted lines respectively represent coordinate axis $X$, $Y$ and $Z$, which is consistent with the definition in OpenSim. It should be noted that the orientation definition of the coordinate axes is different from that in Figure 1.1. The origin of $xyz_{tal}$ with respect to $xyz_{tib}$ was obtained by translation based on ankle anatomy and that of $xyz_{cal}$ by another translation based on $xyz_{tal}$, in which these two translation matrices are denoted as $T_{tib-tal}$ and $T_{tal-cal}$, as shown in (6.2), where $p$ represents the coordinate of a certain point in a certain coordinate frame and $R$ denotes the rotation matrix when the foot is moved to some position. Take pure ankle training in DP for example, the rotation matrix $R$ can be calculated based on (6.3).

\[
\begin{align*}
    p_{3\times1}^{tib} &= R_{3\times3}(\theta_{dp}, \theta_{ie}, \theta_{aa})p_{3\times1}^{tal} + T_{3\times1}^{tib-tal} \\
    p_{3\times1}^{tal} &= R_{3\times3}(\theta_{dp}, \theta_{ie}, \theta_{aa})p_{3\times1}^{cal} + T_{3\times1}^{tib-tal} + T_{3\times1}^{tal-cal} \\
    T_{3\times1}^{tib-tal} &= (0 \quad -430 \quad 0) \\
    T_{3\times1}^{tal-cal} &= (-48.77 \quad -41.95 \quad 7.92)
\end{align*}
\]  
(6.2)

\[
\begin{align*}
    R &= (N_1 + \cos \theta_{dp}(I_{3\times3} - N_1) + \sin \theta_{dp}N_2)^	op \\
    N_1 &= \begin{pmatrix}
        A_{dp}(1)A_{dp}(1) & A_{dp}(1)A_{dp}(2) & A_{dp}(1)A_{dp}(3) \\
        A_{dp}(2)A_{dp}(1) & A_{dp}(2)A_{dp}(2) & A_{dp}(2)A_{dp}(3) \\
        A_{dp}(3)A_{dp}(1) & A_{dp}(3)A_{dp}(2) & A_{dp}(3)A_{dp}(3)
    \end{pmatrix} \\
    N_2 &= \begin{pmatrix}
        A_{dp}(3) & 0 & -A_{dp}(2) \\
        0 & -A_{dp}(3) & A_{dp}(2) \\
        -A_{dp}(2) & A_{dp}(1) & 0
    \end{pmatrix}
\end{align*}
\]  
(6.3)

**Ligament Modelling**

There are seven extrinsic ligaments that control the foot, including medial collateral (deltoid) ligaments: TiNL, TiCL, ATiTL and PTiTL; lateral collateral ligaments complex: ATaFL, CaFL and PTaFL, were included in this model. Five intrinsic ligaments that originate, insert and function within the foot were also included and they are ITaCL, ATaCL, LTaCL, MTaCL and PTaCL, respectively [266]. All 12 ligaments are shown in Table 6.1 and Figure 6.1, in which ligaments are represented as linear elastic elements that can only apply a resistive force while in tension, and the origin/insertion locations were determined based on dissection and anatomical atlases [267, 268].
6.2.2 Model Validation

This model was validated quantitatively. The qualitative movement trends of ankle ligaments were analysed but no published data with which to compare these results were found. Quantitatively, ligament lengths and strains in certain positions were compared with published data to assess the accuracy of this model. The lengths of ankle ligaments depend on the origin and insertion locations on bones and therefore ligament lengths in certain positions can be considered as a measure for model validation. Many researchers have investigated ligament lengths in the neutral position (NP) especially for lateral collateral ligaments through either cadavers sectioning or image-based methods like MRI, as summarised in Table 6.2. Combined statistics were determined based on [269] by including the effect of sample size. It is hypothesized that the data with the sample size no less than 20 are reliable and would be considered as a baseline for validating this computational model. Additionally, Asla et al. [257] measured the lengths of ATaFL and CaFL at maximum dorsiflexion and plantarflexion, which also provides a reasonable measure to validate this model although positions were not specified. Further, an experimental cadaver study performed by Colville et al. [254] investigated the strains of lateral collateral ankle ligaments in dorsiflexion and plantarflexion. Ankles were moved from 20° of dorsiflexion to 30° of plantar flexion while no forces were applied in ankle IE and AA.
Strain was measured continuously in each ligament throughout this ROM. Subtalar joint motion was eliminated in this experimental study by use of two screws passing through calcaneus into talus. To mimic the experimental foot constraint condition, only movement in dorsiflexion and plantarflexion was allowed in this model. Continuous dorsiflexion and plantarflexion was applied to the foot and ligament strains in ATaFL, CaFL and PTaFL simulated from the proposed model were compared to published data. Results involving anterior tibiofibular and posterior tibiofibular were not used for comparisons because the relative motion between tibia and fibula was eliminated in this developed model. To further validate this model, the effects of variations in ligament positions on simulation outcomes were analysed.

Table 6.2: Published lengths of ankle ligaments.

<table>
<thead>
<tr>
<th>Studies</th>
<th>Length (mm) (Means ± S Ds)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>ATaFL</td>
</tr>
<tr>
<td>Siegler et al.</td>
<td>17.81±3.05</td>
</tr>
<tr>
<td>Ozeki et al.</td>
<td>19.80±1.92</td>
</tr>
<tr>
<td>Mkandawire et al.</td>
<td>18.89±2.97*</td>
</tr>
<tr>
<td>Taser et al.</td>
<td>22.37±2.50</td>
</tr>
<tr>
<td>Asla et al.</td>
<td>16.3±3.0</td>
</tr>
<tr>
<td>Combined</td>
<td>20.40±3.28</td>
</tr>
</tbody>
</table>

* represents the sample size is 5; SS represents sample size and values in bold and italic represent reliable results that would be regarded as the baseline for model validation.

6.3. Results

6.3.1 Qualitative Results

The lengths of ligaments spanning the ankle joint are influenced by DP and AA, and five intrinsic ligaments are subject only to ankle IE. The movement tendency of these ligaments in DP, IE and AA is summarised in Table 6.3, where MP represents maximum plantarflexion and MD represents maximum dorsiflexion, MAB represents maximum abduction and MAD represents maximum adduction, ME represents maximum eversion and MI represents maximum inversion, NP represents neutral position in any plane.
Table 6.3: The tendency of movement of ankle ligaments as the foot moves.

<table>
<thead>
<tr>
<th>Ligaments</th>
<th>MP-NP</th>
<th>NP-MD</th>
<th>MAB-NP</th>
<th>NP-MAD</th>
<th>ME-NP</th>
<th>NP-MI</th>
</tr>
</thead>
<tbody>
<tr>
<td>ATaFL</td>
<td>♦</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>PTaFL</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>ATiTL</td>
<td>♦</td>
<td>♦</td>
<td>♦</td>
<td>♦</td>
<td></td>
<td></td>
</tr>
<tr>
<td>PTiTL</td>
<td></td>
<td>♦</td>
<td>♦</td>
<td>♦</td>
<td></td>
<td></td>
</tr>
<tr>
<td>CaFL</td>
<td></td>
<td>♦</td>
<td>♦</td>
<td>♦</td>
<td>♦</td>
<td>♦</td>
</tr>
<tr>
<td>TiNL</td>
<td></td>
<td></td>
<td>♦</td>
<td>♦</td>
<td></td>
<td></td>
</tr>
<tr>
<td>TiCL</td>
<td></td>
<td></td>
<td></td>
<td>♦</td>
<td>♦</td>
<td>♦</td>
</tr>
<tr>
<td>ITaCL</td>
<td>—</td>
<td>—</td>
<td>—</td>
<td>—</td>
<td>♦</td>
<td></td>
</tr>
<tr>
<td>ATaCL</td>
<td>—</td>
<td>—</td>
<td>—</td>
<td>—</td>
<td>—</td>
<td>♦</td>
</tr>
<tr>
<td>LTaCL</td>
<td>—</td>
<td>—</td>
<td>—</td>
<td>—</td>
<td>—</td>
<td>♦</td>
</tr>
<tr>
<td>MTaCL</td>
<td>—</td>
<td>—</td>
<td>—</td>
<td>—</td>
<td>—</td>
<td>♦</td>
</tr>
<tr>
<td>PTaCL</td>
<td>—</td>
<td>—</td>
<td>—</td>
<td>—</td>
<td>—</td>
<td>♦</td>
</tr>
</tbody>
</table>

♦ represents length increases, ♦ represents length decreases, ♦ represents small variations, — indicates no change in length, * indicates strain ≥30%.

6.3.2 Quantitative Results

Ligament Length: Table 6.2 summarised published lengths of ankle ligaments in NP and combined the means and standard deviations (SDs) from multiple studies. Five ligaments (ATaFL, CaFL, PTaFL, PTiTL and TiNL) were used for model validation based on the predefined inclusion criterion. In Figure 6.2, all modelling results fall within the combined means and SDs.

Figure 6.2: Lengths of ATaFL, CaFL and PTaFL, PTiTL and TiNL in NP.
Model-based lengths of ATaFL and CaFL in extreme positions are consistent with those obtained using the MRI technique, as shown in Figure 6.3, where it is assumed that -30° of plantarflexion and 30° of dorsiflexion are consistent with the extreme positions.

**Figure 6.3:** Lengths of ATaFL (left) and CaFL (right) in extreme ankle DP.

**Ligament strain:** Ligament strain changes versus foot movement were also compared with published data. Colville et al. [254] measured strains in lateral collateral ligaments of human ankle using a Mercury-filled Silastic strain gauges that gave zero or positive values. Further, the situation when ligament strains are negative does not make any contribution to joint kinematics and dynamics as ligament is considered as an elastic element. Figures 6.4, 6.5 and 6.6, therefore, only compared the stages with positive strains. While model-based results and linear fitting are consistent with the trend, it is difficult to be confident with any quantitative measure of the model’s accuracy due to the small size of the data from the literature.

**Figure 6.4:** Strain of ATaFL in ankle DP.
6.3.3 Sensitivity Analysis

To analyse the effects of ligament attachment positions, the origin of ATaFL was moved from -3mm to 3mm relative to its initial position respectively along axis-X, Y and Z, a three-dimensional change axis-X, Y and Z at the same time with each -3mm to 3mm was also conducted. Figure 6.7 plots the length variations of ATaFL through MP to MD for each change, where IN represents initial ligament attachment position. Figure 6.8 shows the Means and SDs but only in three important angular positions (MP, NP and MD). Figure 6.9 shows the Means
and SDs of the strains among seven different attachment points in MP and MD since values in NP keep zero. It is assumed that MP is -30° of plantarflexion and MD is 30° of dorsiflexion, as mentioned when comparing ligament lengths. In a similar way, the origin of each ligament was moved from -3mm to 3mm along only axis-x and the corresponding sensitivity is summarised in Table 6.4.

**Figure 6.7**: Sensitivity of lengths of ATaFL in ankle DP.

**Figure 6.8**: Means and SDs of the lengths of ATaFL among seven attachment points.
Figure 6.9: Means and SDs of the strains of ATaFL among seven attachment points.

Table 6.4: Sensitivity of lengths of 12 ligaments in ankle DP to x-axis position.

<table>
<thead>
<tr>
<th>Ligaments</th>
<th>ATaFL</th>
<th>PTaFL</th>
<th>ATiTL</th>
<th>PTiTL</th>
<th>CaFL</th>
<th>TiNL</th>
<th>TiCL</th>
<th>ITaCL</th>
<th>ATaCL</th>
<th>LTaCL</th>
<th>MTaCL</th>
<th>PTaCL</th>
</tr>
</thead>
<tbody>
<tr>
<td>Sensitivity</td>
<td>10.44</td>
<td>7.76</td>
<td>21*</td>
<td>20.89*</td>
<td>13.51</td>
<td>8.5</td>
<td>1.9</td>
<td>7.85</td>
<td>20.44*</td>
<td>8.15</td>
<td>12.35</td>
<td>38.3*</td>
</tr>
</tbody>
</table>

Sensitivity was calculated by \( \frac{L_{\text{Max}} - L_{\text{Min}}}{L_{\text{NP}}} \) \* 100% and *represents significant ligament length variations by origins’ changes along x-axis from -3mm to 3mm.

6.4. Discussion

For evaluation and rehabilitation of ligament related ankle injuries, it is generally accepted that the disability level can be divided into three grades based on how many ligaments get injured [2]. This proposed model aimed to assess the kinematic behaviour of ankle ligaments as the foot rotates during robot-assisted rehabilitation exercises, which could contribute the determination of the sprain level and the optimisation of the robotic rehabilitation programmes. Although this model was proposed for use in robot-assisted therapy, its predefined rotational DOFs do not exactly match those of existing ankle rehabilitation robots [12, 13] where three rotational DOFs are perpendicular to each other. Transformation from robotic measurements to position variables required to drive the model is necessary. An example to transfer robotic posture description to anatomical ankle joint description has been suggested by Tsoi and Xie [273].

This model was initially developed using OpenSim and expanded in MATLAB. The validation of this model occurred as follows. A number of comparisons with published data from cadaver
anatomy or MRI were conducted to evaluate the effectiveness of this model for simulating the kinematics of ankle ligaments. In Figure 6.2, the results simulated from the computational model show a satisfactory accuracy in modelling ankle ligaments by comparing ligament lengths with the average values of published data in NP. Also, the model-based lengths of ATaFL and CaFL in MP and MD were demonstrated acceptable by comparing them with those obtained using MRI, as shown in Figure 6.3. As for the evaluation of this model in terms of ligament strain, it is found from the simulation in OpenSim and MATLAB that strain in the ATaFL increases with ankle plantarflexion and decreases with dorsiflexion, which was in agreement with the data from 10 human cadaver ankles [254] and four ankles of healthy male subjects [257]. Opposite to the ATaFL, simulation also shows that CaFL strain increases with dorsiflexion and decreases with plantarflexion, which corresponded with the results from experimental studies [254, 257]. The results also confirmed the reciprocal function between ATaFL and CaFL [257]. The different elongations of ATaFL and CaFL during the same motion show that in robot-assisted ankle therapy the former may get better rehabilitation training in plantarflexion whereas CaFL is strengthened more effectively in dorsiflexion. It was further found from the model simulation that extreme dorsiflexion and external rotation of talus increased PTaFL strain, which is consistent with the experience of frequently finding tenderness in the PTaFL area in patients with dorsiflexion or external rotation ankle injuries as well as experimental results by Colville et al. [254].

To analyse the effects of ligament attachment positions on evaluation results, sensitivity analysis was conducted in this study by 1): individually varying the origins of ATaFL along the axis-X, Y and Z with ligament length changed accordingly and 2): a three-dimensional change along axis-X, Y and Z at the same time. While effects caused by different attachment points present significantly different, as shown in Figures 6.8 and 6.9, ligament length/strain variations caused by axial movement along axis-X were not as obvious as others. For all 12 ankle ligaments, the lengths of ATiTL, PTiTL, ATaCL and PTaCL are more sensitive to the X-axial movement of their origins, as shown in Table 6.4. However, these differential effects may depend on ligament-specific locations and task-specific joint motions.

Limitations of the proposed model for ankle ligament kinematics assessment in robot-assisted should be noted. The bone geometry in this study was based on a generic ankle-foot complex model of the lower extremity model. Scaling based on this generic model for subject-specific adaptation would contribute to improve the simulation precision. Lewis et al. [274] suggested that the ankle kinematic model with varying rotation axes as the joint rotates would provide a
better description of the foot motion. Thus the issue of fixed axes use in the proposed model can be considered as one limitation. Ligaments being represented as straight lines between origin and insertion points also introduce some simulation errors. Origins and insertion points of these included ligaments were specified based on published anatomical data and the model precision could be improved with subject-specific ligament locations by image-based methods such as MRI. It should be noted here that ligament locations specified by image-based techniques is accurate even in an abnormal state. Future work will focus on developing this model as an in-vivo ankle torque prediction tool in robot-assisted therapy. Although a computational ankle model by Wei et al. [275] was demonstrated to be effective in estimating joint torque by modelling ligaments as linear spring elements, it cannot be used for in-vivo evaluation of joint torque since muscles around the ankle joint were not included. To summarise, future work will extend this model by incorporating muscle modelling and kinematics, ligament and muscle passive properties, muscle activation level, subject-specific adaptation.

6.5. Conclusion of the Paper

This computational ankle model has three rotational DOFs and can fully describe foot motions. The characteristics of three position variables required as the inputs and highly efficient algorithm give it advantages over other methods when combined with robot-assisted ankle therapy. The results from this model show that the kinematics of ankle ligaments compared well with published data. This suggests that the model could be used to analyse ligament kinematics as the foot rotates. This proposed model has the potential to identify length changes of individual ankle ligament that are usually required for quantification of ankle ligament related disability level. These quantitative evaluation results could also provide information for physiotherapists and robots in optimizing rehabilitation programmes.

6.6. Chapter Summary

The Non-PRA-Model was proposed and validated in this chapter. This model uses three position variables that can be measured from sensors built in the AARR as inputs and outputs the kinematics of ankle ligaments. It can be used to quantify the kinematics of ankle ligaments during the robotic training. It also provides the basis for its further development in estimating passive ankle torque, which will be introduced in Chapter 7. It should be noted that the structure of the Non-PRA-Model does not coincide with that of the AARR. The AARR has three perpendicular rotation axes which create a rotation centre while the rotation axes of the model are not perpendicular. To apply this model in conjunction with the AARR, it is assumed that the
ankle joint and subtalar joint orientations of the model respectively approximate the axis-X and -Y of the AARR, or a transformation can be conducted based on the measured orientation of the robotic end effector [273].
Chapter 7. PRA-Model for Estimating Passive Ankle Torque

This chapter presents the PRA-Model by modifying and updating the Non-PRA-Model (Chapter 6). Different from the Non-PRA-Model in estimating ligament kinematics of the ankle joint, the PRA-Model can predict passive ankle torque in subject-specific. The estimation of passive ankle torque can help to understand the mechanical characteristics of the joint. It will be also used to facilitate the implementation of adaptive interaction training on the ankle assessment and rehabilitation robot (AARR). The materials presented in this chapter have been published as [25] in IEEE Transactions on Biomedical Engineering.

These two models are distinguished by their rotation axes: the PRA-Model has perpendicular rotation axes while the three rotation axes of the Non-PRA-Model are not perpendicular. Before introducing the PRA-Model, it is worth using a chart to summarise the similarities and differences between the PRA-Model and the Non-PRA-Model as well as the behind reasons, as in Table 7.1. Two main points: the rotation axes of the Non-PRA-Model are not perpendicular, while the PRA-Model assumes that the ankle rotates about a pivot or about three perpendicular rotation axes; different force elements (FEs) also exist in that the Non-PRA-Model included 12 ligaments while the PRA-Model includes 12 muscles and seven ligaments.

<table>
<thead>
<tr>
<th>Models</th>
<th>Non-PRA-Model</th>
<th>PRA-Model</th>
</tr>
</thead>
<tbody>
<tr>
<td>DOFs</td>
<td>3</td>
<td>3</td>
</tr>
<tr>
<td>Model motion</td>
<td>Three rotation axes that are not orthogonal, with ankle DP and AA between the tibia and the talus, ankle IE between the talus and the calcaneus.</td>
<td>Three orthogonal rotation axes that are located between the tibia and the talus, which is consistent with the structure of the AARR.</td>
</tr>
<tr>
<td>Muscles</td>
<td>No muscles</td>
<td>12 muscles</td>
</tr>
<tr>
<td>Ligaments</td>
<td>12 ligaments</td>
<td>7 ligaments included since these connecting the talus and the calcaneus are excluded due to the model structure.</td>
</tr>
<tr>
<td>Subject-specific adaptation</td>
<td>No</td>
<td>Yes</td>
</tr>
<tr>
<td>Function</td>
<td>Estimating ligament kinematics</td>
<td>Predicting Passive ankle torque</td>
</tr>
<tr>
<td>Contribution</td>
<td>Optimising the robotic training trajectory and providing guidelines in the development of the model of Chapter 7.</td>
<td>Facilitating the implementation of advanced interaction training strategies on the AARR.</td>
</tr>
</tbody>
</table>
This published paper goes as follows. Robot-assisted ankle assessment could potentially be conducted using sensor-based and model-based methods. Existing ankle rehabilitation robots usually use torquemeters and multi-axis load cells for measuring joint dynamics. These measurements are accurate, but the contribution as a result of muscles and ligaments is not taken into account. Some computational ankle models have been developed to evaluate ligament strain and joint torque. These models do not include muscles, and thus are not suitable for an overall ankle assessment in robot-assisted therapy. This study proposed a computational ankle model for use in robot-assisted therapy with three rotational degrees of freedom (DOFs), 12 muscles and seven ligaments. This model is driven by robotics, uses three independent position variables as inputs, and outputs an overall ankle assessment. Subject-specific adaptations by geometric and strength scaling were also made to allow for a universal model. This model was evaluated using published results and experimental data from 11 participants. Results show a high accuracy in the evaluation of ligament neutral length and passive joint torque. The subject-specific adaptation performance is high, with each normalised root mean square deviation (NRMSD) value less than 10%. This study concludes that the PRA-Model could be used for ankle assessment, especially in evaluating passive ankle torque for a specific individual. The characteristic that is unique to this model is the use of three independent position variables that can be measured in real-time as inputs, which makes it advantageous over other models when combined with robot-assisted therapy.

7.1. Introduction

Robots have been developed for the treatment of various ankle injuries over the past decades. Wearable devices are usually aimed at ankle rehabilitation during gait exercises, while platform robots can solely conduct ankle training [10-12, 34, 85, 276]. Zhang et al. [17] systematically reviewed various ankle rehabilitation devices and demonstrated their effectiveness in clinical applications. It was concluded that few existing robots conducted real-time ankle assessments to control the robot. The knowledge of ankle kinematics and dynamics may allow for the alteration of control strategies of robot-assisted therapy based on real-time ankle performance.

Ankle assessment is essential in robot-assisted therapy. Zhang et al. [22] reviewed both qualitative and quantitative assessment techniques. They concluded that most quantitative assessment techniques are reliable in measuring ankle kinematics and dynamics, but are usually only available for the sagittal plane. Few studies conducted ankle assessments in a three-dimensional space where foot motion actually occurs [2]. Most ankle rehabilitation devices use
torquemeters and multi-axis load cells for measuring joint dynamics [21, 220]. These measurements are accurate but expensive. Further assessment at the level of muscles and ligaments is also lacking, although Naito et al. [226] conducted the identification of ankle muscle length using inverse kinematics.

Ankle assessment at the level of muscles and ligaments usually consists of cadaver-based sectioning studies [254, 255], invasive techniques [256] and image-based methods [257]. Cadaver-based and invasive methods are not suitable for rehabilitation. The high computation burden of image-based techniques limits their applications in real-time robotic environment. In a better way, some computational models have been developed for ankle assessment at the level of muscles and ligaments. Liacouras and Wayne [258] proposed a computational approach to model the lower leg to study syndesmotic injury and ankle inversion stability. This model simulated ligament sectioning experiments and actual foot motion was not fully represented. Lindner et al. [259] created an ankle model, but with only lateral ankle ligaments included. Wei et al. [260] established a computational model for determining dynamic ankle ligament strains by inputting kinematic data from a six-camera Vicon MX Capture System. The issues of set-up and time-consuming analysis process impede its application in robot-assisted therapy. More recently, O'Shea and Grafton [182] revised this model with three DOFs at two joints, in which the rotation axis of ankle inversion/eversion locates between the talus and the calcaneus. This model could provide ligament strains and joint moments but was only validated for ankle supination. In general, these evaluated models have been demonstrated to identify the kinematics and dynamics of the foot-ankle complex for some specific applications. However, these models do not include muscles and thus are not suitable for an overall ankle assessment in robot-assisted therapy.

The objective of this study was to develop and validate a novel computational ankle model for use in robot-assisted therapy. This robot-driven model uses three independent position variables as inputs while outputting an overall ankle assessment (kinematics and dynamics of ankle joint, muscles and ligaments). It was developed to allow for subject-specific results by scaling a general musculoskeletal model to enable the application to most of the population. This chapter mainly focuses on the establishment of the ankle model, subject-specific adaptation and model validation.
7.2. Methods

A computational ankle model with three rotational DOFs and 19 FEs was constructed for use in robot-assisted therapy. These 19 FEs include 12 muscles and seven ligaments. Subject-specific adaptations by geometric and strength scaling were made based on participants’ height and weight to enable the application to most of the population. This robot-driven model considers three independent position variables as inputs, and outputs the kinematics and dynamics of individual muscle/ligament and the ankle joint, as shown in Figure 7.1. Two significant assumptions are: 1) ankle motion exists at a single joint between the tibia/fibula unit and the talus/calcaneus. The tibia/fibula is fixed during robot-assisted therapy while the talus/calcaneus rotates in three rotational DOFs; and 2) the rotation axes of ankle dorsiflexion/plantarflexion (DP), inversion/eversion (IE) and abduction/adduction (AA) intersect at one point to facilitate its use in robot-assisted therapy.

![Figure 7.1](image)

**Figure 7.1**: A robotics-driven computational ankle model with muscles and ligaments for assessment with subject-specific adaptation. (The robot could drive this computational ankle model using three angular potentiometers around three rotation axes; the footplate denoted by the red arrow is fixed on the moving platform denoted by the yellow arrow by a six-axis load cell denoted by the green arrow (they comprise the end effector), and the moving platform with three rotational DOFs is driven by four parallel actuators).

7.2.1 Model Formulation

This ankle model was created based on the lower extremity model developed by Delp et al. [263] in OpenSim [262] and analysed in MATLAB. The lower extremity model indicates DP at the ankle joint and IE at the subtalar joint. The AA was added in the proposed model for use in three-dimensional robot-assisted ankle therapy. These three rotation axes were modified in OpenSim to orthogonally intersect at a point considered as the rotation centre of the ankle-foot complex in robot-assisted therapy. Seven ligaments spanning this joint are included in the model and represented by linear elements. The corresponding attachment locations are determined from dissection and anatomical atlases [267, 268], as shown in Figure 7.1 and Table 7.2. Muscles included are the same as that in Gait2392 available open through to OpenSim source.
software [262], but only sections connecting the tibia/fibula and the talus/calcaneus are used, as shown in Figure 7.1 and Table 7.2. The definition of the coordinate axes is consistent with that in OpenSim. Ligament stiffness summarised by Wei et al. [275] based on published data is used in this study and passive muscle force was calculated based on muscle length with respect to neutral length [277].

The definition of the global coordinate system of the ankle joint presented in Chapter 4 is adopted for kinematic analysis. For the specific calculation, same as the model analysis in Chapter 6, the coordinate system used by Delp et al. [263] is adopted to facilitate data extraction. Three coordinate frames respectively denoted as tibia/fibula, talus and calcaneus were defined

### Table 7.2: Position/force parameters of muscles and ligaments.

<table>
<thead>
<tr>
<th>FE</th>
<th>Attachment location [X, Y, Z] (mm)</th>
<th>ST or MF</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Origin</td>
<td>Insertion</td>
</tr>
<tr>
<td>ATaFL*</td>
<td>[3 -419 38]</td>
<td>[9 1 19]</td>
</tr>
<tr>
<td>PTaFL*</td>
<td>[-7 -421 23]</td>
<td>[-13 0 4]</td>
</tr>
<tr>
<td>ATiTL*</td>
<td>[1 -407 -17]</td>
<td>[7 9 -5]</td>
</tr>
<tr>
<td>PTiTL*</td>
<td>[-15 -409 -6]</td>
<td>[-11 5 -9]</td>
</tr>
<tr>
<td>CaFL</td>
<td>[-5 -425 37]</td>
<td>[45 33 2]</td>
</tr>
<tr>
<td>TiNL</td>
<td>[0 -406 12]</td>
<td>[60 39 -26]</td>
</tr>
<tr>
<td>TiCL</td>
<td>[-1 -407 -16]</td>
<td>[50 31 -22]</td>
</tr>
<tr>
<td>EXTDIG</td>
<td>[29 -401 7]</td>
<td>[92 39 0]</td>
</tr>
<tr>
<td>EXTHAL</td>
<td>[33 -398 -8]</td>
<td>[97 39 -21]</td>
</tr>
<tr>
<td>FLEXDIG</td>
<td>[-15 -405 -20]</td>
<td>[44 32 -28]</td>
</tr>
<tr>
<td>FLEXHAL</td>
<td>[-19 -408 -17]</td>
<td>[37 28 -24]</td>
</tr>
<tr>
<td>LATGAS</td>
<td>[-24 -48 24]</td>
<td>[4 31 -5]*</td>
</tr>
<tr>
<td>MEDGAS</td>
<td>[-22 -49 -30]</td>
<td>[4 31 -5]*</td>
</tr>
<tr>
<td>PERBREV</td>
<td>[-14 -430 29]</td>
<td>[47 27 23]</td>
</tr>
<tr>
<td>PERLONG</td>
<td>[-16 -432 29]</td>
<td>[44 23 22]</td>
</tr>
<tr>
<td>PERTERT</td>
<td>[23 -407 16]</td>
<td>[86 23 30]</td>
</tr>
<tr>
<td>SOL</td>
<td>[-2 -153 7]</td>
<td>[0 31 -5]</td>
</tr>
<tr>
<td>TIBANT</td>
<td>[33 -395 -18]</td>
<td>[117 18 -30]</td>
</tr>
<tr>
<td>TIBPOST</td>
<td>[-14 -405 -23]</td>
<td>[42 33 -29]</td>
</tr>
</tbody>
</table>

ATaFL: Anterior talofibular ligament; PTaFL: Posterior talofibular ligament; ATiTL: Anterior tibiotalar ligament; PTiTL: Posterior tibiotalar ligament; CaFL: Calcaneofibular ligament; TiNL: Tibionavicular ligament; TiCL: Tibiocalcanean ligament; EXTDIG: Extensor digitorum longus; EXTHAL: Extensor hallucis longus; FLEXDIG: Flexor digitorum longus; FLEXHAL: Flexor hallucis longus; LATGAS: Lateral gastrocnemius; MEDGAS: Medial gastrocnemius; PERBREV: Peroneus brevis; PERLONG: Peroneus longus; PERTERT: Peroneus tertius; SOL: Soleus; TIBANT: Tibialis anterior; TIBPOST: Tibialis posterior; FE: Force element; ST: Stiffness; MF: Max force.

* represents FE insertion is relative to the talus coordinate system and all others are relative to the calcaneus coordinate system; **represents data from the lower extremity model.
to describe foot motions. The tibia/fibula frame $\mathbf{xyz}_{\text{tib}}$, the talus frame $\mathbf{xyz}_{\text{tal}}$ and the calcaneus frame $\mathbf{xyz}_{\text{cal}}$ keep the same as that in the lower extremity model. The frame $\mathbf{xyz}_{\text{tib}}$ is fixed and no movement exists between $\mathbf{xyz}_{\text{tal}}$ and $\mathbf{xyz}_{\text{cal}}$. The coordinate of these attachment points described in $\mathbf{xyz}_{\text{tib}}$ could be obtained by translation based on ankle anatomy and rotation based on ankle motion, as shown in (7.1), (7.2) and (7.3), where $p$ represents the coordinate of the attachment point with respect to a certain coordinate frame, $R$ is the rotation matrix depending on three independent angular variables denoted as $\theta_{dp}$, $\theta_{ie}$, and $\theta_{aa}$, and $T_{\text{tib-tal}}$ and $T_{\text{tal-cal}}$ are the translation matrices decided by ankle anatomy.

\[
\begin{align*}
\begin{cases}
\mathbf{p}_{\text{tib}} = \mathbf{R}_{\text{tal}} + T_{\text{tib-tal}} \\
\mathbf{p}_{\text{tib}} = \mathbf{R} (\mathbf{p}_{\text{tal}} + T_{\text{tal-cal}}) + T_{\text{tib-tal}}
\end{cases}
\end{align*}
\text{FE inserts at the talus}
\]

FE inserts at the calcaneus

\[
\begin{align*}
\mathbf{R} &= 
\begin{bmatrix}
\cos \theta_{aa} \cos \theta_{ie} & R_{12} & R_{13} \\
\sin \theta_{aa} \cos \theta_{ie} & R_{22} & R_{23} \\
-\sin \theta_{ie} & \cos \theta_{ie} \sin \theta_{dp} & \cos \theta_{ie} \cos \theta_{dp}
\end{bmatrix}
\end{align*}
\text{Subject-Specific Adaptation}

This general model represents a young adult male with a height of 1.8 m and mass of 75 kg. Subject-specific adaptations in evaluating passive ankle torque were conducted on nine young adults (six males and three females) with age $26.11 \pm 2.03$ years old, height $172.67 \pm 11.26$ cm and weight $70.33 \pm 18.48$ kg. These data were collected via the manual ankle assessment device from the same participants as those in Chapter 5. Subject-specific adaptations in estimating ligament lengths were conducted on two additional adults (a male with age 35 years old, height 176 cm and weight 85 kg; and a female with age 27 years old, height 166 cm and weight 51 kg). Participants were required to have no history of severe ankle injuries, no acute ankle sprains within the last year, and no acute symptoms of pain or weakness. Table 7.3 summarises the characteristics of participants and scaling factors used in the proposed model.

The subject-specific adaptation adopted a non-uniform geometric scaling and mass-fat scaling defined by Rasmussen et al. [278], as in (7.4), (7.5) and (7.6). $r$ and $r_b$ represent the actual coordinate and the base of FE attachment points respectively. $F$ and $F_b$ represent actual force
and the base of FEs respectively. $G$ is a $3 \times 3$ scaling matrix for segments, denoted as $G = \text{diag}[G_{11} \ G_{22} \ G_{33}]$, where $G_{11} = G_{33} = \sqrt{k_{m}/k_{l}}$, $G_{22} = k_{l}$, $k_{l}$ represents height and $k_{m}$ weight ratio, the differences between $S_1$ and $S_2$ are if the fat-mass percentage is considered for strength scaling, $R_{fat}$ is obtained based on (7.7) and $R_{fat}^{b}$ is $R_{fat}$ of the base model, and $R_{other}$ equals 0.5.

\[
G = \begin{bmatrix} G_{11} & & \\ & G_{22} & \\ & & G_{33} \end{bmatrix}
\]

\[
r = Gr_{b}
\]

\[
F = S_{i}F_{b}, \ i = 1, 2
\]

\[
\begin{cases}
S_1 = k_m^{2/3} \\
S_2 = \frac{k_m 1 - R_{other} - R_{fat}}{k_l 1 - R_{other} - R_{fat}}
\end{cases}
\]

\[
\begin{cases}
R_{fat}^{\text{for-man}} = -0.09 + 0.0149 * \text{BMI} - 0.00009 * \text{BMI}^2 \\
R_{fat}^{\text{for-woman}} = -0.08 + 0.0203 * \text{BMI} - 0.000159 * \text{BMI}^2
\end{cases}
\]

<table>
<thead>
<tr>
<th>No.</th>
<th>Age (years)</th>
<th>H (cm)</th>
<th>M (kg)</th>
<th>G-Scaling</th>
<th>$S_1$</th>
<th>BMI</th>
<th>$S_2$</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>23</td>
<td>173</td>
<td>80</td>
<td>[1.05,0.96, 1.05]</td>
<td>1.04</td>
<td>26.73</td>
<td>0.97</td>
</tr>
<tr>
<td>2</td>
<td>25</td>
<td>185</td>
<td>103</td>
<td>[1.16,1.03,1.16]</td>
<td>1.24</td>
<td>30.09</td>
<td>1.02</td>
</tr>
<tr>
<td>3</td>
<td>26</td>
<td>186</td>
<td>80</td>
<td>[1.02,1.03,1.02]</td>
<td>1.04</td>
<td>23.12</td>
<td>1.03</td>
</tr>
<tr>
<td>4</td>
<td>26</td>
<td>183</td>
<td>81</td>
<td>[1.03,1.02,1.03]</td>
<td>1.05</td>
<td>24.19</td>
<td>1.02</td>
</tr>
<tr>
<td>5</td>
<td>28</td>
<td>165</td>
<td>60</td>
<td>[0.93,0.92,0.93]</td>
<td>0.86</td>
<td>22.04</td>
<td>0.91</td>
</tr>
<tr>
<td>6</td>
<td>25</td>
<td>180</td>
<td>75</td>
<td>[1.1,1.1]</td>
<td>1</td>
<td>23.15</td>
<td>1</td>
</tr>
<tr>
<td>7'</td>
<td>25</td>
<td>165</td>
<td>61</td>
<td>[0.94,0.92,0.94]</td>
<td>0.87</td>
<td>22.41</td>
<td>0.62</td>
</tr>
<tr>
<td>8'</td>
<td>30</td>
<td>160</td>
<td>49</td>
<td>[0.86,0.89,0.86]</td>
<td>0.75</td>
<td>19.14</td>
<td>0.62</td>
</tr>
<tr>
<td>9'</td>
<td>27</td>
<td>157</td>
<td>44</td>
<td>[0.82,0.87,0.82]</td>
<td>0.70</td>
<td>17.85</td>
<td>0.61</td>
</tr>
<tr>
<td>MRI</td>
<td>35</td>
<td>176</td>
<td>85</td>
<td>[1.08,0.98,1.08]</td>
<td>NA</td>
<td>NA</td>
<td>NA</td>
</tr>
<tr>
<td>MRI'</td>
<td>27</td>
<td>166</td>
<td>51</td>
<td>[0.86,0.92,0.86]</td>
<td>NA</td>
<td>NA</td>
<td>NA</td>
</tr>
</tbody>
</table>

\(^1\)represents female participants and all others are males; No. represents participant number; Age (years), H (cm) is participant height and M (kg) refers to participant mass; BMI is the body mass index based on H and M; $S_1$ and $S_2$ represent two different strength scaling factors and they are defined in (7.5). MRI: Magnetic resonance imaging.

### 7.2.3 Experimental Data

Experimental data regarding the relationship between ankle position and passive torque were collected by a manual ankle assessment device. This device mechanically consists of a handle for manual operation, a moving platform and a footplate rigidly fixed with an ankle orthosis.
Chapter 7. PRA-Model for Estimating Passive Ankle Torque

The components used for sensing include an angular potentiometer and a six-axis load cell (M3715C, Sunrise Instruments), which is the same as the end effector of ‘Robotics’ in Figure 7.1. The rotation axis of the ankle orthosis was made to be consistent with that of the moving platform.

Participants were instructed to sit on a height-adjustable chair with the hip and knee joints in 90° of flexion. The ankle-foot complex was fixed onto the footplate by an ankle orthosis. The rotation axis of the ankle orthosis was visually adjusted to approximate the ankle joint. Participants were verbally encouraged to relax the foot-ankle complex to minimise the effects by muscle activations. Extremes of ankle DP depended on the subjective sensation from participants until the foot is tense. The rotation was considered to be quasi-static with angular velocity less than 8°/s to reduce the effect by velocity. Five cycles (starting from neutral position defined as the ankle position where the foot and the leg are perpendicular to each other in the sagittal plane based on [218], to extreme dorsiflexion, to neutral position, to extreme plantarflexion and getting back to neutral position) were conducted on each participant. This study was approved by the University of Auckland, Human Participants Ethics Committee (9348) for experimentation on human subjects, and consent was obtained from all participants.

Data regarding the neutral lengths of ankle ligaments were obtained from two adults using MRI. The images were obtained using a Siemens Skyra 3T scanner and T1 weighted volume 0.6 x 0.6 x 0.6 mm in a sagittal plane. The lengths of ankle ligaments were measured offline using syngo fastView.

7.2.4 Model Validation

Data from literatures [255, 257, 270-272] and passive ankle torque collected from participants were used for model validation. The lengths of ankle ligaments depend on not only the bone geometry, but also their origin and insertion points on bones. An accurate ankle model should output accurate ligament lengths, and thus ligament lengths could be considered as a measure for model validation. Many researchers have investigated ligament lengths in the neutral position through either cadaver sectioning or image-based methods like MRI, as summarised in Table 7.4. The summary means and standard deviations (SDs) were determined by meta-analysis using the Random Effect Model [279]. The experimental and modelling results were compared for each participant using mean average deviation (MAD), root mean square deviation (RMSD) and NRMSD for evaluating the subject-specific adaptation performance, as respectively shown in (7.8), (7.9) and (7.10). $m_i$ and $e_i$ represent modelling and experimental values at each selected step $i$ respectively, step refers to the moment for each data acquisition,
and \( n \) is the total number of steps. \( \Delta \) is the range of experimental values defined as the difference between the maximum and the minimum values in a data set.

\[
MAD = \frac{1}{n} \sum_{i=1}^{n} |m_i - e_i| / n \quad (7.8)
\]

\[
RMSE = \sqrt{\frac{1}{n} \sum_{i=1}^{n} (m_i - e_i)^2 / n} \quad (7.9)
\]

\[
NRMSD = \frac{RMSE}{\Delta} \times 100\% \quad (7.10)
\]

**Table 7.4**: Published neutral lengths of ankle ligaments.

<table>
<thead>
<tr>
<th>Studies</th>
<th>Length (mm) in Means ± SDs</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>ATaFL</td>
</tr>
<tr>
<td>Siegler et al. [30]</td>
<td>17.81±3.05</td>
</tr>
<tr>
<td>Ozeki et al. [14]</td>
<td>19.80±1.92</td>
</tr>
<tr>
<td>Mkandawire et al. [31]</td>
<td>18.89±2.97*</td>
</tr>
<tr>
<td>Taser et al. [32]</td>
<td>22.37±2.50</td>
</tr>
<tr>
<td>Asla et al. [16]</td>
<td>16.30±3.0</td>
</tr>
<tr>
<td>Combined [279]</td>
<td>19.22±2.50</td>
</tr>
</tbody>
</table>

* represents the sample size is 5, SS represents sample size and values in bold and italic represent the reliable results that would be regarded as the baseline for the validation of this model.

### 7.3. Results

#### 7.3.1 Neutral Ligament Length

Muscle and ligament length for any ankle position is assumed to be obtained from this computational model. Figure 7.2 compares the model-based neutral ligament length with published data summarised in Table 7.4. It was found that modelling results were reasonable and to some extent verify the effectiveness of this model.
7.3.2 Passive Ankle Torque

Ankle torque could be divided into passive torque and active torque. Passive ankle torque was calculated and compared with experimental results. Muscles’ active contribution to ankle dynamics is not in the interest of this study. Figure 7.3 plots the total passive ankle torque in DP, and contributions by muscles and ligaments are also presented separately. Individual FE force distributions with ankle position can be seen from Figures 7.3, 7.4 and 7.5. Figure 7.6 describes passive ankle torque in IE (above) and AA (below), respectively. Passive joint torque at maximum inversion was compared well with published data [280] and no published data regarding the torque evaluation for ankle AA.

**Figure 7.3:** Model-based passive joint torque in ankle DP.
Chapter 7. PRA-Model for Estimating Passive Ankle Torque

The general model used in this study got the similar characteristics (gender, age, height and weight) as participant No. 6. Thus a direct comparison between modelling results and experimental data on participant No. 6 was conducted to validate the effectiveness of this model. To facilitate the comparison, data processing was conducted over five trials on each participant to eliminate random errors caused by participants and the manual operation, see Figure 7.7. The curve in cyan is considered as the baseline for model evaluation. Figure 7.8 shows the direct comparison between modelling results and experimental data on participant No. 6. MAD, RMSD and NRMSD are 0.4926, 0.5326 and 3.0034%, respectively. The individual contribution of each FE is clearly presented.

Figure 7.4: Muscles’ contribution to passive joint torque in ankle DP.

Figure 7.5: Ligaments' contribution to passive joint torque in ankle DP.
Chapter 7. PRA-Model for Estimating Passive Ankle Torque

Figure 7.6: Model-based passive joint torque in ankle IE (above) and AA (below).

Figure 7.7: Experimental passive ankle DP torque over five tests on participant No. 6.

Figure 7.8: Comparison between modelling results and experimental data on participant No. 6.
7.3.3 Subject-Specific Adaptation Performance

This model was mainly developed to be subject-specific in evaluating passive ankle torque based on participants' height and weight by two scaling (geometric and strength) factors. Comparison curves on each participant are shown in Figure 7.9 and statistical data are summarised in Table 7.5. MAD values are less than 1 Nm, except for the one using S1 on participant No. 2. RMSD values are less than 1 Nm, except for the one using S1 on participant No. 2. All NRMSD values are less than 10%. To present the significance of subject-specific adaptation, statistical results with no scaling (NS) are also included in Table 7.5. The results from participant No.’s. 1, 2, 3, 4 and 6 present good modelling accuracy with all NRMSD values less than 4%, while those from participant No.’s. 5, 7, 8 and 9 do not with all NRMSD values greater than 13.8%.

The evaluation of passive ankle torque correlates with the ligament lengths. The performance of the subject-specific adaptation in evaluating ligament lengths is presented in Table 7.6, where the model-based neutral length of each ligament compares well with the data from MRI with all differences less than 1 mm.

7.4. Discussion

The potential of this model for use in robot-assisted ankle therapy is discussed first. The required kinematic inputs to existing computational models are usually obtained from marker-based motion capture systems [281]. This motion analysis technique is not convenient when combined...
Chapter 7. PRA-Model for Estimating Passive Ankle Torque

with robot-assisted therapy due to the issues of set-up and time-consuming analysis. The required inputs to the model are three independent position variables that could be easily sensed from existing ankle rehabilitation robots. The devices described in [12, 34] represent two typical ankle rehabilitation robots, with or without physical rotation axes of end effectors. Both can use linear potentiometers to evaluate the posture of the end effect using forward kinematics or only a multi-axis inclinometer. An alternative method for robots with physical axes is to use angular potentiometers. In general, the required inputs to this model could be easily obtained from robotic devices, which provide the basis for use in robot-assisted ankle therapy.

Table 7.5: Subject-specific adaptation performance on estimation of passive ankle torque for participant.

<table>
<thead>
<tr>
<th>No.</th>
<th>MAD (Nm)</th>
<th>RMSD (Nm)</th>
<th>NRMSD (%)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>NS</td>
<td>S₁</td>
<td>S₂</td>
</tr>
<tr>
<td>1</td>
<td>0.5173</td>
<td>0.8531</td>
<td>0.5318</td>
</tr>
<tr>
<td>2'</td>
<td>0.4583</td>
<td>1.8434</td>
<td>0.7377</td>
</tr>
<tr>
<td>3</td>
<td>0.4514</td>
<td>0.4848</td>
<td>0.4683</td>
</tr>
<tr>
<td>4</td>
<td>0.4387</td>
<td>0.4202</td>
<td>0.4310</td>
</tr>
<tr>
<td>5</td>
<td>1.6334</td>
<td>0.5402</td>
<td>0.6540</td>
</tr>
<tr>
<td>6</td>
<td></td>
<td>0.4926</td>
<td></td>
</tr>
<tr>
<td>7'</td>
<td>1.6596</td>
<td>0.6289</td>
<td>0.7834</td>
</tr>
<tr>
<td>8'</td>
<td>2.3698</td>
<td>0.3858</td>
<td>0.2542</td>
</tr>
<tr>
<td>9'</td>
<td>2.1983</td>
<td>0.4176</td>
<td>0.7467</td>
</tr>
</tbody>
</table>

MAD: Mean average deviation; RMSD: Root mean square deviation; NRMSD: Normalized root mean square deviation; MAD, RMSD and NRMSD are defined in (7.8), (7.9) and (7.10), respectively.

Table 7.6: Subject-specific adaptation performance on two participants.

<table>
<thead>
<tr>
<th>Length (mm)</th>
<th>MRI Modelling</th>
<th>MRI Measured</th>
<th>MRIf Modelling</th>
<th>MRIf Measured</th>
</tr>
</thead>
<tbody>
<tr>
<td>ATaFL</td>
<td>23.6</td>
<td>23.1</td>
<td>19.4</td>
<td>19.2</td>
</tr>
<tr>
<td>PTaFL</td>
<td>23.2</td>
<td>23.1</td>
<td>19.0</td>
<td>19.2</td>
</tr>
<tr>
<td>ATiTL</td>
<td>19.9</td>
<td>19.0</td>
<td>17.3</td>
<td>17.7</td>
</tr>
<tr>
<td>PTiTL</td>
<td>16.5</td>
<td>19.0</td>
<td>15.4</td>
<td>15.5</td>
</tr>
<tr>
<td>CaFL</td>
<td>32.2</td>
<td>32.8</td>
<td>26.6</td>
<td>26.1</td>
</tr>
<tr>
<td>TiNL</td>
<td>43.5</td>
<td>44.1</td>
<td>37.1</td>
<td>37.6</td>
</tr>
<tr>
<td>TiCL</td>
<td>33.3</td>
<td>34.5</td>
<td>31.4</td>
<td>31.4</td>
</tr>
</tbody>
</table>

Existing computational ankle models are usually developed for ligament analysis for certain specific applications. The overall ankle joint torque could not be reliably evaluated without considering the contribution of muscles. However, this proposed ankle model could be used for evaluating ankle joint torque by computing the contribution of each FE. The use of this ankle model may permit a less expensive rehabilitation system, for example by replacing the need to use expensive sensors. Improvement regarding the robotic design for ankle therapy could be further achieved, although a range of robot-assisted ankle rehabilitation techniques have been demonstrated to be effective for individuals with ankle injuries [17]. This model could be used for robot-assisted ankle assessment at the level of joint and FEs.

The model structure is different with ankle anatomy and published models with ankle joint between the tibia/fibula and the talus and subtalar joint between the talus and the calcaneus [263,
Siegler et al. [1] concluded that DP mainly exists at the ankle joint, IE mainly exists at the subtalar joint, and other motion could be the combination of the ankle joint and the subtalar joint. The axis of ankle DP in the proposed model can be approximately considered to be consistent with ankle anatomy, while the IE and AA axes are obviously different with ankle anatomy. Thus direct comparison between modelling results and published data, mainly in terms of ankle DP, was conducted for model evaluation. Figure 7.6 shows passive ankle torque in IE (above) and AA (below), respectively, where the passive joint torque in extreme inversion was well compared with published data [280] on subjects with recurrent ankle inversion sprains, although the maximum positions were not completely consistent.

For model evaluation in DP, both the neutral ligament length and passive joint torque are considered as the measures. The measured ankle ROM (−50° of plantarflexion and 40° of dorsiflexion) is considered to be reasonable with high driving torque applied, which is verified by a physiotherapy expert. Comparisons with published neutral ligament lengths obtained from cadaver anatomy and MRI were conducted. In Figure 7.2, the neutral ligament length calculated from this computational model shows a satisfactory accuracy by comparing them with published data. The main purpose of this model is to estimate passive joint torque as the foot moves for use in robot-assisted therapy. Modelling results were compared with experimental data on participant No. 6 in Figure 7.8 and a high accuracy was achieved, with MAD, RMSD and NRMSD values being 0.4926, 0.5326 and 3.0034%, respectively. The contribution of individual FE, as shown in Figures 7.4 and 7.5, is not validated in this study since no published evidence has been found. Subject-specific adaptations by geometric and strength scaling were made in this model.

Comparisons were conducted on each participant and results show a satisfactory accuracy, with each NRMSD value less than 10%. However, the accuracies vary for different participants, as shown in Figure 7.9 and Table 7.5, which could be caused by various factors like age and gender [282, 283]. Most subject-specific performance on males is close with different strength scaling factors. Strength scaling using S1 on participant No. 2 presents less accuracy than that with S2 and the reason could be that S1 does not consider the influence from the fat percent. However, the subject-specific performance on participants with moderate BMI values presents better estimation accuracy. NRMSD values on participants No’s. 3, 4 and 8 using S2 are less than 3%.

To present the significance of the subject-specific adaptation, the accuracy in evaluating passive ankle torque with respect to NS is also analysed and presented in Table 7.5. There are significant differences among these nine participants regarding the model-based prediction accuracy with
NS. The data from participant No’s. 1, 2, 3, 4 and 6 are encouraging, while those from participant No’s. 5, 7, 8 and 9 do not show a satisfactory prediction accuracy. The potential reason is that the base model used in this study represents a young adult male with a height of 1.8 m and mass of 75 kg, which is closer to those of participant No’s. 1, 2, 3, 4 and 6 than others. Further, it can be seen that the subject-specific adaptation seems more necessary for female subjects than males, which is clearly presented in Table 7.5 where the torque estimation with subject-specific adaptation present significantly improved accuracy with the NRMSD values being 14.7807%, 26.7397% and 19.6852% for participant No’s 7, 8 and 9 with respect to all less than 7% without NS. The reason could be the gender and the body size differences of these three females as the base model. Another interesting point is that estimation with NS is successful and achieves better results than "subject-specific adaptation" method for participant No’s 1, 2 and 4. This behaves abnormal since the subject-specific adaptation does not improve the model-based estimation as expected, from the other hand this is normal because the estimation accuracy has been high with all NRMSD values less than 3.5% when with NS. In general, subject-specific adaptation does not contribute significantly once a high-precision estimation (<10%) has been obtained. However, a more comprehensive subject-specific adaptation law should be investigated in near future with body size, gender, age, health condition, and even ethnic group all considered.

Some other limitations of the model should be also noted. The linear representation of muscles and ligaments as being straight lines between origin and insertion points may affect the estimation of joint torque. The subject-specific adaptation was conducted based on participants' height and weight, and factors like gender, age and health condition should be also included for future improvement. An alternative scaling method can be made based on foot geometry instead of body height and weight. This model assumed normal elastic properties of the ligaments and muscles and thus could reliably predict passive ankle torque for healthy subjects. However, the prediction could be inaccurate (due to abnormal muscle and ligament property) if the robotic training is delivered to ankle-impaired subjects. For example, some pathological conditions have not just stiffer ligaments, but shortened ones. Further improvement will take patients’ health condition into consideration for subject-specific adaption. Although this model can quantify the individual contribution of ankle muscles and ligaments, the estimation accuracy was not validated experimentally. Published data in this field are also limited.

7.5. Conclusion of the Paper

This three-DOF computational ankle model with muscles and ligaments was developed for use in robot-assisted therapy. The use of three independent position variables required as inputs
Chapter 7. PRA-Model for Estimating Passive Ankle Torque

offers an advantage over other models when combined with robot-assisted therapy. This model could evaluate passive ankle torque in DP, IE and AA in robot-assisted therapy for a specific subject, as well as the individual contribution of ankle muscles and ligaments. The information could be helpful for designing a patient-specific therapy program and have more advantages than conventional functional assessment. Future studies could further generalise this model by: 1) incorporating muscles' active contribution; and 2) scaling the strength of muscles and ligaments based on the health condition of patients.

7.6. Chapter Summary

This chapter presents the development and validation of the PRA-Model in predicting passive ankle torque in subject-specific. This model uses three position variables that can be directly measured from sensors built in the AARR as inputs and outputs ligament kinematics and passive ankle torque. Different from the Non-PRA-Model presented in Chapter 6, the PRA-Model is consistent with the structure of the AARR, which allows for its direct use on the robot. Subject-specific adaptations by geometric and strength scaling were also made to allow for a universal model in predicting passive ankle torque. This model was experimentally validated in terms of ligament neutral length and passive ankle torque on the subject-specific basis, and shows a satisfactory accuracy. In general, the PRA-Model could reliably predict passive ankle torque for a specific individual, which will be used to facilitate the implementation of advanced interaction training by identifying the ankle mechanics and the active contribution from the patient (Chapter 9). To summarise, the trajectory tracking performance of the proposed interaction training scheme is satisfactory in the presence of patient-robot interaction, with all NRMSD values less than 2.3%. This interaction training strategy is expected to make the robotic training safe and effective by the consideration of the movement intention of the patient.
Chapter 8. Interaction Control via an Adaptive Predefined Trajectory

Interaction control is crucial to rehabilitation robots for active training. This is usually implemented on the basis of a position controller by which accurate trajectory tracking can be obtained to guarantee safe passive training. The first half of this chapter is dedicated to the development of a cascade position controller with position feedback in the outer loop and force feedback in the inner loop, which aims to implement smooth and safe training on the ankle assessment and rehabilitation robot (AARR). The second half proposes an intelligent training strategy of generating an adaptive predefined trajectory according to the movement intention of the patient. The proposed adaptive interaction training can be implemented by considering the intelligent strategy as a high-level controller and the cascade position controller in the low level. The materials have been submitted to IEEE/ASME Transaction on Mechatronics for possible publication together with the development of the AARR presented in Chapter 4.

This submitted paper goes as follows. This study presents an intrinsically-compliant AARR for the treatment of a variety of ankle injuries. The AARR has a bio-inspired design, devised after a systematic review on existing ankle rehabilitation robots. The AARR is adaptable to subjects with varying ankle ranges of motion (ROMs). It employs four intrinsically-compliant actuators that mimic skeletal muscles for three rotational degrees of freedom (DOFs). The sensing system consists of three magnetic rotary encoders, four single-axis load cells and a six-axis load cell. A cascade controller with position feedback in the outer loop and force feedback in the inner loop is proposed to achieve smooth and safe motion. An intelligent training strategy is also proposed as a high-level controller for adaptive passive-active training for enhanced patient engagement and training safety. To evaluate the control performance, experiments were conducted on a sprained ankle using the AARR. The robot-human interaction was kept as passive while the robot was operated under the cascade controller. The intelligent controller was evaluated through the adaptive passive-active training. The statistical trajectory tracking performance is satisfactory in the presence of patients’ disturbance, with all normalised root mean square deviation (NRMSD) values less than 2.3%, thus indicating the potential of the AARR for clinical applications.
8.1. Introduction

The foot-ankle complex contributes significantly to the function of the lower extremity and has the general functions of balance, support and propulsion [1, 2]. However, it is particularly susceptible to musculoskeletal or neurological injuries. Ankle sprains are common musculoskeletal injuries and involve the overstretching or tearing of the relevant tendons and ligaments. In the United States, more than 23,000 cases are estimated to occur per day [3]. A statistical 230,288 new claims and 278,657 active claims related to musculoskeletal ankle injuries were made to the Accident Compensation Corporation from July 2009 to June 2014, costing over New Zealand $165 million in New Zealand [4]. Neurological injuries may also cause ankle disabilities – a typical example is the drop foot following stroke. Based on an up-to-date report from the American Heart Association, approximately 795,000 people experience a new or recurrent stroke (ischemic or hemorrhagic) in the United States each year, of which about 610,000 are the first events and the remainder are recurrent events [5]. An estimated 60,000 stroke survivors live in New Zealand, and many of them have mobility impairments [6].

A general rehabilitation program for musculoskeletal ankle injuries involves an acute phase treatment by immobilization and sub-acute phase where passive/active ROM and muscle strengthening exercises are normally carried out within the pain-free ankle ROM. Stroke neuro-rehabilitation should include meaningful, repetitive, intensive, and task-specific movement training to promote motor recovery [8]. However, rehabilitation treatments for musculoskeletal and neurological ankle disabilities both require cooperative and intensive efforts from therapists and patients over prolonged sessions [9].

Robot-assisted rehabilitation solutions, as therapeutic adjuncts to facilitate clinical practice, have been actively researched in the past few decades and provide an overdue transformation of the rehabilitation centre from labour-intensive operations to technology-assisted operations [32]. The robot could also provide a rich stream of data using intelligent sensing units to facilitate patient diagnosis, customisation of the therapy, and maintenance of patient records. A systematic review of 29 studies with a total of 164 patients and 24 healthy subjects [17] demonstrated the effectiveness of existing rehabilitation robots in reducing ankle impairments caused by either musculoskeletal or neurological injuries. Moreover, a systematic review conducted by Kwakkel et al. [39] on 10 studies with 218 patients demonstrated that robot-assisted therapy contributes to increase treatment compliance by way of introducing incentives to the patient, such as virtual-reality games.
Chapter 8. Interaction Control via an Adaptive Predefined Trajectory

Existing robot-assisted ankle rehabilitation devices are mainly classified into wearable robots that usually aim at ankle rehabilitation during gait exercises, such as the MIT Anklebot developed by Roy et al. [10], while platform robots focus solely on ankle training [17]. However, platform-based parallel robots have been considered suitable for ankle rehabilitation due to the characteristics of multi-DOFs and large generated torque. A typical instance is the Rutgers Ankle powered by double-acting pneumatic cylinders based on the six-DOFs Stewart-Gough platform [15]. The Rutgers Ankle has been demonstrated the potential for clinical applications by: 1) integrating assessment into rehabilitation; 2) assisting patients in participating virtual-reality games; 3) combining telerehabilitation systems [20]; and 4) having clinical trials on subjects with varying grades of ankle sprains [101], patients post-stroke [100], and children with cerebral palsy [83]. However, this system requires varying positions of the foot-ankle complex and synergic movement of the lower limb from the patient, which could result in difficulties in controlling the robotic training. This robot also has more redundant DOFs than those of the ankle-foot complex. More recently, Saglia et al. [14] developed a redundantly actuated parallel mechanism for ankle rehabilitation using three customised cable-driven linear electric actuators for achieving two DOFs (dorsiflexion and plantarflexion, inversion and eversion). This robotic design has less DOFs than those of the ankle-foot complex. More critically, this robot suffers from the aforementioned issue of varying positions of the ankle joint as the Rutgers Ankle has, although high-performance interaction control algorithms for passive and active exercises have been validated on healthy subjects [21]. While some other parallel robots [35, 64, 284] have been also proposed for ankle rehabilitation, no advanced control strategies that encourage patients’ active participation were implemented and validated. All these robots identified above have similar mechanical structure wherein the moving platform is actuated from the bottom and the rotation centre does not coincide with the ankle joint.

A novel ankle rehabilitation robot was constructed in Mechatronics Lab at The University of Auckland and it fixes the ankle joint centre of rotation during robot-assisted therapy [12]. In addition to the characteristics of the use of light-weight and intrinsically-compliant actuators, satisfactory trajectory tracking was obtained based on a fuzzy-logic controller realised in joint space. However, limitations of this robot (such as its flimsy structure and cable transmission, and the lack of real-time ankle assessments) exist and impede the clinical applications. More significantly, the individual link length-based controller using four linear potentiometers to feedback cannot guarantee all cables in tension, which may not be safe for human users since cable-driven robots may lose controllability if some cables become slack during operations [235]. This robot was further improved with four linear potentiometers (LPs) modified into three
angular potentiometer (APs), as shown in Figure 8.1. Although these modifications allowed for a more compact robotic design and the implementation of inverse kinematics, the robotic functionality and control were not significantly improved.

In all the works mentioned above, existing ankle rehabilitation robots have limitations regarding either the robotic design (such as less robotic DOFs than actual ankle motions, conflicting robotic rotation centre and ankle joint location, and the use of stiff actuators) or the robotic functionality (such as the lack of real-time measurement of the human-robot interaction force and torque). To overcome these limitations, an intrinsically-compliant ankle rehabilitation robot was developed with real-time measurement of the human-robot interaction. An intelligent training strategy was also proposed in combination with a cascade controller for enhanced patient engagement and safety during the robotic training. For simplification, this device is named the AARR on functionality grounds. The scope of this study is focused on the rehabilitative rather than the assessment aspect of ankle therapy. Section 8.2 presents the robotic design and analysis. Section 8.3 introduces the cascade controller and the intelligent training strategy as a high-level controller. Experiments on a sprained ankle were conducted in Section 8.4, followed by discussion in Section 8.5, and the conclusion is summarised in Section 8.6.

### 8.2. AARR

The design of the AARR is presented in Figure 4.1 using a three-dimensional Creo model, and the physical picture can be found in Figures 4.8 and 4.9. This robot is actuated by four parallel
Festo fluidic muscles (FFMs) (Festo DMSP-20-400N) to actuate three rotational DOFs that are for ankle dorsiflexion/plantarflexion (DP), inversion/eversion (IE) and adduction/abduction (AA), respectively, which makes the AARR have bio-inspired design by mimicking the configuration and actuation of the ankle joint by natural muscles. The AARR consists of a fixed platform and a moving platform that in fact is a three-link serial manipulator with three rotational DOFs. The sensing components include three magnetic rotary encoders for measuring angular positions of the robotic end effector, four single-axis load cells, together with four amplifiers for measuring contraction forces of FFMs, and four proportional pressure regulators for the pressure control of the FFMs. These electronic components communicate with the embedded controller (NI Compact RIO-9022) through three independent modules (NI-9401, NI-9205 and NI 9263) for digital input/output (DI/O), analog input (AI) and analog output (AO), respectively. Additionally, a six-axis load cell is used to measure the interaction force and torque between the robot and a human ankle joint based on Equations (4.15 and 4.16). It communicates with the controller through the RS232 port using a customised interface box (SRI M8125).

8.3. Control System

The AARR is intended to perform ankle ROM and muscle strengthening exercises. This requirement can be achieved by deriving a cascade position controller for the three-DOF end effector. To ensure the active participation of patients during robotic training, and hence the effectiveness of their therapy, an intelligent training strategy was proposed to make the robotic motion adaptive based on the assessment from physiotherapists and patients’ intention. Although the AARR is developed for three rotational DOFs respectively for ankle DP, IE and AA, here the DP and IE are controlled while the AA is kept passive to allow for some flexibility. This is because ankle AA is limited and being controlled primarily by rotation of the leg [285]. To facilitate the description of the proposed controllers, ankle DP, IE and AA are also denoted by rotations about axis-X, Y and Z, respectively, as the description in Figures 1.1 and 4.1.

8.3.1 Cascade Control

The posture control of the AARR is the basis of a variety of robot-assisted rehabilitation exercises. This could be achieved by controlling individual FFM length in the joint space. Specifically, the required individual FFM length can be calculated by inverse kinematics based on the required posture of the end effector. The actual individual FFM length can be obtained from the measured posture of the end effector by inverse kinematics, although it cannot be
directly measured. However, this method cannot guarantee all FFMs in tension and may not be safe for human users.

**Figure 8.2**: A cascade controller with position control in the outer loop and individual FFM force control in the inner loop. CT-C: Computed torque controller; PID-C: Proportional-integral-derivative controller; PI-C: Proportional-integral controller; CT-C relates to (4.11) and (4.14); Force distribution refers to the analytic-iterative force distribution method simulated in Figure 4.6(b); and ankle force and torque are calculated using (4.15) and (4.16); the position tracking error is denoted as \( \theta_e = \theta_d - \theta_m \), of which \( \theta_d \) represents the desired position of the end effector while \( \theta_m \) represents the measured position; \( T_d \) represents the desired torque; \( F_m \) represents the contraction force of the FFM; \( L_m \) represents the FFM length; and \( p \) represents the FFM pressure; the FFM model is described in (4.24).

This study proposes a cascade controller with measured posture as feedback in the outer loop, while the measured FFM forces as feedback in the inner loop, as shown in Figure 8.2. This method can guarantee not only trajectory tracking accuracy, but also ensure the FFMs are always in tension for safety. For the forward feedback controller, the individual FFM length can be calculated based on the inverse kinematics. Combining with the measured FFM force, the required pressure can be obtained based on (4.24). However, this open-loop position controller significantly depends on the FFM modelling accuracy. To improve the control accuracy, direct posture feedback in the task space and FFM force feedback in the joint space were compensated to ensure trajectory tracking accuracy and all FFMs in tension while in use.

### 8.3.2 Intelligent Training Strategy

An intelligent training strategy is proposed to derive the maximum benefit from AARR-assisted rehabilitation. Participant self-initiated movements are achieved by allowing adaptation of the reference trajectory when a force or torque threshold is met. A sinusoidal path is selected as the reference trajectory of rehabilitation exercise, since it mirrors manual physiotherapy where the ankle is moved fast near the neutral position and slower at ROM extremes. The force threshold was determined by a physiotherapist who conducted a preliminary assessment and identified a safe threshold that was slightly higher than the actual ankle capacity. During the robotic training, the predefined value was further tuned to allow the robot to reliably adapt its trajectory according to the movement intention of the patient.
The algorithm of the proposed intelligent training strategy is described in Figure 8.3. The initial reference trajectory is defined in step 1 using (8.1), where $A_x$ is the amplitude of the trajectory $X$, $f$ is its frequency, and $t$ represents the time variable. If an interaction force along $X$ is exerted and exceeds the defined threshold $F_0$, a modified trajectory $x_{adap}(t)$ is generated when the time is taken as $t_1$, as shown in step 3. Defining $x_1$ as the current displacement, $x_1'$ as the velocity, $t_1$ as the time of force event, the trajectory adaption law is expressed in (8.2), where the modified phase is obtained based on (8.3) and $F_x$ refers to the interaction force along ankle DP or axis-Y.

In step 4, a new trajectory is generated with the opposite direction of the reference trajectory when the interaction force resists the robot movement. The robot continues to move towards the zero-crossing point, as indicated in step 5. The resulting time is taken as $t_2$ when the robot reaches the zero-crossing point, as shown in (8.4). To continue the robotic training, the participant can initiate a new trajectory by exerting an interaction force. As shown in step 6, for example, a trajectory about axis-Y is initiated when $F_y$ triggers the predefined threshold, where $F_y$ refers to the interaction force along ankle IE or axis-X. The time variable and phase are defined according to (8.5) and (8.6) in steps 7 and 8. In general, the robot trajectory can be adapted and switched freely between the axis-X and Y trajectories based on the movement intention of the patient. Alternatively, steps 9-11 are executed when $F_x$ triggers the predefined threshold. The robot maintains the reference trajectory if human-robot interaction forces remain below the threshold.

---

**Figure 8.3:** Algorithm of the proposed intelligent training strategy.
\[ x_{\text{init}}(t) = A_x \sin(2\pi ft) \]  
\[ x_{\text{adap}}(t) = x_{\text{init}}(t + \varphi_{\text{adap}}) \]  
\[ \varphi_{\text{adap}} = \begin{cases} 
\frac{1}{2f} - \sin^{-1}\frac{|x_1|}{\pi f}, & \text{if } x_1 \geq 0, x'_1 \geq 0, F_x < -F_0 \\
\frac{1}{2f} + \sin^{-1}\frac{|x_1|}{\pi f}, & \text{if } x_1 \geq 0, x'_1 < 0, F_x \geq F_0 \\
\frac{1}{2f} - \sin^{-1}\frac{|x_1|}{\pi f}, & \text{if } x_1 < 0, x'_1 < 0, F_x \geq -F_0 \\
\frac{1}{2f} + \sin^{-1}\frac{|x_1|}{\pi f}, & \text{if } x_1 < 0, x'_1 \geq 0, F_x < -F_0 
\end{cases} \]  
\[ t = t_2, \text{ when } x_{\text{adap}}(t) = 0 \]  
\[ y_{\text{adap}}(t) = A_y \sin\left(2\pi f(t_{y\text{adap}} + \varphi_{y\text{adap}})\right) \]  
\[ t_{y\text{adap}} = t - t_3, \varphi_{y\text{adap}} = \begin{cases} 
0, & \text{if } F_y \geq F_0 \\
\frac{1}{2f}, & \text{if } F_y < -F_0 
\end{cases} \]  

The speed of the robot motion can be also adjusted if a continuous interaction force reaches the predefined threshold. For a sine trajectory, the frequency \( f \) determines its movement speed. For every recalculation loop, the frequency is calculated by (8.7) in step 14, where \( f_0 \) is the initial frequency, and \( k_f \) is a coefficient that reflects the influence of human users. If no human-robot interaction triggers the predefined threshold, the robot follows the reference trajectory in a passive training mode, as in steps 16-18.

\[ f_{\text{adap}} = \frac{(F - F_0)}{k_f} + f_0 \]  

A simulation was then conducted in MATLAB to verify the proposed intelligent training strategy, as presented in Figure 8.4. It can be seen clearly that the proposed intelligent training is able to modify the reference trajectory of the robot in a way that is desired by the patient. The trajectory parameters including phase and frequency are adaptively tuned when the force reaches the defined threshold.
Figure 8.4: Simulation results of the proposed trajectory adaptation control law. The red line is the adapted trajectory in the axis-X and the blue line in the axis-Y. During the period of 20-30s for example, the trajectory is switched from the axis-X axis to the axis-Y axis. As illustrated in the algorithm, when the resisting force is applied at $t_1$ (step 3), the trajectory about axis-X is controlled to reach the zero-crossing point at $t_2$ (step 4-5), and the transition between the two trajectories is completed at $t_3$ when the force along axis-Y is applied (step 6-8).

8.4. Experimental Results

A subject with ankle sprain (male, 29 years) with no history of neurologic disorders gave written informed content and participated in this trial two weeks after the injury when the swelling was gone but the ankle joint was stiff. The sprained ankle was caused by jumping and rolling at basketball and diagnosed as limited ROM, lateral aspect of ankle swollen, and pain on weight bearing. Ethics approval for this protocol was obtained from the University of Auckland, Human Participants Ethics Committee (011904) (Appendix C).

8.4.1 Experimental Protocol

Before the robotic training, a preliminary assessment was conducted to check the appropriate ROM of the sprained ankle. The subject was instructed to sit on a height-adjustable chair with the shank free on the leg holder, as shown in Figure 4.9(c). The ankle-foot complex was strapped into the ankle orthosis. The rotation axis of the ankle orthosis was designed to approximate the ankle joint. Three types of training along the predefined sine trajectory to be applied by the AARR are the single-axis training, the mixed-axis training and the intelligent training. First, the
cascade controller was implemented on the AARR with the single-axis training and the mixed-axis training separately delivered to the participant. The single-axis training refers to the independent ankle DP and independent IE, while the mixed-axis training refers to the simultaneous motions of ankle DP and IE. The amplitude based on the preliminary evaluation is 0.3 rad and 0.2 rad, respectively, for ankle DP and IE of the single-axis training. The mixed-axis training consists of 0.2 rad ankle DP and 0.1 rad ankle IE. Following it, the intelligent ankle training strategy was implemented in combination with the cascade controller for adaptive passive-active ankle training. The defined force threshold is relevant to the ankle disability level of the patient, should be specified by a physiotherapist or a doctor. In the experiments, the force threshold was set high like 100N after consulting with a physiotherapist, and then gradually decreased until the patient can trigger the trajectory adaptation stably. The amplitude of the predefined default trajectory is 0.3 rad for ankle DP and 0.2 rad for ankle IE. All predefined trajectories were set to operate three cycles at 0.05 Hz. The participant was asked to remain relaxed for the single-axis training and the mixed-axis training, while certain active forces were required for the intelligent training.

Figure 8.5: Trajectory tracking responses of the single-axis training.
8.4.2 Cascade Control

Preliminary evaluations of the cascade controller implemented on the AARR are presented in Figures 8.5, 8.6, 8.7 and 8.8. Figure 8.5 plots the trajectory tracking responses of the single-axis training, including X and Y motions. Figure 8.6 presents the individual FFM force tracking responses of the single-axis training. The trajectory and individual force tracking for the mixed-axis training is plotted in Figures 8.7 and 8.8, respectively. The statistical results of the tracking accuracy are summarised in Table 8.1 presented in Section 8.5.

Figure 8.6: Individual FFM force tracking responses of the single-axis ankle DP (left) and IE (right) training.

Figure 8.7: Trajectory tracking responses of the mixed-axis training.
8.4.3 Intelligent Training

The preliminary results of the intelligent training strategy implemented on the AARR in combination with the cascade controller are presented in Figures 8.9 and 8.10. In Figure 8.9, the top and middle plots present the trajectory tracking performance about X and Y, respectively, and the bottom is the force trigger events. It is clear that the trajectory adaptation law could guarantee the robot’s movement keeping coincided with the patient’s intended force. The robot was adaptively controlled based on the movement intention of the patient by using the proposed method, with Case 1 for movement about axis-X during 0-45s, and Case 2 for movement switched to motion about axis-Y during 45-65s. Meanwhile, the continuous patient-robot interaction force resulted in a faster robot movement, which is verified during 65-90s when the trajectory frequency is adjusted. In Figure 8.10, the force tracking performance for each actuator is presented.

Figure 8.8: Individual FFM force tracking responses of the mixed-axis training.
To summarise, the proposed intelligent training strategy is able to implement adaptive passive-active robotic training according to the movement intention of the patient. Specifically, if the patient does not intend to change the predefined trajectory of the robot, or is severely impaired without enough muscle force, the robot operates in a passive mode. If the patient tries to actively participate and provide certain forces during the training, the robot is switched to run in an active mode in which an adaptive trajectory can be obtained by real-time recalculation. The statistical results regarding the trajectory tracking performance are summarised in Table 8.1.
8.5. Discussion

The main shortcomings of existing ankle rehabilitation robots include incompatible robotic design with the ankle joint movement, stiff actuation, non-back-drivability, and lack of flexibility to be adaptive on a variety of human users. The proposed AARR with biologically-inspired design and compliant actuation is safe to human users. It also has flexibility in obtaining varying robotic ROMs by changing the attachment positions of FFMs based on specific applications, as shown in Figure 4.3. Moreover, the main body of the AARR can move along the base depending on varying sizes of the lower extremities. More significantly, the robotic DOFs conform to the anatomical ankle structure, which makes it advantageous over the below-
actuated ankle platforms. However, a limitation of the AARR could be the lack of fluidic source in the medical environment, and can be solved by an actuation solution such as that of Saglia et al. [14] employing customised linear actuators independently designed based on cable-driven systems in developing a parallel ankle rehabilitation robot.

Considering the parallel robot control scheme, two typical frameworks are the joint space controller and the other is in the task space. The joint space controller, as a conventional control idea, is to design a controller to make the actual link lengths conform to the desired lengths computed from the required posture of the manipulator by inverse kinematics [286]. An example is the control of the ankle rehabilitation robot developed by Jamwal et al. [12]. By contrast, the task space controller requires the information from a multi-DOF sensor or multiple sensors that can measure the posture and/or the corresponding velocities of the end effector. Alternatively, direct kinematics that rely on numerical methods like an observer should be conducted to estimate the posture and/or the corresponding velocities. But the direct kinematics of a parallel robot manipulator have always been a difficult, challenging, and open problem. In the proposed AARR, the direct posture measurement of the end effector allows for the use of inverse kinematics.

Zhu et al. [287] developed a parallel manipulator using three FFMs that could be used for ankle rehabilitation, although the potential applications were not specified. With the same issue of the moving ankle joint centre of rotation as the ankle robot developed by Saglia et al. [14], this parallel manipulator if considered as an ankle rehabilitation platform leads to incompatible robotic motion with respect to the anatomical ankle structure. The performance of the proposed adaptive robust controller in the presence of disturbances was not evaluated, although it was demonstrated to effectively deal with the model uncertainties existing in the complex dynamics and the large extent of FFM modelling errors, as well as the strong coupling and inherent nonlinearities of the parallel manipulator dynamics. Additionally, the issue of three-DOF motion achieved by three FFMs is questionable since redundant actuations are a prerequisite for all the cable-based parallel robots, which means n+1 actuators are required to achieve n-DOF motion of the manipulator [235]. This controller does not require position and force sensors for control feedbacks, but the control stability should be further investigated, especially when used on human users.

To ensure the safety of the AARR, a cascade controller with posture feedback in the outer loop and individual FFM force feedback in the inner loop was implemented on the AARR, with satisfactory trajectory and force tracking accuracy achieved for both passive and active training,
as summarised in Table 8.1. Specifically, the trajectory tracking accuracy of the single-axis training presents better than those of the mixed-axis training and the intelligent training, with the NRMSD values of the single-axis training less than 1.4%, while the other NRMSD values being around 2%, which is reasonable since more uncertainties were brought in for the mixed-axis and the intelligent training. For rehabilitation devices, the posture control of the robotic effector is critical for both passive and active training. By contrast, the accurate control of the individual actuator force is not strictly required on condition that all cables are in tension. Based on the statistical results of Table 8.1, most NRMSD values for individual FFM force tracking are greater than 5%, which is acceptable since the FFM force is the secondary control variable with respect to the posture of the robotic posture. Further, the trajectory and FFM force tracking accuracy may be compromised by the robot-human interaction due to the compliance of the FFM. However, the proposed cascade controller was able to not only achieve the trajectory tracking accuracy, but also keep the four FFM tensioned during the robotic ankle training.

Table 8.1: Statistical results of the controlled variables of the AARR.

<table>
<thead>
<tr>
<th></th>
<th>Single-Axis</th>
<th>Mixed-Axis</th>
<th>Intelligent Training</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>X</td>
<td>Y</td>
<td>XY</td>
</tr>
<tr>
<td>F1 (N) RMSD</td>
<td>7.8704</td>
<td>8.0096</td>
<td>3.1579</td>
</tr>
<tr>
<td>F1 (N) NRMSD</td>
<td>13.51</td>
<td>21.47</td>
<td>4.35</td>
</tr>
<tr>
<td>F3 (N) RMSD</td>
<td>3.3823</td>
<td>6.8512</td>
<td>4.6860</td>
</tr>
<tr>
<td>F3 (N) NRMSD</td>
<td>7.62</td>
<td>26.50</td>
<td>7.88</td>
</tr>
<tr>
<td>F4 (N) RMSD</td>
<td>4.6678</td>
<td>7.6453</td>
<td>5.7107</td>
</tr>
<tr>
<td>F4 (N) NRMSD</td>
<td>14.37</td>
<td>19.03</td>
<td>13.40</td>
</tr>
<tr>
<td>X (rad) RMSD</td>
<td>0.0083</td>
<td>—</td>
<td>0.0061</td>
</tr>
<tr>
<td>X (rad) NRMSD</td>
<td>1.39</td>
<td>—</td>
<td>1.53</td>
</tr>
<tr>
<td>Y (rad) RMSD</td>
<td>—</td>
<td>0.0053</td>
<td>0.0044</td>
</tr>
<tr>
<td>Y (rad) NRMSD</td>
<td>—</td>
<td>1.32</td>
<td>2.21</td>
</tr>
</tbody>
</table>

RMSD: Root mean square deviation; NRMSD: Normalized root mean square deviation; RMSD and NRMSD are defined in (7.9) and (7.10), respectively; — represents not applicable.

To further derive the optimal benefit from robot-assisted rehabilitation, it is critical to promote active engagement from the participants during the robotic training. Pehlivan et al. [288] presented a trajectory generation algorithm based on an experimental movement profile, in which the trajectory was determined by piecewise polynomial functions. Regarding the transitions between two different trajectories, Sanz-Merodio et al. [289] adopted a unitary Gaussian function to modulate such a process. However, this kind of trajectory recalculation
method might increase the online computational burden and tend to get a discontinuous desired speed and acceleration levels. In this chapter, a novel intelligent training strategy based on the trajectory adaptation control law is developed with the integration of both passive and active training modes. Adapting the predefined sine trajectory, the controller developed here can modify its phase, frequency and path features according to the participant’s intended force, and the new desired trajectory is thus updated at every step. This strategy is able to encourage the participant’s active efforts by changing the robot movement when the participant achieves a force threshold. Meanwhile, the safety of the intelligent training can also be guaranteed, as the interaction force between human and robot is measured in real-time and the trajectory will be altered to relieve the interaction force as long as it reaches a threshold, which in turn increases the opportunities of the participant being actively involved in the robotic training. Furthermore, the required continuity of the profiles can be ensured to provide continuous levels of robotic training to the user. This intelligent training method also has great potential in providing patients with virtual-reality games, especially those needing simultaneous passive-active commands.

A few points about this study should be noted. First, the present experiments were carried out to evaluate the AARR design, the cascade controller, and the intelligent training strategy, which does not require a large sample of participants. Clinical trials do require a large sample of injured ankles when evaluating the effectiveness of the intelligent training strategy implemented on the AARR. Secondly, this robot was proposed for the treatment of a variety of ankle disabilities caused by musculoskeletal or neurological injuries, although only experiments on a sprained ankle were conducted as preliminary evaluations. Thirdly, this chapter focuses solely on AARR-assisted ankle rehabilitation, although it is also capable of ankle assessment, as its name suggests. Future work will include improving the force tracking performance, controlling the robotic motion for ankle AA training. The effectiveness of the AARR-assisted ankle assessment in evaluating joint ROM and stiffness will also be investigated for providing clinically relevant physiological metrics that can be compared to a benchmark to assess the patient’s recovery and adapt treatment accordingly. Enhancements of the intelligent training strategy, such as adding a forgetting factor combined with visual feedback so that less assistance is provided over time so long as the patient has no problems reaching the desired target, would also be helpful.

8.6. Conclusion of the Paper

This study presents the robotic design, the cascade controller, and the intelligent training strategy of the AARR with the preliminary evaluation on a subject with ankle sprain. According
to a systematic review [17] on existing ankle rehabilitation devices, the intelligent training strategy is proposed for the first time as well as the cascade controller implemented on parallel ankle rehabilitation robots. Their applications on the AARR, actuated by intrinsically-compliant FFM with real-time measurement of human-robot interaction, is also a novel work. Preliminary findings of the AARR on a sprained ankle are promising, thus indicating its potential for clinical applications for the treatment of a variety of ankle injuries.

Scientific contributions of this chapter include: 1) the development of an intrinsically-compliant AARR whose rotation centre coincides with the ankle joint with three DOFs and biologically-inspired actuation scheme; 2) model-based cascaded force and position control of the AARR that can accurately follow reference trajectories while using force distribution algorithm to maintain positive tension in all FFM at all times for enhanced safety; 3) intelligent training strategy that enhances patient engagement and safety by giving them some freedom in continuously modifying the pre-specified reference trajectory based on their interaction with the robot.

8.7. Chapter Summary

This chapter presents the implementation of an intelligent interaction control scheme on the AARR. The interaction controller consists of a high-level intelligent algorithm and a low-level cascade controller. By experiments on a sprained ankle, the cascade controller shows good trajectory tacking performance, and also guarantees all FFM in tension during the operation. The intelligent algorithm can adapt the predefined trajectory based on the movement intention of the patient. If no movement intention is detected, the AARR operates in a passive mode along the predefined trajectory. Otherwise, the AARR runs in an active interaction model along the adapted trajectory, wherein the movement ROM, direction, and velocity are adaptively adjusted. Experiments were conducted on a sprained ankle to evaluate the proposed interaction training scheme.
Chapter 9. Interaction Control via an Adaptive Admittance Law

This chapter proposes an interaction training strategy based on an adaptive admittance law, which is different from the movement intention-based interaction training scheme (Chapter 8). To implement the proposed interaction training on the ankle assessment and rehabilitation robot (AARR), a low-level position controller was developed in joint space by controlling the individual FFM length to achieve trajectory tracking using inverse kinematics, while in the high-level it is an admittance controller that is adaptive to real-time ankle position and patient-robot interaction. It should be noted that the cascade controller presented in Chapter 8 uses the measured contraction force of the Festo fluidic muscle (FFM) as the feedback in the inner loop. This control method is not available if the single-axis load cell for each actuator is removed. By contrast, the position controller in this chapter is easier to be implemented on the AARR even if the single-axis load cell is removed for reducing the cost and the depth of the AARR (the distance between the upper platform and the end effector). This interaction control strategy can conduct adaptive patient-cooperative training to make the AARR comply with the active contribution from the patient. Use of this training scheme allows the AARR to operate in a passive mode if no patient-robot interaction is detected, will otherwise run in an active mode with interaction. The materials have been submitted to IEEE Transactions on Neural Systems and Rehabilitation Engineering for possible publication.

This submitted paper goes as follows. This study proposes an adaptive patient-cooperative training strategy for the treatment of a variety of ankle injuries. The novel patient-cooperative training has been implemented on an intrinsically-compliant AARR. The AARR has three active rotational degrees of freedom (DOFs) actuated by four FFMs located parallel to the calf. Its control scheme consists of a position controller implemented in joint space and an admittance controller in task space. The admittance controller modifies the predefined trajectory of the robot using a novel adaptation law, allowing for the implementation of advanced active training strategies. The adaptation law tunes the parameters of the admittance controller based on ankle assessment in terms of joint position and interaction torque for enhanced safety. Experiments were conducted to evaluate three different patient-cooperative training. These training strategies were 1) a passive mode using a pure joint space position controller, 2) a patient-cooperative mode with a fixed-parameter admittance controller, and 3) a patient-cooperative mode using a variable-parameter admittance controller. The controller's trajectory tracking is accurate, even
when externally disturbed, with a maximum normalised root mean square deviation (NRMSD) value less than 5.4%. The preliminary findings suggest the potential of the patient-cooperative training strategy as a safe and engaging control solution for robots such as the AARR to be used in clinical environments.

9.1. Introduction

Robot-assisted ankle rehabilitation solutions, as therapeutic adjuncts to facilitate clinical practice, have been actively researched in the past few decades, of which wearable robots usually aim at ankle rehabilitation during gait exercises, such as the MIT Anklebot developed by Roy et al. [10] and the bio-inspired soft ankle robotic device developed by Park et al. [11], while platform robots focus solely on ankle training [12-16]. It has been demonstrated by Zhang et al. [17] that robot-assisted rehabilitation robots are effective in reducing ankle impairments caused by either musculoskeletal or neurological injuries. Platform-based ankle robots can be further divided into single-DOF and multi-DOF devices. The single-DOF device is mostly actuated by a single rotating motor while multi-DOF robots are usually actuated in parallel by linear actuators. Further, parallel ankle rehabilitation robots work in different ways depending on the mechanical structure. Robotic systems actuated from the bottom such as [14, 15] require varying positions of the ankle joint and synergic movement of the lower limb from the patient due to the fact that the rotation centre of the robot does not coincide with the ankle joint. By contrast, the patient can keep his/her ankle stationary and fully relaxed on devices actuated from the above [12, 13]. From this point of view, parallel ankle platforms wherein the rotation centre of the robot coincides with the ankle joint are more suitable for rehabilitation purpose. To distinguish the structural differences, I call actuated-from-above parallel ankle robots for A-PARs, while all others for non-A-PARs.

A variety of robot-assisted ankle training strategies have been implemented and evaluated [17]. For single-DOF ankle rehabilitation devices, a typical example is the intelligent stretching device developed by Zhang et al. [16] to treat the spastic/contractured ankle of neurologically impaired patients. Although clinical evidences using combined passive stretching and active movement on children with cerebral palsy [18] and on stroke patients [19] have been demonstrated, its potential applications for ankle rehabilitation are limited since it does not allow for mixed robotic motions of dorsiflexion/plantarflexion (DP), inversion/eversion (IE), and abduction/adduction (AA). Platform-based parallel robots present a comparative advantage for ankle rehabilitation due to the characteristic of multiple DOFs. Non-A-PARs implemented with
advanced control strategies have been extensively researched and evaluated through clinical trials. Rutgers Ankle powered by double-acting pneumatic cylinders were proposed based on the six-DOF Stewart-Gough platform [15]. It has been extensively extended and improved by 1) integrating assessment into rehabilitation; 2) assisting patients in participating virtual-reality games; 3) combining tele-rehabilitation systems [20]; and 4) having clinical trials on subjects with varying grades of ankle sprains [101], patients post-stroke [100], and children with cerebral palsy [83]. More recently, Saglia et al. [14] developed a redundantly actuated parallel mechanism for ankle rehabilitation using three customised cable-driven linear electric actuators for achieving two DOFs (ankle DP and IE). While advanced interaction control algorithms have been implemented with a position control scheme for patient-passive exercises and an admittance control technique for patient-active exercises [21], it does not allow for the adaptation of the parameters of the admittance controller.

A-PARs, by contrast, are still in their early stage in terms of robotic design, advanced control schemes and clinical evaluation, although they have advantages due to compatible robotic structure with ankle anatomy. Tsoi et al. [13] developed a parallel ankle rehabilitation robot using a computed torque impedance controller that regulates the relationship between the applied moments at the joint and the ankle rotary motion. To improve the adaptability of the robot, an online parameter estimation algorithm for biaxial ankle kinematics and a generic computational ankle model were developed to facilitate the real-time adjustment of the robot stiffness so that actuating effort of the robot is reduced in stiff foot configurations to prevent application of excessive forces/moments. Jamwal et al. [12] constructed a wearable ankle rehabilitation robot using four PAMs, and proposed an adaptive fuzzy-logic controller for smooth training. Experimental data were obtained on a neurologically intact subject with a satisfactory trajectory tracking performance in the presence of unpredicted human-robot intervention, use of compliant and nonlinear actuators, and parallel kinematic structure. The proposed adaptation scheme, however, was for the control of PAMs rather than the robotic interaction with human users. To promote the development of A-PARs, patient-cooperative training strategies should be further investigated, which requires robots to guide the motion of the affected ankle, and simultaneously comply with the forces exerted by the patients for enhanced safety.

Patient-cooperative training strategies were defined by Riener et al. [290] as that the robotic system takes into account patients' intention and voluntary efforts rather than solely imposing predefined motions. This kind of cooperative robotic training is expected to improve the
Chapter 9. Interaction Control via an Adaptive Admittance Law

therapeutic outcome compared to classical rehabilitation strategies by introducing a certain amount of robot compliance, and adapting robot behaviour to patients' motor abilities. Blaya and Herr [82] developed a variable-impedance ankle-foot orthosis, and clinical data suggested potential benefits for the treatment of drop-foot compared to conventional ankle-foot orthoses. More recently, Schück et al. [291] tried to validate the feasibility and effects of patient-cooperative robot-assisted gait training through a four-week pilot trial. While the subjects trained more actively and with more physiological muscle activity than in a passive mode, the limited number of subjects did not permit valid conclusions on the effect of patient-cooperative robotic training. In general, the patient-cooperative robot-assisted rehabilitation is feasible in clinical practice and enables patients to train in an active, variable and more natural way, which was also demonstrated by Duschau-Wicke et al. [292] on 11 patients with incomplete spinal cord injury.

![Image of AARR](image)

**Figure 9.1:** An intrinsically-compliant AARR. (a) The AARR; (b1) The robotic end effector with a six-axis load cell for the measurement of patient-robot interaction; (b2) The control box with the layout of electronic components; and (c) The AARR in use on a subject with ankle sprain. The red and blue dotted lines in (a) represent the axes of ankle DP and IE while the pink dotted line in (b1) represents the axis of ankle AA.

Our lab has developed an intrinsically-compliant ankle assessment and rehabilitation robot (AARR) that integrates real-time ankle assessment in terms of joint kinematics and dynamics into three-dimensional robot-assisted rehabilitation, as shown in Figure 9.1. In the present work, a patient-cooperative training strategy has been proposed to encourage active engagement of the patient and achieve enhanced safety during the robotic training. This patient-cooperative robot-assisted ankle training is realised using a joint space position controller and a task space admittance controller. The parameters of the admittance controller is adaptively adjusted based
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on real-time ankle assessment and predefined robotic ROM. Preliminary experiments on a sprained ankle were also performed to evaluate the performance of the adaptive patient-cooperative training on the AARR. This chapter is organized as follows. Section 9.2 presents the robotic design and sensing units of the AARR. Section 9.3 describes the controller design of the joint space position controller, the task space admittance controller, and the novel adaptation law. Section 9.4 presents the preliminary experimental results on a sprained ankle, followed by the discussion and conclusion in Sections 9.5 and 9.6, respectively.

9.2. AARR

The proposed adaptive patient-cooperative training is implemented on an intrinsically-compliant AARR. This robot is shown in Figure 9.1 and more details can be found in Figure 4.8. The AARR is actuated in parallel by four FFMs (Festo DMSP-20-400N) for having three rotational DOFs that are ankle DP, IE and AA, respectively. Four proportional pressure regulators (Festo VPPM-6L-L-1-G18-0L6H) are used for the pressure control of the FFM. The AARR, as a parallel mechanism, consists of a fixed platform and a moving platform, of which the moving one is a three-link serial manipulator with three rotational DOFs. For the robotic sensing function, three magnetic rotary encoders (AS5048A) are installed along each axis for measuring the three-dimensional angular positions of the robotic end effector and the patient’s ankle, a six-axis load cell (SRI M3715C) is located between the footplate and the robotic end effector for measuring the patient-robot interaction forces/torques, as shown in Figure 9.2. It is hypothesized that there is no relative motion between the footplate and the patient’s foot during the robotic training, thus the measured position of the end effector equals that of the involved foot.

![Figure 9.2: Sensors installation in the AARR. (a) the magnetic rotary encoder for measuring the angular position of a single axis; (b) the six-axis load cell (SLC) for detecting patient-robot interaction. FP: Footplate, EE: End effector.](image)

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9.3. Control System

Ankle rehabilitation programs usually involve ROM and muscle strengthening exercises. Active assistance exercises are primarily aimed at neuro-motor training of patients suffering from neurological disorders, while resistive exercises are generally used for strength training of patients due to musculoskeletal disorders. However, adaptive patient-cooperative robotic training are highly relevant for general rehabilitation tasks. To allow the patient to fully regain his/her ankle ROM and strength, the patient-robot interaction is encouraged. This can be achieved by deriving an adaptive patient-cooperative training strategy implemented on the AARR, see Figure 9.3. The proposed adaptive patient-cooperative training strategy consists of a low-level position controller and a high-level admittance controller that are implemented in joint space and task space, respectively. To facilitate the description of the proposed adaptive patient-cooperative training scheme, ankle DP, IE, and AA are also denoted by rotations about axis-X, Y and Z, respectively, as the description in Figures 1.1 and 4.1.

**Figure 9.3:** An adaptive patient-cooperative training strategy implemented on the AARR with a position controller in joint space and an admittance controller in task space. M refers to the inertia matrix of the three-link end effector of the AARR; B and K represent the damping and stiffness of the admittance mechanism respectively; DFFML: Desired Festo fluidic muscle length; MFFML: Measured Festo fluidic muscle strain; PID-C: Proportional-integral-derivative controller; Adaptation law refers to (9.15) and (9.16), and is also shown in Figure 9.4; and ankle force and torque are calculated using (4.15) and (4.16); θr is the reference position, and the position tracking error is denoted as θe, θe=θd-θm, of which θd represents the desired position of the end effector while θm is the measured position; Ti represents the patient-AARR interaction torque.

9.3.1 Joint Space Position Controller

A position controller is the basis of a variety of robot-assisted rehabilitation exercises, while it serves as a low-level controller for the patient-cooperative robotic ankle training in this study. The position control of the robotic end effector of the AARR could be achieved by controlling individual FFM length in joint space. As in Figure 9.3, the desired individual FFM length is calculated by inverse kinematics based on the desired position of the end effector, while, as the feedback to the PID controller, the actual individual FFM length is obtained by inverse kinematics based on measured position of the end effector. This joint space position controller outputs pressure that directly goes to VPPMs to actuate the FFM.
More specifically, the desired trajectory is predefined by a physiotherapist and denoted as $\theta_d(t)$ in (9.1). The measured trajectory is obtained from three magnetic rotary encoders and denoted as $\theta_m(t)$ in (9.1). Individual FFM length can be calculated using (9.2) based on inverse kinematics and AARR configuration, where $l_{4x1}^d(t)$ and $l_{4x1}^m(t)$ respectively represent desired and measured FFM lengths, $\mu$ is a coefficient that relates the FFM length to the link length and depends on the AARR configuration, $\kappa_{4x3}$ relates the link length to the position of the robotic end effector and depends on the inverse kinematics of the AARR. Lastly, the error $e_{4x1}(t)$ shown in (9.3) is the input of the PID controller, and the desired individual FFM pressure can be calculated according to (9.4) with well-tuned $K_p$, $K_i$, and $K_d$.

$$\begin{align*}
\left\{ \begin{array}{l}
\theta_d(t) = [\theta_{dp}^d(t) \ \theta_{de}^d(t) \ \theta_{AA}^d(t)]^T \\
\theta_m(t) = [\theta_{dp}^m(t) \ \theta_{de}^m(t) \ \theta_{AA}^m(t)]^T
\end{array} \right. \quad (9.1)
\end{align*}$$

$$\begin{align*}
\left\{ \begin{array}{l}
l_{4x1}^d(t) = \mu \kappa_{4x3} \theta_d(t) \\
l_{4x1}^m(t) = \mu \kappa_{4x3} \theta_m(t)
\end{array} \right. \quad (9.2)
\end{align*}$$

$$e_{4x1}(t) = l_{4x1}^d(t) - l_{4x1}^m(t) \quad (9.3)$$

$$p_{4x1}(t) = K_p e_{4x1}(t) + K_i \int_0^t e_{4x1}(t) dt + K_d \frac{de_{4x1}(t)}{dt} \quad (9.4)$$

### 9.3.2 Task Space Admittance Controller

The interaction tasks cannot be handled by pure motion control that rejects forces exerted by human users as disturbances. Impedance and admittance control schemes are usually considered as the basis of robotic interaction training. Impedance controllers accept a displacement as input which is measured and react with a force. For use on the AARR, the impedance control law can be considered as (9.5). The desired driven torque can be calculated based on (9.7) in combination with (9.6). An analytic-iterative technique proposed by Taghirad and Bedoustani [237] can be used to distribute the desired robotic torque to the individual desired FFM force. However, it is challenging to be implemented due to the high requirement of the robotic assembly precision. Moreover, engagements from patients may induce sudden force distribution changes on the FFM, which makes the control unstable.

$$T_i(t) - T_i^d(t) = M(\theta_d \ddot{t} - \theta_m \ddot{t}) + B(\theta_d \dot{t} - \theta_m \dot{t}) + K(\theta_d(t) - \theta_m(t)) \quad (9.5)$$

$$T_r(t) - T_i(t) = M\theta_m \ddot{t} + C\theta_m \dot{t} + C_f \theta_m \dot{t} + G \quad (9.6)$$
In admittance control mode, the robot assumes the behaviour of an admittance and its movements are determined by the external force from the patient. Under this control mode, the AARR deviates from the reference trajectory in the presence of patient-robot interaction but is otherwise following the reference trajectory. One of the important functionality of rehabilitation robots is to guide the patient’s affected ankle joint through certain training trajectories. For severely affected joints, the effort required to realise the motion will be completely provided by the rehabilitation robot, and the patient’s ankle will act as a passive environment. As the rehabilitation progresses, a common operation of rehabilitation robots could involve adaptive patient-cooperative training to encourage active engagement.

An adaptive admittance controller is implemented in the task space of the AARR for patient-cooperative training. The admittance control law is proposed as in (9.8), where \( \theta_r(t) \) and \( \theta_d(t) \) represent the reference trajectory and the recalculated desired trajectory, respectively, and \( T_i(t) \) is the patient-robot interaction torque, \( B \) and \( K \) respectively represent the damping and stiffness coefficients. In addition, the end effector of the AARR is a three-link serial manipulator whose inertia tensor \( M \) was obtained from a three-dimensional model of the AARR created in Creo. The specific calculation of the global inertial tensor is based on (9.9), where \( M \) is the global inertial tensor that is the function of the position of the end effector, \( I_i \) is the inertial tensor of the ith link, \( R_i \) is the rotation matrix relating the ith link coordinate frame to the global frame. The global coordinate system is located at the rotation centre of the moving platform with the X-axis pointing to the right, the Y-axis pointing forwards and the Z-axis pointing upwards with respect to human users.

\[
T_i(t) = M(\theta_d \ddot{t} - \theta_r \ddot{t}) + B(\theta_d \dot{t} - \theta_r \dot{t}) + K(\theta_d(t) - \theta_r(t)) \tag{9.8}
\]

\[
M(\theta_{DP}, \theta_{IE}, \theta_{AA}) = \sum_{i=1}^{3} M_i = \sum_{i=1}^{3} R_i I_i R_i^T \tag{9.9}
\]

The integration of the feed forward measured patient-robot interaction torque allows for a variable admittance controller for adaptive training. The patient-robot interaction forces/torques are calculated from the readings of the six-axis load cell using (4.15) [234], where \( F \) and \( T \) represent forces and torques respectively, the script a and slc represent the ankle joint and the
six-axis load cell, $\mathbf{R}_{slc}^a$ is a $3 \times 3$ rotation matrix from frame $a$ to frame $slc$, $P_{slc}^a \times \mathbf{R}_{slc}^a$ is a $3 \times 3$ skew matrix from frame $a$ to frame $slc$, and $P_{slc}^a$ is defined in (4.16).

To ensure the safety of the proposed patient-cooperative training strategy, bounded input and bounded output (BIBO) stability of the admittance controller is conducted. Equation (9.8) can be rewritten in (9.10), and (9.11) is then obtained, where $T_i(t) = M(t)\dot{\theta}_d(t) + B(t)\dot{\theta}_d(t) + K(t)\theta_d(t)$. System transfer function (13) is obtained through Laplace transformation (12). Based on (14), this system is BIBO stable, since all eigenvalues are in the open left half plane with $B > 0$ and $M > 0$.

$$M(t)\ddot{\theta}_d(t) + B(t)\dot{\theta}_d(t) + K(t)\theta_d(t)$$

$$= T_i(t) + M(t)\dot{\theta}_r(t) + B(t)\dot{\theta}_r(t) + K(t)\theta_r(t)$$

(9.10)

$$T_i(t) = T_d(t) - M(t)\ddot{\theta}_r(t) - B(t)\dot{\theta}_r(t) - K(t)\theta_r(t)$$

(9.11)

$$T_i(s) = -[M(s)s^2 + Bs + K]X(s)$$

(9.12)

$$\frac{X(s)}{T_i(s)} = \frac{-1}{M(s)s^2 + Bs + K(s)}$$

(9.13)

$$s = \frac{-B \pm \sqrt{B^2 - 4MK}}{2M}$$

(9.14)

### 9.3.3 Adaptation Law

While the AARR can be programmed to carry out fundamental passive and active training with the joint space position and task space admittance controllers, an adaptive interaction control scheme is required for enhanced safety and rehabilitation efficacy. Adaptation of rehabilitation robots can be realised through modification of either the magnitude of feed-forward force component or the parameters of the interaction controller in response to assessment criteria such as trajectory tracking error [117]. The focus of such adaptation algorithms is that the rehabilitation torques provided by the robot should vary over time to continuously challenge the patient to exert their own effort and thus actively engage in the robotic rehabilitation treatment [293].

It is likely that the stiffness of the patient's ankle joint varies over its ROM, resulting in a rapidly changing ankle torque during training. Therefore, to guarantee the patient's safety, it is important for the rehabilitation robot to provide assistive torques with adjustable compliance. In this study,
an adaptation law is proposed to allow variable admittance parameters based on real-time ankle position and interaction torque measured from the AARR. Specifically, $B$ and $K$ are adapted by rules (15) and (16), where $B_0$ and $K_0$ are base values and $a_1$, $a_2$, $a_3$, $b_1$, $b_2$, and $b_3$ are weighting coefficients that adjust the influences of the angular position and the interaction torque. To reduce the effect of the angular dependency of the patient's passive ankle stiffness, the interaction torque is normalized by a model prediction of the ankle's passive elastic torque. Both stiffness and damping parameters are bounded by a saturation function to guarantee the system stability and training safety. The adaptation method reduces the stiffness and damping with increasing robot angular displacement and increasing interaction torque. This ensures that the patient is able to safely backdrive the robot if his/her ankle is moved into an uncomfortable position, and so prevents a large and potentially harmful contact force.

$$B = \begin{cases} B_1 & \text{if } B < B_{l1} \\ \frac{1}{a_1 e^{a_2 \theta}} & \text{if } B_{l1} \leq B \leq B_{l2} \\ B_{l2} & \text{if } B > B_{l2} \end{cases} \frac{T_{\text{modeling}}}{T_{\text{measured}}} \quad (9.15)$$

$$K = \begin{cases} K_1 & \text{if } K < K_{l1} \\ \frac{1}{b_1 e^{b_2 \theta}} & \text{if } K_{l1} \leq K \leq K_{l2} \\ K_{l2} & \text{if } K > K_{l2} \end{cases} \frac{T_{\text{modeling}}}{T_{\text{measured}}} \quad (9.16)$$

To verify the effectiveness of the admittance controller and its adaptation law, a simulation experiment in MATLAB was conducted and the results are presented in Figure 9.4. It is found that the interaction torque in the admittance controller is able to effectively deviate the robot's movement from the initial reference trajectory. This increases patient motivation and participation in the rehabilitation process as his/her movement intention is reflected by the robot trajectory adaption and thus promotes the notion of progress. Modification of the reference trajectory is also influenced by damping and stiffness parameters that enable the robot to provide assistance with adjustable compliance. When the ankle is highly extended or exerts a large interaction torque, controller compliance is high so the robot movement is much easier to change, and the trajectory deviation is prominent. On the other hand, the trajectory deviation is slight and the robot is running in a more passive way when the robot is in low compliance situation. Meanwhile, constraints on the robot movement range and the controller compliance are enforced by using saturation functions to prevent endangerment to human safety.
Chapter 9. Interaction Control via an Adaptive Admittance Law

![Figure 9.4](image)

**Figure 9.4**: Simulation experiment results of the admittance controller and impedance adaptation law. (a) The recalculated desired trajectory $\theta_d$ is determined by the initial reference trajectory $\theta_r$ and the deviated value $\Delta \theta$, which is calculated by (9.8). Here $-0.1 \leq \Delta \theta \leq 0.1, -0.35 \leq \theta_d \leq 0.35$. The saturation function is denoted by the red and blue dotted lines, respectively. (b) The damping $B$ and stiffness $K$ are determined by the robot’s angular position, interaction torque, and the modeled ankle stiffness. They are adapted by using (9.15) and (9.16). Here, $0.6 \leq B \leq 6, 1 \leq K \leq 10$. The saturation function is denoted by the blue and red dotted lines, respectively. $M$ is the inertial parameter calculated by dynamic model of the AARR. To get a clear view of its changes, $M$ is multiplied by 100 times.

### 9.4. Experimental Results

A subject (male, 29 years; height: 1.7m; weight: 68 kg) with an ankle sprain and no history of neurologic disorders participated in this trial two months after the injury. At this time the swelling was gone but the ankle joint still relatively stiff. The sprained ankle was caused by jumping and rolling during basketball and diagnosed as limited ROM and torn ligaments. The subject gave written consent to participate in the trial according to ethics approval obtained from the University of Auckland, Human Participants Ethics Committee (011904) (Appendix C).

#### 9.4.1 Experimental Protocol

Although the AARR is developed for three rotational DOFs for ankle DP, IE and AA, only training in DP and IE are conducted. Two reasons include 1) the naturally small ROM of ankle
AA that is primarily controlled by leg rotation [285], and 2) the low ROM of the AARR's AA DOF, of which the limited robotic ROM can be improved by optimizing the FFMs' mounting locations in later versions of the AARR.

Before the robotic training, a preliminary assessment was conducted to check the appropriate ROM of the sprained ankle by a physiotherapist. The participant was instructed to sit on a height-adjustable chair with the shank free on the leg holder, as shown in Figure 4.9(c). The ankle-foot complex was strapped into the ankle orthosis. The rotation axis of the ankle orthosis was designed to approximate the ankle joint.

Three types of training are conducted using the AARR: 1) passive mode using pure position control in joint space, 2) patient-cooperative training using a fixed-parameter admittance controller, and 3) patient-cooperative training using a variable-parameter admittance controller. The pure position control is implemented on the AARR along a mixed trajectory of axis-X and Y, while the patient-cooperative training is solely conducted about axis-X. All predefined trajectories were set to operate three cycles at 0.05 Hz. The participant was asked to remain relaxed for the passive training, while active patient involvement was required for the patient-cooperative training. Range of motions for each training are summarised in Table 9.1.

### Table 9.1: Three kinds of training (sinusoid) to be applied to the participant.

<table>
<thead>
<tr>
<th>Training Modes</th>
<th>X (rad)</th>
<th>Y (rad)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Dorsiflexion</td>
<td>Plantarflexion</td>
</tr>
<tr>
<td>PT</td>
<td>0.3</td>
<td>0.3</td>
</tr>
<tr>
<td>PCT with FA</td>
<td>0.2</td>
<td>0.2</td>
</tr>
<tr>
<td>PCT with VA</td>
<td>0.2</td>
<td>0.2</td>
</tr>
</tbody>
</table>

PT: Passive training; PCT: Patient-cooperative training; FA: Fixed admittance; VA: Variable admittance.

### 9.4.2 Model-Based Ankle Torque

The estimated passive ankle torque can be obtained from a computational ankle model that has three rotational DOFs with 12 muscles and seven ligaments, which has been validated in Chapter 7. The modelling results for this participant are presented in Figure 9.5. To facilitate its use with the real-time variable-parameter admittance controller, an approximation equation (9.17) was fitted to the numerical model to evaluate the passive elastic ankle torque, where $p_1 = 5.1992$, $p_2 = 1.6238$, $p_3 = 9.7792$, and $p_4 = 0.2927$. 

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9.4.3 Position Control

One of the important functions of rehabilitation robots is to guide the patient’s affected ankle joint through certain position trajectories. In this study, the position controller was developed in joint space. Experimental results on the sprained ankle are presented in Figures 9.6 and 9.7.

Specifically, the trajectory tracking performance in task space is shown in Figure 9.6, and those of individual muscle length in joint space are plotted in Figure 9.7. The statistical results of the
trajectory tracking accuracy are summarised in Table 9.2 at the end of the Section 9.4, with all NRMSD values less than 6%. The trajectory tracking performance in both task space and joint space is satisfactory in the presence of patients’ disturbance, with all NRMSD values in task space less than 5.4% and that in joint space less than 4.7%.

![Graphs showing muscle length tracking responses](image)

**Figure 9.7:** Joint space length tracking responses of the FFM during passive training.

### 9.4.4 Patient-Cooperative Training

The patient-cooperative training was first evaluated using the fixed-parameter admittance controller in combination with the joint space position controller. The admittance controller's B
and $M$ were fixed at 0.03. The patient-cooperative training was conducted along a sinusoid with an amplitude of 0.2 rad for ankle DP and ankle IE fixed.

Experimental results on the sprained ankle are presented in Figure 9.8, where a satisfactory performance is obtained for the position tracking of the robotic end effector and the real-time patient-cooperative torque is recorded. The statistical results of the trajectory tracking accuracy about X-axis are summarised in Table 9.2, with the NRMSD value being 0.0326. It is illustrated from Figure 9.8 that by using the admittance controller the recalculated desired trajectory deviated from the initial reference path in accordance with the interaction torque about X-axis, while the movement about Y-axis remained stable around its neutral position. The deviation of the reference trajectory was only determined by the interaction torque, without consideration of the patient’s ankle status such as stiffness, position, or passive joint torque. For instance, the recalculated trajectory deviated from its reference path to move towards plantarflexion when the
interaction torque was in negative direction during 0-10s. The robot deviated and moved towards dorsiflexion with respect to its reference during 12-18s when positive interaction torque was detected. The overall AARR ROM was also bounded to prevent ankle hyperflexion or hyperextension, which was reflected between 54 and 57s in Figure 9.8. Specifically, while the interaction torque kept negative and increasing during this period, the recalculate trajectory was bounded to -0.35 rad.

![Figure 9.9: Trajectory tracking responses of the patient-cooperative training with variable admittance.](image)

The patient-cooperative training was then evaluated using the adaptive admittance controller with $B$ and $K$ being adjusted based on (9.15) and (9.16). The patient-cooperative training was also conducted along a sinusoid whose amplitude is 0.2 rad for only ankle DP.
results on the sprained ankle are presented in Figure 9.9. The real-time adaptive damping and stiffness coefficients are recorded as well as the measured patient-robot interaction torque and modelling passive ankle torque. Table 9.2 summarises the statistical results of the trajectory tracking accuracy about X-axis, with the NRMSD value being 3.77%. Furthermore, the ROM of the AARR is bounded between -0.35 and 0.35 rad to avoid injuring the patient when continuous interaction torque exists. In the same way, the deviated movement for a certain moment was limited to between -0.1 and 0.1 rad to prevent sudden changes of the trajectory. The robot damping and stiffness coefficients remain inside a constrained space between 0.01 and 0.1 to guarantee the system stability, again ensuring the patient safety.

It was noticed that the model predicted passive ankle torque was only determined by the angular position of the robotic end effector, which is small when the ankle is close to its neutral position and increases when it deviates from the neutral position. This finding is in accordance with the results presented in Section 9.4.2. The patient-robot interaction torques were obtained based on (4.15) and (4.16) from the six-axis load cell. It can be seen from Figure 9.9 that the interaction torque depends on both the ankle position and the patient's active engagement with the AARR. The interaction torque tends to be larger for a higher ankle extension, although certain rapid changes exist due to the patient's subjective intention. $B$ and $K$ are adapted using (9.15) and (9.16) according to the ankle position, the modelling passive ankle torque as well as the patient-robot interaction torque. It is clearly presented in Figure 9.9 that $B$ and $K$ increased as the interaction torque decreased while decreased as the interaction torque increased during 15-20s. However, the adaptation law is also influenced by the modelling torque and the ankle position. For instance, $B$ and $K$ were small at 4-5s when the interaction torque was low, which can be accounted for by the large ankle flexion and model predicted torque. This novel adaptation law allows the AARR to be more compliant in the extended position compared to the neutral ankle position. Specifically, it is easier for the patient to change the robot movement with a lower $B$ and $K$, for example, during the periods of 30-35s and 52-57s. In contrast, during periods of 7-10s and 38-41s the robot is operating in a passive mode with high $B$ and $K$, limiting backdrivability.

As experimental results presented in Figure 9.8 and 9.9 show, the patient-cooperative training with fixed admittance or variable admittance are able to realise both passive and active training. If a patient does not exert interaction torque above the defined threshold, the robot will operate in a passive mode to assist the injured ankle to track a predefined reference trajectory. Above that interaction torque threshold, the robot will run in the patient-cooperative mode with its
trajectory recalculated based on the admittance control law. The torque threshold was set at 0 Nm in the experiments. The real-time adaptation of the damping and stiffness coefficients of the admittance controller is determined by the measured interaction torque, the modelling passive ankle torque, as well as the angular position of the involved ankle. The adaptive patient-cooperative training strategy, however, presents advantages compared to that with fixed admittance since people naturally think that the robot should be more compliant with a stiffer joint. For instance, adapting robot compliance according to joint stiffness increases robot backdrivability at joint ROM limits, thus making patients safe and comfortable. This realises the remark of Riener et al. [120] who suggested that robot-assisted training with robot-assisted assessment will likely make therapy more comfortable, safe and effective.

<table>
<thead>
<tr>
<th></th>
<th>PT</th>
<th>PCT with FA</th>
<th>PCT with VA</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>X (rad)</strong></td>
<td>RMSD</td>
<td>0.0247</td>
<td>0.0223</td>
</tr>
<tr>
<td></td>
<td>NRMSD</td>
<td>4.12</td>
<td>3.26</td>
</tr>
<tr>
<td><strong>Y (rad)</strong></td>
<td>RMSD</td>
<td>0.0107</td>
<td>0.0016</td>
</tr>
<tr>
<td></td>
<td>NRMSD</td>
<td>5.36</td>
<td>Inf-L</td>
</tr>
<tr>
<td><strong>MS of FFM 1 (%)</strong></td>
<td>RMSD</td>
<td>0.0028</td>
<td>—</td>
</tr>
<tr>
<td></td>
<td>NRMSD</td>
<td>4.34</td>
<td>—</td>
</tr>
<tr>
<td><strong>MS of FFM 2 (%)</strong></td>
<td>RMSD</td>
<td>0.0073</td>
<td>—</td>
</tr>
<tr>
<td></td>
<td>NRMSD</td>
<td>4.71</td>
<td>—</td>
</tr>
<tr>
<td><strong>MS of FFM 3 (%)</strong></td>
<td>RMSD</td>
<td>0.0025</td>
<td>—</td>
</tr>
<tr>
<td></td>
<td>NRMSD</td>
<td>3.77</td>
<td>—</td>
</tr>
<tr>
<td><strong>MS of FFM 4 (%)</strong></td>
<td>RMSD</td>
<td>0.0065</td>
<td>—</td>
</tr>
<tr>
<td></td>
<td>NRMSD</td>
<td>4.20</td>
<td>—</td>
</tr>
</tbody>
</table>

RMSD: Root mean square deviation; NRMSD: Normalized root mean square deviation; RMSD and NRMSD are defined in (7.9) and (7.10), respectively; MS: Muscle strain; PCT: Patient-cooperative training; FA: Fixed admittance; VA: Variable admittance; Inf-L: Infinitely large; — represents not applicable.

### 9.5. Discussion

Considering the control schemes of parallel robots, two typical frameworks are the joint space controller and the task space one. The joint space controller, for position control, is to make actual link lengths conform to desired lengths computed from the required position of the manipulator based on inverse kinematics [286]. In a similar way, a joint space force controller is also achievable if force distribution can be conducted for a given robotic torque. This method, however, is subject to numerical optimisation for minimum energy consumption [237]. In this study, the joint space position controller was selected as the basis of the adaptive patient-
cooperative training strategy, which can be accounted for by two points. One is that the force distribution of the parallel AARR with minimum energy consumption for given robotic torques cannot be obtained in real time. The other is the relative simple inverse kinematics of parallel robots.

Robot-assisted passive training can be achieved by the pure position control of the end effector while patient-cooperative training are commonly implemented by interaction controllers, similar to the work presented by Saglia et al. [21]. The pure trajectory tracking control is to force patients’ limbs on predefined trajectories without taking into account real-time disability level that varies among patients. By contrast, patient-cooperative training strategies usually require the identification of patient-robot interaction force and torque, and is mostly achieved by using impedance or admittance controllers. It is well known that the basis of adaptive interaction control is to modify the robot motion in a way that is desired by the patient, which is believed to be the most appropriate for rehabilitation. However, the issue of reference trajectory adaptation has some drawbacks, for example, the extent of the trajectory adaptation cannot be well determined and the changes in trajectory may result in an un-physiological pattern. To tackle this problem, this kind of interaction control is often companied by adaptation laws for adjusting the impedance or admittance of the robots. The robot can be adjusted from stiff to compliant for patients with varying disability levels. Specifically, the robot behaviour can be set more compliant if the patient has less severe impairment so that the patient can contribute more effort to the robot-assisted training. The robot behaviour can be also made stiff if the patient cannot achieve the required rehabilitation exercises.

Patient-cooperative training strategies of rehabilitation robots can be implemented based on the trajectory tracking error between the target location and the proximity of the subject’s ankle. A typical example is the Anklebot control system developed Roy et al. [10] with positive clinical findings [84, 85, 103]. Adaptive interaction controllers have been developed in a different way. Meng et al. [294] developed an adaptive interaction controller implemented on a six-DOF parallel ankle rehabilitation robot with impedance adaptation according to the patient’s muscle activity level. While this adaptation law was validated on healthy subjects, it relates the muscle activity ratio to only the damping parameters of the impedance controller, also is subject to electromyography signals. Wilkening and Ivlev [295] developed two adaptive model-based assistive controllers for a soft elbow training robot. The patient’s disability level and active engagement have been also considered for advanced adaptation law. Hussain et al. [296] developed an adaptive impedance controller that adjusts the robot assistance according to
disability level and voluntary participation of human subjects, and was implemented on a lightweight robotic gait training orthosis. While it has been demonstrated that real-time estimation of voluntary participation of human subjects could result in adaptive robotic assistance, the identification of the active joint torque is subject to inverse dynamics and estimation of passive joint torque of the lower extremities [297].

The form of therapy may be more important than its intensity: muscle strengthening offers no advantage over movement training [120]. Experienced rehabilitation therapists advocated the assistance-as-needed training strategy [121]. Recent studies have investigated this novel training strategy to keep a challenging assistance level for the patient to avoid slacking. Emken et al. [298] proposed a tracking error-based controller with a forgetting factor for providing only as needed. Wolbrecht et al. [293] introduced the forgetting factor to the adaptive control law to decay the robotic assistance when errors in task execution are small. Banala et al. [299] gradually increased the difficulty by reducing the robotic assistance and increasing the treadmill speed when patients got better tracking performance. These robotic interaction controllers behave like: the robot creates a restoring force if the participant deviates from the desired trajectory, while the robot should not intervene if the tracking error is acceptable. More recently, a novel assistance-as-needed control algorithm was proposed by Pérez-Rodríguez et al. [300] to provide anticipatory actuation to the patients. To tailor the therapy to each patient, Metzger et al. [301] adapted the exercise difficulty for robot-assisted hand training based on the assessment-driven selection.

As each algorithm has its own specific advantages related to the platform on which it is implemented, there is no obvious golden standard for online difficulty adaptation of the robotic training. The proposed patient-cooperative training strategy presents major advantages when combined with the AARR. On one hand, the AARR can conduct real-time ankle assessment. On the other hand, the AARR can drive a computation ankle model to estimate subject-specific passive ankle torque. In this study, the novel adaptation law was proposed to adjust the parameters of the admittance controller based on real-time ankle assessment. The assessment index involves the measurements of the ankle position of the patient, the patient-robot interaction forces/torques, as well as the model-based estimation of the subject-specific passive ankle torque. This novel adaptation law based on real-time patient-robot interaction and joint position is expected to be safer with respect to model-based adaptation schemes. The robotic ROM, the ranges of stiffness and damping coefficients of the admittance controller are also bounded for enhanced safety.
As Krebs et al. [32] suggested, there is no “one-size-fits-all” optimal robot-assisted training strategy and therapy should be tailored to each patient. However, the proposed patient-cooperative training can be considered as a "one-size-fits-most" scheme that is able to conduct both passive and active training. Further, this "one-size-fit-most" training scheme considers real-time ankle assessment in terms of kinematics and dynamics. The proposed patient-cooperative training strategy in this study considers real-time assessment for adaptively adjusting the admittance law.

Although the proposed patient-cooperative training have behaved excellent in the presence of random patient-robot interaction, some limitations should be noted. First, one patient was recruited in the present experiments is to simulate clinical human-robot interaction, which does not require a large sample of participants. Clinical trials do require a large sample of participants with injured ankles when evaluating its therapeutic effects. Second, this patient-cooperative training strategy was proposed for the treatment of a variety of ankle disabilities caused by musculoskeletal or neurological injuries, although only experiments on a sprained ankle were conducted. Thirdly, it is worth exploring what the most optimised parameters of the admittance controller are, although an appropriate set of coefficients have been obtained from experiments. Future work will further the optimisation of the proposed adaptation law with a specific set of adaptation coefficients for each subject.

9.6. Conclusion of the Paper

This study presents a novel adaptive patient-cooperative training strategy that promotes patient involvement and safety on the AARR. Preliminary findings on a sprained ankle are encouraging, demonstrating robust and accurate trajectory tracking, and thus establishing its efficacy for robot-assisted ankle therapy. To the best of the authors’ knowledge, the admittance control of actuated-from-below parallel ankle robots with multiple DOFs has not been reported in literature. Additionally, the proposed adaptation law for adjusting admittance parameters based on real-time ankle position, model predicted ankle torque, and patient-robot interaction torque is a novel work that enhances patient involvement and safety during the robotic ankle training. This will aid more effective and comfortable ankle therapy in future robotic rehabilitation devices.
Chapter 9. Interaction Control via an Adaptive Admittance Law

9.7. Chapter Summary

This chapter presents an adaptive interaction training strategy on the AARR for the treatment of a variety of ankle injuries. By the proposed adaptive admittance law together with the joint space position controller, the AARR can deliver adaptive training based on the real-time patient-robot interaction. If no active ankle torque is detected, the AARR operates in a passive mode along the predefined trajectory. If active torque is identified, the AARR runs in an active interaction mode, where the performance of the admittance controller is adaptively adjusted according to real-time ankle position and interaction torque. One point worth mentioning is that the joint space position controller proposed in this chapter does not require the use of single-axis load cells for measuring contraction force of the FFMs, while the one presented in Chapter 8 does. Experiments were conducted on a sprained ankle to evaluate the proposed interaction training scheme. To summarise, the trajectory tracking performance of the proposed interaction training scheme is satisfactory in the presence of patient-robot interaction, with all NRMSD values less than 5.4%. As the same as the control scheme proposed in Chapter 8, this interaction training strategy is also expected to make the robotic training safe and effective by the consideration of real-time ankle position and patient-robot interaction.
Chapter 10. Conclusions and Future Work

This chapter seeks to summarise the main outcomes and conclusions of the research presented within this thesis. The development of the ankle assessment and rehabilitation robot (AARR), robot-assisted ankle assessment techniques including both sensor-based and model-based methods, and adaptive interaction training schemes, have been presented and discussed in previous chapters. This chapter also provides recommendations for future work that should be further explored.

10.1. Outcomes and Contributions

This thesis begins with an independent chapter of Introduction (Chapter 1) to give the readers an overview of the significance of this research and the outline. A published literature review (Chapter 2) was then conducted to evaluate the effectiveness of various ankle rehabilitation devices to identify research gaps. To better understand the integration of the function of real-time assessment into ankle rehabilitation robots, another review was conducted (Chapter 3) to identify potential ankle assessment techniques in robot-assisted therapy.

The next six chapters lead to three major outcomes. The first outcome, presented in Chapter 4, is the development of an intrinsically-compliant AARR that provides advantages over other ankle rehabilitation devices, including compliant actuation, a compatible mechanical structure that allows the ankle to be stationary during the training, three degrees of freedom (DOFs) for three-dimensional exercises, and the integration of real-time biomechanical assessment to allow for adaptive interaction training. The second outcome is the development of robot-assisted assessment techniques. The built-in sensors of the AARR and computational models can be used to estimate the position and torque of the ankle joint, also provide biomechanical information of ankle muscles and ligaments. Specifically, Chapter 5 presents an assessment technique in measuring ankle position and patient-robot interaction torque via sensors, Chapters 6 and 7 introduce model-based assessment techniques for estimating ligament kinematics and passive torque of the ankle joint. The third major outcome is the implementation of adaptive interaction training strategies on the AARR. In Chapter 8, an intelligent training strategy was proposed and implemented for adaptive training together with a cascade controller. This interaction training was achieved by defining an adaptive predefined trajectory based on the movement intention of the patient. Chapter 9 introduces another type of interaction training scheme via an adaptive
admittance controller. The parameters of the admittance law are adaptively adjusted based on real-time ankle position and patient-robot interaction for enhanced training safety and efficacy. To make the major outcomes and contributions of this research clear, Table 10.1 is presented and details are provided in the following subsections.

**Table 10.1**: Summarised outcomes and contributions of this research

<table>
<thead>
<tr>
<th>No.</th>
<th>Outcomes and Contributions</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>Development of the AARR</td>
</tr>
<tr>
<td>2</td>
<td>Robot-assisted ankle assessment</td>
</tr>
<tr>
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</tr>
<tr>
<td>3</td>
<td>Adaptive interaction training</td>
</tr>
<tr>
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</table>

### 10.1.1 Development of the AARR

A novel parallel robot has been proposed and developed for the treatment of musculoskeletal and neurological ankle injuries. While a range of robots have been developed and demonstrated to be effective for ankle rehabilitation [17], existing ankle devices suffer from a variety of limitations, including the incompatible robot structure that requires varying ankle positions during the robotic training, stiff actuation, a non-three-dimensional workspace, and the lack of real-time ankle assessment and adaptive interaction training schemes.

This research overcomes the above-mentioned limitations by developing an intrinsically-compliant AARR. First, this robot has a compatible movement structure that allows the ankle to be stationary during the robotic training, which is achieved by mimicking the configuration and actuation of the ankle joint by natural muscles. Secondly, four intrinsically-compliant Festo fluidic muscles (FFMs) are adopted for soft actuation to make the robotic training safe and comfortable for patients. Thirdly, this robot has three rotational DOFs that are redundantly actuated by four FFMs. These three DOFs are for ankle dorsiflexion/plantarflexion, inversion/eversion and adduction/abduction, respectively. Lastly, but importantly, as the name AARR suggests, real-time assessment function is integrated into the AARR to facilitate the implementation of advanced interaction training strategies. The ankle assessment technique used on the AARR was proposed based on a recent review [22], including both sensor-based and model-based methods. Additionally, the AARR has flexibility in adjusting the robotic range of motion (ROM) and placement for use on different patients. The developed AARR in this research has advantages over others, including intrinsic compliance, a compatible robot
structure that allows the ankle to be stationary during the training, three DOFs for three-dimensional rehabilitation exercises, and the integration of real-time ankle assessment.

10.1.2 Ankle Biomechanical Assessment via Sensors

The measurement of ankle position and torque/stiffness allows for effective monitoring during a rehabilitation program. Existing assessment techniques have a variety of limitations when used in robot-assisted ankle therapy [22]. For example, dynamometer-based methods rely on manual manipulation and torquemeter-based methods are mostly used in single-DOF devices. To conduct a biomechanical assessment of the ankle joint during the robotic training, this research proposed a novel assessment technique that is integrated on the AARR with three rotary sensors measuring ankle orientation and a six-axis load cell detecting real-time patient-robot interaction.

Moreover, to evaluate the validity and reliability of the proposed assessment technique, a manual device was constructed for ankle assessment in dorsiflexion and plantarflexion, and experiments were conducted on nine healthy subjects. It was demonstrated that the measured ankle position was consistent with those from the NDI Polaris optical tracking system, and the measured ankle torque/stiffness compared well with published data. Using repeatable trials by the same operator, the test-retest reliability was demonstrated to be high, with the intraclass correlation coefficient (ICC$_{2,1}$) values greater than 0.846 and standard error of measurement (SEM) less than 1.38. A novel assessment technique was proposed and validated in this research for measuring ankle position and torque/stiffness (using rotary sensors and six-axis load cells), and its potential to be combined with robot-assisted ankle rehabilitation has been demonstrated.

10.1.3 Ankle Biomechanical Assessment via Computational Models

Ankle biomechanical assessment via sensors has been identified to be suitable and effective in measuring the overall characteristics of the ankle joint. However, the commonly-used sensors built in the robot cannot do ankle assessment at the level of ligaments and muscles. While the six-axis load cell is able to estimate real-time patient-robot interaction during the robotic training, its passive component cannot be extracted. Under these circumstances, model-based techniques can be adopted to conduct ankle assessment at the level of ligaments and muscles and can also estimate passive joint torque. Two typical approaches are finite element analysis-based methods and multi-body kinematics/dynamics-based methods. Multi-body-based methods are more suitable for real-time assessment due to their highly efficient algorithms [258] with respect to the finite element method. Further, while multi-body-based models have been
developed for a variety of applications [259-261], they mostly rely on real-time inputs from the motion analysis system allowing continuous updating, and at a significant cost.

To allow for real-time estimation of ankle ligament kinematics and passive joint torque during the robotic training, this research developed a Non-PRA-Model and a PRA-Model. The Non-PRA-Model presented in Chapter 6 has three rotational DOFs and 12 ankle ligaments. This model uses inputs for three position variables that can be measured from sensors built in the robot and outputs the kinematics of ankle ligaments. Model validation was conducted by comparing the results of the model with published data from cadaver anatomy and magnetic resonance imaging (MRI) in terms of ligament length and strain as the foot rotates. It was demonstrated that this computational model has the potential to be used in robot-assisted ankle therapy for real-time assessment of ligament kinematics. This information can be used for the quantification of ligaments related to disability and also for optimising the robotic training trajectory. Further, in Chapter 7, the Non-PRA-Model was updated by including 12 muscles and using a different model structure for predicting passive ankle torque. This model assumes that the ankle movement rotates around a pivot or three perpendicular rotation axes, which is consistent with the motion centre of the AARR. In general, the proposed model-based assessment techniques have been demonstrated to be able to estimate ankle ligament kinematics and passive joint torque during the robotic training.

10.1.4 Interaction Control via an Adaptive Predefined Trajectory

Position control is the basic requirement of rehabilitation robots to conduct training exercises. Satisfactory trajectory tracking performance cannot only guarantee safe and effective passive training, it also provides the basis for advanced interaction training. In this research, a cascade controller with position feedback in the outer loop and force feedback in the inner loop was developed and implemented on the AARR. This method can guarantee good tracking of the predefined trajectory, as well as allowing four FFMs to all be in tension during operations, which is safe for patients since robots may lose controllability with certain cables being slack during the robotic training. It must be noted that the implementation of the cascade controller involves the force distribution of each FFM for a given robotic torque. I adopted an analytic-iterative method that has less computation burden with respect to numerical algorithms and leads to less energy consumption with respect to analytic methods. Further, considering the cascade position controller in the low level, a high-level intelligent algorithm was proposed for adaptive interaction training. This intelligent strategy is implemented by defining an adaptive predefined trajectory based on the movement intention of the patient.
To evaluate the performance of the cascade position controller as well as the intelligent algorithm, experiments were conducted with an individual who had a sprained ankle using the AARR. The patient-robot interaction was kept as passive when the robot was operated only by the cascade position controller. On the basis of this position controller, the intelligent training algorithm was then evaluated as a high-level controller when the patient’s movement intention was encouraged for adaptive interaction training. All training shows a satisfactory trajectory tracking performance in the presence of patient-robot interaction, with all normalized root mean square deviation (NRMSD) values less than 2.3%. To summarise, the proposed intelligent training algorithm together with the cascade position controller have the potential for clinical applications of ankle rehabilitation, especially when implemented on the AARR. With the consideration of the movement intention of the patient, this novel interaction training scheme can adaptively adjust the predefined trajectory to be in a safe range, and comfortable and effective for the patient.

10.1.5 Interaction Control via an Adaptive Admittance Law

Chapter 8 proposed an interaction training scheme via an adaptive predefined trajectory based on the movement intention of the patient, while the other interaction training scheme presented in Chapter 9 takes into account real-time ankle position and patient-robot interaction for adaptation. The proposed patient-cooperative training strategy in Chapter 9 is implemented by a high-level adaptive admittance controller and a low-level position controller. The low-level position controller is different from the cascade position controller presented in Chapter 8 as the cascade controller can guarantee both the trajectory tracking of the end effector in task space and force tracking of the FFMs in joint space, while the one in Chapter 9 achieves the trajectory tracking of the robotic end effector by controlling the individual length of the FFMs in joint space using inverse kinematics. In this situation, the Chapter 9 position controller is simple to be implemented and suitable when the contraction force of the FFMs cannot be directly measured.

To make the proposed patient-cooperative training safe and comfortable for patients, a high-level variable-parameter admittance controller was proposed to be combined with the low-level position controller. Specifically, the admittance law-based interaction controller can be adaptively adjusted using a novel adaptation law that tunes the controller parameters according to real-time ankle position and patient-robot interaction torque. Through experimental evaluations (passive mode using the low-level position controller, patient-cooperative mode with a fixed-parameter admittance controller, and patient-cooperative mode using a variable-
Chapter 10. Conclusions and Future Work

parameter admittance controller) on a sprained ankle using the AARR, it was found that the trajectory tracking accuracy is satisfactory with NRMSD values of all less than 5.4%. As with the interaction training scheme proposed in Chapter 8, this interaction training strategy is also expected to make the robotic training safe and effective by the consideration of real-time ankle position and patient-robot interaction.

While these two interaction training schemes are both capable of conducting adaptive passive and active training, the one presented in Chapter 8 is more suitable for passive training and the one in Chapter 9 is better for active training. However, further research should be conducted to compare the effectiveness of these two interaction training schemes.

10.2. Future Work

10.2.1 Optimisation and Improvement of the AARR

While the developed AARR in this research is flexible in generating varying ROMs, driven torques and the overall placements of the robot, it suffers from a few limitations in terms of the mechanical design. Three of these limitations are discussed further and potential improvements during future work are proposed. The distance between the upper platform and the robotic end effector is too long; this distance is not conveniently and easily adjusted. Pure robotic training for ankle adduction/abduction is impracticable based on the current layout of the FFMs. Finally, the installation of the FFMs is not robust. Therefore, future work can focus on the optimisation and improvement of the current AARR in relation to the following aspects.

Mechanically, 1) designing more flexibility to allow for the adjustment of the height and inclination angle of the rehabilitation robot, to allow the robot to be set to any posture according to the requirements of different patients; 2) replacing the cables used for connecting FFMs by ball joints or universal joints to make the actuation system more stable and accurate; and 3) decreasing the distance between the upper platform and the robotic end effector to allow the patient to easily interface with the robot (may be achieved by using shorter actuators, removing single-axis load cells, or other methods such as structure optimisation). Functionally, the achieved ROM and driven torque of the robot can be improved by optimising the connection point of the FFMs.

In brief, in the process of the commercialisation of the AARR, mechanical design optimisation should be conducted to allow the robot in an optimal performance. A friendly appearance of the robot will also help with obtaining buy-in from patients and physiotherapists.
Chapter 10. Conclusions and Future Work

10.2.2 Ankle Assessment in a Three-Dimensional Space

The proposed sensor-based ankle assessment technique in the AARR is implemented by three rotary sensors for measuring ankle position and a six-axis load cell for measuring the patient-robot interaction torque. To evaluate the validity and reliability of this technique, a single-DOF manual device was constructed for ankle assessment in dorsiflexion and plantarflexion. Experimental results from nine healthy subjects demonstrated the feasibility of this sensor-based ankle assessment for robot-assisted therapy. However, this assessment technique was validated only for ankle training in dorsiflexion and plantarflexion in a manual mode. Future work should focus on the evaluation of this proposed ankle assessment technique in a three-dimensional space using the AARR.

As presented in Chapter 6 and Chapter 7, the Non-PRA-Model and the PRA-Model were developed to measure ligament kinematics and passive joint torque, respectively. However, the Non-PRA-Model was only validated by comparing simulation results with published data, and further evaluation should be conducted the consideration of subject-specific adaptation. The subject-specific ligament length can be measured using a MRI technique or ultrasound scan. Further, understanding how ligament length changes when the ankle is moved away from its neutral position will also contribute to the model validation. The PRA-Model presented in Chapter 7 is slightly different from the Non-PRA-Model presented in Chapter 6. The PRA-Model includes both ligaments and muscles and was developed to predict passive ankle torque with subject-specific adaptation. The model validation exists only in ankle dorsiflexion and plantarflexion using the data collected in Chapter 5. A three-dimensional validation of this model should be further conducted using the data collected from the AARR.

10.2.3 Determination of the Optimal Training Trajectory

While the model presented in Chapter 6 has been demonstrated to be able to estimate ankle ligament kinematics and provide the basis for the optimisation of the robotic training trajectory, the optimisation algorithm has not been investigated yet in this research. The optimal robot-assisted rehabilitation trajectory for a specific injury can be determined based on real-time assessment of ankle ligament kinematics. Specifically, different training trajectories will lead to different kinematics of ankle ligaments, thus there must be a theory-based optimal rehabilitation trajectory that can challenge the ligament tension to the utmost. For a single-ligament injury, an iteration technique can be used to identify the optimal rehabilitation trajectory with the objective function being the maximisation of the ligament strain. For ankle injuries of multiple ligaments, the objective function can be considered as in (10.1) or (10.2), where n represents the amount...
of the injured ligaments, $S_i$ and $k_i$ represent the strain and the weight factor of the $i$th ligament, respectively.

$$\max \sum_{i=1}^{n} (k_1S_1 + k_2S_2 + \cdots + k_nS_n)$$  \hspace{1cm} (10.1)

$$\max \sum_{i=1}^{n} \sqrt[k_1]{(S_1)^2} + \sqrt[k_2]{(S_2)^2} + \cdots + \sqrt[k_n]{(S_n)^2}$$  \hspace{1cm} (10.2)

10.2.4 Computational Models to be Used on Patients

While the Non-PRA-Model and the PRA-Model have been demonstrated to be valid in estimating ankle ligament kinematics and passive joint torque (Chapters 6 and 7), they are only experimentally validated on healthy subjects. One point to be noted is that rehabilitation robots are developed mostly for use on patients rather than healthy users. The current model assumed normal elastic properties of the ligaments and muscles and thus it could reliably predict passive ankle torque for healthy subjects. The prediction could be inaccurate (due to abnormal muscle and ligament property) if the robotic training is delivered to ankle-impaired subjects. For example, some pathological conditions have not just stiffer ligaments, but shortened ones. Therefore, the computational models developed in this research should be further improved by considering patients' ankle disabilities caused by musculoskeletal or neurological injuries.

10.2.5 Optimisation of the Adaptation Law

Chapter 9 presents an adaptive admittance controller to adjust the robotic training trajectory based on real-time patient-robot interaction. The proposed adaptation law allows the robot to provide assistive torques with adjustable compliance for enhanced training safety. The method reduces the stiffness and damping of the AARR as the angular position of the robotic end effector and the interaction torque increase. However, the base values $B_0$ and $K_0$, and weighting coefficients $a_1, a_2, a_3, b_1, b_2, \text{ and } b_3$ have not been optimised for optimal control performance of the AARR. Future work will focus on the optimisation of these parameters to adjust the influences of the angular position and the interaction torque for an individual.

10.3. Summary

This chapter summarises the major outcomes and contributions achieved in this research, and points out some work that can be further explored in future. In general, this research has significantly contributed to the improved effectiveness of robot-assisted ankle rehabilitation.
techniques in relation to three aspects. First, an intrinsically-compliant AARR was developed. This robot is considered to be an optimal ankle rehabilitation platform since it has advantages including compliant actuation, three DOFs for three-dimensional rehabilitation exercises, and the integration of sensor-based ankle assessment. Secondly, assessment techniques via sensors and computational models were developed to facilitate the implementation of advanced training on the AARR. Thirdly, adaptive interaction training schemes were implemented on the AARR by the consideration of the movement intention of the patient and real-time patient-robot interaction. By using the rehabilitation techniques developed in this research, the treatment of ankle injuries can be more comprehensive, safe and effective.
### Appendix A. Data of Ankle Assessment via Sensors

#### Table A.1: MPD and MPP of each participant.

<table>
<thead>
<tr>
<th>Subjects</th>
<th>With the same rater</th>
<th>Trial 1</th>
<th>Trial 2</th>
<th>Trial 3</th>
<th>Trial 4</th>
<th>Trial 5</th>
</tr>
</thead>
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<td></td>
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<td>MPD</td>
<td>MPP</td>
<td>MPD</td>
<td>MPP</td>
<td>MPD</td>
</tr>
<tr>
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<td>41.3</td>
<td>-44.3</td>
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<tr>
<td>2</td>
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<td>-42.2</td>
<td>39.2</td>
<td>-42.1</td>
<td>37.2</td>
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<td>-46.7</td>
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<tr>
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<td>41.1</td>
<td>-44.5</td>
<td>38.5</td>
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<td>41.2</td>
<td>-46.0</td>
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</tr>
<tr>
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</tr>
<tr>
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<td>39.6</td>
<td>-61.2</td>
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<td>49.31</td>
<td>39.61</td>
<td>49.27</td>
<td>38.87</td>
</tr>
<tr>
<td>SD</td>
<td></td>
<td>4.09</td>
<td>5.44</td>
<td>3.29</td>
<td>7.45</td>
<td>3.02</td>
</tr>
</tbody>
</table>

MPD: Maximum position of ankle dorsiflexion; MPP: Maximum position of ankle plantarflexion; SD: Standard deviation.

#### Table A.2: PPTD and PPTP of each participant.

<table>
<thead>
<tr>
<th>Subjects</th>
<th>With the same rater</th>
<th>Trial 1</th>
<th>Trial 2</th>
<th>Trial 3</th>
<th>Trial 4</th>
<th>Trial 5</th>
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</thead>
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<tr>
<td></td>
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<td>PPTD</td>
<td>PPTP</td>
<td>PPTD</td>
</tr>
<tr>
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<tr>
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<td>9.38</td>
<td>8.46</td>
<td>9.16</td>
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<tr>
<td>SD</td>
<td></td>
<td>2.55</td>
<td>1.55</td>
<td>2.620</td>
<td>1.44</td>
<td>2.96</td>
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</table>

PPTD: Peak passive torque of ankle dorsiflexion; PPTP: Peak passive torque of ankle plantarflexion; SD: Standard deviation.
Table A.3: PS-20°D and PS-30°D of each participant.

<table>
<thead>
<tr>
<th>Subjects</th>
<th>Trial 1</th>
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<th>Trial 3</th>
<th>Trial 4</th>
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</thead>
<tbody>
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<td>20°D</td>
<td>30°D</td>
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Mean: 0.1942, 0.2914, 0.1862, 0.2787, 0.1885, 0.2897, 0.1814, 0.2733, 0.1887, 0.2739
SD: 0.0706, 0.1235, 0.0632, 0.1314, 0.0752, 0.1543, 0.0573, 0.1247, 0.0681, 0.1158

PARTICIPANT INFORMATION SHEET
(Participants)

Project title: Ankle assessment techniques and assistance-as-needed control for ankle rehabilitation robot
Name(s) of Researcher(s): Mingming Zhang, Prof. Shane Xie

Researcher Introduction

This research will be conducted by Mingming Zhang, a PhD student in the Department of Mechanical Engineering. The research is supervised by Prof. Shane Xie, a professor in the Department of Mechanical Engineering.

Project Description and Invitation

This project aims to develop an effective control strategy for a robotic device designed to facilitate ankle rehabilitation. The device is intended for use with patients suffering from musculoskeletal injuries at the ankle joint or neurological diseases which impair ankle and foot movement. In particular, the part of the research where human participation is required is designed to evaluate the performance of the developed ankle assessment techniques and assistance-as-needed control strategy for ankle rehabilitation robot. At the same time, information regarding the physical characteristics of the ankle such as stiffness and ankle kinematics will also be identified to facilitate development of the controller and the robotic device.

You are invited to participate in this research by carrying out a series of exercises using a prototype of this ankle assessment and rehabilitation device. Potential participants in this research are chosen from staff and students from the department of mechanical engineering who are healthy individuals over the age of 16 and have no existing ankle or foot impairment. It is preferred that you are familiar or comfortable with the operation of robotic devices. You will be shown the experimental setup and briefed about the operation of the prototype prior to your commitment to participation in this research. Your identity will be kept confidential from third parties.

Project Procedures

You will only be required to take part in no more than five experimental trials. The duration of each trial is expected to span approximately one hour or less. The experimental trial will be carried out in the Mechatronics laboratory at the University of Auckland’s School of Engineering.

Before the experimental trials begin, you will be given an explanation of the operation of the robotic devices, in particular the safety measures put in place to allow termination of the robot operation should an emergency situation arise. A brief demonstration of the prototype will also be given. Your age, gender and body weight will also be collected so that the results can be normalized with respect to the body weight and related to the corresponding demographic groups.

After the briefing and collection of information, your foot and lower limb will be strapped in place on the prototype device. The experimental trials consists of two parts, the first involves the passive movement of the ankle and foot structure by the robotic device. During this part of the trial, you should relax your foot on the robotic device and allow the robot to move the foot passively along a predefined motion path. During this time, information regarding the orientation and position of the foot and the forces applied on the robot will be logged and analysed to allow computation of the ankle kinematic and stiffness parameters.

In the second part of the trial, you will be required to move your foot actively in the directions given by the user interface. The robot will provide a resistance against the motion during this part of the trial. Again, sensor information on the robotic device will be logged and analysed. This part of the trial will allow determination of the active forces applied by the foot to the robot.

You may experience a small level of physical discomfort during a normal trial. However, should the level of discomfort exceed that of your liking, you can terminate the experimental trial by either indicating to the researcher or by using the emergency stop button provided. Since the actuators used in the prototype can produce significant
Appendix B. Ethics Approval (9348)

forces, there is a possibility of injury should the hardware or software of the prototype malfunctions. This will be very unlikely as the prototype would have been tested extensively before the commencement of the experimental trial. In addition to that, the researcher or another individual will be present to provide aid if required. To ensure that prompt medical attention is available during emergencies, the trials will be conducted during the operating hours of the University Health Services clinic at the city campus.

Data Storage/ Retention/ Destruction/ Future Use

As discussed in the previous section, data collected will be in the form of sensor readings which are used to compute the motion and force/moment observed during the experimental trial. This information will also be used to identify properties of your ankle joint. Additionally, your age, gender and body weight will also be recorded. All such data will be stored electronically on a computer hard drive kept in a secure location. The data will be stored in such a way that a third party will not be able to identify you through the information stored on the data file. The information collected will be kept for a period of up to six years as reference for current and possibly future research, which include, but not limited to presentations, conference and journal papers, theses, and device calibration. When no longer required, such data files will be destroyed through permanent deletion. If you are interested, you can also arrange with the researcher a suitable time to have a discussion about any information derived from offline analysis of the collected data.

With your permission, a photograph/video of the experimental trial may also be taken using an electronic device. The photograph/video of the trial will also be stored electronically on a hard drive kept in a secure location. The video will be taken in such a way that it will provide minimal features which can be used to reveal your identity. You are able to request the recording process of both photographs and videos to be stopped at any time during the experiment. You are able to view and edit, or withdraw your video and/or photographic recording data from the research within one month from the date of the experiment. Videos and photographs recorded will be used by the research team to further develop the device in the future and may also be used in publications including, but not limited to, theses, conference presentations and journal papers.

All data (including sensor data, videos and photographs) released outside of the research team will only be presented in a way that does not identify you.

Right to Withdraw from Participation

You may withdraw from participation at any time. You may also withdraw the data collected during your experiment from the research within one month from the date of the experiment.

Anonymity and Confidentiality

Your identity will be kept confidential from all third parties. If the data collected is used in publications, you will be referred to using a generic identifier such as “participant A”. However as videos or photographs recorded will contain items of clothing you are wearing at the time of the trial, anonymity may not be guaranteed. However recordings will be performed in the manner as to minimise the risk of identification.

Participation in this research is entirely voluntary. Assurance from the head of department has been obtained such that neither your grades nor your relations with the university will be affected by agreement/refusal to participate in this research.

Incidental Findings

There is a small chance where an anomaly may be found in the data collected regarding the characteristics of your foot and ankle. Should this occur, you will be informed and should consult a qualified physiotherapist or your general practitioner to verify such findings.

Compensation

In the unlikely event of a physical injury as a result of your participation in this study, you may be covered by ACC under the Injury Prevention, Rehabilitation, and Compensation Act 2001. ACC cover is not automatic, and your case will need to be assessed by ACC according to the provisions of the Injury Prevention, Rehabilitation, and Compensation Act 2001. If your claim is accepted by ACC, you still might not get any compensation. This depends on a number of factors, such as whether you are an earner or non-earner. ACC usually provides only partial reimbursement of costs and expenses, and there may be no lump sum compensation payable. There is no cover for
mental injury unless it is a result of physical injury. If you have ACC cover, generally this will affect your right to sue the investigators.

If you have any questions about ACC, contact your nearest ACC office or the investigator.

You are also advised to check whether participation in this study would affect any indemnity cover you have or are considering, such as medical insurance, life insurance and superannuation.

**Contact Details and Approval Wording**

Researcher:
Mingming Zhang  
Email: mzha130@aucklanduni.ac.nz  
Phone: 022 0949-220

Supervisor:
Prof. Shane Xie  
Email: s.xie@auckland.ac.nz  
Phone: 09 373-7599 extn. 88143

Head of Department:
Prof. Brian Mac  
Email: b.mace@auckland.ac.nz  
Phone: 09 373-7599 extn. 88148

For any queries regarding the ethical concerns you may contact the Chair, The University of Auckland Human Participants Ethics Committee, The University of Auckland, Office of the Vice Chancellor, Private Bag 92019, Auckland 1142. Telephone 09 373-7599 extn. 83711.

APPROVED BY THE UNIVERSITY OF AUCKLAND HUMAN PARTICIPANTS ETHICS COMMITTEE  
REFERENCE NUMBER 9348

**Participants wanted for experimental trials on ankle rehabilitation robot**

**Project Title:** Ankle assessment techniques and assistance-as-needed control for ankle rehabilitation robot  
**Names of Researchers:** Mingming Zhang, Prof. Shane Xie

Human participants are currently being sought to take part in experimental trials for a research project. This research involves the development of an ankle assessment and rehabilitation robot. Invitation to participate in this research is extended to postgraduate students and staff members who are healthy individuals with no ankle and foot impairment. The experimental trial will involve physical interaction between the participant and a prototype ankle assessment and rehabilitation device.

The experimental trials are aimed at evaluating the effectiveness of an ankle assessment and rehabilitation device. The trials will require the participant to undergo two types of exercises on the robot, the first involves the passive movement of the foot by the robotic device and the second requires the participants to actively move their foot against resistance provided by the robotic device. In terms of time commitment, participants are required to attend no more than five experimental trials where the duration of each of these trials are expected to be about 60 minutes.

If you are interested in participating in this research and/or would like to learn more about the experimental trials, please contact the researchers through the contact details listed below.

**Researcher (PhD student):**  
Mingming Zhang  
Email: mzha130@aucklanduni.ac.nz  
Phone: 022 0949-220

**Supervisor:**  
Prof. Shane Xie  
Email: s.xie@auckland.ac.nz  
Phone: 09 373-7599 extn. 88143
CONSENT FORM
(Participants)

THIS FORM WILL BE HELD FOR A PERIOD OF 6 YEARS

Project title: Ankle assessment techniques and assistance-as-needed control for ankle rehabilitation robot
Name(s) of Researchers(s): Mingming Zhang, Prof. Shane Xie

I have read the Participant Information Sheet, have understood the nature of the research and why I have been selected. I have had the opportunity to ask questions and have them answered to my satisfaction. I have chosen to participate in this research on a voluntary basis.

- I agree to take part in this research.
- I understand that participation in this research will require physical interaction with a prototype robotic device.
- I understand that there is a small possibility of injury should the prototype device malfunction, and I accept this risk.
- I understand that my identity will not be revealed in publications derived from this research.
- I understand that the information related to the experimental trial which I am involved in will be recorded using sensors on the prototype device and stored electronically with secure access.
- I understand that I am free to withdraw participation at any time, and to withdraw any data traceable to me up to one month after the collection of the data.
- I understand that participation in this research will not affect my grades or my relations with the university (where applicable) as assured by the head of department of Mechanical Engineering.
- I agree / do not agree to be videotaped and photographed.
- I understand that I can request to stop the recording process at any time.
- I understand that I am free to view, edit or withdraw any photographic or video recordings of the trial involving me for up to one month after the collection of the data.
- I understand that the data will be kept for 6 years, after which they will be destroyed.
- I understand that I am only required to attend one 60 minute experimental trial.
- I agree to be informed of any incidental findings that arise from this research.
- I do / do not wish to have a follow up discussion of the results obtained from my experimental trial. Contacting email address ________________________________

Name ________________________________
Signature ________________________________ Date ________________________________

APPROVED BY THE UNIVERSITY OF AUCKLAND HUMAN PARTICIPANTS ETHICS COMMITTEE
REFERENCE NUMBER 9348
Appendix C. Ethics Approval (011904)

PARTICIPANT INFORMATION SHEET
(Participants)

Project title: Robot-Assisted Ankle Assessment and Rehabilitation
Name(s) of Researcher(s): Mingming Zhang, Prof. Shane Xie

Researcher Introduction

This research will be conducted by Mingming Zhang, a PhD student in the Department of Mechanical Engineering. The research is supervised by Prof. Shane Xie, a professor in the Department of Mechanical Engineering.

Project Description and Invitation

This project aims to develop an effective control strategy for a robotic device designed to facilitate ankle assessment and rehabilitation. The device is intended for use with patients suffering from musculoskeletal injuries at the ankle joint or neurological diseases which impair ankle and foot movement. In particular, the part of the research where human participation is required is designed to evaluate the effectiveness of the developed ankle assessment and rehabilitation robot. At the same time, information regarding the physical characteristics of the ankle kinematics and dynamics will also be identified to facilitate the development of the control of the robotic device.

You are invited to participate in this research by carrying out a series of exercises using a prototype of this ankle assessment and rehabilitation robot. Potential participants in this research are chosen from individuals who are over the age of 16 and have existing ankle injuries, like stroke patients. More specifically, they will be selected by the researcher with the help of a doctor/physiotherapist based on the predefined criteria (Participants who would like to improve their ankles' motor function and should be able to conduct a certain ankle range of motion (≥10°) in one direction without any pain will be chosen to participate in this study). It is preferred that you are familiar or comfortable with the operation of robotic devices. You will be shown the experimental setup and the operation of the prototype prior to your commitment to participation in this research. Your identity will be kept confidential from third parties.

Project Procedures

You will only be required to take part in five to ten trials. The duration of each trial is expected to span approximately two hours or less. The experimental trials will be carried out in the Mechatronics laboratory in the University of Auckland.

Before the experimental trials begin, you will be given an explanation of the operation of the robotic devices, in particular the safety measures put in place to allow termination of the robot operation should an emergency situation arise. A brief demonstration of the prototype will also be given. Your age, gender and body weight will also be collected so that the results can be normalized with respect to the body weight and related to the corresponding demographic groups.

After the briefing and collection of information, your foot and lower limb will be strapped in place on the prototype device. The experimental trials involve the passive rehabilitation training of the ankle and foot structure by the robotic device. During this part of the trial, you should relax your foot on the robotic device and allow the robot to move the foot passively along a predefined motion path. During this time, sensor-based information will be logged and analysed to allow computation of the ankle kinematics and dynamics. Further, you will be required to move your foot actively in the directions given by the user interface. The robot will provide assistance or resistance based on real-time ankle capacity. In the same way, sensor-based information will be logged and analysed to allow more advanced control strategies, like assistance-as-needed control.

You may experience a small level of physical discomfort during a normal trial. However, should the level of discomfort exceed that of your liking, you can terminate the experimental trial by either indicating to the researcher or by using the emergency stop button provided. Since the actuators used in the prototype can produce significant
forces, there is a possibility of injury should the hardware or software of the prototype malfunctions. This will be very unlikely as the prototype would have been tested extensively before the commencement of the experimental trial. In addition to that, the researcher will be present to provide aid if required. To ensure that prompt medical attention is available during emergencies, the trials will be conducted during the operating hours of a certain clinic or hospital. Each participant will be required to fill in a questionnaire involving the subjective feedback of the use of such a robot-assisted ankle assessment and rehabilitation device right after the whole training.

**Data Storage/ Retention/ Destruction/ Future Use**

As discussed in the previous section, data collected will be in the form of sensor readings which are used to compute the motion and force/moment observed during the experimental trial. This information will also be used to identify properties of your ankle joint. Additionally, your age, gender and body weight will also be recorded. All such data will be stored electronically on a computer hard drive kept in a secure location. The data will be stored in such a way that a third party will not be able to identify you through the information stored on the data file. The information collected will be kept for a minimum of six years as reference for current and possibly future research, which include, but not limited to presentations, conference and journal papers, theses, and device calibration. When no longer required, such data files will be destroyed through permanent deletion. If you are interested, you can also arrange with the researcher a suitable time to have a discussion about any information derived from offline analysis of the collected data.

With your permission, a photograph/video of the experimental trial may also be taken using an electronic device. The lower part of the body of each participant will be photographed or video-recorded as well as the robotic device. The photograph/video of the trial will also be stored electronically on a hard drive kept in a secure location. The video will be taken in such a way that it will provide minimal features which can be used to reveal your identity. You are able to request the recording process of both photographs and videos to be stopped at any time during the experiment. You are able to view and edit, or withdraw your video and/or photographic recording data from the research within one month from the date of the experiment. Videos and photographs recorded will be used by the research team to further develop the device in the future and may also be used in publications including, but not limited to, theses, conference presentations and journal papers.

All data (including sensor data, videos and photographs) released outside of the research team will only be presented in a way that does not identify you.

**Right to Withdraw from Participation**

You may withdraw from participation at any time. You may also withdraw the data collected during your experiment from the research within one month from the date of the experiment.

**Anonymity and Confidentiality**

Your identity will be kept confidential from all third parties. If the data collected is used in publications, you will be referred to using a generic identifier such as “participant A”. However as videos or photographs recorded will contain items of clothing you are wearing at the time of the trial, anonymity may not be guaranteed. However recordings will be performed in the manner as to minimise the risk of identification.

**Incidental Findings**

There is a small chance where an anomaly may be found in the data collected regarding the characteristics of your foot and ankle. Should this occur, you will be informed and a qualified physiotherapist / doctor will be recommended to verify such findings.

**Compensation**

In the unlikely event of a physical injury as a result of your participation in this study, you may be covered by ACC under the Injury Prevention, Rehabilitation, and Compensation Act 2001. ACC cover is not automatic, and your case will need to be assessed by ACC according to the provisions of the Injury Prevention, Rehabilitation, and Compensation Act 2001. If your claim is accepted by ACC, you still might not get any compensation. This depends on a number of factors, such as whether you are an earner or non-earner. ACC usually provides only partial reimbursement of costs and expenses, and there may be no lump sum compensation payable. There is no cover for mental injury unless it is a result of physical injury. If you have ACC cover, generally this will affect your right to sue the investigators.
Appendix C. Ethics Approval (011904)

If you have any questions about ACC, contact your nearest ACC office or the investigator.

You are also advised to check whether participation in this study would affect any indemnity cover you have or are considering, such as medical insurance, life insurance and superannuation.

Contact Details and Approval Wording

Researcher (PhD student):
Mingming Zhang
Email: mzha130@aucklanduni.ac.nz
Phone: 022 0949-220

Supervisor:
Prof. Shane Xie
Email: s.xie@auckland.ac.nz
Phone: 09 373-7599 extn. 88143

Head of Department:
Prof. Brian Mace
Email: b.mace@auckland.ac.nz
Phone: 09 373-7599 extn. 88148

For any queries regarding the ethical concerns you may contact the Chair, The University of Auckland Human Participants Ethics Committee, The University of Auckland, Office of the Vice Chancellor, Private Bag 92019, Auckland 1142. Telephone 09 373-7599 extn. 83711.

APPROVED BY THE UNIVERSITY OF AUCKLAND HUMAN PARTICIPANTS ETHICS COMMITTEE
REFERENCE NUMBER 011904

CONFIDENTIALITY AGREEMENT

Project Title: Robot-Assisted Ankle Assessment and Rehabilitation
Names of Researchers: Mingming Zhang, Prof. Shane Xie

I was invited to do the screening to sort out the qualified participants for this research project based on the predefined criteria (Participants who would like to improve their ankles' motor function and should be able to conduct a certain ankle range of motion (≥10°) in one direction without any pain will be chosen to participate in this study).

I promise not to disclose any participant Information like gender, age, height, body weight and any others.

Print name of the Physiotherapist / Doctor: _________

Signature: _________________________________
Date: ____________________________________

Contact details:
________________________________________________________________________
________________________________________________________________________

Researcher (PhD student):
Mingming Zhang
Email: mzha130@aucklanduni.ac.nz
Phone: 021 1499-817

Supervisor:
Prof. Shane Xie
Email: s.xie@auckland.ac.nz
Phone: 09 373-7599 extn. 88143
Appendix C. Ethics Approval (011904)

Your help is very much appreciated.

APPROVED BY THE UNIVERSITY OF AUCKLAND HUMAN PARTICIPANTS ETHICS COMMITTEE
REFERENCE NUMBER 011904

CONSENT FORM
(Participants)

THIS FORM WILL BE HELD FOR A PERIOD OF 6 YEARS

Project title: Robot-Assisted Ankle Assessment and Rehabilitation
Name(s) of Researchers(s): Mingming Zhang, Prof. Shane Xie

I have read the Participant Information Sheet, have understood the nature of the research and why I have been selected. I have had the opportunity to ask questions and have them answered to my satisfaction. I have chosen to participate in this research on a voluntary basis.

- I agree to take part in this research.
- I understand that participation in this research will require physical interaction with a prototype robotic device.
- I understand that there is a small possibility of injury should the prototype device malfunction, and I accept this risk.
- I understand that my identity will not be revealed in publications derived from this research.
- I understand that the information related to the clinical trials which I am involved in will be recorded using sensors on the prototype device and stored electronically with secure access.
- I understand that I am free to withdraw participation at any time, and to withdraw any data traceable to me up to one month after the collection of the data.
- I understand that participation in this research will not affect my grades or my relations with the university (where applicable) as assured by the head of department of Mechanical Engineering.
- I agree / do not agree to be videotaped and photographed.
- I understand that I can request to stop the recording process at any time.
- I understand that I am free to view, edit or withdraw any photographic or video recordings of the trial involving me for up to one month after the collection of the data.
- I understand that the data will be kept for 6 years, after which they will be destroyed.
- I agree / do not agree to be informed of any incidental findings that arise from this research.
- I agree / do not agree that a qualified physiotherapist / doctor will be recommended to verify this if any incidental findings.
- I do / do not wish to have a follow up discussion of the results obtained from my experimental trial. Contacting email address ____________________________

Name ______________________
Signature ______________________ Date ______________________

APPROVED BY THE UNIVERSITY OF AUCKLAND HUMAN PARTICIPANTS ETHICS COMMITTEE
REFERENCE NUMBER 011904
Appendix D. Bill of Materials

Materials including mechanical parts and electronic components for the AARR came from many sources. Some of mechanical parts were fabricated in the Machine Shop in the Department of Mechanical Engineering, University of Auckland; some of them were made using 3D printer. Following is the list of main electronic components that are for actuation, position/force sensing and data communication/acquisition, respectively.

<table>
<thead>
<tr>
<th>Functionality</th>
<th>Parts</th>
<th>Specification</th>
<th>Quantity</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Actuation</strong></td>
<td>Fluidic muscle</td>
<td>Festo DMSP-20-400N-RM-CM</td>
<td>4</td>
</tr>
<tr>
<td></td>
<td>Pressure regulator</td>
<td>Festo VPPM-6L-L-1-G18-0L6H</td>
<td>4</td>
</tr>
<tr>
<td><strong>Position sensing</strong></td>
<td>Rotary encoder</td>
<td>AS5048A</td>
<td>3</td>
</tr>
<tr>
<td><strong>Force sensing</strong></td>
<td>Single-axis Load cell</td>
<td>Futek LCM 300 250lb</td>
<td>4</td>
</tr>
<tr>
<td></td>
<td>Amplifier</td>
<td>Futek CSG110</td>
<td>4</td>
</tr>
<tr>
<td></td>
<td>Six-axis Load cell</td>
<td>SRI M3715C</td>
<td>1</td>
</tr>
<tr>
<td></td>
<td>Interface box</td>
<td>SRI M8125</td>
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<tr>
<td><strong>Control</strong></td>
<td>Real-time controller</td>
<td>NI Compact RIO-9022</td>
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</tr>
<tr>
<td></td>
<td>Digital input/output</td>
<td>NI-9401</td>
<td>1</td>
</tr>
<tr>
<td></td>
<td>Analog input module</td>
<td>NI-9205</td>
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</tr>
<tr>
<td></td>
<td>Analog output module</td>
<td>NI-9263</td>
<td>1</td>
</tr>
</tbody>
</table>
Appendix F. Publication List

Peer Reviewed Journal Papers


Appendix F. Publication List


**Peer Reviewed International Conference Papers**


References


176

References


References


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183

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References


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References


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References


References


J. L. Emken, J. E. Bobrow, and D. J. Reinkensmeyer, "Robotic movement training as an optimisation problem: designing a controller that assists only as needed," in *IEEE 9th International Conference on Rehabilitation Robotics*, 2005.


