

Chapter 1

General Introduction

1.1 Introduction

Human movement is essential for many daily-life activities, and has an important impact on human functioning. Rehabilitation medicine aims to minimize symptoms and disability as a result of movement disorders [1]. Therefore, the study of human movement and the treatment of movement disorders is an important aspect of rehabilitation medicine. In clinical practice, medical history, physical examination and clinical gait analyses are carried out to assess the clinical problem and the level of functioning of a patient, to define treatment goals, and to aid clinical decision-making.

In the past century, movement analysis techniques have been developed to analyse the movement that is related to activity. These techniques can play an important role in objectifying and quantifying clinical motor function assessment, and may optimize clinical treatment decisions [2]. Nowadays, gait analysis can be performed very accurately with laboratory-based systems such as 3D optical motion capture systems and force plates [3]. However, in the Netherlands, these systems are often not used, due to their complexity or the lack of availability of well-equipped gait laboratories. Furthermore, physical examination is mainly based on the subjective experience of examiners. There is still a lack of focus on accurate performance and objective and quantitative measurements.

Recently, ambulatory movement analysis systems have become available. The term ambulatory refers to systems that are not restricted to a gait laboratory. Specifically, they are camera-less, wearable on the subject itself, and easily transportable. These systems have the potential to assist or improve objective and quantitative motor function assessment in clinical practice. This thesis aims to evaluate the application of ambulatory movement analysis systems in the motor function assessment of children with cerebral palsy (CP) and adults with osteoarthritis (OA) of the knee.

This general introduction starts with a description of the theoretical framework used in rehabilitation medicine. Subsequently, an overview is given of the history of human movement analysis, clinical problems, and ambulatory movement analysis systems. Finally, the aim and outline of this thesis will be described in more detail.

1.1.1 The International Classification of Functioning, Disability and Health

The International Classification of Functioning, Disability and Health, known as the ICF (WHO, 2001), is used as a theoretical framework in rehabilitation medicine to describe and measure health and disability (Figure 1.1). In 2007, the ICF for children and youth was developed (WHO, 2007).

The domains in the ICF are classified from body, individual and social perspectives. 'Body functions and structures' refers to physiology and the physiological functioning of body systems and structures (e.g. anatomical parts such as organs and limbs). A malfunction in a body function or malformation of a structure is called impairment. 'Activity' is defined as the execution of a task or action by an individual, who might be disabled to perform that activity. 'Participation' refers to an individual's involvement in a life situation. A difficulty at the personal level is referred to as an activity limitation, and at the social level as a participation restriction. Environmental and personal factors have an impact on all components of functioning, activity and participation.

In rehabilitation medicine, the framework of the ICF assists the clinician to make a diagnosis and a prognosis, and to plan the treatment [4]. Physical examination and gait analysis are measurements at the level of body function and activity (Figure 1.1).

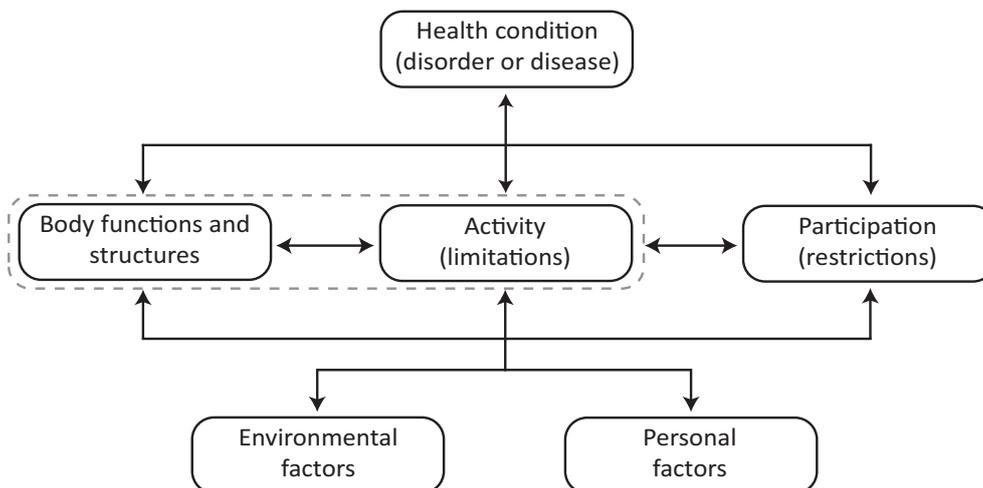


Figure 1.1. The International Classification of Functioning, Disability, and Health (ICF)

1.1.2 History of human movement analysis

Human movement has been of interest to scientist for many centuries. History shows how the present understanding of human movement has developed as a series of steps, each based on previous developments in the field and on the scientific and cultural environment of the contributors [5]. Particularly in the past two centuries, the development of photography, force plates and modern computers has made major contributions to the understanding of human movement, and opened up the way to clinical movement analysis. This section gives an overview of the major contributors to human movement analysis in the past, and describes the impact that new techniques have had on the measurement of and insight into human movement and the field of biomechanics.

The first description of (animal) motion can be traced back to Aristotle (384-322 BC) in his 'De Motu Animalium' [6]. Aristotle considered movement to be generated by an active 'mover', while the natural state of any system was at rest. He defined speed to be linearly dependent on the exerted force of 'the mover'. Several centuries later, Leonardo da Vinci (1452-1519) was the first person to study anatomy within the context of mechanics [7]. He was followed by Vesalius (1514-1564), who wrote one of the most influential books on human anatomy, 'De humani corporis fabrica' [8]. It included an accurate anatomical description of human musculature, and he demonstrated that human movement is due to the contraction of muscles.

The greatest pioneer of scientific motion analysis was Borelli (1608-1679), often labelled as 'the father of biomechanics'. He described in 'De Motu Animalium I and II' the study of animals, and related them to machines [9]. In biomechanics, the body is viewed as a structure (skeleton), comprised of levers (bones) with pivots (joints) that are actuated by the forces produced by pairs of agonist and antagonist muscles [7]. Moreover, Borelli used mathematics to prove his theories about muscles and locomotion. He was the first to determine the position of the human centre of gravity. His work was influenced by Descartes (1596-1650) [10] and Galileo (1564-1642) [11].

Descartes launched the idea that all living systems, including the human body, are simply machines ruled by the same mechanical laws. Galileo studied the motion of bodies, using mathematics, and can be considered as a precursor of Newton (1643-1727), who published the laws of motion in 1687 [12]. Euler (1707-1783) generalized Newton's laws of motion to representations that are still used extensively to describe rigid body motion in the biomechanics of human motion [7]. In the 19th century, knowledge about muscular

function increased further. The study of muscle activity, using electromyography (EMG), was improved by Duchenne de Boulogne (1806-1875), based on the first work concerning bioelectricity by Galvani (1737-1798). In the Netherlands, Swammerdam (1637-1680) made a major contribution to the understanding of nerve-muscle function by introducing nerve-muscle preparation. [13,14]

The study of human motion was strongly influenced by the development of photography in the 19th century. Muybridge (1830-1904) and Marey (1830-1904) were pioneers in using photography to capture image sequences which revealed the details of human and animal locomotion (especially the pattern of limb movements in horses) [5,14-16]. Marey also invented a pneumatic system which he placed in a special shoe to measure foot pressure: the first instrumented shoe was born. He further developed a dynamometric table which can be considered as the first force plate, to measure the ground reaction forces acting on the foot. The ground reaction force reflects the net mass-acceleration products of all body segments, and is the net result of all active muscular forces. Marey was the first to use kinematic data (description of movement) and kinetic data (forces and moments involved with the movement) and combined them to calculate the mechanical work during a movement.

The first three-dimensional research focusing on gait was carried out by Braune (1831-1892) and Fischer (1861-1917) [17]. They determined the centre of gravity and moments of inertia of the body, using cadavers. They then performed an extensive gait analysis, using four fixed cameras and light tubes positioned on a subject, indicating the centres of the joints and the centres of gravity of the body parts. From the data they calculated the positions of the joints, the centre of gravity, displacement, velocity, and acceleration, and subsequently calculated the forces acting on the body. Fischer published their work between 1895 and 1904. The methodology of Braune and Fischer is still the basis for gait and human movement analysis today.

The next main contributor to the study of locomotion was Bernstein (1896-1966), who was one of the first to use the term biomechanics. He studied a large number of subjects, and improved the methodology of photogrammetry. Bernstein, who originally published all his work in Russian, was the first to mention the indeterminacy problem of the human musculoskeletal system towards task achievement, leading to the concept of movement and muscle co-ordination. [14]

After the 2nd World War, the University of California carried out extensive research into human motion, as part of the federal rehabilitation program for veterans of the war [5,14,18]. Their goal was to obtain fundamental data on locomotion in the normal human subject, as well as in the amputee, in order to provide information to improve the design of artificial limbs. They used a number of measurement methods and devices, including pins in prominent bony landmarks, EMG, the interrupted light technique (with light markers on the body), the glass walkway to study segment angles, and a force transducer incorporated in a prosthesis. Furthermore, they improved the force plate, based on the mechanical transducer designed by Elftman (1938) [19].

In the 1960s and 70s, several major contributions were made in reports documenting the kinematics of gait, using different tools such as electro-goniometry, cinematography, interrupted light photography, and television systems [20-23]. In 1987, Winter (1930) published a widely used and referenced book on human gait entitled 'The biomechanics and motor control of human gait' [21].

The development of the computer in the 2nd half of the 20th century marked the introduction of modern computer-based motion analysis systems. This was a significant step forward, and opened up the way to instrumented gait analysis with a fast data-processing and output that was clinically meaningful [5].

Currently, gait laboratories in universities, hospitals and rehabilitation centres are equipped with video camera systems, optical motion capture systems, force platforms embedded in the floor, and EMG systems linked to computers. The kinematics of every stage of the gait cycle can be obtained by means of optical markers placed on the subject (passive light reflecting or active light emitting), which are captured by infrared cameras. Different protocols have been developed to obtain useful joint kinematics, based on the location of anatomical landmarks, such as the Helen Hayes model [24] and the CAST protocol (Calibration Anatomical System Technique) [25]. The video data assist in the interpretation of the kinematics. Kinetics are obtained from force plate data (including the magnitude, direction, and location of ground reaction forces) and an inverse dynamics approach is used to calculate the net forces and moments acting on each joint. Muscle function is studied from the EMG data. [26]

1.1.3 Clinical movement analysis

The clinical application of gait analysis was mainly driven by the treatment of problems caused by cerebral palsy (CP). Orthopaedic surgeons advocated the use of gait analysis to assist clinical decision-making [27-29]. Clinical gait analysis is currently embedded in the rehabilitation process of children with CP for the planning and evaluation of treatment options to prevent deformity and increase mobility. In the Netherlands, this is often based on observational gait analysis, using video and EMG data to measure abnormal segment movement, to observe the function of the muscles during walking, and to understand the underlying mechanisms of gait deviations.

Apart from (observational) gait analysis, medical history, physical examination and medical imaging (e.g. X-rays or Magnetic Resonance Imaging) are very important for the assessment of motor disorders in rehabilitation medicine today. Physical examination is carried out to determine the strength and selective motor control of isolated muscle groups, to evaluate the degree and type of muscle tone, and to estimate the degree of static deformity and/or muscle contracture at each joint. Furthermore, torsional and other deformities of the bone, fixed and mobile foot deformities, balance, equilibrium responses, and standing posture are also assessed in a physical examination [27]. An important part of the physical examination in children with CP is a spasticity test to measure the level of spasticity in specific muscle groups [30].

Furthermore, biomechanical factors of the gait cycle are becoming increasingly relevant in the management of diseases such as knee osteoarthritis (OA) [31-34]. Although clinical gait analysis is not yet widely used in these patients, measurement of net joint moments (particularly the knee adduction moment) can be carried out to demonstrate abnormal joint-loading. This may assist with diagnostics and the estimation of disease progression, as well as in the evaluation of the effects of interventions, such as orthopaedic surgery or implants.

1.2 Clinical problems

1.2.1 Measurement of spasticity in patients with cerebral palsy

Spasticity is a clinical phenomenon that is frequently present in cerebral palsy (CP), as well as stroke, spinal cord injury, and multiple sclerosis. CP describes a group of permanent disorders of the development of movement and posture, causing activity limitation, that are attributed to non-progressive disturbances that occurred in the developing fetal or infant brain [35]. The motor disorders of CP are often accompanied by disturbances of sensation, perception, cognition, communication, and behaviour, by epilepsy, and by secondary musculoskeletal problems. CP is the most common cause of physical disability in childhood, affecting 1.5-3.0 per 1000 living births in Europe [36]. In premature births, or babies with very low birth weight, the prevalence increases to 40-100 per 1000 living births [37]. Spastic CP is the most common form of CP (77%) [38].

The definition of spasticity has been subject to debate for a long time, due to the complex aetiology and expression at joint level. One of the commonly used definitions for spasticity is described by Lance in 1980: *“a motor disorder characterized by a velocity-dependent increase in tonic stretch reflexes (muscle tone) with exaggerated tendon jerks, resulting from hyper excitability of the stretch reflex, as one component of the upper motor neurone syndrome”* [39]. A stretch response is a normal protective mechanism that occurs when a resting muscle is lengthened rapidly or forcefully. Spasticity means that the threshold for this response is lowered and can be elicited much more readily, even on passive examination.

The Support Programme for the Assembly of a database for Spasticity Measurement (SPASM) consortium redefined spasticity in 2005 as: *“disordered sensori-motor control, resulting from an upper motor neurone lesion, presenting as intermittent or sustained involuntary activation of muscles”* [40,41]. This definition describes spasticity more generically than Lance’s definition, encompassing the entire range of signs and symptoms that are collectively described as the excess features of the upper motor neurone syndrome, and not exclusively the hyper excitability of the stretch reflex.

Spasticity is a primary impairment, a direct result of the upper motor neuron syndrome in spastic CP. Other primary impairments are hypertonia (non-velocity-dependent increased resistance to passive movement), enhanced reflexes, poor selective control and paresis (leading to weakness) [27,42,43]. Due to abnormal balance in the muscles, secondary

impairments develop over time, such as muscle contractures (i.e. reduced muscle-tendon length), changes in muscle cells, increased muscle stiffness or bone or joint deformities. Coping mechanisms (i.e. compensation strategies) are referred to as tertiary impairments [27]. The impairments could limit the performance of motor tasks [44]. The majority of children with CP experience limitations in mobility and gait. Mobility can be classified according to the Gross Motor Functional Classification System (GMFCS I-V) [45,46].

Approximately 70% of children with CP are able to walk at the age of five [47], but their gait deviates from that of typically developing children. Classification of gait patterns in CP have been described by Becher et al. [48] and Rodda et al. [49] and focus on gait deviations in the sagittal plane. The classification of mobility and gait pattern assists optimal communication, the determination of a prognosis, and the choice of treatment options.

Treatment of abnormal posture and gait, due to spasticity and weakness, is very important in children with CP, to prevent bone deformity and to increase mobility. Spasticity assessments and clinical gait analysis are used to assist clinical decision-making and the evaluation of treatment, such as orthopaedic surgery, selective dorsal rhizotomy, botulinum toxin type A injections in muscles to decrease spasticity, and assessment of ankle foot orthosis [42].

Several physical examination tests have been developed to assess spasticity (measurement at the ICF-level of body function and structure). The most commonly used test to assess spasticity is the Modified Ashworth Scale [50]. Other tests are the Modified Tardieu Scale and the Spasticity Test (SPAT), based on Lance's definition of spasticity [30,51-53]. In these tests, passive joint motions are performed at one or more different stretch velocities to observe the range of motion of the joint and the joint angle of catch in a specific muscle group. The catch has been defined as "*a sudden appearance of increased muscle activity in response to a fast passive stretch, which leads to an abrupt stop or sudden increased resistance during the movement*" [30,43]. The angle of catch is measured with goniometry after the fast passive stretch.

Although the information obtained from physical examination is considered to be important to support clinical decision-making, there is a lack of focus on accurate performance and objective and quantitative measurement. Physical examination is mainly based on the education and experience of the examiner. Although clinical guidelines have been formulated, test performance and interpretation of test results can be subjective

and inaccurate [30,54-56]. More specifically, in the currently used tests for spasticity in children with CP, such as the Ashworth Scale, the Tardieu Scale or the SPAT, the test-retest reliability can be low. This is partly due to: (i) the difficulty in achieving standardization; (ii) the subjective encountering of a catch; and (iii) the lack of objective results of the tests [30,54,57,58]. Although movement analysis systems can assist objective and quantitative measurement, they are generally not used (except for goniometry) due to the practical problems that are experienced and the complexity of the technical systems.

Clinical gait analysis is performed to identify gait deviations in children with CP of GMFCS class I-III (measurement at the ICF-level of activity). This includes children who are able to walk, although limited in balance, co-ordination, stair-climbing, and walking long distances. When using optical motion analysis systems in gait laboratories, movement can be measured with a reliability of less than 5 degrees [3]. Many studies have improved the integrity of movement analysis systems, the protocols of anatomical calibration, and the interpretation of gait data. Using the technical systems that are available, gait analysis data is objective, quantitative, accurate, and easy to document.

However, in clinical gait analysis, the techniques of optoelectronic markers and force platforms that are needed for 3D measurements are not always used, due their complexity, the lack of availability of gait laboratories or special equipment, and high costs. Optical markers have 'line of sight' problems, resulting in missing data, and measurement volume is limited depending on the camera placement. Furthermore, patients adapt their gait pattern due to the need for (multiple) constrained foot placements on the force plate, the limited space in the laboratory, and the feeling that they are being observed. These disadvantages limit the clinical use of such laboratory-based systems [59-62].

1.2.2 Measurement of mechanical loading in patients with osteoarthritis of the knee

In recent years, there has been increasing awareness of the importance of biomechanical factors in the pathogenesis and progression of osteoarthritis (OA) [31-34]. OA is a degenerative joint disease, which occurs in a substantial percentage of the elderly population [33,63,64]. Although mechanical loading on the joints appears to be necessary to maintain healthy cartilage, abnormal joint-loading increases the risk of OA, since cartilage has a limited ability for self-repair [63,65].

OA occurs most often in the knee joint, is more common in women than in men, and incidence and prevalence increases with age. 1.5% of adults over the age of 55 suffers from painful, severe knee OA, and up to 10% has mild to moderate knee OA. Risk factors for OA are age, gender, obesity, joint injury, laxity, and malalignment. With the increasing in age and obesity in the general population, the prevalence and incidence of knee OA is likely to increase in the near future. [33,63,64]

The aetiology of OA is very complex, and involves biological, mechanical and structural factors [63]. OA includes cartilage destruction, subchondral bone-thickening, and new bone formation [64]. Traditionally, radiographs have been used in the diagnosis of OA to identify joint-space narrowing, sub-chondral sclerosis, and osteophyte formation [66]. OA can be classified with the Kellgren/Lawrence scale, according to the location within the joint and the severity of the disease [67].

People with knee OA experience limitations in daily physical functioning. The most common and dominant symptom is knee pain [68]. Other symptoms are joint stiffness and swelling [33]. OA results in a decline in mobility, and could ultimately lead to joint replacement surgery. Common interventions for knee OA are exercise, weight management, bracing, heel wedges, muscle-strengthening, high tibial osteotomy, and pharmacologic treatment [69].

Various studies have investigated knee joint-loading in relation to disease severity of knee OA. Many interventions focus on minimizing mechanical loading on the affected compartment of the knee. An important parameter that is used to identify the mechanical loading on the knee joint is the net external frontal knee moment, also called the knee adduction moment (KAdM). The KAdM reflects the distribution and magnitude of the load transferred through the medial versus the lateral compartment of the tibiofemoral joint, and is mainly defined by the direction and magnitude of the ground reaction force and its lever arm with respect to the knee joint centre [70]. In patients suffering from medial

compartment knee OA, increased KAdMs have been observed during gait, compared to asymptomatic subjects [71-74]. Gait analysis, based on force plates and optical marker systems, can be used to calculate the frontal knee moments during gait.

However, clinical gait analysis is often not performed at all for patients with OA, despite the increasing interest in evaluation of the effect of biomechanical factors such as the joint moments [31-34]. There are no gait laboratories in many rehabilitation centres and hospitals, and when they are available, the practical problems that are encountered limit the use of movement analysis systems.

1.3 Ambulatory movement analysis systems

New technology for the ambulatory measurement of human movement has recently been developed. This technology overcomes the restrictions and limitations of laboratory-based movement analysis systems, and could potentially play a role in physical examination and gait analysis.

An inertial and magnetic measurement system (IMMS) has been developed to measure the orientation of body segments [59,75,76]. An IMMS sensor unit contains miniature MEMS sensors (microelectromechanical systems), including a 3D accelerometer, a 3D gyroscope, and a 3D magnetometer. These inertial sensors are small and lightweight, and can easily be attached to a body segment. A data-logger, worn on the body, can wirelessly be connected to a notebook. Segment and joint orientations are calculated by fusion of acceleration, angular velocity, and heading of magnetic field in fusion algorithms, and by using anatomical calibration procedures based on reference postures and/or rotation axes. IMMS may be a good alternative to conventional 3D optical motion capture systems. It is not restricted to a certain measurement volume, and can be used to collect data on multiple movements. Protocols have recently been developed for IMMS anatomical calibration procedures to analyse the gait of children with CP [77].

Similarly, instrumented force shoes (IFS) have recently been developed for the ambulatory assessment of ground reaction force (GRF) and centre of pressure (CoP), as an alternative to force plates in gait laboratories. The IFS is an orthopaedic sandal that is equipped with two 6-degrees-of-freedom force/moment sensors under the heel and forefoot. An IMMS

sensor unit is attached to each force sensor at the lateral side of the shoe. Studies that have been carried out to evaluate the application of the IFS for the measurement of GRF, CoP and ankle moments in adults have reported promising results. [60,61,78]

Application of the IMMS and IFS could assist or improve objective and quantitative motor function assessment in clinical practice. The IMMS could be applied to objectify joint angle measurements of spasticity tests in children with CP. To analyse the gait of children with CP, IMMS can be used to measure multiple strides of natural gait, with no restriction in the measurement volume or adaptation of the gait pattern due to limited space. The IFS opens up the way for kinetic measurements of multiple strides during gait in patients with knee OA when no gait laboratory is available.

1.4 Aim and outline of the thesis

The general aim of this thesis is to evaluate the feasibility and quality of the new ambulatory movement analysis systems, motivated by application in clinical motor function assessment.

Three different applications are addressed in this thesis:

- the inertial and magnetic measurement system to assess spasticity in children with cerebral palsy (Chapters 2 and 3)
- the inertial and magnetic measurement system to analyse gait in children with cerebral palsy (Chapter 4)
- the instrumented force shoe to estimate the knee adduction moment in adults with osteoarthritis of the knee (Chapters 5 through 8)

Chapters 2 to 4 describe the application of the IMMS in the clinical assessment of children with CP. **Chapter 2** evaluates the precision and accuracy of the angle of catch measurement in the assessment of spasticity of the medial hamstrings, soleus and gastrocnemius muscles in twenty children with CP. Joint angles measured with goniometry are compared to joint angles measured with the IMMS. A method to obtain accurate joint angles from IMMS orientation is also described.

Chapter 3 assesses whether the angle of catch is the consequence of a sudden velocity-dependent increase in muscle activity, as described by Lance (1980) in the definition of spasticity. Surface electromyography is used to observe muscle activity during the spasticity tests in the same group of children. Simultaneously, joint angle and stretch velocity is measured with the IMMS.

The application of the IMMS in the gait analysis of children with CP is addressed in **Chapter 4**. The protocol Outwalk [77], developed as an easy way to measure 3D kinematics during gait by means of IMMS, is evaluated in six children with CP and one healthy child. Joint angles measured with the IMMS, based on the Outwalk protocol, are compared with outcomes of a conventional laboratory-based system and protocol [25].

In Chapters 5 to 7 the IFS is applied to the gait analysis of twenty adults with knee OA, to estimate the net external knee adduction moment. **Chapter 5** describes the effects of wearing an IFS on gait characteristics in patients with knee OA. Chapters 6 and 7 report on a method to determine the KAdM, based on GRF and CoP measurements of the IFS, in combination with IMMS segment orientation measurements. Since it is difficult to measure positions with the IMMS, the joint centre positions required to estimate the KAdM must be obtained by means of an orientation-based linked-segment model.

Chapter 6 evaluates the performance of (i) the IFS in measuring the 3D GRF and CoP, (ii) the linked-segment model based solely on segment orientations to determine joint positions, and (iii) the combination of IFS and the linked-segment model to determine the KAdM. Segment orientations obtained from an optical motion analysis system are used for this evaluation.

Chapter 7 extends the evaluation of the ambulatory movement analysis system for KAdM estimation, using segment orientations from the IMMS as input to the linked-segment model. In this way, the performance of the entire ambulatory movement analysis system, consisting of IFS and IMMS, is evaluated in the estimation of the KAdM in the gait of patients with knee OA, whereas Chapter 6 mainly focuses on the performance of the IFS and the linked-segment model.

Clinically, interventions in knee OA often aim to reduce the load on the diseased compartment, in order to decrease the rate of progression of the OA and the pain experienced by patients. Hence, gait modifications have also been assumed to minimize the loading on the affected side of the knee. Reduction in walking speed, in-toeing or out-

toeing, and medio-lateral trunk sway, are compensations that are observed in patients with knee OA [79-82]. However, there is still a lack of knowledge about the influence of these different manipulations.

Chapter 8 describes the effects of these gait modifications on the 3D knee moments (mainly the KAdM) in fourteen healthy subjects. This chapter will provide more insight into the importance and possible clinical application of the KAdM measurement.

Finally, **Chapter 9** summarizes the main findings of the studies described in this thesis. This chapter also reflects on the feasibility of each of the above-mentioned studies in clinical practice, and recommendations are made for potential future research in the field of ambulatory systems and their clinical application.

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