Analysis of body motions based on optical markers

Accuracy, error analysis and clinical applications

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ABSTRACT

This thesis aims to evaluate the influence of soft-tissue artefacts on analyses of body motions based on optical markers. A second aim is to apply instrumented gait analysis in clinical situations.

Introduction: Instrumented gait analysis has been used since 1960 as a clinical evaluation/investigative tool at orthopaedic clinics. The technique is based on a number of reflective spherical markers attached to the skin. The position of the skin markers is recorded as the subject walks through the measurement volume. Recorded data form the basis when monitoring movements of body segments. The crucial and still not completely explored issue is to what extent these systems are able to reproduce the movement of the body segment that is being studied.

Material and methods: In *Study I,* the skin movement at the foot was studied using skin markers and radiographs. The subjects stood on one foot in three positions, 20° dorsal flexion, a neutral position and 30° plantar flexion, while radiographs were exposed. In *Study II,* the aim was to study problems with soft-tissue movement along the lower extremity. Skin and underlying structures were provoked partly by anterior-posterior and longitudinal strain and partly by being put into vibration to investigate their stiffness and damping characteristics. The aim of *Study III* was to examine the accuracy of the optical tracking system used throughout *Studies IV-V* by simultaneous recording using skeletal markers and radiostereometry (RSA). Nine patients with a total knee arthroplasty (2 males/7 females, median age: 63.1 years; range 59-72) were included in *Study III*. In *Study IV,* 20 patients with bilateral spastic cerebral palsy (15 males/5 females median age: 12.9 years, range 9.4-15.3) and 20 controls (13 males/7 females, median age: 13.0 years; range 10.2-15.7) were included. For *Study V*, nineteen unilateral transfemoral amputee patients (9 males/10 females, median age: 46.5 years; range 19.9-62.3) and fifty-seven matched controls were included.

Results: Studies of soft tissue motions on the foot revealed marker movement in relation to the bone up to 4.3 mm at the ankle, which decreased gradually to 1.8 mm at the first inter-phalangeal joint. Soft-tissue movements mainly occurred in the anterior-posterior direction of the leg and pronounced self-oscillations were recorded when markers were placed on wands. The results from comparisons between RSA and OTS showed good agreement regarding extension/flexion motions. For abduction/adduction and in-/external rotation, significant differences between the two systems were observed. The group with cerebral palsy was weaker in all muscle groups in the lower limbs and they walked at a slower speed. A significant relationship between plantar flexing torque and the strength of six of the eight investigated muscle groups could be detected in patients with cerebral palsy.

An even stronger relationship (rho=0.58-0.76) was found between generating power and muscle strength in all eight muscle groups. Two years after conversion from a conventional to bony anchored leg prosthesis, femoral amputees improved their hip extension and reduced their anterior pelvic tilt.
Conclusion: Instrumented gait analysis is a non-invasive and valuable tool to study body motions. Knee motions in the sagittal plane (flexion/extension) are close to data obtained from RSA based on skeletal markers, whereas the resolution of rotations in the two other planes is poorer, probably due to soft tissue motions and geometrical reasons. Further comparative studies with simultaneous use of skeletal and superficial skin markers are needed to explore this issue further; not least concerning the hip and ankle joint.
Keywords: Instrumented gait analysis, Motion analysis, Skin markers, Optical tracking system

LIST OF PAPERS

This thesis is based on the following studies, referred to in the text by their Roman numerals.

I. **The relative skin movement of the foot: a 2-D roentgen photogrammetry study.**
Tranberg R, Karlsson D.
Clinical Biomechanics, 1998, 13, 1998, 71-76

II. **On skin movement artefact-resonant frequencies of skin markers attached to the leg.**
Karlsson D, Tranberg R.
Human Movement Science, 1999, 18, 627-635

III. **Simultaneous measurements of knee motion using optical tracking system and Roentgen stereo photogrammetric analysis (RSA).**
Tranberg R, Saari T, Zügner R, Kärrholm. J
Conditionally accepted

IV. **Muscle strength and kinetic gait pattern in children with bilateral spastic CP.**
Nyström Eek M, Tranberg R, Beckung E.
Conditionally accepted

V. **Improvements in Hip- and Pelvic motion for patients with Osseo-Integrated trans-femoral prostheses.**
Tranberg R, Zügner R, Kärrholm J.
Conditionally accepted

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ABBREVIATIONS

ASIS	Anterior superior iliac spine
CP	Cerebral palsy
GMFCS	Gross Motor Function Classification System
MCU	Motion capture unit
OI	Osseointegrated
OTS	Optical tracking system
PSIS	Posterior-superior iliac spine
RSA	Radiostereometric analysis (same as roentgen stereo-photogrammetric analysis)
SMS	Skin-based marker set
TFA	Transfemoral amputee

DEFINITIONS IN SHORT

Calibrated volume	The volume (height, width and depth) in which measurement can take place.
Centre of mass (COM)	A point at which the entire mass of a segment could be concentrated, while still having the same mechanical effect.
Cluster	A plastic shell equipped with three or more reflective markers that are used to track a body segment.
Digitiser	A device for transforming analogue data into digital data. For example, manually drawn lines on a radiograph and reference points can be transferred to co-ordinates in terms of figures of x- and y- positions with the aid of a digitiser.
Force plate	A device that measures force, commonly in three dimensions, i.e. vertical and horizontal (forward and side).
Inverse dynamics	A process by which intersegmental forces and moments are calculated by applying Newton's equations of motion. This process includes measured data, i.e. kinematics and ground reaction forces, as well as the estimated inertial properties of involved segments.
Mann-Whitney U test	Non-parametric rank sum test for differences between two independent variables, making no assumption regarding the distribution of data. The test can be used when testing between a control group and an experimental group.
Marker/s	A polystyrene hemisphere, with a minor flat surface, covered with a retro-reflective material.

Piezoelectric effect	A device that responds to applied mechanical strain by producing an electric potential.
Pointer	A device used for creating landmarks. The pointer is used to identify anatomical landmarks without having to place markers at the location.
Reference object	L-shaped metal profile used together with a calibration wand during calibration. Defines the global co-ordinate system with its three axes.
Rho (Spearman's rho)	A measure of statistical dependence. The value of rho varies between 0 and 1. A rho with a value of 1 indicates an absolute dependence between the two variables that are being studied.
Spearman's rank correlation	Non-parametric rank test for correlation between two variables, making no assumption regarding the distribution of data.
Stiffness	A way to express how firm the soft tissue is. The stiffness, *k,* is calculated as the force that is applied, divided by the displacement caused by this force. Unit: N/m.
Strain gauge	A device in which exposure to mechanical strain causes its electrical resistance to change.
Virtual marker	A marker that is created in the software. For example, a virtual marker could be created as the midpoint (50% of the distance of the line between two skin markers). Virtual markers are also referred to as landmarks.
Wilcoxon signed-rank test	Non parametric rank sum test for differences between two dependent variables making no assumption regarding the distribution of data. The test can be used for differences between two different follow-ups in the same group of subjects.

1 INTRODUCTION

1.1 Instrumented gait analysis

The following chapter will discuss the basics of gait analysis with the main emphasis on analyses of body movements using optical markers. Other devices, such as goniometers, accelerometers and systems based on the video image technique, will not be taken into account in this thesis.

1.2 Historical aspects

The history of gait analysis is commonly divided into two main parts, i.e. before and after the introduction of computerisation [1]. The earliest findings of descriptions of gait analysis in humans have been reported as early as in ancient Greece by Aristotle (384-322 BC) [2]. In addition to the work of Aristotle, a few milestones have to be mentioned. In a recently published review, Baker [1] takes us on an exceptional journey through the history of motion analysis with the emphasis on humans. The journey starts with Aristotle and continues with Giovanni Borelli (1608-1679) and his work on muscle and tendon biomechanics, even if his contribution to gait analysis was not directly substantial. The invention of the photographic technique by Niépce (1765-1833) in 1822 was revolutionary, as it made it possible to save a motif without sketching or painting it. After Niépce's sudden death in 1833, his co-worker Louis Daguerre (1787-1851) continued the work and made further developments. At this early stage of the photographic era, the exposure time was, however, tremendously long; it took several hours, making it impossible to capture any motions, so photographs from this time were only taken of static objects, such as landscapes. Work on the further development of the photographic technique led to reduced exposure time, which made it possible to capture the movement of both animals and humans. In 1878, Eadweard Muebridge [3] published his work on trotting horses using this technique and a set-up of multiple cameras. The first people to conduct a three-dimensional gait analysis were Otto Fischer and Wilhelm Braune and their experiment took place in the summer of 1891, when they studied the human movement of a German infantry soldier [4]. At the same time as Braune and Fischer published their work, Louis Le Prince (1842-1890) invented the technique of motion pictures, which we now call film. The earliest recordings that still remain are the "Roundhay Garden Scene", which was recorded at 12 frames per second. Movie cameras were used throughout the 1960s, when television imaging cameras were introduced [5]. The introduction of digital cameras in the 1980s made motion analysis systems faster and easier to use. Following the development of motion capture systems in recent decades, interest has primarily focused on higher sampling rates and higher resolution, which have made it possible to record rapid movements and use markers with smaller dimensions.

Instrumented gait analysis is not only for kinematic studies, kinetics are also needed. The development of devices for recording three-dimensional force components during gait, i.e. force plates, can originally be attributed to Jules Amar (1879-1935) in 1916. Amar used a pneumatic technique to record forces. A mechanical version of a three-dimensional force plate was described in 1938 by Elftman [6]. The force plates used in gait analysis today are usually electronically operated devices using strain gauges or piezo-electric effect [7].

1.3 General aspects

Instrumented gait analysis, more commonly known as *gait analysis,* is a frequently used method in which one or more pieces of technical equipment are used to obtain objective data relating to a specific gait trial. In most cases, instrumented gait analysis takes place at a gait laboratory, even though some equipment can be used outside the gait laboratory. Different techniques, such as standard video recordings [8-10] and goniometers [11, 12], have been used and are still being used to record kinematics, even though high-speed video cameras, together with passive retro-reflective markers, are most frequently used. Several of the systems on the market provide active markers, mainly for outdoor measurements. The data that are obtained from gait trials were originally divided into kinematic data (i.e. data of motion) and kinetic data (i.e. data of force), even though they are combined in the latter part of analysis. Furthermore, the kinematic data can be recorded as 2-dimensional (2-D) or 3-dimensional (3-D) data. The easiest way to record motion data is to record 2-D data, i.e. height and width, since this only requires one camera. It will, however, result in a lack of information about the third dimension, i.e. depth, during the recording. So, to record kinematic data in 3-D requires at least 2 cameras that are calibrated together (Figure 1).

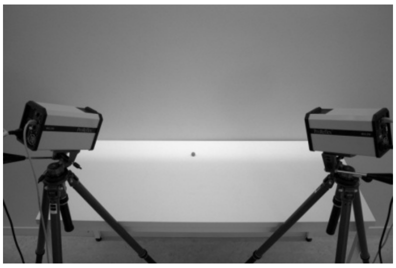

Figure 1. *A two-camera set-up for the 3-D capture of a marker.*

1.4 Sources of errors

Numerous sources of error can, and will, affect the result of a gait analysis. To make this more understandable, a structured way of displaying this problem is needed. Several steps (Figure 2) have to be passed before any result of a gait analysis can be presented.

Figure 2. *Possible sources of errors in gait analysis*

1.4.1 System set-up

To obtain kinematic data of high quality, which can serve as the basis for further calculations, the cameras need to be arranged so that they surround the entire volume in which the measurements will take place. Furthermore, the cameras should be positioned so that at least two cameras see each marker at the same time. Ideally, two cameras positioned with a camera-to-camera angle of 60° should be used. In some cases however, this is difficult to organise and smaller angles have been used; this can be compensated for by adding more cameras.

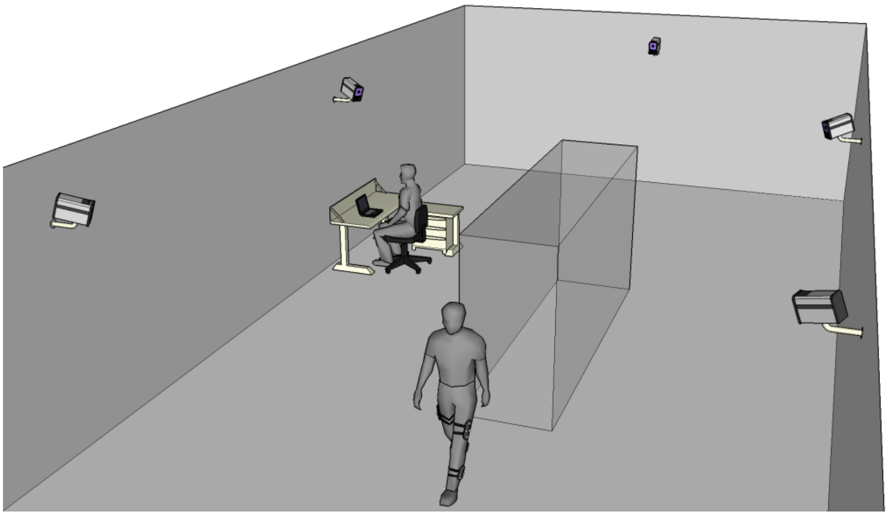

Figure 3. *System set-up with five cameras and a calibrated volume, shaded in the middle of the figure (with permission from Qualisys AB).*

1.4.2 System calibration

To perform 3-D measurements, the camera system needs to be calibrated. For the calibration used in *Studies I-II*, a fixed calibration frame was used, while, for *Studies III-V,* this procedure was performed using a reference object and a wand (Figure 4), i.e. a dynamic calibration. The reference object, an L-shaped metal profile equipped with four markers with a predefined distance between the markers, is placed on the floor, usually at the centre of the conceivable measurement volume. The reference object defines the origin of the global co-ordinate system, together with the x-, y- and z-axis. To perform the calibration procedure, a wand equipped with two markers with a predefined distance between them is randomly moved around in the volume to be calibrated. When the calibration process is complete, the result is presented as the residuals of each camera and as the SD, expressed in mm, of the predefined wand. The size of the calibration set depends on the size of the calibrated volume that is needed and, in *Study III*, a 300 mm calibration set was used, while a 750 mm calibration set was used in *Studies IV-V*. When force plates are used, as in *Studies IV-V,* their positions have to be determined to orient them in the global coordinate system.

Figure 4. *Reference object and wand used for dynamic calibration (with permission from Qualisys AB).*

1.4.3 Marker models

A marker model is a set of spherical retro-reflective markers that make it possible to trace one or more body segments. Marker models are either based solely on a set of skin markers, so-called skin-based marker sets (SMS), or on a combination of skin markers and cluster(s) consisting of a plastic shell and at least three markers attached to the shell (Figure 5). Skin markers are generally attached to the skin by double-adhesive tape, in contrast to clusters that are usually fastened with an elastic strap.

Figure 5. *Skin markers and double-adhesive tape on the left, cluster markers with an elastic strap on the right (with permission from Qualisys AB).*

A number of different marker models are currently available; they not only differ by whether or not they use a cluster, a different number of markers and procedures for recording are also used. In some models, additional markers are required for the static measurement on which the model is based. In some models, measurements such as ASIS-to-ASIS distance are required [13, 14]. Whether a skin marker-based marker model or a cluster-based model should be used is one of the fundamental questions that gait laboratories have to deal with, in addition to many other questions, during their start-up period. The main reason for this is the large amount of work associated with a change of marker model. From a clinical point of view, the SMS model like the one used in *Studies III, IV* and *V* may be preferable, as it is easy to use, the application of markers is strictly based on well-known bony landmarks and no additional measurements are needed. Comparisons between several marker models have been published by Ferrari et al. [15]. They compared five different protocols including skin markers, markers on wand and clusters and concluded that all five protocols show good intra-protocol repeatability. Most of the gait variables showed good correlation. The determinations of knee abduction/adduction and ankle inversion/eversion angles turned out to be inconsistent [15].

The skin-marker model used in *Study III* consisted of 15 markers, including the pelvis and the lower extremities. For the marker model used in *Studies IV-V*, three markers were added to include the trunk segment.

1.4.4 Calculations

Calculations of joint angles are based on the assumptions that all body segments can be considered as rigid bodies. Skin-markers and/or clusters are used for defining locations on body segments and to track body segments as they are moved throughout the measurement volume. In the same manner as markers are being used for defining location solely, they can also be used for definition of landmarks. The main difference of marker and landmark is that a landmark is always based on one or several skin-markers and never the opposite way. From the positions of the 15 skin-markers; researchers are able to create a model of the lower extremity including the pelvis. Definition of the model starts with the pelvis followed by the shanks which also serves as base for defining the knee and the ankle joints respectively. Finally, it is completed with the thigh segments which are based on the hip joint; knee joint and the supra-patellar marker together with the foot segments that are composed of the three markers at the lateral malleolus; the heel and the forefoot (Figure 6).

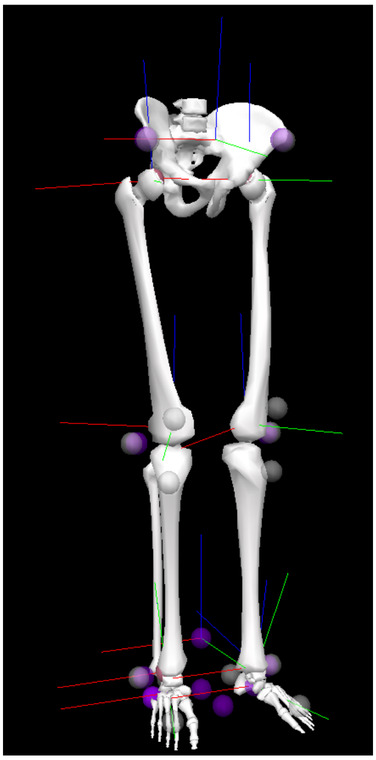

Figure 6. *An overview of the 15 skin-marker model.*

Calculations of joint angles follow the concept of Euler angels [16] which have been adopted in biomechanical field of orthopaedics. This referees to rotation between two coordinate systems, one located in the stationary segment and one located in the moving segment [17]. In general the proximal segment is chosen as stationary segment and the proximal segment as the moving segment. For angels in the knee, the thigh segment is chosen as stationary segment and the shank as the moving segment. Definition of angle sequence starts with flexion/extension which is the rotation about the x-axis and follows by the abduction/adduction which is the rotation about the y-axis and ends with the axial rotation that takes place the z-axis.

Calculation of quantitative values of joint moments is resolved with inverse dynamic and it can briefly be described as follows. In inverse dynamic assumption about that the body segments are linked to each other and the property of body segment are taken into account, i.e. each body-segments end-points; mass of segment; centre of mass for segment; acceleration of each segments end-points and the ground reaction force. End-points of segments are resolved of either skin-marker positions or landmarks. Centre of mass (COM) assumes to be located along the segments longitudinal axis with a distance expressed in percent of the total length of the segment. These distances is historically based on cadaver tests [18] or approximated to that the segments are of a geometrical shape, e.g. cylinder, ellipsoid or a cone with a solid mass [19] (Figure 7). The ground reaction force and its point of origin are obtained using a force plate.

Finally joint power is calculated by multiplying the joint moment by the angular velocity.

A fuller description of Euler angles and a more detailed description of inverse dynamics can be found in textbooks, for example, *Biomechanics and Motor Control of Human Movement* by Winter and *Human Walking* by Rose as well as *Clinical Gait Analysis Teory and Practice* by Kirtley.

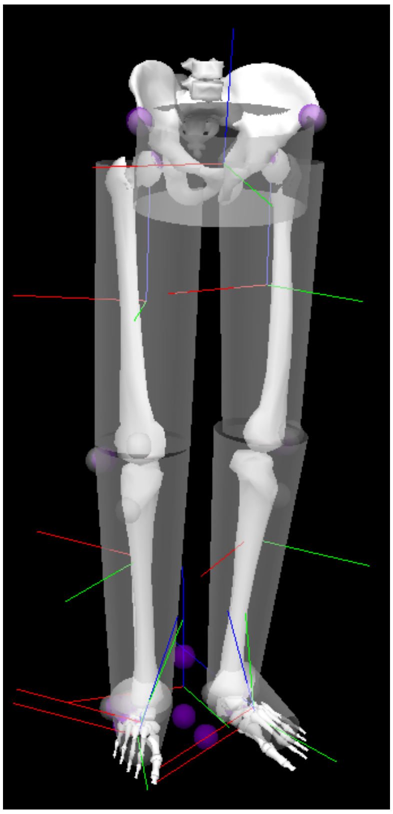

Figure 7. *Model showing the geometrical shape of segments. N.B. Coordinate systems in this figure are placed with their origin in the location of each segments COM.*

1.4.5 Presentation

Even when it comes to presentation of data from gait analysis watchfulness must be taken. In general gait laboratories present kinematic and kinetic data as percent of gait cycle, i.e. the time it takes to perform a gait cycle is set to 100%. This must be taken into account when subjects show great variation in their gait velocity. Moreover, when kinetic data are to be presented this is not only normalized to percent of gait cycle for time, but also to the subject's weight concerning joint moment and joint power given as Nm/bodyweight and W/bodyweight respectively. Inclusion of the subject's height into the normalization of moment, Nm/bodyweight x height, have been debated, but is not widely accepted. Moisio et al. [20] found that this would reduce differences of the moment in the sagittal plane due to gender. No difference was reported concerning hip moments in the frontal plane [20].

1.5 Gold standard

To achieve any authentic motion that occurs between two body segments invasive methods are needed. Intracortical pins have been used during the last two decades [21-26]. This method is still the most accurate way to obtain motion between two, or several, bony segments. Briefly described, bone pins are inserted into the bone under the infiltration of local anaesthesia. Clusters are then attached to the intracortical pins in order to be traced with the optical tracking system. This method is, however, associated with limitations due to the risk of infection and ethical considerations and should most probably only be used as an evaluation method.

Use of roentgen stereophotogrammetric analysis (RSA) with the use of bony anchored tantalum balls is a different method for detection of intra-segmental motion with high precision. The method originates from Selvik [27] who presented it in 1974. In the first period the technique was static. In 1988, Kärrholm and co-workers published the first study on knee kinematics in which they used a dynamic version of RSA [28]. The dynamic RSA consists of two ceiling-mounted radiographic tubes with a common angle exposed at the same time on two film exchangers. Prior to the dynamic investigation a reference position of the subject's knee was obtained. All subjects' performed their reference radiograph lying supine with the knee of study aligned in 0° extension. While this reference procedure was carried out all subject's knee were placed in a calibration cage so that tibia was parallel to the longitudinal axis of the cage and the rotation of the leg. The transverse axis of the tibia was kept parallel to the posterior edge of the femoral condyles. During the study of knee kinematics all subjects were instructed to stand with the knee in a maximally flexed position with the foot in a neutral position, but still be in the measurable volume.

Fifteen reflective skin-markers were applied following the model described earlier. A reference measurement was then recorded with the OTS.

Subjects were given the opportunity to test the trial procedure which including a slowly extend the knee to full extension during 3-4 seconds before the actual measurement took place. When subjects felt comfortable with the procedure, the motion were recorded with sequential exposures at a rate of two to four exposures per second. At the same time recordings with the OTS was obtained at 240 exposures per second.

2 PROBLEM AREAS

Retro-reflective markers that are attached to the skin with double-adhesive tape are prone to move in relation to the underlying bone structure. *The extent to which this occurs in the case of markers placed on the ankle and foot is addressed in Study I.*

The soft tissue on the leg moves in different ways in relation to the bony structure when exposed to a provocation. These movements depend on their composition in terms of skin, muscles, tendons and fascias. *This topic is addressed in Study II.*

In clinical gait analysis, knee kinematics is one of the most important variables. Measuring knee kinematics is, however, associated with a number of difficulties. *This topic is addressed in Study III.*

Clinical gait analysis is commonly used when studying patients with gait disorders. Studies of the relationship between muscle weakness and gait pattern, in terms of moments and power, in children with cerebral palsy have been only rarely reported. *This topic is addressed in Study IV.*

Patients with a transfemoral amputation (TFA) that are traditionally treated with socket prostheses frequently report problems with low back pain. This is often related to the limitation of extension in the hip on the affected side which, in its prolongation, might cause an excessive anterior tilt in the pelvis. *This topic is addressed in Study V.*

.

3 AIM OF THE STUDIES

Study I	To investigate the degree of displacement between skin-mounted markers and the underlying bony structures of the foot.
Study II	To examine the soft-tissue stiffness of the lateral side of the leg and to explore the oscillation characteristics of wand markers on the thigh and shank.
Study III	To compare the kinematics of the knee measured simultaneously using an optical tracking system and skeletally fixed markers using radiostereometric analysis (RSA).
Study IV	To evaluate the relationship between muscle strength and gait pattern with special emphasis on kinetics in children with cerebral palsy (CP).
Study V	To evaluate the effects on the gait pattern in patients with a transfemoral amputation after changing from a socket to an osseointegrated (OI) prosthesis.

4 SUBJECTS AND METHODS

4.1 Subjects

A total of 132 individuals were included in the five studies (Table 1).

Table 1. Summary of patients/subjects participating in the five studies.

Study	I	II	III	IV	V	All
Patients	*	*	9	20	19	48
Subjects/controls	6	1	0	20	57	84
Males/females	4/2	1/0	2/7	15/5 (13/7)	40/36	132

Study I
Six healthy subjects, two females and four males, participated in the study. The median age of the subjects was 34.5 (30-51) years.

Study II
A healthy male adult subject, mass 85 kg, height 1.78 m and age 35 years, with no surgical history, was chosen as the subject.

Study III
A total of nine patients, with a median age of 63 years ranging from 59 to 72 years, who had undergone total knee replacement (TKR), were studied. During the TKR operation, five to seven tantalum markers were inserted into the tibia and femur respectively. Seven patients were studied at the 1-year follow-up visit and two at the 2-year follow-up visit.

Study IV
Twenty patients (15 males/5 females) with bilateral spastic cerebral palsy (CP) and 20 normally developing children (13 males/7 females) were included in this study. The median age (range) was 12.9 (9.4-15.3) for the CP group and 13.0 (10.2-15.7) for the control group.

Study V
Nineteen patients, 10 females and 9 males, with a unilateral transfemoral amputation, were studied on two occasions. The median age (range) for the group at study inclusion was: 46.5 years (19.9-62.3). Of the 19 patients:

- Thirteen were amputated due to trauma
- Four were amputated due to a tumour
- One was amputated due to infection
- One was amputated due to an arterial embolus

The median (range) time since the primary amputation was 7.7 years (1-42).

Fifty-seven (57), age-, side- and gender-matched, healthy subjects served as controls. The median age (range) for the controls was 46.1 years (20.3-69.8).

4.2 Methods

Study I
Six spherical lead markers with a diameter of 2 mm were glued with silicone-based skin glue (Uro-Bond® III Brush-On Adhesive, UROCARE Products, Inc. 2735 Melbourne Avenue, Pomona, CA 91767-1931, USA) to the skin directly above the following landmarks:

- The medial malleolus
- The navicular bone
- The medial part of the calcaneus
- The head of the first metatarsal bone
- The base of the first metatarsal bone
- The base of the fifth metatarsal bone

A special platform (Figure 8) was constructed to make it easier for subjects to keep their foot in the three fixed positions; 20 degrees upwards (towards dorsal flexion), horizontal and finally 30 degrees downwards (towards plantar flexion), throughout the radiographic measurements.

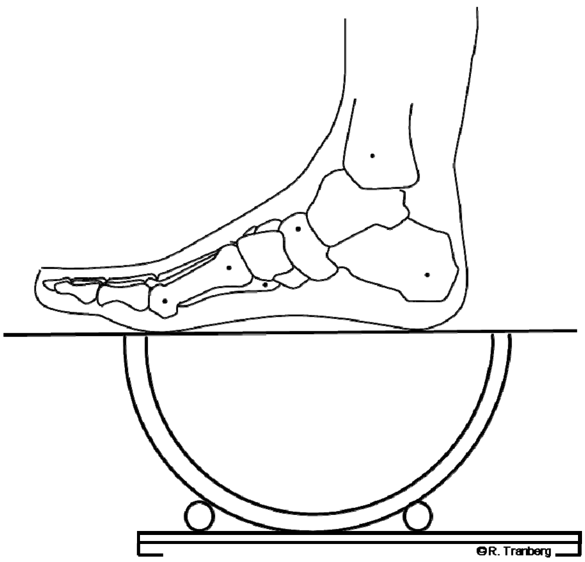

Figure 8. *Adjustable platform allowing subjects to maintain their foot position during the measurement session.*

While radiographs were taken, the subjects were instructed to stand on the exposed foot only. A co-ordinate system was marked out over each bony segment (Figure 9). Digitisations of all radiographs were carried out by one person using a Podscat™ digitiser (Logotech Graphic System International GmbH, Hilden, Germany). Before any calculations were made, all marker position data were normalised to a foot of size 42. The movement of each marker in the x- and y-directions was calculated relative to its local co-ordinate system for the three different foot positions described above.

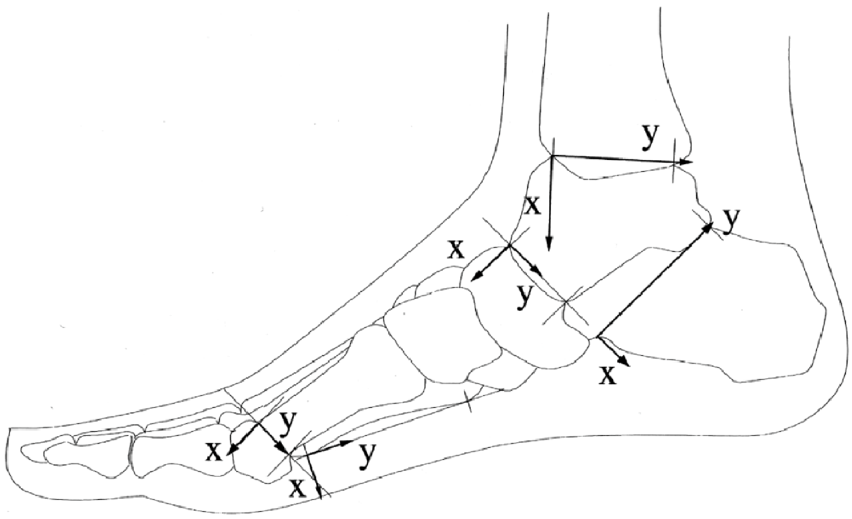

Figure 9. *Local coordinate systems applied on each bony structure.*

Study II
Soft-tissue stiffness
A tangential load of 2.5 N was applied to an adhesive tape attached to the skin surface at nine locations. The total contact area of the adhesive tape was 500 mm^2 during all the tests. A reflective marker was attached to the tape and its direction was measured by a five-camera motion analysis system at 240 Hz (ProReflex™ MCU240, Qualisys AB, Sweden). Additional markers were attached to locations on the same segment (thigh or shank). As a result, the deflection of the marker during study could be expressed using a segment-fixed co-ordinate system. The thigh and the shank segments were examined at nine different locations, which were equally distributed, along the lateral side during loading in the anterior/posterior direction (Figure 10).

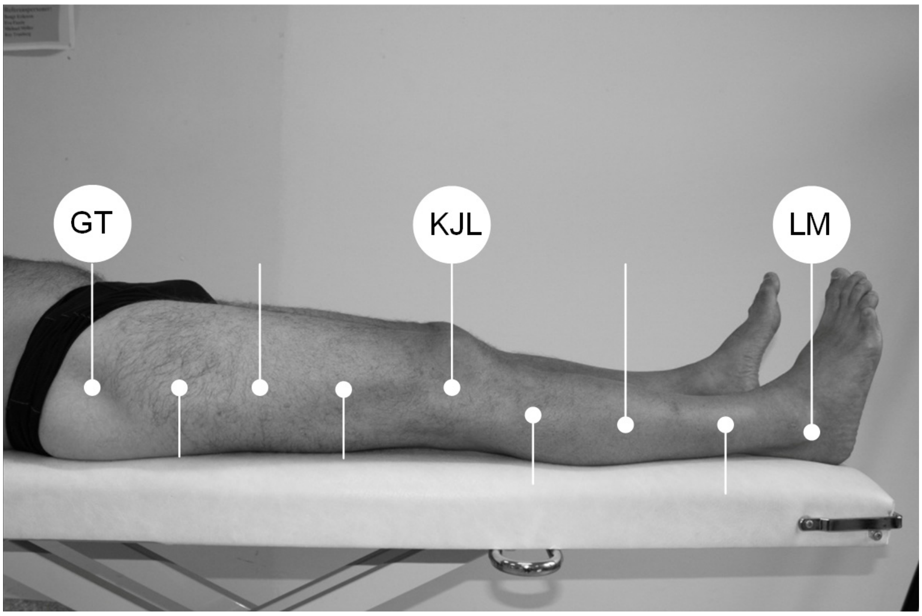

Figure 10. *Markings for the nine examination locations. Great trochanter – GT, knee joint line – KJL and lateral malleolus – LM.*

For loading in the proximal/distal direction, the same locations were used, apart from the 25% and the 75% locations on the thigh and shank respectively.

The magnitude of the force was chosen to be limited so that it deflected only the local tape marker and not the other markers on the segment. During the first part of the measurements, the subject lay on his back on an examination couch positioned with the leg at the edge of the couch, thereby allowing the applied force to hang free. During the second part of the measurements, the subject kept his leg vertical. For both parts of the test, the subject was instructed to keep his leg relaxed during one trial and then, during the next trial, to activate the muscles as much as possible, still keeping his leg in the same position.

The stiffness, *k* (N/m), was calculated from the equation

$$k = \frac{F}{\Delta}$$

Where *F* is the force (N) applied to the body and Δ (m) is the displacement produced by the force.

During the test of marker movement, the force was applied just before the camera system started to record. When recording began, the load was released within two seconds and the soft tissue was able to return to its original position.

Oscillation measurements
Free oscillations of wand markers following release from the initial displacement were recorded. The steel wand has a length of 100 mm and a diameter of 1.6 mm. On the tip of the wand, a small aluminium disc was mounted where the markers were attached with double-sided adhesive tape. The base of the wand was fixed to an aluminium plate. This plate was fixed to an elastic strap to attach the wand marker to the thigh or shank (Figure 11).

Excluding the strap and the marker, the total mass of the wand and aluminium plates was 15 g. Two marker sizes were used during the testing; a 30 mm diameter marker with a mass of 3.4 g (marker 1) and a 19 mm diameter marker with a mass of 0.8 g (marker 2).

Marker positions were sampled at 240 frames/second using a five-camera ProReflex™ motion analysis system. During all the measurements, the subject was instructed to hold his leg vertically and the wand marker was displaced initially (approximately 20 mm) and then suddenly released.

Figure 11. *Subject with a wand marker on the thigh and skin markers on the great trochanter major and lateral side of the knee.*

The initial displacement was in the anterior/posterior direction during the first recordings, followed by the proximal/distal directions during the remaining measurements. Recordings were made for both relaxed muscles and when muscles were activated. Vibration responses of the wand markers alone were examined with the base plate mounted on a rigid surface. In order to obtain information on resonant frequency, spectral analysis using Fast-Fourier-Transform (FFT) was performed on the displacement samples. The FFT calculations were made using Acqknowledge III™ software (BIOPAC Systems Inc, Santa Barbara, CA, USA).

Study III

Dynamic RSA with two film exchangers placed parallel to one another was used. The two film exchangers were set to expose in an order of 4-4-3-3-2-2 exposures per second. A uni-planar calibration cage, specially constructed for this purpose (Figure 12) (RSA Biomedical™, Umeå, Sweden), was attached in front of the film exchangers. Both radiographic tubes were placed symmetrically with a film to a focus distance of 1.5 metres and an angle of 20° in relation to an axis perpendicular to the calibration cage respectively [29-34].

Figure 12. *A uni-planar calibration frame mounted in front of the two film exchangers. In this case, the frame was equipped with six retro-reflective markers to make it possible to trace in the OTS.*

An optical tracking system (OTS), consisting of eight cameras (MCU 240, Qualisys AB, Göteborg, Sweden), was used to record the skin-marker positions. Cameras were placed to surround the subject, without interfering with the radiographic equipment. Dynamic calibration was then performed, resulting in a total measurable volume of 2.4 m^3 (1.6 x 1.0 x 1.5 metres). After the calibration of the OTS system was completed, a static recording of the position of the RSA calibration cage was made to obtain the systematic differences between the two co-ordinate systems. Prior to the dynamic RSA examination, a static RSA examination was performed with the subject lying supine. In this exposure, the knee was aligned with the RSA cage co-ordinate system in a standardised way [31]. The position of the knee in this examination constituted the "starting" or calibration position for the subsequent dynamic RSA measurements. In order to record the knee kinematics, a total of 15 spherical reflective markers with a diameter of 19 mm were attached to the skin using double-adhesive tape (Figure 13). Markers were attached to the skin on the sacrum, anterior superior iliac spine, lateral knee-joint line, proximal to the superior border of the patella, tibial tubercle, heel, lateral malleolus and between the second and third metatarsals [29].

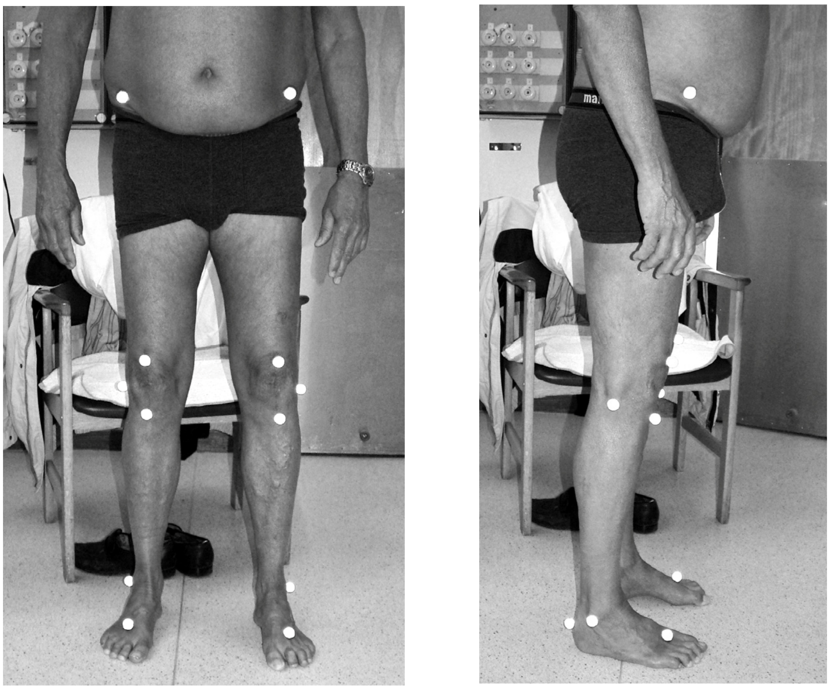

Figure 13. *Subject equipment with the 15 reflective skin markers.*

A physiotherapist with more than 10 years' experience of marker placement for this marker model performed all the marker attachments on all subjects. The subjects were then asked to enter into the measurement volume and a static reference recording was obtained with the OTS while the subjects were standing in an upright position aligned with the x-axis of the global co-ordinate system (Figure 14).

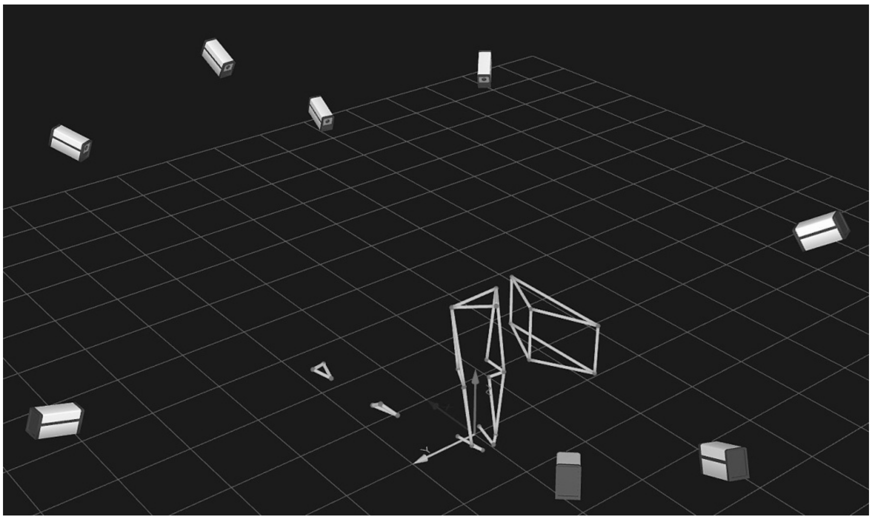

Figure 14. *Measurement set-up with the OTS cameras surrounding one subject placed in the middle, together with the two radiographic tubes on the left-hand side and the film exchanger with the calibration frame on the right- hand side.*

Prior to the simultaneously recorded measurements, all the subjects were instructed to stand with their knee in what was for them a maximum flexed position and, at a given signal, slowly extend their knee as much as possible.

To ensure that both systems were recording at the same time, opto-sound synchronisation was used. This made it possible to synchronise each measurement within 1/25 s.

The stereo radiographs were scanned (Scan Maker 9800XL, Microtek International, Inc, Taiwan) and each of the cage and patient markers was measured using digitised images. Knee-joint movements were evaluated using the UmRSA® software (RSA Biomedical, Umeå, Sweden). For the RSA and OTS systems, Euler angles were used to express joint angles. All joint angles were calculated in the same order; flexion/extension, internal/external rotation and varus/valgus respectively to avoid discrepancy between the two systems. In this situation, and in previous evaluations of knee kinematics, we have used the femur as a fixed reference segment and the tibia as the moving segment [34-37].

Before any calculations of data obtained from the OTS were made, all marker position data were filtered using a Butterworth low-pass filter with a cut-off frequency set at 15 Hz. This was followed by calculations of hip-, knee- and ankle-joint kinematics, based on data from the optical tracking system. All seven body segments, i.e. the pelvis together with the thigh, shank and both feet, were created on the basis of 3-D co-ordinates derived from the 15 skin markers.

Defining the pelvis segment and hip joints
A modified Coda pelvis [38] was used to define the pelvic segment and calculate the hip joint centres. The modification consisted of a reduction of the two posterior markers on the left and right posterior superior iliac spine (PSIS) into one marker on the sacrum, positioned at the mid-point between the left and right PSIS (Figure 15).

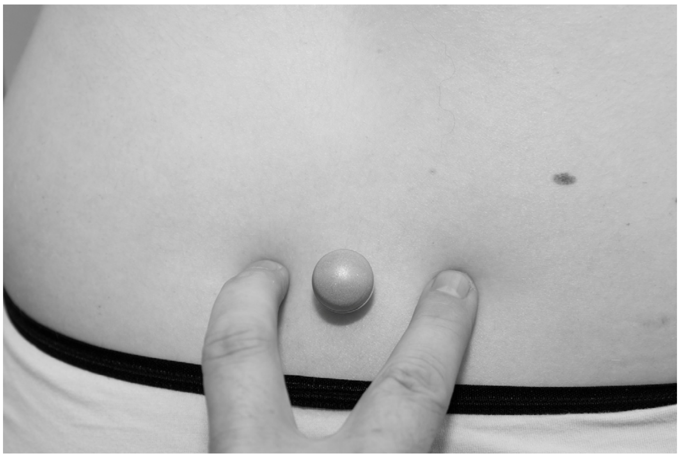

Figure 15. *Palpation of PSIS bilaterally for position of sacrum marker.*

One marker on the left and one on the right anterior superior iliac spine respectively completed the pelvic segment (Figure 16).

Figure 16. *Reflective markers placed at ASIS bilaterally.*

The local co-ordinate system of the pelvis segment originates at the mid-point of an axis defined by the right and left anterior superior iliac spine (ASIS) markers and its positive x-axis pointing towards the right ASIS marker. To define the y-axis, the sacrum marker is used, but it is rotated 180 degrees and thus points forward, i.e. away from the sacrum marker. Perpendicular to the x- and y-axis, the positive z-axis points upwards (Figure 17).

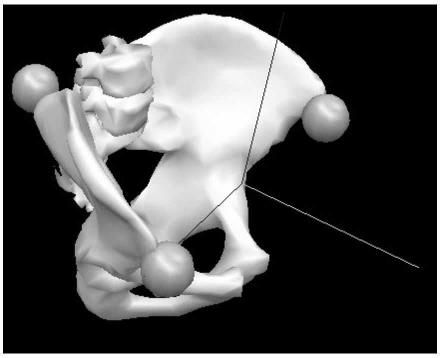

Figure 17. *Animated model of Pelvis used in studies III and V.*

Hip-joint centres are defined in relation to the pelvis segment, according to Bell and co-workers [38, 39] recommendations for right and left hip-joint centres (RHJC and LHJC).

The hip-joint centres in the x, y and z co-ordinates are expressed as follows:

$$RHJC = (0.36 \cdot ASIS\ Dist.), (-0.19 \cdot ASIS\ Dist.), (-0.3 \cdot ASIS\ Dist.)$$

$$LHJC = (-0.36 \cdot ASIS\ Dist.), (-0.19 \cdot ASIS\ Dist.), (-0.3 \cdot ASIS\ Dist.)$$

Defining the shank segments, knee joints and ankle joints:
The shank segment is used as base for definitions of the knee and ankle joint. For definitions of the knee joint, the two markers – lateral knee-joint line and lateral malleolus – together with the marker at the tibial tubercle serve as the origin of further transformations. First, a correction of marker off-set, i.e. the distance from marker centre to skin surface, was made. For the 19 mm markers used throughout Studies III, IV and V, this distance was generally set at 0.009 m (9 mm). This resulted in two new virtual, also called landmarks, which in this case will be called knee-joint line moved (Figure 18) and ankle moved for the left and right side respectively.

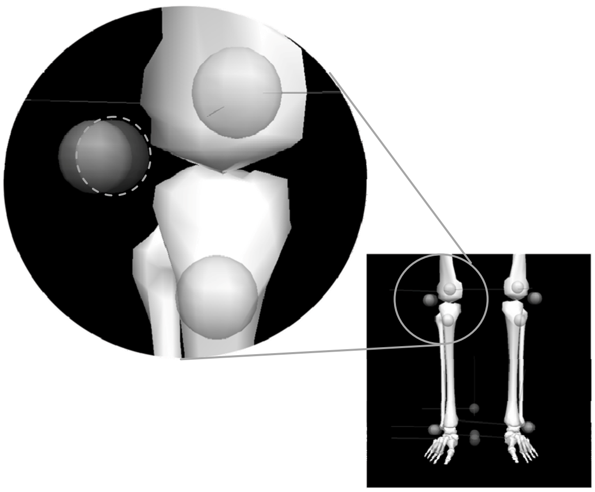

Figure 18. *A virtual marker (dotted circle) generated with an off-set of 0.009 m from the position of a physical marker, in this case the lateral knee joint line maker.*

The distance between these two virtual markers is used as a scaling factor for the two new landmarks; in this case, the knee joint and further the ankle joint. The joint positions are expressed as a percentage of the scaling factor and are based on definitions proposed by Gardener and Clippinger [40], together with empirical knowledge.

Euler angles, expressed as joint angles, were calculated using Visual 3-D™ Professional ver. 3.99.25.8 (C-Motion, Inc, Germantown, USA), consistently using the proximal segment as the reference segment. All joint angles were calculated using the Cardanic sequence [17], but with a change in the order of the calculation of angles from x-y-z to x-z-y. This means that the angles are expressed in the order of flexion/extension, internal/external rotation and varus/valgus angulations respectively in order to comply with the order of calculations used in the RSA system.

Study IV

Participants were identified from the medical records at the gait laboratory in Gothenburg and from habilitation centres around Gothenburg. The inclusion criteria were bilateral spastic CP, age nine to 15 years and walking without assistive devices. Normally developing children of the same ages were recruited from a group made up of the children of staff and friends. Exclusion criteria were orthopaedic surgery during the past 12 months or botulinum toxin injection within the last six months. The children were tested for muscle strength with a hand-held myometer (Figure 19) (Chatillon Force Measurement Systems, Florida, USA) in eight muscle groups in the lower extremities: hip extensors, hip flexors, hip abductors, hip adductors, knee extensors, knee flexors, dorsiflexors and plantar flexors. The examination procedure used in this study was similar to the procedure used by Eek and co-workers, using the “make” technique and with standardised positions [41]. After instruction and familiarisation with the procedure, three attempts were made and the recording with the maximum value was used for further analysis. The lever arm for each muscle group was measured with a tape measure and the joint moment (Nm) was calculated.

Figure 19. *Hand held myometer used in study IV.*

All the values were then normalised to Nm/kg by dividing them by each subject’s body weight in order to make comparisons with kinetic gait analysis data possible. Gross motor function in the children with CP was classified according to the Gross Motor Function Classification System (GMFCS) [42].

All the children performed an instrumented gait analysis with a 6-camera motion capture system (ProReflex MCU 240, Qualisys AB, Gothenburg, Sweden), together with two force plates (Kistler 9182C, Kistler Group, 8408 Winterthur, Switzerland). The cameras were arranged in order to enclose a measurement volume of 13.8 m^3. Before measurements took place, subjects were equipped with 18 retro-reflective skin markers. These markers were attached to the skin using double-adhesive tape. All the subjects had the opportunity to test the walking path before the measurements took place. Three measurements were recorded and the measurement that differed least from the other two was used for further analysis. Kinematic data from the skin markers served as a basis for calculating joint centres. Joint kinetics, in terms of joint moments and joint power, were calculated using inverse dynamics. Moment and power were normalised to the subject's body weight. For joint moments, the maximum value exerted by each of the muscle groups was used for analysis. Generating and absorbing gait power respectively at the hip and ankle joints in the sagittal plane was calculated. The area enclosed by the positive section of the graph and the x-axis represented generating power and the negative section of the graph and the x-axis represented absorbing power as described by Riad et al. [43].

Study V

Kinematic data were acquired with a computerised video motion capture system utilising eight infrared cameras recording at 240 samples per second (Qualisys MCU 240, Qualisys AB, Gothenburg, Sweden). Eighteen reflective spherical markers were attached to the skin over bony landmarks (acromion, 12th thoracic vertebra, sacrum, anterior superior iliac spine, lateral knee-joint line, proximal to the superior border of the patella, tibial tubercle, heel, lateral malleolus and between the second and third metatarsals) [13]. The markers were placed on left and right side respectively. The cosmetic foam on the prosthetic components was removed to enable the placement of markers. The mirroring of marker positions from the intact side to the prosthetic side was based on individually adjusted distances (Figure 20). Prior to the recording of the kinematic data, all patients and controls had an opportunity to familiarise themselves with the walkway and define a comfortable walking speed. Three recordings were acquired from each patient and control. The most representative measurement was chosen for further analyses.

Figure 20. *Patient with osseointegrated prosthesis equipped with reflective markers.*

4.3 Statistical methods

Study I

Excel for Windows was used for descriptive statistics, the normalisation of data and calculations of mean and standard deviations (SD).

Study II

Descriptive statistics of subject

Study III

Data from the RSA system were interpolated at 5° intervals using a linear approach. Corresponding values from the OTS system were then extracted for comparison. Data from the RSA system and the OTS system were presented as mean values (SEM). For the comparison and interpretation of data on a subject basis, scatter plots with linear regression with a 95 per cent confidence interval were used. SPSS ver. 17.0 was used for all statistical calculations.

Study IV

Non-parametric tests were used as not all the data could be shown to be normally distributed. Comparisons between groups were analysed with the Mann-Whitney U-test. Correlations between muscle strength, gait moments and power were tested with Spearman's rank correlation coefficient. p-values of 0.05 or less were regarded as evidence of statistically significant findings.

Software packages Statview and SPSS (version 15.0), were used for statistical analysis.

Study V

Non-parametric tests were consistently used, as some of the data were not normally distributed. Comparisons within the amputee group, first and second occasion, were analysed using Wilcoxon signed-rank test. Comparisons between groups, patients and controls, were analysed using the Mann-Whitney U test. p-values of 0.05 or less were considered to be statistically significant. SPSS version 16.0 was used for statistical calculations.

4.4 Ethics

Study I

All the subjects received oral information about the purpose, procedure and risks of the study. No ethical approval was obtained.

Study II

The subject was a member of the research group and could be regarded as having full knowledge of the procedure and risks of the study. No ethical approval was obtained.

Study III

Informed consent was obtained from all nine subjects and the study was approved by the regional ethics committee in Gothenburg, Sweden, by 14 June 1999. Document ID: R 301-99.

Study IV

The Regional Ethical Review Board in Gothenburg approved the study on 15 October 2001. Written informed consent was obtained from the parents of each participant. Document ID: Ö 478-01

Study V

The Regional Ethical Review Board in Gothenburg approved the study on 14 June 1999. Document ID: R 031-99

5 SUMMARY OF PAPERS

Study I: *The relative skin movement of the foot: a 2-D roentgen photogrammetry study.*

The purpose of this study was to investigate the size of the displacement between skin-mounted markers and the underlying bony structures of the foot.

INTRODUCTION: Markers that are mounted at the skin are thought to follow the underlying structures. This is, however, not correct, due to skin movement artefacts, which can be referred to the movement of the skin in relation to the underlying bone structures. This movement will, at least to some extent, influence the outcome of a motion analysis when skin-mounted markers are used.

METHOD: Six healthy subjects, two females and four males with a median age of 34.5 (30-51), participated in the study. Prior to the examination, six spherical lead markers were glued to the skin over the following bony structures: the medial malleolus, the medial side of the calcaneus, the base and the head of the first metatarsal bone and the base and the head of the fifth metatarsal bone. 2-D radiographs were taken from a medial aspect of the foot in three different positions, starting at 20° dorsal flexion, at 0° and ending in 30° of plantar flexion.

RESULT: Markers were found to move up to a mean of 4.3 mm (SD 2.47). The most relative movements were found for the most proximally placed markers. Less marker movement was seen for markers placed more distally with a mean displacement of 1.8 mm (SD 0.52).

CONCLUSIONS: Markers mounted on the skin of the foot moved from 1.8 to 4.3 mm, while a dorsal-plantar range of movement of 50° was performed. The largest movement was seen for the most proximally attached markers.

Study II: On skin movement artefact-resonant frequencies of skin markers attached to the leg.

The purpose of this study was to examine the soft-tissue stiffness along the lateral aspect of the leg and to explore the oscillation characteristics of the wand marker system on the thigh and shank.

INTRODUCTION: Skin movement artefacts may lead to substantial errors when calculating bone motion from data obtained from markers attached to the skin. For slower movements, the values of skin movement artefacts might be relatively small and regarded as measurements of static postures. However, during faster activities the transient motions of the soft tissue might produce higher artefact values that will affect the measurements.

METHOD: A healthy male adult subject (mass 85 kg, height 1.78 m, age 35 years) was studied. For the examination of soft-tissue stiffness, a total of nine locations along the lateral side of the leg were investigated. A tangential load of 2.5 N was applied to the location of interest using adhesive tape. To explore the oscillation characteristics, two markers with different sizes were used, together with a wand of 100 mm.

RESULT: The tangential stiffness varied between 330 and 1900 N/m, with the lowest stiffness in the anterior/posterior direction on the proximal thigh. The highest stiffness was in the proximal-distal direction at the knee. The resonant frequencies of wand markers placed on the leg were found to be typically 17-24 Hz for the 30 mm marker and 41-45 Hz for the 19 mm marker. The effect of keeping the muscles tensed was also measured and increased muscle tension provided a more stable base, producing increased stiffness. For the values of the smallest marker (diameter 19 mm), combined with the relaxed muscle case, the vibration was damped too quickly to derive enough data for spectral analysis.

CONCLUSIONS: A study of the transient parts of the skin movement artefact indicates that attachment sites located on the thigh should be avoided and, if wands are going to be used, markers with the lowest mass are preferable.

Study III: *Simultaneous measurements of knee motion using an optical tracking system and roentgen stereo photogrammetric analysis (RSA).*

The purpose of this study was to compare two methods for measuring the 3-D kinematics of the knee joint.

INTRODUCTION: Invasive methods are more reproducible and accurate than non-invasive ones when it comes to recording knee-joint kinematics, but they are usually less accessible and less safe. For this reason, non-invasive methods with passive markers are widely used. With these methods, varying marker sets based on a number of single markers, or sets of markers, known as clusters, are used to track the body segments.

METHOD: Nine subjects (ten knees) were investigated simultaneously with a dynamic RSA system and a motion capture system while performing an active knee extension.

RESULT: Comparisons of flexion/extension revealed good agreement between the two systems, with a slight overestimation by the OTS system, starting with a difference of 2° at 0°of knee flexion and gently switching to a moderate underestimation, resulting in a difference of 5° at a more pronounced flexion value. The external/internal rotation angle showed fairly good correspondence during the first 15° of flexion, with a mean difference within one degree. However, an increasing divergence was observed from 20° of flexion, ending in a maximum difference of 11.4° at 50° of knee flexion. Finally, for abduction/adduction, a systematic difference with a variation of 2° to 4° was present throughout the measured range of motion. Linear regression analysis showed a clear dependency between extension-flexion angles, as recorded with the two methods, with an r^2 of 0.96. Data for abduction-adduction and internal-external rotation respectively showed a lack of dependency, with r^2 0.036 and 0.00014 respectively.

CONCLUSIONS: The skin-marker model presented here produced reliable data for flexion/extension. Recordings of internal/external rotation and abduction /adduction were less accurate on an individual basis than at group level, most probably due to soft-tissue motion and the presence of small movements that occur in these planes.

Study IV: *Muscle strength and kinetic gait pattern in children with bilateral spastic CP.*

The purpose of this study was to evaluate the relationship between muscle strength and gait pattern with the emphasis on kinetics in children with cerebral palsy (CP).

INTRODUCTION: Cerebral palsy is often associated with an abnormal gait pattern. The relationship between muscle strength and kinetic gait pattern in children with bilateral spastic cerebral palsy has only been noticed by a few authors. A review of the literature reveals that the relationships between muscle weakness and gait pattern is not fully explored or understood.

METHOD: A group of twenty patients, age ranging from 9 to 15 years, with bilateral spastic cerebral palsy, were investigated and their data were compared with those of a control group consisting of twenty normal children. The children were tested for muscle strength and underwent gait analysis.

RESULT: Children with CP were not only significantly weaker in all muscle groups but also walked at a slower speed and with a shorter stride length when compared with the reference group. Gait moments differed at ankle level, with significantly lower moments in children with CP. Gait moments were closer to the maximum muscle strength in the group of children with CP. Furthermore, a correlation between plantar flexion gait moment and muscle strength was observed in six of the eight muscle groups in children with CP; a relationship that was not found in the reference group. A similar pattern was seen between muscle strength and generating ankle power with rho values between 0.582-0.766.

CONCLUSIONS: The present study confirms previous observations that there is a correlation between muscle strength and gait pattern now also described in terms of kinetics. The results indicate that plantar flexors are important muscles which are often compromised in children with CP with weakness that affects the kinetic gait pattern. Furthermore, other muscle groups in the leg are also needed in order to produce an effective ankle plantar-flexing moment and generate power.

Study V: Improvements in hip and pelvic motion for patients with osseointegrated transfemoral prostheses.

The purpose of this study was to evaluate changes in the hip and pelvic kinematics in patients with a transfemoral amputation after changing from a socket to an osseointegrated (OI) prosthesis.

METHOD: Nineteen patients with transfemoral amputations were studied with 3-dimensional gait analysis, walking with a socket prosthesis within two days before the osseointergration procedure. Post-operative gait analysis was carried out at the 2-year follow-up visit. Fifty-seven (3 controls to 1 patient) age-, side- and gender-matched healthy subjects constituted the control group.

RESULT: Postoperative data revealed that patients with an osseointegrated transfemoral prosthesis increased their hip extension by 7.3° ($p=0.007$), changing from -2.6° (range -13.4°—10.7°) to -9.9° (range -29.4°—5°). Moreover, the pre-operative anterior pelvic tilt was reduced by 4.0° ($p=0.016$), changing from 21.7° (range 11.9°—34.8°) to 17.7° (range 5.5°—25.7°). Not only the values for hip extension but also the values for pelvic tilt changed toward the values of the controls.

CONCLUSION: We were able to confirm previous observations that patients walking with a socket prosthesis do so with reduced hip joint extension and increased anterior pelvic tilt. We also found that conversion to a limb prosthesis fixed to the bone improved this walking pattern by increasing the hip extension and reducing the pelvic tilt.

6 DISCUSSION

Study of body motions (Studies I-III)

Studying the movements of humans is a demanding task. Analysing movements that look simple is in fact a fairly difficult proposition, as soon as we take a more serious look at all the details of this movement. For centuries, technical devices have been used to document human movement. Moreover, it is clear that, as soon as we attempt to measure something, no matter what, we also bring errors into the measurement in question. When analysing data from gait analysis, it is of fundamental importance that the clinicians and engineers who are taking care of subjects, regardless of whether this takes place in the research laboratory or in a clinical setting, must be aware of the way these errors influence the final outcome. Another important topic is the interpretation of what is being measured. One of the most frequently discussed errors affecting the result of measurements of human motion is the movement of soft tissues in relation to the underlying bony structures, generally known as *"the soft tissue artefact"* (STA) and in particular on the foot, known as *"the skin motion artefact"* (SMA).

In *Study I*, this error was studied for the foot segment at six different locations, starting at the medial malleolus and ending at the first metatarsal joint. In general, a larger displacement was noted for the proximally placed markers, with less displacement as distal positions were studied. Taking this into account and transforming this result into clinical terminology, the artefact would result in an error of approximately ± 2.5° within the 50° range of motion studied.

The foot model used in Studies I and III-V assumes that the foot is one rigid segment, but it is clear that this is not always the case.

Attempts to divide the foot into several segments have been presented and, in 1999 a five-segment foot model, including the shank, calcaneus, mid-foot, the first metatarsal bone and the hallux, was presented by Leardini and co-workers [44]. This model was based on a rigid cluster attached to the skin surface with tape, together with landmarks defined by a pointer. A consistent and repeatable pattern of rotations within the group of subjects was reported for several of the joints studied. Moreover, in 2001, Carson and co-workers [45] published their work on a three-segment foot model divided into hind-foot, fore-foot and hallux, a model that was fairly similar to that of Leardini and co-workers but based on skin markers. This foot model has also been called the Oxford foot model. Questions about whether movements of the foot segment should be detected with skin markers, as in the Oxford foot model, or clusters still remain. This question has been raised by Nester and co-workers [46]. They concluded that, in comparison with skin markers and clusters on the foot with bone-anchored markers, there was no clear answer to the question of whether skin markers or clusters were preferable to reproduce the actual bone movement of the foot.

In *Study II*, we studied the soft tissue artefact (STA) in terms of stiffness and response to a vibrating provocation of the thigh and shank. This is a well-known problem, especially when the emphasis is placed on the thigh [47-52]. The results of the present study revealed relatively small displacements of the skin markers along the thigh and shank, as well as at the knee and the ankle, and there were generally no large differences between the nine locations.

This examination was, however, performed in two static positions with muscles relaxed or tensed and the risk of additional displacements due to movement must be taken into account.

Due to the construction of the thigh, with large muscle masses surrounding the femoral bone, it can be expected that it will be sensitive to soft tissue artefacts in general. Values of high STA have been reported, with magnitudes varying from 2.5 mm to 42 mm [52-57]. It would therefore be preferable to reduce the number of markers attached to the thigh to a minimum. Most models use one thigh marker, a marker placed on a wand on the lateral side of the thigh or a supra-patellar skin marker. The use of a thigh marker as an indicator of the axial rotation of the thigh segment originates from the assumption that a wand would increase the distance between the marker and the axis of rotation and would therefore predict the rotation with greater accuracy. This could, however, be contradictory due to the fact that a large STA is present. Wren and co-workers [58] investigated the effectiveness of a patella marker to track hip rotation in comparison with thigh markers on wands at two locations. The results show that the patella marker was clearly better in both static measurements and measurements during walking and was not changed by different gait speeds [58].

In Study III, the 3-D knee kinematics derived from the optical tracking system and a marker model based on 15 skin markers were studied and compared with kinematic data obtained using a roentgen stereophotogrammetric analysis (RSA) system. Nine subjects, a total of ten knees, participated in this study.

All the subjects had undergone total knee arthroplasty prior to the examination.

Seven subjects were studied at their one-year follow-up and two at their two-year follow-up. During the measurement, all the subjects started in a flexed position and, on a given signal, they slowly extended their knee under weight-bearing. This movement took place in a standardised movement during RSA measurements to obtain maximum range of motion. Prior to the measurement, a few patients experienced difficulty completing the extension movement of their knee and an adjustment of the starting position was made. Possible reasons for this can be lack of balance, high BMI (mean 29 ±5.2) and limited range of motion due to the TKA.

The findings of the studies in the present thesis show generally good conformity on flexion/extension up to 30° of flexion, when a fairly small divergence in the OTS compared with RSA occurs. This divergence slowly increases during the extension phase and results in a underestimation of 5° at 50° of flexion. The reason for this divergence is not completely clear, but one likely cause is the so-called "cross-talk" which originates from differences in the alignment of the co-ordinate axis to the body axes.

In terms of the internal/external rotation, a similar pattern was observed with good conformity throughout the first 20° of flexion. With an increasing flexion angle, the differences between the two systems gradually increased to a maximum of more than 10° at 50° of flexion. The natural range of rotation about the longitudinal axis is approximately 6° [59] and the large difference seen at a large angle of flexion may be related not only to the misalignment of the co-ordinate axis but also to soft tissue artefacts related to the high muscle activity of the quadriceps muscle that might influence the supra-patellar marker.

When it comes to abduction/adduction translation based on group values, a divergence was seen at the beginning and at the end of the measurements, with the best agreement between 20 and 35° of flexion. The observed differences were probably caused by soft tissue artefacts that became pronounced when patients activated their muscles to initiate the extension.

Furthermore, when they reach the end of the extension movement, patients/subjects have to balance on one leg and compensate with hip and ankle movements to maintain knee stability. The artefacts caused by underlying soft tissues are probably so pronounced that they totally over-ride the small movements that may be present. Interestingly, there was a systematic difference between the two methods. In fact, this has a major bearing on the clinical usefulness of these methods. The OTS system consistently showed more abduction than the RSA system, which may be due to the fact that the knee-joint centre in the present marker model is not perfectly aligned with the RSA system and therefore produces this systematic offset, which may be designated as a systematic error. Furthermore, this offset may also be an effect of the incorrect determination of the hip joint centres[47, 60-63], which are defined in the calculations by the three markers on the pelvis [38]. Since the comparisons on an individual basis showed no correlation at all, it appears that the noise caused by the soft tissue movement is far too high for any relevant analysis of abduction/adduction during a step-up. It could be that the same problem will appear during walking, but this issue is not confirmed in the present thesis and has to be addressed in future studies. It is, however, more likely that the patient population studied here would walk with a more controlled motion pattern on flat ground.

Clinical applications (Studies IV-V)

Instrumented gait analysis as a tool for evaluating gait pattern and as a contribution to decision-making before and after surgery has frequently been used in recent decades. One of the main areas in which gait analysis has become a more or less standard procedure is in the decision-making relating to the treatment of patients with cerebral palsy (CP), [64-71]. In *Study IV*, which is one of the first studies comparing muscle strength with kinetic gait data in children with CP, it was shown that children with CP might have normal or near-normal joint moments in gait, but, to obtain this, they used a higher percentage of their maximum muscle strength. This could be one explanation of why they more easily became tired, with reduced endurance.

Muscle strength was measured with a hand-held myometer and it served as the output of voluntary muscle contraction. The co-operation of the children and good selective control is essential to perform a test like this. All the children in this study were able to take part in the full muscle strength tests and were able to activate every muscle group in isolation. Measuring plantar flexor strength and comparing these data between the groups created some difficulties, due to the fact that the children in the reference group produced a force that exceeded the capacity of the hand-held device that was being used. As a consequence of this, it is only possible to establish that the group with CP was weaker than the reference group when it came to the plantar flexors. The findings in the present study of muscle weakness in children with CP are consistent with earlier reports [72, 73] and the observed shift in generating power from ankle plantar flexors to hip has been previously described by Riad and co-workers in children with unilateral spastic cerebral palsy [43].

When studying gait moments in relation to muscle strength, it is possible that the joint moments in gait not only represent the active muscle contraction but also the torque generated by ligaments and joint structures. It was not, however, possible to separate these variables in the present study and a shortened Achilles tendon may therefore be a reason why the plantar-flexing gait moments in the children with CP exceeded voluntary muscle strength.

Previously, muscle weakness has been reported in children with CP and it has been found to correlate with kinematic gait variables, walking ability as measured with the Gross Motor Function Classification System (GMFM), walking velocity and stride length [72-74]. In a group of able-bodied children, Fosang and Baker made an estimation of grade 3 muscle strength/weakness, which is equal to lifting the extremity against gravity, and found that grade 3 strength was less than normal gait moments [75]. This indicates that grade 3 is not enough for a normal gait pattern. Previous studies by Eek and co-workers show that, in a group of children with bilateral spastic CP, muscle strength over 50% of normal in the legs was needed to be able to walk without aids [76]. Furthermore, Goldberg and Neptune [77] used a computer simulation technique of gait pattern to study compensatory strategies in the legs. Plantar flexors were shown to be able to compensate for several musculoskeletal deficits at hip and knee level, but the reverse was, however, not possible, i.e. weak plantar flexors were always connected with an abnormal gait pattern.

Although weakness in all muscle groups was found, it was only reflected in the kinetic gait pattern by reduced moments and power around the ankle, resulting in a reduced push-off. Moreover, there was a significant correlation between muscle strength in almost all the muscle groups that were measured and the plantar-flexing gait moments and ankle-generating power in children with CP, a correlation that was not seen in the reference group.

This pattern of correlation indicates that not only muscle weakness in the plantar flexors results in reduced ankle plantar-flexing moment and reduced push-off in gait but also that other muscle groups can affect the output at the ankle. One possible explanation for this may well be that good stabilisation is needed in both the hip and knee joints to be able to produce the necessary push-off at the ankle.

A previous study of children with CP related to muscle strength training and gait analysis revealed that increased strength around the hips improved the plantar-flexor generating power at push-off [78]. Ross and Engsberg showed that the muscle groups that explained the largest variance in gait and gross motor function in a group of children with CP, walking with or without aids, were the hip abductors, followed by the ankle plantar flexors and ankle dorsiflexors [79]. These facts indicate that it may be important to focus on strengthening hip muscles as well as plantar flexors in order to improve the gait pattern.

Another problem is that, although the children with CP were found to have normal or near-normal moments in gait, they used a higher percentage of their maximum muscle strength to obtain this. Working near maximum level may be tiring for muscles in the long run and may subsequently result in fatigue and reduced endurance.

Several of the results in the present study focus on the importance of ankle plantar flexors for gait pattern. As the foot is the point of contact between the body and ground, where the resulting force from the body can act to move itself in the desired direction, it is logical to focus on the kinematics and kinetics at the ankle joint. It is, however, clear that, if the soft tissues above the ankle are not stable, it is difficult to obtain full effect from the plantar-flexing forces at the ankle. Furthermore, it is well known that ankle plantar flexors are normally the largest contributors to forward propulsion and are vital to a normal gait pattern [80], as well as being important for limb support [30].

As many interventions such as stabilising braces, botulinum toxin injections and orthopaedic procedures are suggested to patients with CP on a number of occasions and, as they frequently involve the ankle, it is of great interest to understand the kinetics at the ankle when planning interventions.

Study V describes the changes in the hip and pelvic kinematics in 19 transfemoral amputees, who were treated with an osseointegrated femoral prosthesis. It is well known that limitations in gait pattern due to a TFA depend on several factors. Loss of, or reduced capacity in the hip-joint extensors might result in reduced hip extension during gait.

Compensatory mechanisms such as asymmetries in step length and lateral bending of the trunk have been reported in earlier studies [81-85]. These are mechanisms that in the long run may be one of the important underlying factors for commonly reported problems with low back pain (LBP) in this group of subjects [84, 86-91]. Several attempts to find a solution to this problem have been reported. In 2003, Lee and co-workers described the influence of the length of lower-limb prostheses upon changes in spinal kinematics. They were not, however, able to conclude that there was any correlation between leg length and the LBP reported [88]. Another contribution that could lead to LBP is the enclosure of the residual limb and lateral part of the pelvis by the prosthetic socket. This results in restrictions in the natural movement of hip extension. Rabuffetti and co-workers [84] reported that, in a group of eleven TFA subjects, reduced hip extension occurred in the late part of the stance phase on the prosthetic side, while the contralateral side is in the phase of foot strike. In the eleven TFA subjects studied in this study, a mean hip-extension reduction of 7.5° was found. They also observed that these subjects walked with an 8.2° increase in anterior pelvic tilt. The subjects in the present study displayed a similar pattern when using their socket prosthesis. They did, however, improve their hip extension by approximately 7° and also reduced the anterior tilting of the pelvis by about 4.0°, corresponding to changes towards a more normal walking pattern. However, slightly more than 4° of hip extension, together with approximately 10° of pelvic tilt, still remain to reach the level of the controls. The question of whether it is fully realistic to believe that this could be attained due to limitations related to insufficient hip muscles and the mechanical prosthetic components that are present must be taken into consideration. Unpredictably, almost no changes in hip extension were found on the non-amputated side. The present study reflects changes in kinematic patterns in the hip and pelvis occurring during the first two years after osseointegration and it is possible that further improvements could occur with time.

7 LIMITATIONS

Study I

Consideration must be taken of the fact that all the subjects performed the test in static positions and the limited sample size would reduce the opportunity to draw any definitive conclusions about what occurs during gait.

Study II

As the result is based on a static trial and as this is a case study, no generalisations about how this will influence a dynamic situation can be made.

Study III

The selection of subjects and the restricted motion studied is a limitation and the potential for transferring the obtained results to gait in general is difficult to prove.

Study IV

The opportunity to measure plantar-flexion strength needs to be further investigated. There was difficulty ensuring that the subjects only contracted the muscles that were being studied.

Study V

The opportunity to transfer a skin-marker-based model to a combination of human body segments and prosthetic components has to be further investigated. A specially developed model for prosthetics might be a solution.

8 CONCLUSION

Skin-marker displacements at the foot do not exceed 4.1° within 50° of dorsal-to-plantar flexion and the reported displacement decreases as more distal positions are studied.

The reported data indicate that markers on wands placed on the thigh and shank should be considered, especially on the proximal section of the thigh, due to soft tissue artefacts and low stiffness. Moreover, if wands are used, markers with the lowest mass are preferable.

The fifteen skin-marker model used in Study III provides reliable data for flexion/extension. However, kinematic data relating to internal/external rotation and abduction/adduction were less accurate on an individual basis than at group level.

The plantar flexors are important muscles that are often compromised in children with CP, with weakness affecting the kinetic gait pattern.

Muscle groups other than the plantar flexors in the leg are also needed to make it possible to produce an effective ankle plantar-flexing moment and generate power.

There is a correlation between muscle strength and gait pattern and this is best described in terms of kinetics.

Patients with a transfemoral amputation walking with socket prostheses do so with reduced hip-joint extension and increased anterior pelvic tilt.

Conversion to a limb prosthesis fixed to the bone improved the walking pattern by increasing the hip extension and reducing the pelvic tilt.

9 FUTURE PERSPECTIVES

Further investigations of knee kinematics with a specific emphasis on additional markers are needed, as well as evaluations of different models for calculating the determination of joint centres. For example, link-based models, six degrees of freedom and functional joint centres require further study.

Further analysis of the opportunities to track the pelvic segment is important, as well as the validation of different techniques for determining hip-joint centres.

Further investigations of kinematics and kinetics in the use of prostheses with the emphasis on specific marker models and procedures for calculating the use of prostheses are needed.

Further research on kinetic and kinematic gait pattern in patients treated with an osseointegrated transfemoral prosthesis is necessary.

SAMMANFATTNING

Instrumenterad gånganalys har sedan 1960 talet används som kliniskt utvärdering/utrednings verktyg vid ortopediska kliniker inom- och utomlands. Tekniken baseras på att ett antal reflekterande sfäriska hudmarkörer som fästs på huden. Under rörelsemätningen befinner sig försökspersonen inom en definierad volym, den så kallade mätvolymen. Mätvolymen är det utrymme som inom vilket mätningar kan förtas och som innesluts av videokamerornas (3 eller flera) bildfält. Hudmarkörernas position registreras under inspelningen av försökspersonens rörelsemönster och bildar tillsammans underlag för att följa de olika kroppssegmentens rörelser. Forskningsarbetet fokuserar i de tre första arbetena på metodologin kring att visualisera fot- och knäledens kinematik med hjälp av metoden med hudmarkörer och videokameror. Resultat från studie 1 avseende hudrörlighet på foten visar att markör rörligheten i förhållande till underliggande benstrukturer som mest rör sig 4,3 mm vid ankeln för att successivt avta till 1,8 mm vid stortåleden. I studie 2 belyses problematiken med mjukdels-rörligheten längs nedre extremiteten. Hud och underliggande strukturer provoceras dels genom antero-posterior och longitudinell belastning och dels genom att sättas i svängning för att undersöka töjningen respektive dämpningen i strukturerna. Förloppen registrerades med hjälp av hudmarkörer och videokameror. Resultaten från studien visar på att den huvudsakliga hudrörligheten förkommer i antero-posterior led samt att betydande egensvängningar förekommer då markörer placeras på utriggare. I tredje delarbetet undersöks noggrannheten i det videobaserade rörelseanalyssystemet med hjälp av ett röntgen stereofotogrammetriskt system (RSA).). Nio patienter försågs i samband med protesoperation med intra corticala tantalum kulor i under- och lårben. I samband med mättillfället applicerades reflekterande hudmarkörer enligt den modell som används kliniskt. Mätningar av knäextension utförs simultant med de två systemen. Resultaten visar på en god överensstämmelse vad avser extension/flexion på individuell och gruppnivå. Beträffande abduktion/adduktion samt inåt-/utåtrotation ses betydande skillnader mellan de två systemen.

I de två följande arbetena, arbete fyra respektive fem, sätts fokus på hur tekniken fungerar ur ett kliniskt perspektiv där tekniken appliceras för klinisk utvärdering. I studie fyra undersöks muskelstyrka samt gångförmåga med avseende på moment- och effektuttag hos20 patienter med cerebral pares med en matchad kontrollgrupp på 20 friska barn. Resultaten visar på att gruppen med cerebral pares var svagare i alla muskelgrupperna i de nedre extremiteterna samt att de gick med lägre hastighet. Momentuttaget vid gång var närmare det maximala momentet för gruppen med cerebral pares. Ett

signifikant samband mellan plantarflekterande momentet och styrkan i sex av de åtta undersökta muskelgrupperna kunde påvisas i gruppen med cerebral pares. Ett än tydligare samband (rho=0.58-0.76) kunde konstateras mellan genererande effekt och muskelstyrkan för alla åtta muskelgrupper. I den femte och avslutande studien undersöks gångmönstret med avseende rörligheten i bäcken och höft på 19 transfemoralt amputerade. Data från gång med hylsprotes (pre-op) jämförs med data från gång med osseointegrerad protes (2 år post-op). Som kontrollgrupp användes 57 köns och åldersmatchade friska individer. Gång med osseointegrerad protes visar vid 2-års uppföljning på signifikanta förbättringar i ökad höft extension och minskad anterior tilt av bäckenet.

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