Motion Analysis and Postural Stability of Transtibial Prosthesis Users

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To Edward Alexander Maltby and Gertrude Rusaw

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ABSTRACT

The AIMS of the thesis were to critically evaluate motion analysis methods used during investigations of transtibial prosthesis users, and to propose improvements to these methods. Additionally, the aim was to evaluate if vibratory feedback could be used to improve postural stability in transtibial prosthesis users and how being a prosthesis user influenced muscular response to postural perturbations.

MATERIALS AND METHOD *Study I* systematically analyzed 68 peer-reviewed articles investigating lower-limb kinematics in transtibial prosthesis users. *Study II* evaluated motion of prosthetic feet using a functional joint centre (FJC) method. *Study III* evaluated the influence of a vibratory feedback device on postural stability in 24 transtibial prosthesis users. *Study IV* investigated how the prosthetic limb affected EMG response latency in the prosthetic- and intact-limb of 23 transtibial prosthesis users when compared to a matched able-bodied control group (n=23).

RESULTS *Study I* showed a general low level of evidence and low quality in the studies under review and that there were methodological problems which made comparison of studies difficult. *Study II* found that sagittal position of FJCs for prosthetic feet were different between types of prosthetic feet as well as compared to an intact ankle. *Study III* showed vibratory feedback based on pressure under the prosthetic foot caused increased deviations of the centre of pressure in the mediolateral direction, and decreased reaction times in fast voluntary movements of the centre of gravity. *Study IV* showed the EMG response latencies of transtibial prosthesis users were increased in both the intact limb and the prosthetic limb. Increased latencies were found in the contralateral limb when the perturbation was received through the prosthesis.

CONCLUSIONS Methodological issues make interpretation of kinematics of transtibial prosthetic users difficult and motion of the prosthetic foot is not the same in different designs of prosthetic feet or compared to an intact limb. Vibratory feedback can be used to improve some aspects of postural stability, and automatic postural responses are slower in transtibial prosthesis users than in able-bodied controls. These findings contribute to the understanding of how researchers model motion of transtibial prosthesis users and how this group maintains postural stability with a prosthesis.

Keywords: Artificial limb, Balance, Electromyography (EMG), Motion analysis, Postural stability.

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LIST OF PAPERS

This thesis is based on the following studies, referred to in the text by their Roman numerals (I-IV).

Study I

Motion-analysis studies of transtibial prosthesis-users: a systematic review.
Rusaw D., Ramstrand N.
Prosthetics and Orthotics International, 2011, 35(1), 8-19.

Study II

Sagittal plane position of the functional joint centre of prosthetic foot-ankle mechanisms.
Rusaw D., Ramstrand N.
Clinical Biomechanics, 2010, 25(7), 713-720.

Study III

Can vibratory feedback be used to improve postural stability in persons with transtibial limb loss?
Rusaw D., Hagberg K., Nolan L., Ramstrand N.
Submitted

Study IV

The contribution of the prosthesis and weight-bearing on EMG response latency following platform perturbation in transtibial prosthesis users.
Rusaw D., Hagberg K., Nolan L., Ramstrand N.
In manuscript

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1 ABBREVIATIONS

AMP_{AP}	Anteroposterior Sway Amplitude
AMP_{ML}	Mediolateral Sway Amplitude
AP	Anteroposterior
APR	Automatic Postural Response
AV	On-axis Velocity
BoS	Base of Support
CoG	Centre of Gravity
CoM	Centre of Mass
CoP	Centre of Pressure
DCL	Directional Control
EMG	Electromyography
FHA	Finite Helical Axis
FJC	Functional Joint Centre
GRF	Ground Reaction Force
IC	Initial Contact
ICR	Instant Centre of Rotation
LoS	Limits of Stability
ME	Maximum Excursion
ML	Mediolateral
MV_{AP}	Mean Anteroposterior Velocity
MV_{ML}	Mean Mediolateral Velocity

PPS	Path length per second
RMS_{AP}	Anteroposterior Root-mean-square
RMS_{ML}	Mediolateral Root-mean-square
RoM	Range of Motion
RT	Reaction Time
RWS	Rhythmic Weight Shift
SB	Standing Balance
SD	Standard deviation
SR	Stretch reflex
SSR	Support Surface Rotation
TO	Toe-off
TTA	Transtibial amputation

2 DEFINITIONS IN SHORT

Automatic Postural Response (APR)	The unconscious muscular response ($\approx \geq$ 100 milliseconds (ms)) to a sudden movement of the support surface, or other sufficiently large postural perturbation.
Balance	The relationship of the body's centre of mass (CoM) to the base of support (BoS). A state of *unbalance* would be one where the CoM is outside of the BoS. The measure of state of *balance* can be assessed using many tests of postural stability.
Base of Support (BoS)	The area contained within the perimeter of contact and the support surface.
Centre of Gravity (CoG)	The vertical position of the centre of mass.
Centre of Mass (CoM)	The net three-dimensional position of the weighted average of all mass segments in a body.
Centre of Pressure (CoP)	The calculated mean bi-planar position of all vertical forces applied to the top surface of a forceplate.
EMG Onset Latency	The length of time for a muscular reaction to reach a predetermined threshold.
Feedback	Describes a scenario where, within a closed-loop system, results from an elicited control signal are used to influence a future output.
Forceplate	A tool consisting of multiple force transducers used to measure net forces and locations of objects on the forceplate.

Functional Joint Centre (FJC)	A joint location used in motion analysis which is analytically determined from previously captured motion data.
Ground Reaction Force (GRF)	The vector sum of the individual x-,y-,z-components of all the forces applied by an object to the surface of a forceplate. The origin of the GRF is the CoP.
Initial Contact	The first instance of contact of a foot against the support surface during walking. Is normally made with the heel, but in pathological gait can be with other parts of the foot.
Instant Centre of Rotation (ICR)	The calculated 2-dimensional centre of rotation at any point in time. Requires knowing the position of two segments in relation to each other at two subsequent points in time.
Kinematics	The area of mechanics which describes the translations and rotations of bodies without description of the forces or moments producing movements.
Limits of Stability (LoS)	The maximum distance a person is able to shift their CoG from a central position without falling or shifting foot position.
Marker	The basic building block of motion analysis. These are the objects attached to body segments and/or joints in order to describe the position of the object in relation to some previously determined frame of reference. These markers can be active or passive.
Motion Analysis	The field of study which focuses on describing/analyzing how things move.

Postural Stability	The dynamic process which monitors and maintains upright stance. The process of not falling. The term used to describe the relative stability of a person in an upright position.
Postural Perturbation	An externally applied challenge to a postural task. Can include physical, cognitive, optical, vestibular, or pharmacological perturbations.
Stretch Response	The unconscious muscular response (≈ 30-50 ms) to a sudden movement of the support surface, or other sufficiently large postural perturbation. Elicited by external stretch stimuli.
Surface Electromyography	The area of physiology and/or biomechanics measuring muscular/electrical phenomena without breaking the skin barrier.
Toe-off	The last instance of contact of a foot before it begins the swing-phase of gait.
Transtibial Amputation	An amputation which bisects the tibia. Can be due to trauma or disease. Results in the total removal of the ankle, but leaves some remnant of the tibia.
Vibratory Tactor	A device used to convert electrical charge via a controller into a mechanical vibration.

3 INTRODUCTION

Individuals with a unilateral transtibial amputation (TTA) have had a complete removal of the anatomical ankle. This lack of an ankle joint presents many challenges in physical function as they must conduct the same tasks as able-bodied individuals, but with a prosthesis. Although advances in prosthetic technology mean that transtibial prosthetic users can perform many of the activities able-bodied individuals are able to, they must compensate as a result of the prosthetic limb.

As part of the process of improving performance researchers are often interested in quantifying physical function of prosthetic users. One common method used to evaluate physical function as it relates to physical movement is three-dimensional motion analysis. The first two studies in this thesis have dealt specifically with how researchers use motion analysis in studies of transtibial prosthesis users. *Study I* systematically reviewed motion analysis methods used in studies involving transtibial prosthetic users and provided recommendations for future improvement. *Study II* specifically evaluated how a prosthetic foot/ankle moves if we use the same constraints as those that are used on an intact ankle in motion analysis.

Studies *III* and *IV* further investigated physical function of transtibial prosthetic users by evaluating postural stability. *Study III* evaluated the effectiveness of a feedback device to improve various measures of postural stability. *Study IV* explored the muscular response to support surface perturbation in individuals with a unilateral TTA.

The following thesis summarizes these four studies and presents results which contribute the understanding of what methods researchers use in motion of transtibial prosthetic users, and the potential problems of this method when used on a prosthetic foot/ankle mechanism. The results also reveal how the prosthetic limb can influence postural stability in these same individuals.

4 BACKGROUND

4.1 Motion analysis

In a clinical or research setting, motion analysis often refers to the study of motion of the human body. This can be accomplished using many different technologies. In the context of this thesis *motion analysis* refers to stereophotogrammetry [1], in which multiple video cameras capture the motion of markers placed on an individual whilst a motor task is conducted. By using a number of cameras it is possible to analytically determine three-dimensional position of markers based on the two-dimensional coordinates provided by individual cameras. This coordinate data is then used individually, or combined with further variables (such as kinetics and electromyography — EMG) to make clinical decisions regarding:

- a diagnosis of disease
- assessment of disease severity
- the progress of an intervention
- prediction of the outcome of an intervention [2].

4.1.1 System and random error

As the goal of motion-analysis is model motion of the muskuloskeletal system, it is important to recognize there are relevant sources of error inherent to the process. A thorough description of the sources of error has been described elsewhere [2-4]. These can be classified as *random error* and *systematic error*. The random error is confined to high-frequencies and is typically caused by electrical interference, ambient lighting conditions which can cause inaccuracy when converting the video images to numerical marker points [4]. Random errors are typically dealt with by using appropriate filtering techniques discussed later. Systematic errors can result from optical distortion of camera lenses, improper calibration of capture volume, improper placement of cameras, or other variables not considered random in nature. Systematic errors are reduced by using factory calibrated cameras, proper calibration techniques and appropriate lab set-up [4].

4.1.2 Marker placement

In order to model three-dimensional human movement researchers must first record the three dimensional position of markers placed on the body. Markers used can be active (powered transmitter) or passive (reflective). They can be placed directly on the skin with double sided tape or attached as rigid clusters of markers on a backing plate which is subsequently fixed to the body using elastic or velcro (Figure 1). Once marker position has been established in three dimensional space, the next step is to define body segments and to define where the joints, connections between these segments, are located (Figure 2). As the movement of interest is actually that of the skeletal structures within the body, and it is not always possible to directly mount markers to the skeleton, it is necessary to model the motion utilizing movements from the surface of the body. For example, markers could be on the skin, clothing or, in the case of many orthopaedic applications, on a device such as a prosthetic limb.

Figure 1 – Cluster-sets of reflective markers. Image courtesy of: Qualisys AB, Sweden.

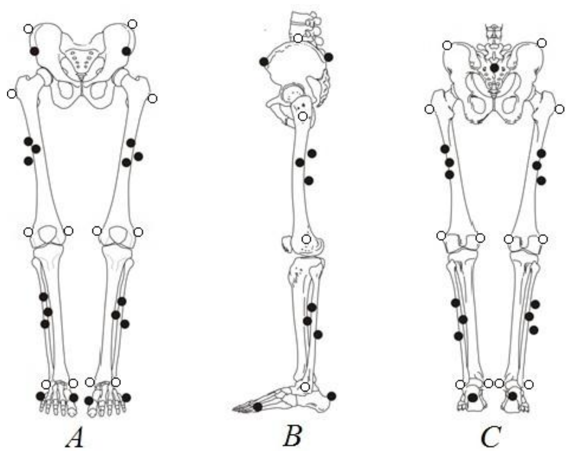


Figure 2 – An example of marker placements for defining joints and segments (white markers) and tracking the motion of segments (black markers) of the lower extremity. Based on the biomechanical model defined by Capozzo et al. [5].

4.1.3 Biomechanical models

When modeling motion, researchers and clinicians must apply a biomechanical model to be used for the calculation of the variables of interest (joint kinematics, temporospatial parameters, etc.). Biomechanical models are the means by which motion of the markers are given meaning. By defining a biomechanical model researchers define where limb segments (foot, shank, thigh, etc.)and joints between these segments exist, the motions that can be elicited (2D vs. 3D-motion), and the degrees-of-freedom each segment is able to move in (translation, rotation, translation and rotation). There are many biomechanical models which have been validated in the literature [5-8]. They all have their strengths and weaknesses, depending on the purpose of the research [9]. The validity of each model is measured by how well it matches the true motion of the segments involved. This is not always a simple feat, particularly in instances where multiple joints are present within a predefined segment. The foot and shank, for example, are typically defined as two connected rigid segments. In reality the foot contains 26 bones and 33 subsequent joints while the shank consists of 2 bones (Tibia and Fibula) which do not move as a rigid segment. This means that there is often incongruence between the biomechanical model and reality. There are also other sources of deformation which violate the so called *rigid-segment-model*. There is motion of the soft tissue over the segments and joints, such that motion of the skeleton is not reflected by motion of the skin overlying it [10, 11] in addition to equipment based error inherent in the motion analysis systems [4]. Multi-segment models of the foot have been proposed, both for an intact foot [12] as well as in one investigation of prosthetic feet [13]. These efforts help to reduce the incongruence between the model and reality, though there still remain several sources of error that must be considered when using these methods in practice. If researchers are interested in defining the foot and ankle as a series of connected rigid segments, it is important to understand the effect of the difference between the model and reality.

4.1.4 Processing of data

Once capturing of motion data is completed, meaningful information must be extracted from it. This process of extraction involves filtering the raw data of unwanted signals, processing the data to extract variables of interest (joint angles, temporospatial parameters, etc.), and interpreting the results [2].

The first step in processing the data involves filtering the raw data. Within the raw data there are many sources of random error which filtering is used to attenuate. These include the amount of ambient light (in the case of reflective markers) and electrical interference. These *noise* components are confined to the high end of the frequency spectrum in the raw signal. Filtering of this high frequency noise from the relevant motion data contained in the low-frequency content is accomplished using of a low-pass filter [4]. The low-pass cutoff frequency is dependent on factors such as the activity being performed, where on the body the marker is located and the environmental conditions of the laboratory (electrical interference, light, etc.). Although frequency content changes for markers placed on different location of the body, frequency analysis has shown that the relevant motion data is confined to frequencies below 10 Hz [14].

With filtered data the processing which extracts meaningful information about the movement captured can begin. This can include, but is not limited to, the joint kinematics (angular-position, -acceleration and -velocity) and temporospatial parameters (gait velocity, step/stride length, etc.). From this information it is possible to draw conclusions about the individual's, or group of individuals', movement.

4.2 Challenges of motion analysis in individuals with transtibial amputation

4.2.1 Transtibial amputation and prostheses

Transtibial amputation refers to the surgical or traumatic removal of the foot and ankle, leaving some tibial-remnant. The intact knee anatomically and functionally separates a TTA from a more proximal amputation level such as knee-disarticulation or transfemoral amputation. The overall incidence of lower-limb amputation (all amputation distal to the pelvis) rates vary greatly between countries and regions with Europe, with reports between 16 and 34 cases per 100,000 inhabitants [15, 16]. The proportion of TTA of all lower-limb amputation has been reported to be between 28 and 74% depending on the cause of amputation and the region of the publication [15-19].

The amputation rates and the rates of those who have been successfully fitted with a prosthesis differ greatly. If the cause of amputation is due to disease, successful fitting can be expected in between 50-65% of cases [20] [21], while in those individuals who have had an amputation due to trauma, the likelihood of a functional recovery is higher [22].

A transtibial prosthesis is typically constructed of a number of common components (Figure 3).

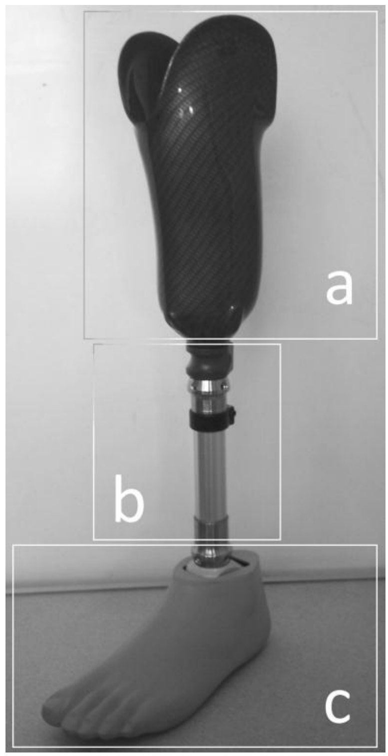


Figure 3 – A transtibial prosthesis where (a) is the prosthetic socket, (b) is the pylon, and (c) is the prosthetic foot. Prosthetic components by Otto Bock , GmbH (Duderstadt, Germany)

The prosthetic socket is the main component to which a prosthetist has influence over the design. This is the main structural interface between the residual limb and prosthesis with forces being transmitted between the prosthetic limb and socket via this interface [23]. The socket can be made of different materials including plastic and various forms of fibre-composite (carbon-fibre, glass-fibre, etc.).

The structural link between the socket and the prosthetic foot is the prosthetic pylon. This component can be rigid, or dynamic offering both rotational and translational shock absorption [24].

There are many different designs of prosthetic feet available commercially and classification of these feet can be difficult. This is due to the fact that classification based on a *structural* criteria can belong to multiple groups based on a *functional* criteria. The classification system used in this thesis is that proposed by Hafner et al.[25] in which there are four main classifications for prosthetic feet. These classifications are: conventional (CV), single-axis (SA), multi-axis (MA) and energy-storing-and-response (ESAR). Many modern prosthetic foot/ankle mechanisms are constructed from either a foam/plastic inner mass of varying densities (in the case of a SACH foot or other CV-foot) or a fibre-composite spring and a shell (as in an ESAR-foot) (Figure 10). The prosthetic feet may have a cosmetic cover for the foot componentry or provide the foot shape as an integral part of the foot construction. Prosthetic foot/ankle complexes do not necessarily contain a joint in the sense of a rigid ball-and-socket or fixed-axis joint commonly used in motion analysis models. Therefore describing the position of the joint required for biomechanical modeling can be difficult (Figure 4). A rigid-segment model used to describe an intact limb (itself subject to errors) may be even less appropriate for a prosthetic foot, which may not have a defined joint. Additionally, there are many different types of prosthetic feet and direct comparison of one to another may also be inappropriate.

4.2.2 Marker placement on prosthesis

When conducting instrumented gait analysis of prosthesis users it is common to position the markers on the prosthetic limb based on the position of the markers of the intact limb. Sometimes this has been made through approximation [26, 27] and sometimes through a direct measurement from the remaining foot [28, 29]. This creates a source of error at both the knee and the ankle. As the prosthetic socket proximally in many cases prevents the attachment of reflective markers directly to the skin, it is necessary to attach markers to the outside of the prosthetic socket (Figure 4). As there is a degree of relative movement between the prosthetic socket and the residual limb, the recorded three-dimensional movement does not necessarily reflect the true motion and presents an additional source of error [30]. In addition to the prosthetic socket, there are problems associated with determining marker placement and joint position on the prosthetic foot based on the anatomy of the intact foot. The markers on the foot assume motion of the prosthetic foot will closely match that of the intact foot. It is not known if this is true.

Figure 4 - Reflective marker set-up for a prosthetic limb seen from the anterior direction.

A common method for determining joint location in motion analysis has been to locate the joint centre based on anatomical landmarks [7, 31]. In the case of the ankle joint this would result in an ankle joint located at a midpoint between the markers placed on the medial and lateral malleoli. For the above reasons this method may not be sufficient for a prosthetic foot as the actual joint centre could be in a different location. Other efforts to determine a joint centre based on actual motion of two segments in relation to each other have been made [32-35]. These results have been encouraging as they describe the joint centre of rotation based on actual motion and not on assumptions based on marker locations. However, the methods are sensitive to rigid-body assumptions, noise in the data and the RoM used in determination of the joint centre. While many methods perform well when the RoM is large (approximately 45 degrees), a smaller number have been shown to have acceptable accuracy at ranges of around 20 degrees [36]. One method which satisfies this accuracy at reduced RoM is the functional joint centre (FJC) method as proposed by Schwartz et al. [35].

An understanding of the methods researchers have used in describing kinematics of transtibial prosthesis users might identify possible shortcomings and/or strengths of the methods for future research. A better understanding of how a prosthetic foot moves, if rigid segment theory is applied to the movement, might accommodate for any systematic error in the calculations. The FJC method represents a promising method to evaluate the motion of a prosthetic foot.

4.3 Laboratory based assessment of postural stability

4.3.1 Definitions of postural stability

Postural stability, in the context of this thesis, is defined as the measure of how capable a person is in *not falling*. This definition encompasses many different mechanisms depending on the postural task and the way in which postural stability is quantified. In investigations of quiet standing a very common method of evaluating postural stability is to quantify motion of the centre of pressure (CoP) and extract various measures from this motion [37]. If the postural task is more challenging (one that actively attempts to cause the participant to lose stability), it can be more useful to look at postural adaptations and muscular response to perturbations, via EMG analysis [38].

4.3.2 Forceplates and postural stability based on the centre of pressure

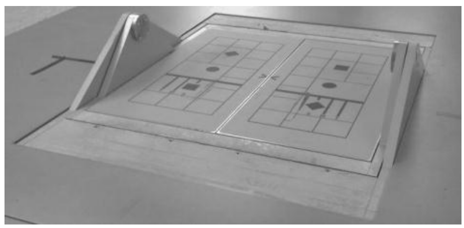


Figure 5 - Forceplate commonly used in assessment of postural stability.

One of the most common objective analyses of postural stability involves the use of forceplates (Figure 5). A forceplate provides electrical voltage output from force-transducers (through the use of strain-gauges or piezoelectric crystals) typically located under the platform. These are used to calculate forces exerted on the surface of the platform (Figure 5). Depending on the design of the platform the resultant forces can be separated by their component forces (x-y-z) and expressed individually or combined to describe the force vector in three-dimensions. They describe the mediolateral, anteroposterior and inferosuperior forces exerted by a person or object on the forceplate. In some cases it is only important to export the z-component of the forces exerted on the forceplate. The z-component component is required in order to extract CoP information. In situations where the mass applied to the forceplate is sufficient and proper calibration has been carried out, the CoP is the origin of the ground-reaction-force (GRF) vector and has an origin on the support surface. A common method of analysis of postural stability involves extracting information about the motion of the CoP. In quiet standing this can be calculated using only the vertical force (z-component) applied to the platform via four force-transducers and (z-component) and two moments arms [39, 40]. A common clinically relevant question is how motion of the CoP can be used to identify the risk a person has of falling in the future [41-44]. To these ends various measures have been able to identify those individuals who are at risk of falling. Stability in the mediolateral plane (root-mean-square (RMS) of CoP excursion, mean mediolateral velocity of the CoP (MV_{ML}), mean amplitude of mediolateral excursion (AMP_{ML}), and mean velocity of CoP (MV) have been linked to increased fall risk [41-44]. However, all these investigations were on individuals without lower-limb amputation so the conclusions cannot be directly applied to prosthetic users.

4.4 Postural stability in individuals with TTA

In upright posture the body has been shown to behave like an inverted pendulum [45, 46]. Some have argued that this may be an oversimplification as it misses significant contributions from the hip and knee [47], while others suggest the model is valid [48]. The inverted pendulum model states that the largest controlling factor for keeping the body upright is the ankle. By definition this is called the ankle strategy [49]. As the motion of the centre of gravity (CoG) moves anteroposteriorly the ankle is the major control factor acting to bring the CoG back into a position of stability in quiet stance (Figure 6). When the ankle strategy is insufficient to maintain postural stability there is an increased reliance on what is called the hip strategy [50, 51]. This strategy states that a greater proportion of maintenance of postural control is coming from the hip, and not the ankle. Transtibial prostheses users lack an anatomical ankle, including all sensorimotor structures, and are subsequently unable to maintain postural stability with an ankle strategy on the prosthetic side. To maintain postural stability they must therefore compensate using the remaining structures and a modified postural strategy, with a greater hip strategy component.

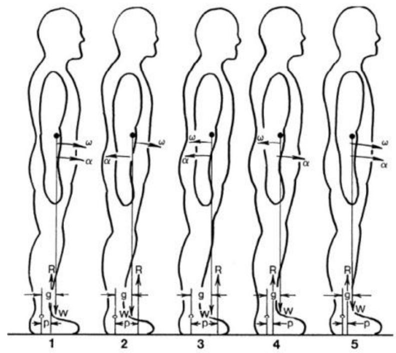


Figure 6 - Postural stability in the sagittal plane can be modelled using an inverted pendulum. Image modified from: Winter et al. [46].

It is well known that lower limb prosthesis users in general have challenges in their ability to maintain postural stability [52, 53]. Studies have reported decreased balance confidence [54-56] and falling more frequently [57, 58]. Some clinical outcome measures have been useful in identifying prosthesis users who will fall [59]. Though, most understanding regarding postural stability of individuals with amputation comes from laboratory based outcome measures.

4.4.1 Centre of pressure and postural stability in transtibial prosthesis users

It has been shown that prosthesis users perform worse than able-bodied individuals in postural tasks, or investigations which evaluate postural stability [60-75]. These investigations have shown that unilateral transtibial prosthesis users load their intact limb more than their prosthetic limb and that the anteroposterior (AP) motion of the CoP under the prosthetic foot is smaller in magnitude than under the intact foot [71-73, 75]. Prosthesis users have increased excursion of the CoP in both the mediolateral (ML) and AP directions [61], and increased root-mean-square (RMS) of the ML and AP velocity of the CoP [64]. When the postural task becomes more challenging (for instance by standing on a moving platform), prosthesis users have increased measures of instability and excursion in the AP direction when compared to able-bodied controls [61, 75]. Those with amputation due to vascular disease have increased AP and ML excursion [66], though more recent studies have found this increase disappears as the users become more skilled with their prosthesis [68]. To maintain postural stability prosthesis users rely more on vision than able-bodied controls [62]. However, this reliance has been shown to diminish with time from amputation [63, 65] and to be influenced by the amount of attention the person can give to the balance task [64].

There have been a number of investigations involving EMG in relation to postural adaptations. These have shown that for transtibial prosthesis users a shift in the ML direction in order to lift one leg causes an earlier burst of more proximal muscles (tensor-fascia-latae) [69, 74]. Aruin et al. has shown that in response to catching a falling ball prosthesis users had increased activity of the muscles on the intact side of the body indicating a postural adaptation [60].

Individuals with amputation have been shown to have decreased measures of postural stability as defined by motion of the CoP and an altered postural adaptation as shown by EMG responses [60, 69, 74]. With prosthetic users it is possible that altered EMG responses are a passive mechanism due to mechanical constraints of the prosthesis (inefficient movement). It could also be that there is a sensorimotor interaction which is contributing. For instance, decreased sensory feedback from the side with a prosthesis could be such a contributing factor. In a study which subjected unilateral transtibial prosthetic users to a *tether-release* evoked fall, and in which recovery required a step (defined as time to toe-off), it was shown they responded slower when the step was lead with the intact limb (prosthetic foot remained on support surface) [76]. In the same study, the authors state that the pooled-data indicated the TTA-group responded faster (regardless of side) than the matched control-group. As the response time was determined using kinematics, simultaneous EMG data during this study could have helped to further explain the differences between the groups.

4.4.2 Feedback utilized by prosthesis users

As the ankle contains sensory and motor structures that contribute to postural stability, it is clear that a prosthetic user has significant limitations not faced by able-bodied individuals. Mouchnino et al. [70] suggested that at least a portion of the postural reorganization that prosthesis users have after the limb loss is the result of decreased feedback from the affected limb. They proposed this feedback mechanism to be the pressure sensed on the supporting foot, and how this is used to orient the centre of mass (CoM) and determine an appropriate position after the proposed movement. This has been supported by Isakov et al. [67] who proposed the reduction of postural stability is directly related to the inability of the prosthesis user to access proprioceptive information from the affected limb. Lower-limb sensitivity, specifically poor vibration sense, has been shown to be a strong indicator of previous falls and increased AP excursion of the CoP in transtibial prosthesis users [77].

4.4.3 Mechanical stability of a prosthetic foot

It is thought that mechanical characteristics can play a role in how stable a prosthetic foot is in gait, though this has not been shown or proposed as a mechanism in postural stability or quiet stance. One theory [21, 22] states that a rigid prosthetic forefoot keel provides an external torque to the knee joint which acts to keep it stable. In this theory the stability of the knee is relying less on the internal torque provided by the knee muscle extensors. A second theory suggests that stability is facilitated by the prosthetic foot's ability to accommodate to uneven surfaces by maintaining contact with the floor for a longer period of time [23]. This theory was supported by Hafner et al. [24] who suggested that the perception of stability is influenced by the ability to extend the amount of time spent in mid-stance without heel off. A recent study specifically investigated how the stiffness of the prosthetic foot influenced dynamic balance control, defined as the ratio of ankle torques between the intact and prosthetic limb in response to CoM movement [78]. The results showed a positive correlation between increasing stiffness of the prosthetic foot and dynamic balance control.

4.4.4 Sensory feedback used to improve postural stability

Efforts with other groups of patients to supplement sensory information to individuals with poor postural stability have been encouraging. Vibratory feedback applied to the trunk has been shown to reduce measures of instability (RMS of CoP excursion, RMS of body tilt) in persons with reduced vestibular function [79, 80] and in a healthy sample [81]. Because in quiet stance the body behaves as an inverted pendulum [45, 46] it is possible the shifts of the CoP could be an equally beneficial source as the trunk tilt information. Sienko et al. [79] found the CoP excursion results "mirrored" the trunk tilt results in a sample of persons with reduced vestibular function.

4.4.5 Sensory feedback used in transtibial prosthesis users

Investigations have been conducted to try to supplement the missing afferent information in prosthetic users with another feedback modality. To the author's knowledge, these investigations have uniformly chosen weight-distribution in quiet stance as the chosen outcome variable to assess the efficacy of sensory feedback. The results have indicated that weight-distribution and gait symmetry can be improved by utilizing interventions applied unilaterally on the prosthetic side. Published studies have included the use of electrical feedback [82], vibratory post-effects [83, 84] and feedback via pneumatic air-balloons [85, 86]. Lee et al. [87] also showed that unilaterally applied sub-sensory stochastic stimulation improved measures of quiet standing balance. To date it is unknown if sensory feedback can be used to improve postural stability, as defined by motion of the CoP, in transtibial prosthesis users.

4.5 Response to support surface perturbations

Postural stability can be investigated by rapidly moving the support surface and investigating how the individual responds to this perturbation. It can involve rapid movements of the support surface through rotation [88-90], translation [91], or rotation and translation [91-93] (Figure 7). The rapid support surface movements elicit muscular responses which then can be classified based on their latency (time to onset) after the perturbation is elicited. The first responses which can occur are reflex responses, due to external stretch stimuli. These occur between ≈ 30-40 milliseconds (ms) after support surface movement [90, 94]. Reflex responses are then followed by the automatic postural response (APR) which starts at ≈ 100 ms [90, 91, 94, 95] and (depending on definition) extends to 180 ms [94], 250 ms [91], or 325 ms [95]. In the case of rotational perturbations, the responses are elicited when the rotation is of sufficient amplitude and velocity (minimum 4 degrees at 50 degrees/second). Commonly, researchers are interested in the EMG response latency to perturbations as this is indicative of the ability to recover to sudden perturbations [96]. Various groups of patients have increased latencies following support surface perturbations including those with peripheral neuropathy, muscular sclerosis, and the elderly [38].

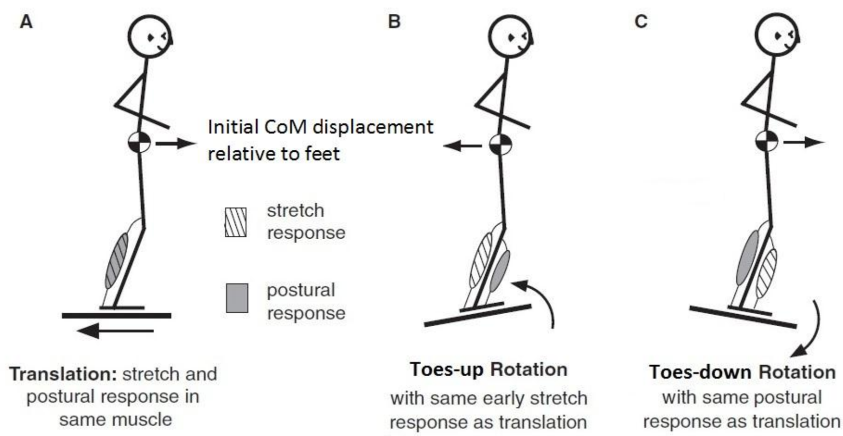


Figure 7 – The organization of earlier responses to platform perturbations based on the type of perturbation. Translational perturbations (A) have similar temporal responses but the organization cannot be determined entirely by what is happening at the ankle. In (B) and (C) the stretch reflexes cannot be used to determine what is happening with the body as the CoM in (A) and (B) are moving in the same direction, but the stretch response is in opposing antagonistic muscles. Image modified from Ting [95].

4.5.1 Contributing factors in eliciting automatic postural responses

The mechanisms which elicit an APR following a platform perturbation are complex. The sensory receptors of the lower-extremity, for instance those in the plantar surface of the foot [97] and ankle [98] contribute to the ability to respond to the perturbation. It is important to note that there are other sensory contributions from more proximal joint levels [93, 99, 100], as well as from other sensory modalities such vestibular and vision [101, 102]. This is referred to as the multi-sensory contribution to postural perturbations (Figure 8). It is this multi-sensory contribution which is received and interpreted at various levels in order to elicit an appropriate response to a perturbation. This is likely the reason certain individuals with reduced distal sensation can elicit similar postural reactions utilizing afferent information from more proximal signals [93, 99, 100]. Apart from the sensory contributions there are also other influential factors including anxiety [104], previous experience [105], attention [106], and joint position [107] which have been shown to influence the APR following support surface rotations.

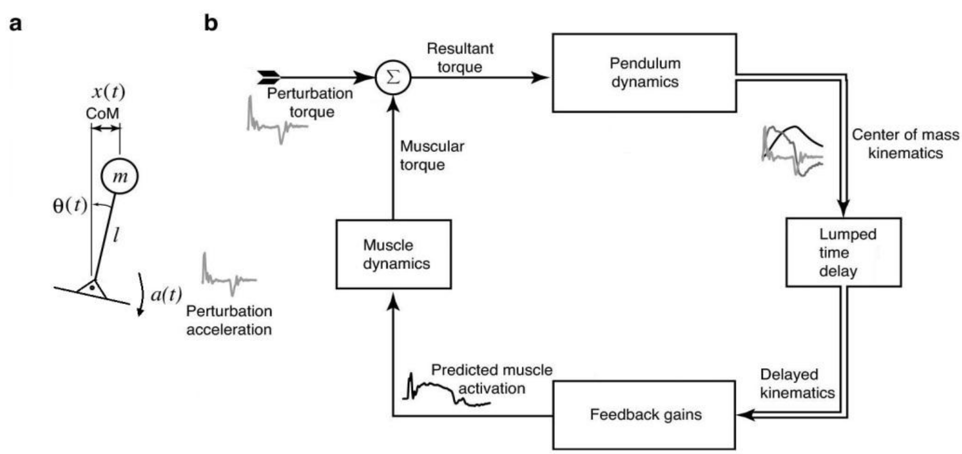


Figure 8 - Simple feedback model showing the relationship between joint torques, coupling delays, CoM motion, and muscular response interact to maintain postural stability following a perturbation. Figure modified from Ting et al. [103]

4.5.2 Potential influences of being a prosthetic user on APRs after platform perturbation

The consequence of having an amputation is a total lack of sensorimotor structures distal to the amputation. This would cause a change in the afferent sensory information and altered motor control. The constraints of the prosthesis (rigidness of the foot, etc.) would also have an effect on the structures of the residual limb (joints and structures proximal to the amputation). This would result in altered sensory information from remaining structures and reduced effectiveness of motor structures attempting to accomplish movement with a reduced lever arm. As individuals who had reduced distal sensation are able to compensate with more proximal structures [93, 99, 100], it is reasonable to assume that transtibial prosthesis users may also be able to compensate in this way. In lateral shifts required to lift one leg, prosthetic users have earlier activation of more proximal muscles [69, 74]. Though, these reactions are volitional and do not give an understanding of the automatic postural response to a perturbation, themselves unconscious.

Considering the movement of the CoM in transtibial prosthesis users, it is possible that a perturbation would give different effects than in persons with an intact ankle. The motion of the CoM is the major mediating factor in which muscles become active following a perturbation [97, 100]. An able-bodied individual is able to dorsiflex the ankle following a toes-up rotation, something prosthesis users are less able to accomplish. Therefore, it is reasonable to assume that in prosthetic users the CoM would have an altered excursion as a result of the platform rotation. Similarly one can postulate how this would affect response in a toes-down direction.

It is known that postural adaptations result as a consequence of an amputation and that these result in altered (non-symmetric) weight-bearing distributions in transtibial prosthesis users [67, 72, 73]. These postural adaptations not only result in altered position of the CoM but also in the load-tension relationship of remaining musculature and tissues. Currently there is a lack of knowledge about how a TTA might affect automatic postural responses following support surface rotations when compared to able-bodied individuals. There is a need of better understanding in how transtibial prosthesis users compensate for the limb loss and integrate their prosthetic limb into a sensorimotor response to a platform perturbation.

5 PROBLEM AREAS

Study I

Little is known about what methods researchers are using when investigating kinematics of transtibial prosthesis users. A systematic review of the methods used to capture, calculate, and report kinematic variables would help to identify limitations and strengths of the methods chosen.

Study II

Researchers commonly model kinematic motion of prosthetic feet based on the assumption that they move in the same fashion as an intact ankle. It is not known how differently prosthetic foot/ankles move if the same modeling techniques are used for them as those on intact ankles.

Study III

Prosthesis users have decreased values of postural stability. Vibratory feedback relaying information of postural orientation has been shown to improve postural stability in some patient groups. It is not currently known whether similar feedback can improve postural stability in persons with TTA.

Study IV

Delayed EMG response latency increases the risk for falls and fall related injury. A prosthetic limb is likely to influence a person's EMG response latency to rapid movements of the support surface. Currently we do not know how using a prosthetic limb affects this EMG response.

6 AIMS

Study I

To critically examine the methods and techniques used by researchers in collecting and reporting three-dimensional kinematic data related to transtibial prosthetic users, including the independent and dependent variables utilized. To propose recommendations for future direction of research in this area.

Study II

To identify the functional joint centre (FJC) of a selection of commonly used prosthetic feet. Analysis will determine if the FJCs of the prosthetic feet differ from the FJC of an intact control foot. Additionally, analysis will compare how the FJC method compares with the commonly used method of estimating joint parameters based on the intact side (anatomical method).

Study III

To evaluate the effects of a vibratory feedback system on static and dynamic balance in persons with unilateral transtibial limb loss.

Study IV

To understand how weight-bearing and limb-position affect EMG response latency of transtibial prosthesis users. Analysis will investigate how the intact- and affected-limb differ when subjected to support surface rotations in the pitch plane.

7 RECRUITMENT AND PARTICIPANTS

Participant characteristics for the studies involving human testing are listed in detail in Table 1. The TTA-group (24 individuals) were recruited using the following inclusion criteria (studies *II*, *III* and *IV*):

- individuals who had unilateral TTA
- primary cause of amputation not due to diabetes or peripheral vascular disease
- no current concomitant health issues (including problems with residual limb or neurological disease)
- no problems regarding fit or function of their current prosthesis
- had been a regular prosthetic user for at least one year

TTA-group participants were recruited in one of two ways:

1. from a participant database at the School of Health Sciences, Jönköping University. This database provided seven individuals in the TTA-group.
2. from 4 prosthetic clinics in southern Sweden (Jönköping, Borås, Gothenburg, and Kungsbacka). Clinics provided the remaining 17 individuals in the TTA-group.

For those participants recruited through prosthetic clinics, first contact was made through the prosthetist currently working with the patient. Follow-up contact by the author was made only after approval of the patient.
Participants in the matched control-group were recruited among staff at The Lundberg Laboratory for Orthopaedic Research at Sahlgrenska Academy in Gothenburg, The School of Health Sciences at Jönköping University, and friends/family of the staff at these institutions.

Table 1 - Summary of participant characteristics for studies *II*, *III*, and *IV*. TTA-group characteristics for *Study II* (dark shaded row), *Study III* (all participants in TTA-group), and *Study IV* (all except last participant in TTA-group and matched Control-group participants). Summary statistics [mean and (SD)] given for years since amputation (years), height (m), mass (kg), and age (years).

TTA-Group															Control-Group			
Sex	Cause of amputation	Years since amp	Height (m)	Mass (kg)	Age (years)	Residual Limb Length	Socket	Liner	Suspension	Foot	Dominant (L/R)	Side Amp (L/R)	Tobacco User	Phantom Sensation	Sex	Height (m)	Mass (kg)	Age (years)
M	Trauma	8	1.78	74	46	Ordinary	Hard	Gel	Pin	Flexfoot	L	R	N	Y	M	1.78	84	52
M	Infection	5	1.8	67	27	Ordinary	Hard	Gel	Pin	Flexfoot	R	R	N	N	M	1.8	64	38
M	Trauma	4	1.9	78	40	Ordinary	Hard	Gel	Pin	Flexfoot	L	R	Y	Y	M	1.93	78	40
M	Trauma	12	1.84	97	65	Long	Hard	Gel	Pin	Flexfoot	R	R	N	N	M	1.84	108	63
M	Trauma	3	1.86	82	53	Ordinary	Hard	Gel	Pin	Reflex VSP	L	L	Y	N	M	1.86	92	51
M	Trauma	3	1.79	68	65	Long	Hard	Gel	Pin	C.P. Trustep	R	R	N	N	M	1.79	76	62
F	Trauma	33	1.59	64	60	Long	Hard	Gel	Pin	Elation	R	R	N	Y	F	1.62	62	52
M	Trauma	2	1.78	87	51	Ordinary	Hard	Gel	Sleeve	Flexfoot	L	R	N	N	M	1.75	85	52
M	Trauma	19	1.8	83	33	Ordinary	Hard	Gel	Pin	C.P. Trustep	R	R	N	N	M	1.74	72	31
M	Trauma	34	1.78	85	47	Ordinary	Hard	Gel	Pin	Flexfoot	R	R	N	Y	M	1.84	85	43
M	Trauma	3	1.78	88	72	Long	Hard	Gel	Pin	Flexfoot	L	L	Y	N	M	1.75	78	72
M	Trauma	44	1.8	91	63	Ordinary	Hard	Gel	Pin	Flexfoot	R	R	N	N	M	1.81	84	63
M	Trauma	5	1.7	76	49	Ordinary	Hard	Gel	Pin	Sureflex	L	R	Y	Y	M	1.7	76	49
F	Osteosarcoma	28	1.71	73	47	Long	Hard	Gel	Sleeve	C.P. Trustep	L	R	N	Y	F	1.68	66	47
M	Trauma	12	1.8	112	26	Long	Hard	Gel	Pin	Single Axis	L	R	N	Y	M	1.85	102	34
M	Trauma	21	1.78	81	37	Long	Hard	Gel	Pin	Renegade	L	R	N	N	M	1.75	80	37
F	Congenital	25	1.54	64	25	Long	Hard	Gel	Sleeve	Flexfoot	R	R	N	N	F	1.6	53	24
M	Trauma	13	1.8	78	60	Ordinary	Hard	Gel	Vacuum	Renegade	L	R	N	N	M	1.82	87	59
F	Infection	5	1.82	57	30	Ordinary	Hard	Gel	Pin	Elation	R	R	N	N	F	1.8	70	27
F	Congenital	45	1.64	64	45	Long	Hard	Gel	Pin	Elation	R	R	Y	N	F	1.64	65	42
M	Thrombosis	8	1.79	60	56	Long	Hard	Gel	Pin	Flexfoot	L	R	Y	Y	M	1.82	80	59
M	Trauma	3	1.85	88	51	Long	Hard	Gel	Vacuum	Flexfoot	R	R	N	Y	M	1.88	90	53
M	Trauma	19	1.76	101	62	Long	Hard	Gel	Vacuum	Duralite	R	R	N	N	M	1.72	95	59
M	Trauma	4	1.76	100	53	Long	Hard	Gel	Pin	Flexfoot	R	L	Y	Y				
Study 4	*Mean (SD)*	*15.4 (13.6)*	*1.77 (0.08)*	*79.0 (13.8)*	*48.2 (13.8)*										*Mean (SD)*	*1.77 (0.08)*	*79.7 (13.1)*	*48.2 (12.6)*
Study 3	*Mean (SD)*	*14.9 (13.5)*	*1.77 (0.08)*	*79.9 (14.2)*	*48.5 (13.5)*													

Study I

No experimental participants were recruited for the study.

Study II

One participant with TTA (*darkened row in Table 1*) (male, 176 cm, 98 kg, 60 years at time of capture for *Study II*) participated in the study. The participant served as his own control using intact contralateral leg.

Study III

A power calculation using anteroposterior sway amplitude data of the CoP (ΔCOPy) from a previous study [75] established that a minimum sample size of n=24 was required to detect a statistically significant difference ($p<0.05$) between two paired-groups, given a statistical power of 0.8 and a true difference between groups of 1.00 meter/20 sec.

24 participants with TTA (19 male/5 female; mean height: 1.77 m (SD=0.08); mean weight: 79.9 kg (SD=14.2); mean age: 48.5 (SD=13.5) participated in the study.

Study IV

A power calculation using EMG response latency times from a previous study [88] established that a minimum sample size of n=23 was required to detect a statistically significant difference ($p<0.05$) between two paired-groups, given a statistical power of 0.8 and a true difference between groups of 20 milliseconds (ms).

23 participants with TTA (TTA-group) (*all except last row in Table 1*) [(18 male/5 female; mean height: 1.77 m (SD=0.08); mean weight: 79.0 kg (SD=13.8); mean age: 48.5 (SD=13.5)] and matched-group (height × mass × age) of 23 control participants (Control-group) [(18 male/5 female; mean height: 1.77 m (SD=0.08); mean weight: 79.7 kg (SD=13.1); mean age: 48.2 (SD=12.6)] participated in the study.

8 METHODS

Study I

A systematic review was conducted in June 2009 on literature published in English between January 1984 and June 2009. The search was within the Cochrane, Medline and Cinahl databases. Inclusion criteria for the search were that the articles must: have employed an experimental research design, collected three-dimensional kinematic data of the lower-extremity, and have transtibial prosthesis users as experimental participants.

Articles which met the inclusion criteria were classified according to level of evidence [108] (Table 2) and quality of study design [109] (Figure 9).

Level of evidence	Type of study
Level I	A. Randomized trial i) Statistically significant difference ii) No statistically significant differences but narrow confidence intervals B. Systematic review of Level I RCTs (homogeneous studies)
Level II	A. Prospective cohort study B. Poor quality RCT (eg.<80% follow-up) C. Systematic review (i) Level II studies (ii) Non-homogeneous Level 1 studies
Level III	A. Case-control studies B. Retrospectived cohort study C. Systematic review of Level III studies
Level IV	A. Case-series studies B. Case-report studies
Level V	Expert opinion

Table 2 – Level of evidence classifications according to Bhandari et al. [108].

1. Purpose clearly stated
2. Relevant literature review conducted
3. Study design appropriate for the study aims
4. No obvious biases present
5. Sample size described in detail
6. Sample size justified
7. Informed consent given
8. Reported using valid outcome measures
9. Reported using reliable outcome measures
10. Intervention described in sufficient detail
11. Results reported with statistics
12. Appropriate statistical analysis
13. Results reported with clinical importance
14. Conclusions are appropriate to aims
15. Clinical implications reported
16. Limitations acknowledged

Figure 9 - Quality criteria used in review according to Law et al. [109]

The three critical analyses focused on: 1) the methods of data capture; 2) the independent variables in the analyses; and 3) the dependent variables researchers were investigating. The variables of interest were quality of the study [108], the level of evidence [109], the number of participants (prosthesis users), age (years), sex distribution (male/female), primary intervention, activity conducted under analysis, number of trials per activity, type of feet utilized, the marker placement protocol, number of markers utilized, biomechanical model defined, motion capture system used, and the capture frequency during data collection.

Study II

A repeated measures study design was used to investigate the functional joint centres (FJCs) [35] for each of a selection of prosthetic feet and the intact foot of a single participant. Analysis compared the position of the FJCs within the prosthetic feet, and in comparison to the control foot. An analysis of inter-trial reliability of the FJC method was conducted utilizing confidence intervals of the x- and y-coordinate positions within two testing occasions.

Six prosthetic feet were chosen (Figure 10) and fit to one participant (*at time of Study II*: age: 60 years, mass: 98 kg) on two separate occasions. The same process was carried out with each of the six prosthetic feet and included:

- Fitting and alignment of the prosthetic foot
- Ten minute practice session
- Attachment of the reflective markers (Figure 11)
- Data collection

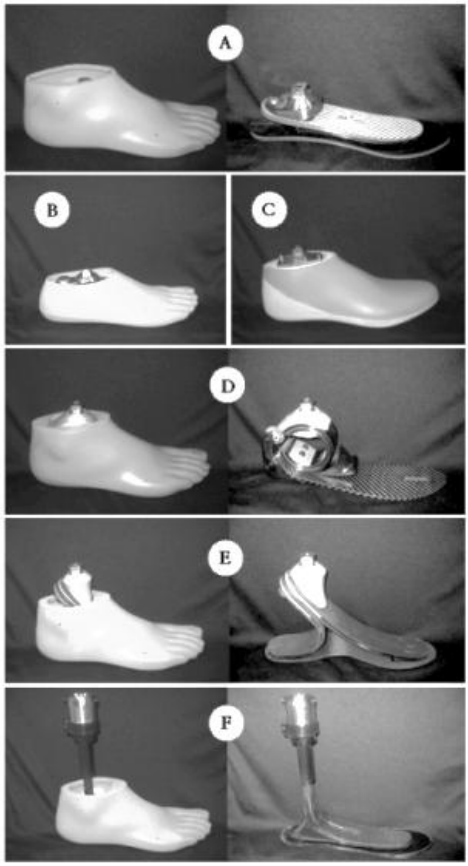


Figure 10 - The six prosthetic feet used in this study. As classified by Hafner et al. [27]. A, D, E and F belong to the ESAR category, F belongs to SA, and C belongs to CV. Image from *Study II.*

The data collection protocol required the participant to walk the length of a 10-metre walkway during which three-dimensional coordinate data was captured using an eight-camera motion analysis system (Qualisys AB., Sweden). Ten trials were collected for each of the prosthetic feet, with a total of 60 trials in total. A second testing occasion was conducted two weeks after the first in which the identical testing protocol was followed. The participant's intact limb served as the control limb for all analyses.

Reflective Marker Placement

As only one participant was used in *Study II*, the marker positions were the same throughout all testing protocols, on all prosthetic feet. Marker positions on the prosthetic limb were determined using the measured positions of the reflective markers from the intact limb. The positions are presented in Figure 11.

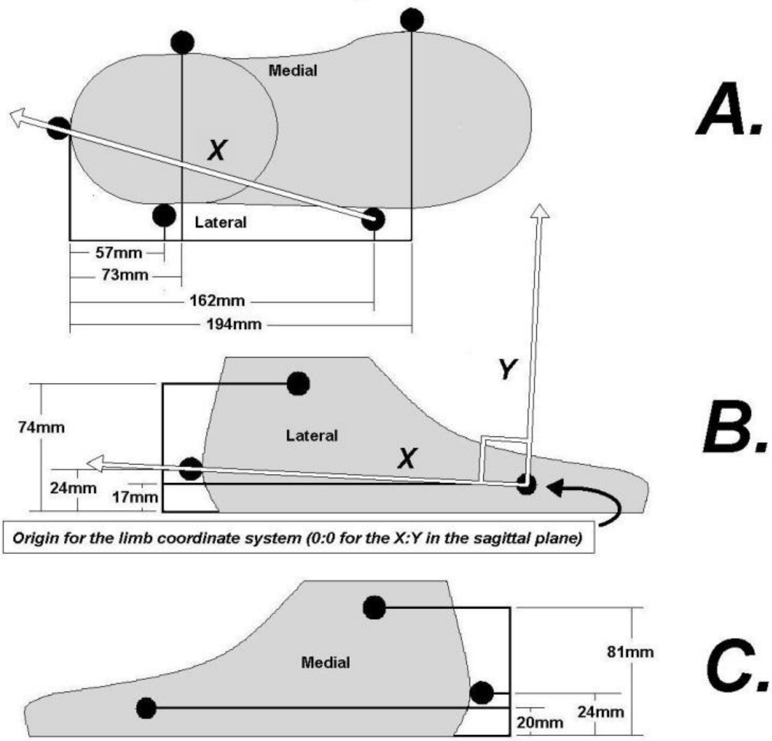


Figure 11 - Marker placement was determined by measuring the anatomy of the intact foot. Placement on the prosthetic foot from above (A), lateral (B), and medial (C) is matched based on the corresponding measurements from the intact limb. x:y coordinates used in analysis are defined in (B) with an origin at the marker signifying the 5th metatarsal head, or in the case of the prosthetic foot, the position matching that of the 5th metatarsal of the intact foot. Position shown includes the heel-height of the shoes worn during data capture. Image from *Study II.*

Data Processing

Data was processed offline using Visual 3D (C-Motion Inc., USA). Data was first low-pass filtered using a second-order Butterworth filter with a cutoff frequency of 6 Hz. Coordinate data then was transformed from a lab-based coordinate system to one with an origin located at the reflective marker placed on the 5th metatarsal on the intact limb, and the marker representing the 5th metatarsal on the prosthetic limb (Figure 11). The FJC algorithm is based on the method developed by Schwartz and Rozumalski [35] and is provided here in full from *Study II*:

Consider all frames between $IC \rightarrow TO$

1) $(A, B, C \dots n)$

For the two segments, shank (S) and foot (F), at frame A find the vector $\overrightarrow{J^A}$ which represents the ankle joint position at frame A (Eq.2),

2) $\overrightarrow{J^A} = \overrightarrow{S^A} + \overrightarrow{S_f^A} = \overrightarrow{F^A} + \overrightarrow{F_f}$

where each of the variables in Eq. 2 is a vector quantity describing the position of the limb-coordinate systems $[(F)\ and\ (S)]$ in relation to the lab-coordinate system (L).

Given Eq. 2, $\overrightarrow{J^A}$ is common to both segments S and F. Though, because $\overrightarrow{J^A}$ can be a number of points along a finite helical axis (AoR), further reduction is required. Therefore, for all combinations of 3 frames within the phase $(A, B, C \dots n)$, compute the finite helical axes for intervals $A \rightarrow B$, $A \rightarrow C$, and $C \rightarrow B$:

3) $AoR^{\ A\rightarrow B}$, $AoR^{\ A\rightarrow C}$, $AoR^{\ C\rightarrow B}$

4) Accept helical axes where a minimum ROM of 5 degrees is attained. Compute each individual joint center candidate (JCC) as the intersection ∩ of the finite helical axes for each pair of intervals:

5) $JCC^A = AoR^{\ A\rightarrow B} \cap AoR^{\ A\rightarrow C}$
$JCC^B = AoR^{\ A\rightarrow B} \cap AoR^{\ C\rightarrow B}$
$JCC^C = AoR^{\ A\rightarrow C} \cap AoR^{\ C\rightarrow B}$

Define the FJC as the mode of a random selection (2,000,000) of all possible JCCs:

6) $FJC = mode\ (JCC^A, JCC^B, JCC^C, \dots JCC^{2{,}000{,}000})$

Additional Methods
Study II: Pilot Testing

In preparation for the execution of *Study II*, various unpublished methods were tested in a series of pilot trials. The pilot trials are described in the order they were carried out:

- Mechanical Pilot
- Gait Pilot
- FJC Validation

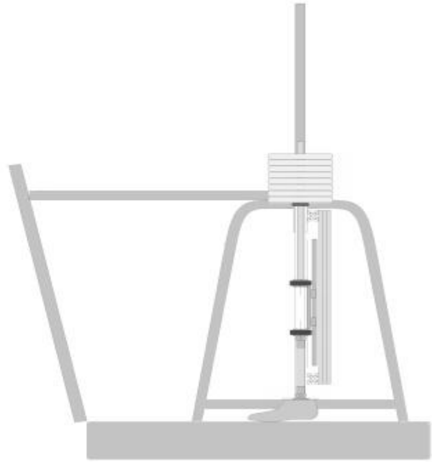


Figure 12 - Mechanical device built to test the ICR method. Image by: Mr. Kjell-Åke Nilsson.

The *Mechanical Pilot* describes efforts to use a mechanical device (Figure 12) to move the prosthetic feet through a RoM in order to calculate the centre of rotation.

The *Gait Pilot* used a transtibial prosthetic user to move the prosthetic foot through the required RoM.

Both the *Mechanical* and *Gait Pilot* used a geometric method called the Reauloux Method to calculate what is referred to as the Instantaneous Centre of Rotation (ICR) (Figure 13).

The *FJC Validation* utilized a rigid two-segment linked-model with a joint capable of a single-degree of freedom rotation about a known axis of rotation. This pilot used the same FJC algorithm employed in *Study II* (Figure 14).

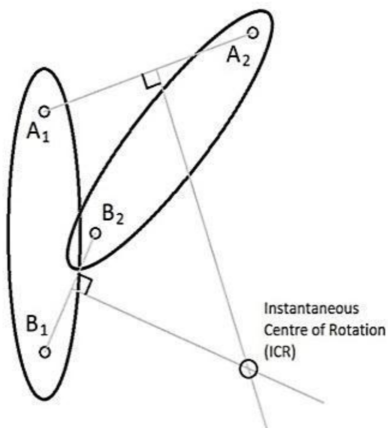


Figure 13 - The Reauloux Method for calculation of the ICR. The two-dimensional coordinate positions (x-y) of two rigid segments captured at two consecutive instances in time (A_1,B_1) and (A_2,B_2). The ICR is the intersection of two lines extending at right angles from the bisection of the line joining each point from one instant in time to the next.

Mechanical Pilot

In this pilot a custom-made frame was constructed which held a prosthetic foot in place above a surface which rotated in the pitch direction (toes-up/toes-down). The foot was mounted on a sliding track which moved in the inferosuperior direction and was loaded with a mass of 80 kg (Figure 12). With a prosthetic foot mounted in the frame, and having positioned reflective markers on the prosthetic 'shank/foot' (Figure 11), a pitch rotation of the prosthetic foot was elicited in the sagittal plane to rotate it through a RoM. The Reauloux Method was used to calculate the position of the instantaneous centre of rotation (ICR).

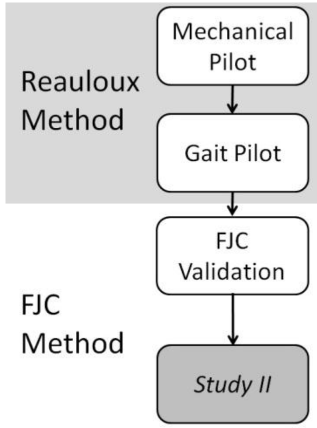


Figure 14 – Flowchart of the pilot testing and the algorithms used in calculating for each pilot. Reauloux and FJC methods refer to the algorithm used in calculating the centre of rotation for each pilot testing scenario.

Gait Pilot

The geometric Reauloux Method used the x-/y-component position for the markers for filtered data points from two consecutive instances in time (120 Hz) in the pilot testing. A line connecting the two consecutive points is bisected, with a line projecting at a right angle from this point. For this calculation two points from the same segment are required to be tracked. The full algorithm was written in Visual Basic for Applications (Microsoft Corporation Inc., USA)

One transtibial prosthetic user (same individual as in the *Study II*) (Table 1) was recruited for the test. Reflective markers were positioned on the prosthetic limb (Figure 11) and 10 consecutive trials of a 10-metre walkway wearing the current prosthetic limb were conducted. Using the same markers for designation of the limb-based model, an ICR was calculated for each consecutive data interval for the entire data collection for each pass over the force-plate. The ICR was determined as the mean position (x/y) for all intervals for all ten passed.

FJC Validation

Validation of the FJC method was conducted using a rigid model with a known single-axis rotation (one degree of freedom). This rigid model was tested using a full-marker set-up required to track motion of a foot and shank segment (Figure 15). A series of ten trials were collected in which an investigator moved the model through a RoM of approximately 20 degrees in toes-up and toes-down directions, for an approximate angular excursion of 40 degrees. FJC position was then calculated for each of the trials and x- and y-coordinate positions were averaged for the 10 trials. Means and SDs were used to evaluate the method.

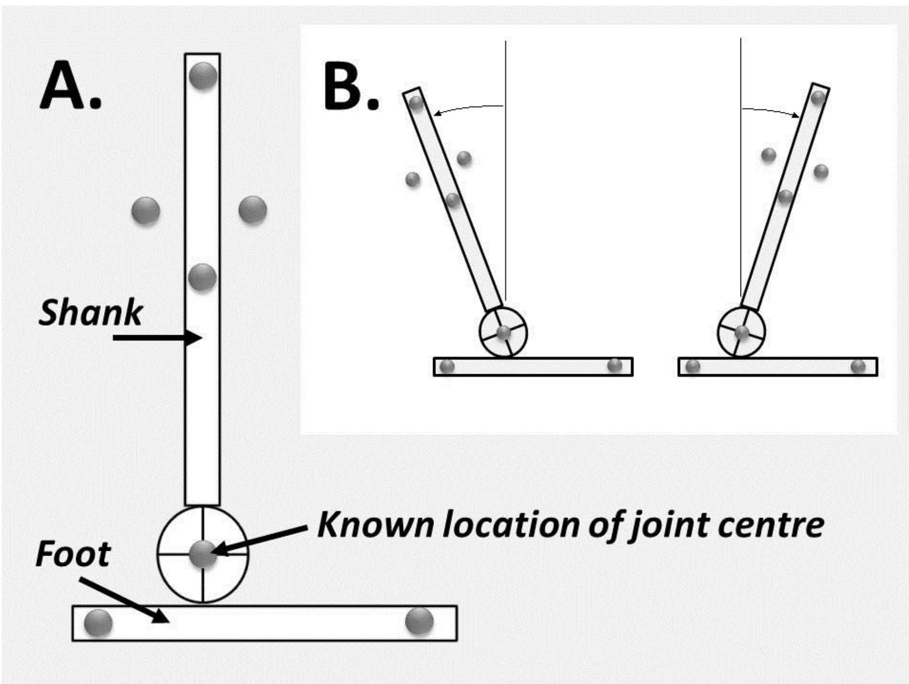


Figure 15 – Validation of the FJC method on a rigid model with a known joint centre location (A). The two rigid links (shank and foot) were moved through a RoM and the calculated FJC was compared to the known location of the mechanical joint (B).

Study III

In this study a series of postural stability tests were conducted using a Pro Balance Master (NeuroCom International Inc., Oregon, USA). The system incorporates a 46 cm × 46 cm forceplate which is capable of sagittal plane pitch rotations in toes-up and toes-down directions. Sampling frequency for force-platform data was 100 Hz.

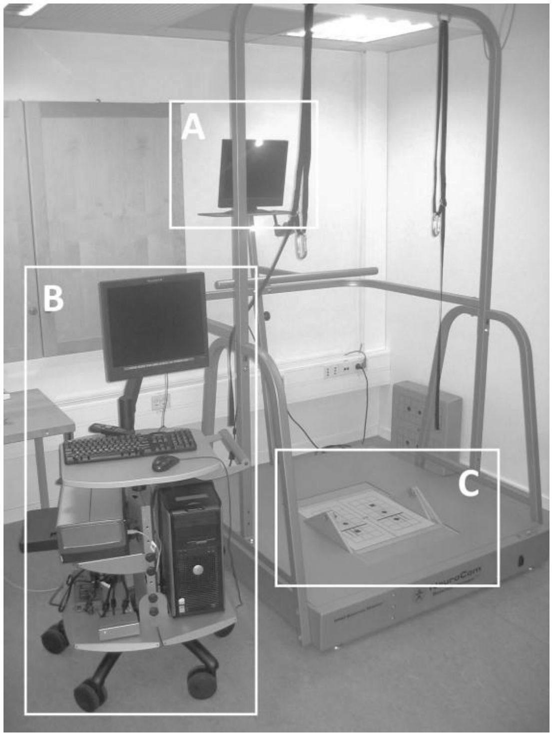


Figure 16 – Pro Balance Master (Neurocom Inc., USA) platform where: *A* is the monitor which participants use to monitor CoG motion and prompts for testing protocols, *B* is the system computer for collecting data and allowing the investigator to monitor testing protocols, and *C* is the forceplate which collects raw pressure data.

Prior to testing, the participants were fitted with a safety-harness in case of a fall. During testing they stood on the forceplate (Figure 16 C) facing a computer screen (Figure 16 A) which prompted them through the testing protocol. During some tests the participants were required to follow an icon on the computer screen which displayed motion of their CoG. The investigator followed the movements, and prompted appropriate tests, via a separate computer screen (Figure 16 B).

Participants were fit with a device capable of providing vibratory feedback to the thigh of the limb with TTA. Four individual Flexiforce transducers (Tekscan, Inc., Boston, USA) were positioned under the prosthetic foot (Figure 17). Pressure applied to these force transducers produced a signal which was transmitted to a 4-channel controller with on-board microprocessor. After receiving voltage output from individual force transducers, a sine-wave signal (230 Hz) current output was transmitted to provide power output to individual tactors located on the participant's thigh. The channels were independently controlled and capable of producing an output of 350 mArms (milliampere root-mean-square) at 250 Hz to power the individual tactors.

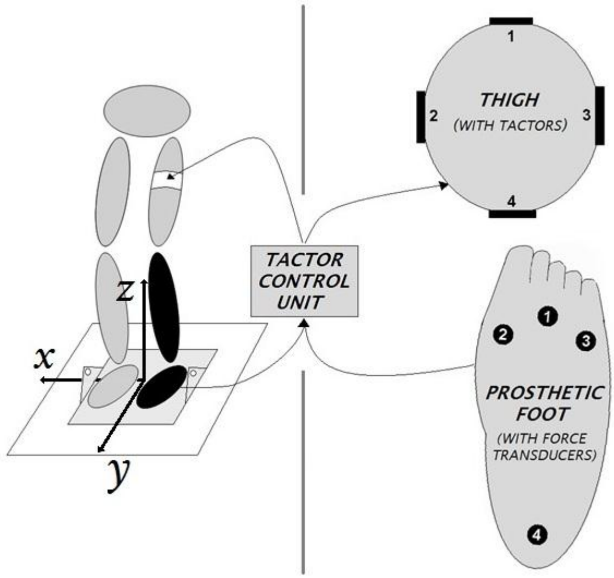


Figure 17 - Tactor Control Unit and channel descriptions 1, 2, 3 and 4 (corresponding to anterior, posterior, medial and lateral). Black limb represents the prosthetic limb. Each channel was individually controlled with a force transducer located under the prosthetic foot linked to a tactor on the thigh. Global reference frame to the left with x-y-z corresponding to mediolateral-anteroposterior-inferosuperior directions respectively. Limb reference frame to right with x-y corresponding to mediolateral-anteroposterior directions respectively. Image from *Study III*.

Three separate tests of postural stability were investigated: *Standing Balance (SB)*, *Limits of Stability (LoS)*, and *Rhythmic Weight Shift (RWS)*. Definitions for dependent variables are taken from *Study III*. The SB test was conducted under 4 surface and vision conditions: eyes-open/stable surface, eyes-closed/stable surface, eyes-open/unstable surface, eyes closed/unstable surface (Figure 20).

Standing Balance

In this test the participant stood upright on a force-platform (Figure 18) where a series of 20 second trials were completed in which raw forceplate data was collected. Following collection, raw data was filtered using a low-pass zero-phase lag 4th-order Butterworth filter at a cutoff of frequency of 10 Hz. Coordinate position of the CoP was calculated using the anteroposterior component of the CoP (CoP*y*) and in the mediolateral component (CoPx) (Equation 1) for each of the vision and surface conditions (Figure 19).

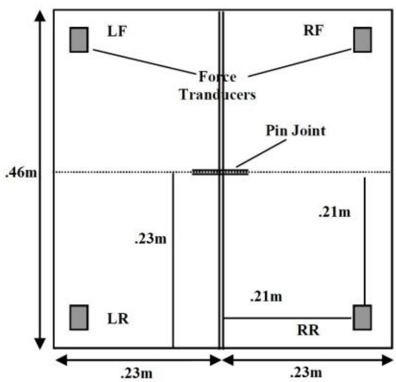


Figure 18 – Forceplate design. The position of the four force transducers (LR, LF, RF, RR) allowed the calculation of the CoP based on the magnitude of the vertical force applied. Image modified from [40].

$$CoPx = \frac{(RF + RR) - (LF + RR)}{RF + RR + LF + LR} \times 10.2$$

$$CoPy = \frac{(RF + LF) - (LR + RR)}{RF + RR + LF + LR} \times 10.7$$

Equation 1 – The equations for calculating the x- and y-component positions of the CoP (equations 1 and 2). Where (RF) is right-front, (RR) is right-rear, (LF) is left-front and (LR) is left-rear. Coordinate positions given in centimetres (cm).

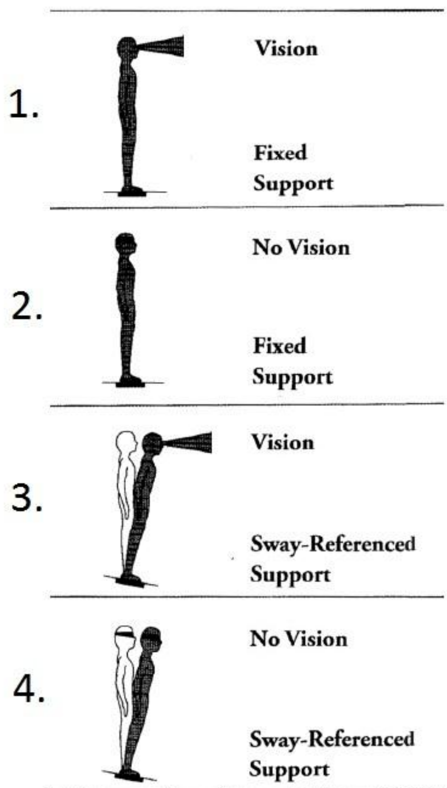


Figure 19 - Testing conditions for the Standing Balance (SB) test. Modified from [40].

Following the completion of the 4 standing conditions, 9 dependent variables were calculated:

- mediolateral and anteroposterior CoP-RMS (RMS_{ML}, RMS_{AP}) (Equation 2)
- path length per second of the CoP (PPS) (Equation 3)
- mediolateral and anteroposterior sway amplitude (AMP_{ML} and AMP_{AP}) (Equation 4)
- mediolateral and anteroposterior path length (P_{ML}, P_{AP}) (Equation 5)
- mediolateral and anteroposterior mean velocity (MV_{ML},MV_{AP}) (Equation 6).

Equations 21-25 from Study III

$$RMS_{AP} = \sqrt{\frac{{y_1}^2 + {y_2}^2 \ldots + {y_n}^2}{n}}; RMS_{ML} = \sqrt{\frac{{x_1}^2 + {x_2}^2 \ldots + {x_n}^2}{n}}$$

Equation 2 - Root mean squared (RMS) of the anteroposterior (RMS_{AP}) and mediolateral (RMS_{ML}) CoP, where $[x]$ and $[y]$ are the respective x-direction and y-direction coordinate positions of the instantaneous CoP for consecutive frames, and $[n]$ is the number of frames.

$$PPS = \frac{f}{n-1} \Sigma i = 1\ to\ n - 1 \sqrt{\{(xi + 1 - xi)2 + (yi + 1 - yi)2\}}$$

Equation 3 - Path length per second (PPS) of CoP, where [f] is the samply frequency, [n] is the number of frames, and([x],[y]) are instantaneous CoP coordinate positions.

$$AMP_{AP} = \sqrt{(y_A - y_B)^2}; AMP_{ML} = \sqrt{(x_A - x_B)^2}$$

Equation 4 - Sway amplitude (AMP), where($[x_A]$,$[x_B]$) and($[y_A]$,$[y_B]$) are the respective maximum and minimum x- and y-direction coordinate positions of the instantaneous CoP.

$$P_{AP} = y_1 + y_2 \ldots + y_n; P_{ML} = x_1 + x_2 \ldots + x_n$$

Equation 5 - Anteroposterior and mediolateral path length (P), where [x] and [y] are the distances travelled in the x- and y-directions between successive frames.

$$MV_{AP} = \frac{P_{AP}}{t}; MV_{ML} = \frac{P_{ML}}{t}$$

Equation 6 - Anteroposterior and mediolateral velocity (MV_{AP}, MV_{ML}), where $[P_{AP}]$ and $[P_{ML}]$ are the anteroposterior and mediolateral path length (Eq. 4), and $[t]$ is time.

Limits of stability

This test was an evaluation of the participant's ability to voluntarily shift CoG towards goals which represent their maximum distance from a central position in 8 directions (Figure 20). Following completion of the conditions 3 dependent variables were calculated: maximum excursion (ME), directional control (DC), and reaction time (RT).

- ME is defined as the angular difference between the angle of inclination at trial initiation and the maximum angle of inclination towards the goal.
- DC is defined as the total angular distance travelled by the CoG expressed as a percentage of the shortest possible distance (Equation 7)
- RT is defined as the length of time for a participant to voluntary shift their CoG in an intended direction following a visual cue using a threshold value of 5% of the total angular distance to the goal.

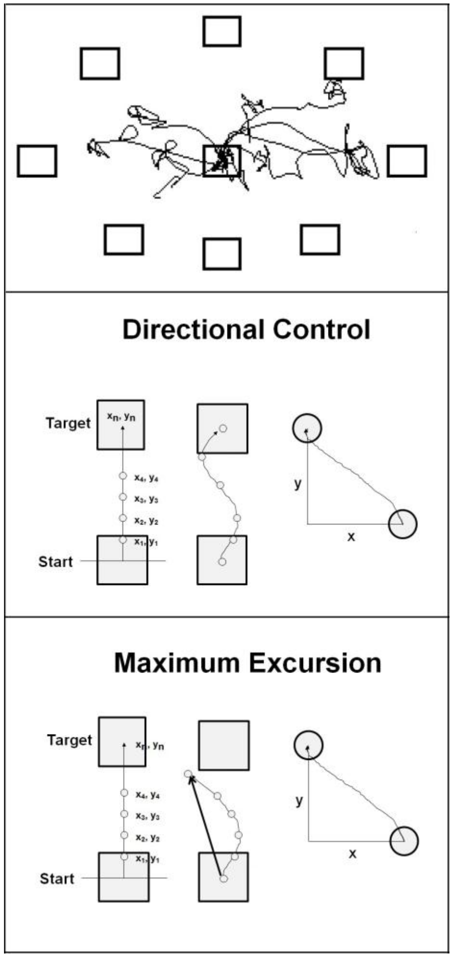


Figure 20 - Testing conditions for the Limits of Stability (LOS) test. Images modified from [40].

$$\left[\frac{(CoG\ Path\ On-axis)-\ (CoG\ Path\ Off-axis)}{(CoG\ Path\ On-axis)}\right]\times 100$$

Equation 7 - Directional control (DC) for the LOS test.

Rhythmic Weight Shift

This test was an evaluation of the participant's ability to shift their CoG in the anteroposterior and mediolateral directions by following a cursor moving at three predetermined velocities (slow, moderate, fast) (Figure 21). Following completion of the test two dependent variables were calculated: directional control (DC) and on-axis velocity (AV):

- DC is defined similarly in the RWS test as it is in the LoS test except that it uses the SD of the CoG path (Equation 8).
- AV is defined as a participant's ability to match the velocity of a moving target (Equation 9).

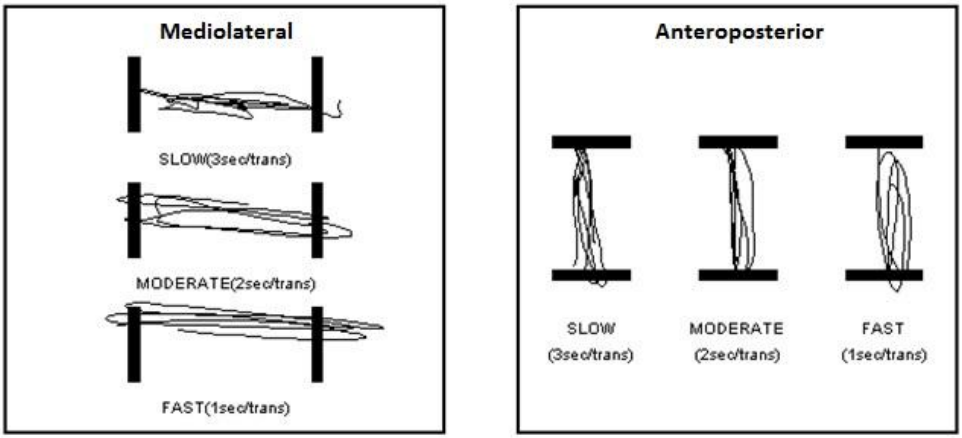


Figure 21 - Testing conditions for the Rhythmic Weight Shift (RWS) test. Image modified from [40].

$$\left[\frac{(CoG\ SD\ On-axis) - (CoG\ SD\ Off-axis)}{(CoG\ SD\ On-axis)}\right] \times 100$$

Equation 8 - Equation for directional control (DC) in the RWS test.

$$\frac{Total\ angular\ displacement\ in\ the\ intended\ direction}{Total\ length\ of\ time}$$

Equation 9 - Equation for on-axis velocity (AV).

Study IV

Participants were required to stand on a platform (Figure 22) capable of rapid support surface pitch rotations (toes-up/toes-down) (Figure 23). This is the same core-structure as in *Study III* with modifications to the surface, support-harnessing, and signal acquisition (compare Figures 16 and 22). Conditions were manipulated to investigate how limb-position and weight-bearing affected automatic postural response (APR) in a TTA-group compared to a control-group. There were three limb-positions evaluated (prosthetic limb on platform, intact limb on platform , both limbs on platform) (Figure 24 a-c). As the two analyses in the study investigated the intact limb and prosthetic limb individually, variables named ON, OFF, and BOTH refer to the position of the limb of interest. For instance, in the analysis of the intact limb, OFF refers to the position of the intact limb. There were three weight-bearing conditions (reduced weight-bearing [25% of total body-weight], equal weight-bearing [50% of body weight], increased weight-bearing [75% of total body-weight]).

Figure 22 - Force-platform with participant standing facing digital display of CoP and mass-distribution.

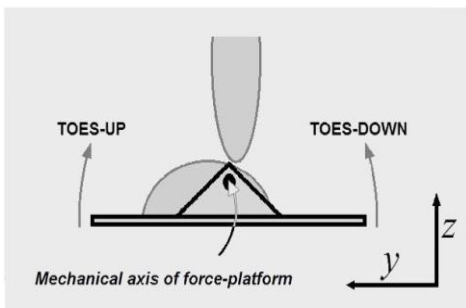


Figure 23 - Image explaining the type of perturbation utilized. Pitch plane rotations of the support surface were elicited corresponding to toes-up/toes-down directions. Image modified from *Study IV*

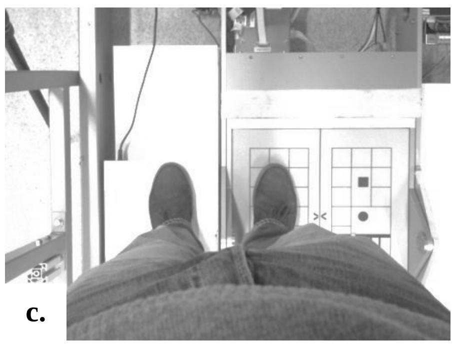
Surface EMG signals were collected from lower-extremity musculature. For the TTA-group, electrodes were placed over the tibialis anterior (TA) and gastrocnemius medialis (GM) muscles on the non-amputated limb, and bilaterally on vastus lateralis (VL) and biceps femoris (BF). For the Control-group, electrodes were placed bilaterally over the tibialis anterior (TA), gastrocnemius medialis (GM), vastus lateralis (VL) and biceps femoris (BF).

Using visual feedback from monitors the participants were required to maintain CoP position and predetermined weight-bearing distributions between trials. Following confirmation of these goal positions/weights the investigator triggered a series of rapid surface perturbations. A total of 99 perturbations were elicited, split into randomized subgroups for weight-bearing, limb-position, and direction of perturbation.

A threshold value to determine onset of EMG activity was determined using background activity for each muscle collected for 100 ms prior to platform movement. These EMG onset latencies were then compiled to give a mean latency for each condition. Analysis was conducted to compare the APR EMG onset latency times for the TTA-group to the Control-group.

a.

b.



c.

Figure 24 - The three limb-position conditions used in the experimental protocol. a: both limbs on support surface; b: One limb on support surface; c: Opposite limb on support surface. Picture is of a able-bodied control, TTA-group participants had the same conditions.

Additional Methods *Study IV:* Pilot Testing

Weight-bearing Pilot

As part of the preparation and pilot phase of *Study IV*, the question of weight-bearing distributions was posed. The aim of the pilot was to evaluate how close the actual weight-distributions were to the 25-50-75% distributions intended in the aims. A follow-up to this aim was to see if there was a way to decrease the variability of these distributions.

The weight-bearing pilot study investigated three separate methods aimed at ensuring consistency for the 25-50-75% weight-distributions that were required as part of the study's methodology. They were as follows:

Method 1:

Involved calculating the 25-50-75% distributions and providing the participant a practice period prior to testing in which they received digital feedback. This visual feedback was not available during the real testing protocol.

Method 2:

Similar to method 1, but after every 11-trial clusters the participant had a short reminder session of practice with feedback. They then moved back to testing without the addition of real-time feedback.

Method 3:

The participant received real-time feedback about their weight-bearing distributions, in addition to the 5 minute practice session prior to data collection.

As an addition to Method 3, the variability of the data was also evaluated following filtering out the trials that were ≥5% from the intended 25-50-75% distributions.

9 STATISTICAL METHODS

In all four studies descriptive statistics were used to show variations in the data. All further analyses employed parametric methods after having determined their appropriateness using tests for normality. Statistical significance was determined using a critical alpha level of α=0.05 for all tests unless otherwise stated. A summary of the specific methods employed for each study is provided in Table 3.

Table 3 - Summary of statistical tests used in each of studies *I-IV*.

	Studies			
Methods	**I**	**II**	**III**	**IV**
Determination of Sample Size				
a-priori Power-Calculation			X	X
Tests of Normality				
Shapiro-Wilk test		X		X
Kolmogorov-Smirnov test			X	
Descriptive statisitcs				
Mean	X	X	X	X
SD	X	X	X	X
Range	X			
95% CI		X	X	X
Main Analyses				
One-way repeated-measures ANOVA		X		
Two-way multi-factorial ANOVA			X	
Three-way multi-factorial ANOVA			X	X
Test-retest Reliability [1] and Difference within control data [2]				
Paired t-test		X[1]		X[2]
Tests of Correlation				
Spearman's Correlation Coefficient	X			
post-hoc Analyses				
Pair-wise comparisons		X	X	X
Tukey's post-hoc			X	X
Bonferroni adjustment		X	X	

Study I

Descriptive statistics are presented and a Spearman's Correlation analysis was utilized to investigate the relationship between time (in years) and the level of evidence [108] and quality of research [109].

Study II

Study II contained two primary analyses:

1) The FJC of the control foot compared to the FJCs of the prosthetic feet; and
2) The anatomical method compared to all the FJCs (both intact and prosthetic feet).

To address the aims, two one-way repeated-measures ANOVAs were conducted using the x- and y-coordinate positions. The first ANOVA contained all x-coordinate positions, the second contained all y-coordinate positions (from both testing occasions). Comparisons of FJCs were made using pooled FJC data across all trials for each prosthetic foot compared to pooled data of a representative trial from the control foot. Comparisons of the anatomical method with the FJC method were analyzed using pooled data from all available trials. To determine where significant differences existed, post-hoc pair-wise comparisons were performed. Confidence intervals of mean absolute differences between testing occasions were additionally performed. Test–retest reliability of the FJC sagittal plane position was evaluated using a paired t-test for each of the two testing occasions for each foot. A Bonferroni adjustment ($p=0.025$) was used as the analysis contained two comparisons per foot (mean x- and y-coordinate positions). Inter-trial reliability of the FJC sagittal plane position was investigated using the confidence intervals of the x- and y-coordinates within the 2 testing occasions.

Study III

An *a-priori* power calculation to determine sample size was conducted.

Study III contained three primary analyses:

1) If the addition of vibratory feedback affected SB;
2) If the addition of vibratory feedback affected LoS; and
3) If the addition of vibratory feedback affected RWS.

To address these aims, three separate analyses were conducted:

SB was analyzed using a three-way MANOVA with three independent variables (Vibration × Support-Surface × Vision) and nine dependent variables.

LoS test was analyzed using a two-way MANOVA with two independent variables (Vibration × Direction) and three dependent variables.

RWS was analyzed using a three-way MANOVA with three independent variables (Vibration × Direction × Velocity) and two dependent variables.

Study IV

An *a-priori* power calculation to determine sample size was conducted.

Study IV contained two primary analyses:

1) Are there differences between the EMG latencies following perturbations of the intact leg of transtibial prosthesis users compared to able-bodied controls.
2) Are there differences between the EMG latencies following perturbations of the leg with a prosthesis of transtibial prosthesis users compared to able-bodied controls.

To address the aims, two three-way ANOVAs were conducted. Bilateral control-data was combined after a paired t-test showed no significant difference ($p>0.05$).

The first analysis was on the intact limb using the intact limb of the prosthetic user and the combined values for both limbs of the control participants. The intact analysis had three independent variables (Group × Limb-position × Mass) and eight dependent variables (latency of each of the four EMG channels in two rotation directions).

The second analysis was on the side with the prosthesis using the thigh of the limb with an amputation with the corresponding combined signals of the control participants. The prosthetic limb analysis had three independent variables (Group × Limb-position × Mass) and four dependent variables (latency of each of the two EMG channels in two directions).

10 ETHICS

Study I

This review article involved no experimental participants and therefore no ethical approval was required.

Study II

The participant provided written informed consent to the study which was approved by the regional ethics committee in Linköping, Sweden on the 13th of August, 2008. Document number: DNR-07, T52-08.

Study III & Study IV

Written informed consent was provided by all 48 participants according to the research application approved by the regional ethics committee in Linköping, Sweden on the 28th of September, 2007. Document number: DNR-07.

11 RESULTS

Study I

The literature search yielded a total of 54 articles from the Medline and Cinahl databases. Searching the reference lists of review articles also identified an additional 14 articles that met the inclusion criteria. A total of 68 articles satisfied the inclusion criteria and were included in the analysis (Figure 25).

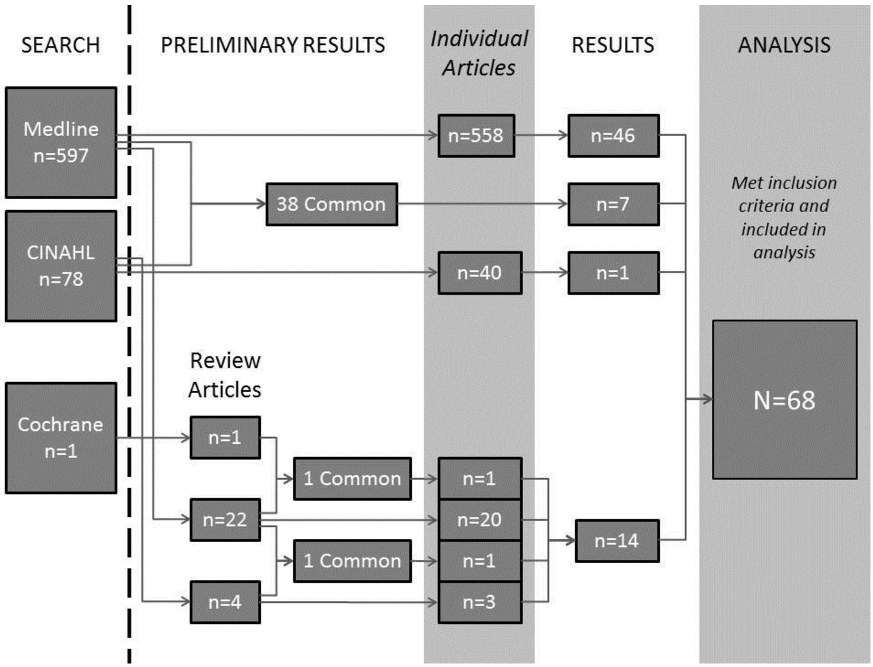


Figure 25 – Flowchart outlining process of exclusion for found and excluded articles.

The distribution of level of evidence for the included studies was: 3 x Level II (sub-category B; poorly designed randomized controlled trial), 23 x Level III studies (sub-category A; case-control studies), and 42 x Level IV studies (34 sub-category A, case-series studies; 8 sub-category B, case-report studies) (Table 2; *Study I*).

The mean quality score of the included studies was 10.1 (SD 2.5) out of a possible 16 points. The quality scores ranged from 2 to 15 points. The included studies had a mean sample size of 8.9 participants (SD 6.4), age of 42.5 years (SD 14.0), and sex distribution of male/female = 0.85/0.15 (Table 2; *Study I*).

The results of the review highlighted a number of methodological differences which make direct comparison of results between studies difficult. Variability in the data collection methods were considerable including the frequency of data capture, marker placement protocol, biomechanical model utilized, and activity conducted as part of the experimental protocol.

Given the kinematic results of the analyzed studies, and the known error associated marker misplacement [110], prosthetic ankle dorsiflexion had a potential error of 27% (range 18-34%).

The Spearman's correlation analysis resulted in a weak positive correlation between increasing year of publication and higher level of evidence (r_s=.360, p=.002) and a weak (non-significant) positive correlation between increasing year of publication and increased research quality (r_s=.236, p=.051).

Study II

Comparison of the FJC of the control foot with the prosthetic feet

The control foot was significantly different ($p<0.05$) in the x-direction from all the prosthetic feet investigated. The control foot was significantly different ($p<0.05$) than all feet in the y-direction, except for the Advantage foot ($p=0.462$) (Figure 26).

Comparison of the FJC to the anatomical position

The anatomical ankle position was significantly different ($p<0.05$) than all FJC positions in both the x- and y-directions, with two exceptions. These exceptions were a non-significant difference between the anatomical position and the FJC of the control foot in the x-direction ($p=0.547$) and the anatomical position and the Advantage foot in the y-direction ($p=0.012$) (Figure 26).

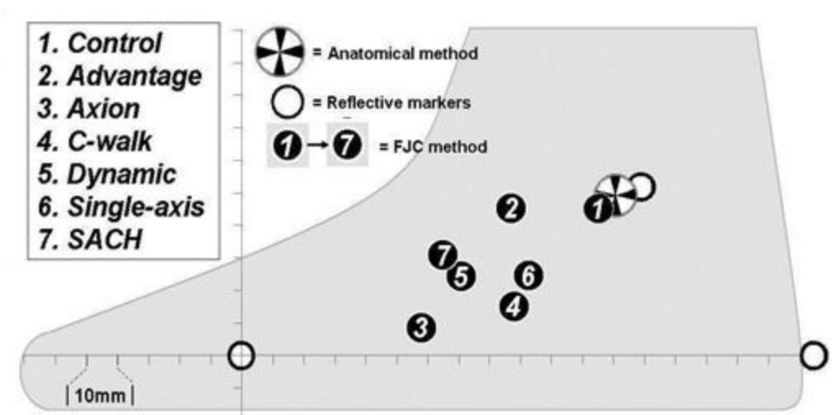


Figure 26 - The position of the FJCs for 6 prosthetic feet and the control foot. Black and white cross designates the position of the ankle given the anatomically based method of ankle position. Image modified from *Study II*.

Reliability

Pairwise comparisons showed statistically significant differences existed for all the feet tested, including the control foot, except for the x-direction for the control foot and the SACH foot (Table 4). Mean difference between testing occasions for pooled-foot data was 5.9mm and 10.9mm for the x- and y-directions respectively.

Table 4 - Reliability of the FJC method used on the six prosthetic feet and the control foot. Mean difference between testing occasion 1 and occasion 2 with 95% CI, SD and p-values. Coordinate data in millimetres (mm). Table from *Study II*.

		Mean Diff.	95% CI	SD	p-value
Control	x	3.7	0.2 — 7.1	4.8	0.041*
	y	-5.8	-9.6 — -2.0	5.3	0.007
Advantage	x	7.8	5.0 — 10.6	3.9	0.000
	y	12.7	9.1 — 16.2	4.9	0.000
Axtion	x	8.5	6.6 — 10.4	2.7	0.000
	y	8.2	6.8 — 9.7	2.0	0.000
C-walk	x	4.6	1.3 — 7.9	4.3	0.012
	y	11.5	6.1 — 16.9	7.0	0.001
Dynamic	x	6.9	5.7 — 8.2	1.7	0.000
	y	10.2	6.2 — 14.1	5.5	0.000
Single-axis	x	-4.3	-7.4 — -1.1	4.4	0.013
	y	8.7	6.0 — 11.5	3.8	0.000
SACH	x	3.3	-0.1 — 6.7	4.8	0.058*
	y	14.2	11.4 — 16.9	3.8	0.000

Additional Results: Pilot Testing *Study II*

Mechanical Pilot

The mechanical pilot testing failed to produce large enough deflections of the prosthetic foot to warrant continued use of the method. The maximum dorsiflexion angle attained from the neutral position was 6.7 degrees. Plantarflexion was difficult to attain as the fore-foot simply rose from the support surface. It was not possible to attain plantarflexion angles greater than 2 degrees.

Gait Pilot

The results of the Gait Pilot showed that the ICR was in a position which was not the same as the marker denoting the lateral malleolus (approximate position of the anatomical ankle position). The mean x-/y-positions of the lateral marker and ICR were 156.05/81.75mm and 76.89/90.84mm respectively. The results clearly show the Reauloux Method had a variability which made using it questionable (Table 5).

FJC Validation

The results of the rigid model testing utilizing the FJC method produced mean x-/y-positions of the lateral marker and FJC with similar values (Ankle marker [119.12/52.76mm] and FJC [119.56/52.20mm]) and SDs which were small with respect to the inherent error of the capture system (Table 6).

Table 5 - Gait Pilot results. Mean and SD of x and y coordinate positions in millimetres (mm).

	Ankle Marker		ICR	
	x	*y*	*x*	*y*
Mean	156.05	81.75	76.89	90.84
SD	2.14	30.25	505.00	885.20

Table 6 - Validation of the FJC method using a rigid model. Mean and SD of x and y coordinate positions in millimetre s(mm).

	Ankle Marker		FJC	
	x	*y*	*x*	*y*
Mean	119.12	52.76	119.56	52.20
SD	0.04	0.03	0.09	0.67

Study III

Standing Balance (SB)

The results showed a statistically significant main effect (VIB – NOVIB) (p=0.001) in the SB test. This difference indicated increased AMP_{ML} in the VIB condition (mean diff. 0.010 m, 95%CI 0.004 – 0.016 m). This indicates the total excursion range in the ML direction was greater with the addition of vibration.

Limits of Stability (LoS)

The results showed a statistically significant main effect (VIB – NOVIB) for the independent variable RT (p=0.013). The RT was faster with vibration (869 ms [SD 29]) than without vibration (982 ms [SD 33]). Mean difference (VIB – NOVIB) was 113 ms (95%CI -202 – -24). This indicates that the participants responded quicker to voluntary movements of the CoG with the addition of vibration.

Rhythmic Weight Shift (RWS)

The results indicated no significant main effect (VIB – NOVIB) for any of the independent variables.

Study IV

Prosthetic Limb

There was a statistically significant main effect for Group (TTA—Control). This was in the biceps femoris muscle in the toes-up direction ($p=0.023$) indicating increased EMG latency for the TTA-group (180 ms) compared to the Control-group (129 ms) (Figure 27-1, *left side of figure*).

Intact Limb

There was a statistically significant main effect for Group (TTA—Control). This was in the gastrocnemius muscle in the toes-up direction ($p=0.021$) and indicated increased mean EMG latency times for the TTA-group (182 ms) compared to the Control-group (116 ms) (Figure 27-2a, *middle of figure*).

There was a statistically significant main effect for Limb-position (ON—OFF—BOTH). This was in the vastus lateralis muscle in the toes-down direction between ON—OFF (98—147 ms) and ON—BOTH (98—130 ms). There was a statistically significant interaction effect in the Group—Limb-Position comparison ($p=0.018$) indicating increased mean latency times for the TTA-group (195 ms) compared to the Control-group (126 ms) in the OFF position (Figure 27-2b, *right side of figure*).

There was a significant main effect for Weight (INCREASED—REDUCED—EQUAL). This was in the vastus lateralis muscle in the toes-down direction between INCREASED—REDUCED (100—138 ms) and INCREASED—EQUAL (100—137 ms) (Figure 27-2b, *right side of figure*).

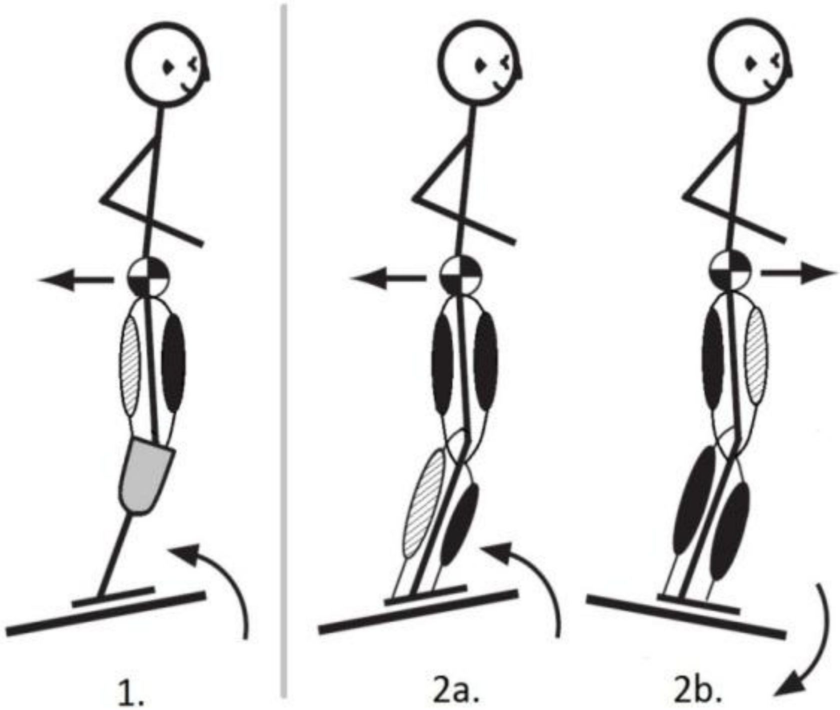


Figure 27 - Significantly different EMG response latencies when comparing the TTA-group and the control-group. (1) the prosthetic limb in the toes-up direction compared to the control-group; (2a) the intact limb of the TTA-group in the toes-up direction compared to the control-group; and (2b) intact limb of the TTA-group in the toes-down direction compared to the control-group. Significant differences denoted by stratified pattern over muscles. Image modified from Ting [95].

Additional Results: Pilot Testing *Study IV*

Weight-bearing Pilot

Method 1

The results for method 1 showed the mean and (SD) values for the intended distributions to be 36% (12), 54% (9), and 77% (11) (Figure 28).

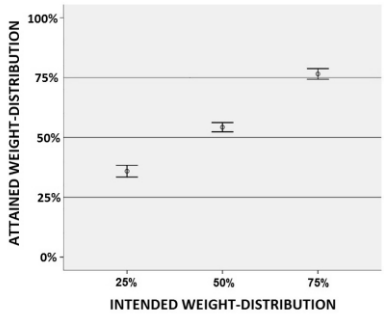


Figure 28 – Method 1: Pooled weight-distributions (25-50-75%) for 3 selected prosthesis users.

Method 2

The descriptive results for the combined trials show mean differences in the 25-50-75% weight-bearing scenarios of 2.3% - 0.6% - 0.7% (Figure 29).

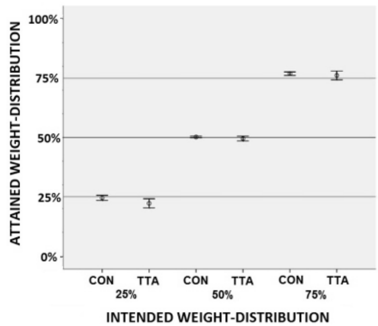


Figure 29 – Method 2: The weight-distributions (25-50-75%) for an able-bodied control (CON) and one prosthetic user (TTA) utilizing a practice/reminder session after every 11 trials cluster.

Method 3

The descriptive results for the cleaned data show mean differences in the 25-50-75% weight-bearing scenarios of 2.2% - 0.3% - 1.2% (Figure 30).

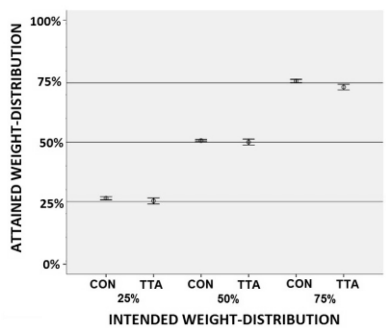


Figure 30 – Method 3: The weight-distributions (25-50-75%) for an able-bodied control (CON) and one prosthetic user (TTA) utilizing real-time feedback and cleaning of trials where results deviated ≥5%.

12 DISCUSSION

The results of this thesis have clearly shown that transtibial prosthesis users have a number of differences in relation to motion analysis and postural stability when compared to able-bodied individuals.

The quality of the evidence that researchers are presenting in studies which utilize three-dimensional kinematics of transtibial prosthesis users is generally low, but is improving with time. Efforts can be made in this area of research to make a positive systematic shift in the quality of the research presented. Additionally, there are large systematic errors present when rigid-body assumptions derived from the intact musculoskeletal system are applied to the motion of a prosthetic foot. These errors suggest that motion of prosthetic feet are different from each other and from an intact ankle.

Previous research has shown that transtibial prosthetic users have decreased values in postural stability measures. This thesis has shown that a simple feedback device, as part of the prosthetic limb, can positively improve the ability of transtibial prosthesis users to make rapid voluntary shifts of their centre of gravity. Additionally, prosthetic users make use of information about the pitch plane rotations of the support surface via the prosthetic limb. When these rotations are received through only the prosthetic limb they cause delayed reactions in the limb with the prosthesis and in the intact limb, indicating bilateral effects of unilateral amputation. Increased weight-bearing on the intact limb reduces the latency of response on the intact limb, but has no significant effect on the prosthetic side.

The following discussion has a study-by-study structure. Each section is concluded with the implications of the study individually, and how it relates to the overall thesis.

Study I

The aims of *Study I* were to critically examine the methods and techniques used in collecting and reporting three-dimensional kinematics of transtibial prosthesis users. It is clear that there are methodological issues that prevent valid comparison of the results between studies and these should be addressed to permit easier communication between researchers.

Evidence and Quality

The highest level of evidence identified in the reviewed studies was Level II (b) (three studies). An encouraging note is that the level of evidence is increasing with time, suggesting that researchers are adopting better research designs. It should also be noted that there were very small methodological changes that could have been made that would have caused a shift in the level of evidence results. For instance, with the inclusion of a control group many of the studies which were classified as Level IV evidence would have been changed to Level III evidence. Although it may not be appropriate for all studies to use able-bodied controls, it is reasonable to assume that a large proportion of the 42 studies in Level IV could have been moved to Level III by simply including an appropriate control group.

Biomechanical Models

One major issue that was identified in the reviewed studies was the variability in the choice of biomechanical models applied to kinematic data. The variety of biomechanical models used in the literature is of concern given the effect this can have on results. In a more recent review, Kent and Franklin-Miller [111] investigated specifically which biomechanical models were utilized by researchers. It is interesting to note that they have a very similar conclusion to that in *Study I*. They conclude that there needs to be additional focus on research to identify if a possible definitive solution exists for modeling transtibial lower-limb kinematics. Though, the authors concede this is not likely possible, and that a more realistic goal would be for researchers to have a clearer picture of the error inherent in the methods they employ. They state additionally, as much detail as possible should be provided when describing the methods employed in a study. This ambiguity was also stated in the conclusions of *Study I* as a limiting factor of the included studies.

The foot as an independent variable

The most common independent variable used in the studies reviewed was the experimental manipulation of the prosthetic foot. It is not surprising that investigators choose to investigate the prosthetic foot as this is arguably the major structural component of the transtibial prosthetic limb. Moreover, it is easy to control as it is manufactured in a systematic way and less likely to be influenced by confounding factors, as would be the case with a prosthetic socket which is manufactured under less controlled conditions. Of interest is the choice of the researchers to investigate SACH feet in the reviewed articles (34 of the studies included the SACH foot, 20 since year 2000). This seems to be in contrast to the more advanced types of prosthetic feet that clinicians want to provide clinically [25, 112]. In addition to clinician preference, accurate sales figures of prosthetic feet would further help to focus research resources based on need. It is possible the use of the SACH foot simply reflects the researchers' preference for a mechanically 'simple' prosthetic foot in their investigations. A methodological issue identified in many of the studies was the investigators' description of the prosthetic feet which was found to be incomplete, making repeatability of the study problematic. With a more detailed description of the components the repeatability of studies can be improved in the future.

One of the main conclusions of *Study I* was to state the product name and number for all components used in a research project. This was easily accomplished in *Study II* which involved placing a number of new components on the same prosthetic socket and a pylon (tube adapter). In contrast, *Study III* and *Study IV* name only the product names without giving the serial number or product number as suggested in *Study I*. As studies *III* and *IV* utilized the participants' currently functioning prosthesis it was in some cases impossible to determine the product description in such detail. In hindsight the conclusion named in *Study I* should instead read "name as much detail of the components as possible" or "serial numbers and product name and numbers should be recorded if practically possible."

Implications

The results of *Study I* provided justification for execution of *Study II* based on two factors. First, it was clear there was no definitive solution for the placement of reflective markers on the prosthetic limb. Secondly, modeling the kinematic variables meant placing markers over the ankle to designate the ankle joint. It was therefore decided to investigate if it was possible to determine an appropriate location for the reflective markers.

Study II

The aim of *Study II* was to identify and compare the position of the FJCs within a selection of commercially available prosthetic feet and in an anatomical ankle. It was clear that there were not only differences evident between the prosthetic feet and an intact ankle, but also within the selected group of prosthetic feet. When the anatomical method of joint centre location and the FJC method are compared in the intact foot the differences are quite small (mean difference x/y=2.6 mm/5.4 mm). When the same two methods are compared in the prosthetic feet it is clear that the motion of the prosthetic feet is significantly different to that of an intact foot (mean pooled difference x/y=39.6 mm/27.0 mm).

Reliability

The FJC method was demonstrated to have sufficient reliability to justify its use clinically. The pooled-mean differences for the selected feet were 5.9mm and 10.9mm for the x- and y-directions respectively. The standard deviation of error differences for the x- and y-coordinate positions for testing occasions one and two (3.7/4.5 mm) indicate adequate inter-session reliability. These magnitudes are comparable to the data presented by the developer of the algorithm in the study validating the method [35].

Mechanical Pilot and Gait Pilot

As the preparatory work for *Study II* was being executed a number of steps were completed in order to test various algorithms for calculating the joint centre and physically moving the prosthetic foot/ankle through a given RoM. The two algorithms were the Reauloux Method and the FJC Method. There were clear improvements in the results following the switch from the Reauloux Method to improved FJC Method, evident by more valid and repeatable results (Table 5).

The methods of moving the foot through the necessary RoM also had a development. This included a Mechanical Pilot employing a device which rotated the limb at the ankle, physically manipulating the limb with the investigators own body weight in the motion analysis lab, validation of the method using a rigid-model, and using an individual prosthetic user who walked with the given components in the motion analysis lab. The mechanical method proved to be of no practical use as the total RoM of approximately 8-9° was too small to justify further use [36, 113]. It was abandoned for a method using a single transtibial prosthetic user in the *Gait Pilot*. Given the variability of the results using the *Gait Pilot* (as evident by the large SDs) it was clear that the method was not going to work for the intended purposes. It was then decided to evaluate the FJC method, but it was necessary to first assure the method was valid by testing it on a rigid structure with a known rotational axis, and to decrease the inter-trial variability as evident in the SD.

Validation of FJC method on rigid model

The FJC method was tested in the *FJC Validation* and had much more encouraging results than the Reauloux Method. The validity of the method was ensured by the congruency between the known joint centre (reflective marker) and the FJC's coordinate positions matching. The variability was also improved as evident in the marked reduction of the inter-trial SD (Table 6).

The mediolateral results for the FJC

Early in the investigation it became apparent that the frontal plane position of the FJC (Figure 31) meant that the use of the three-dimensional position of the FJC was not possible. Recommendations for RoM in calculating the FJC state that an angular excursion of *at least* 5 degrees is necessary as the error reached acceptable levels at this magnitude, with the slope of the error reaching a plateau at approximately 20 degrees [36, 113]. This RoM is greater than the RoM attained in a prosthetic foot and may explain the variability. In hindsight, this *axis* formed by the variability (Figure 31, arrow) may be useful in defining an axis of rotation. It is similar in form to the method employed more recently by Sawers and Hahn [114]. They investigated the trajectory of the centre of rotation of ESAR prosthetic feet. Their results validate the results of *Study II* in that the position lies anteriorly and inferiorly to the anatomically based ankle markers. Though, they showed that over the course of the stance phase the position deviated substantially in both the horizontal and vertical directions. In their conclusion they state that their method, finite helical axis (FHA) method is potentially discouraging to those working in gait analysis of prosthesis users as it requires custom algorithm programming and implementation. The FJC method is far less resource demanding and incorporates the mode of multiple FHAs instead of a single FHA as the previous researchers incorporate. Rather than defining one FJC for all motion data, one could specify a range of motion for which the FJC would then be based upon such as specific phases of single-limb stance. This would result theoretically in subsequently greater error (variability) of each FJC as the RoM is decreased [36, 113]. Though, the benefits of the more accurate FJC position representing a specific phase of the gait cycle may outweigh this. Further research is warranted to address this question.

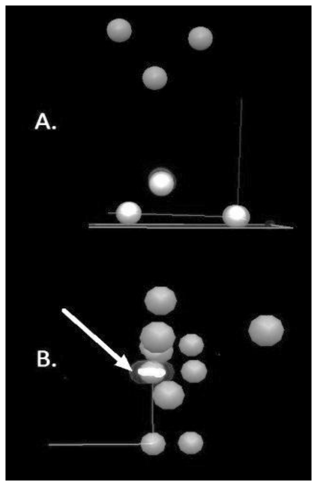


Figure 31 - Visual representation the FJC method on the rigid model during validation. Top (A) shows the sagittal view and bottom (B) shows the transverse view. Image taken from Visual 3D (C-Motion inc., USA). Image modified by the author to make markers more visible.

What does the FJC position say about *real* motion?

One can understand a great deal about the *general* pattern of motion of a prosthetic foot by looking at the position of the FJC. This is explained graphically in Figure 32 [115]. Although the given example uses vector addition to illustrate the point, it is still useful for visualization. What is shown is that as any object rotates in relation to another it is possible to calculate the centre of rotation based on the relationship of rotation and translation of the object. In the image, when pure rotation occurs the centre of rotation of the wheel will be at the surface at which it is rotating over. When the wheel begins to have an element of translation, for example slipping over the surface, the centre of rotation then moves superiorly such that it lies within the object. If the object had 100% slip and no rotation over the surface, the centre of rotation would be at the geometric centre of the wheel.

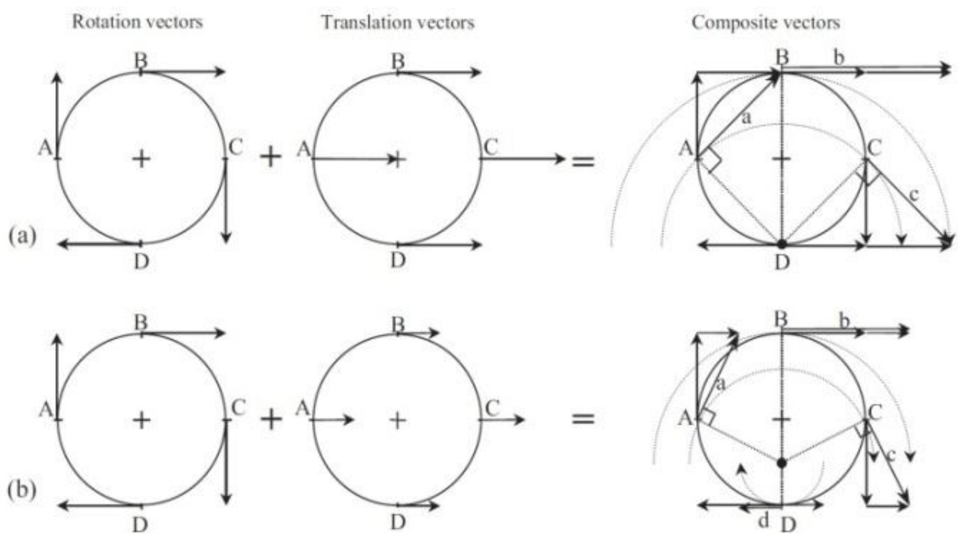


Figure 32 - Graphical representation of how the relationship between rotation and translation of an object can affect the two-dimensional position of the ICR. This method uses vector addition, different to the FHA method used in the FJC algorithm and image is only for visualization. Image from: Moorehead et al. [116].

What do the results say about the motion of the prosthetic foot?

One can use the illustrated point of the previous section to discuss how this would relate to the results in *Study II*. In the current investigation the stationary segment was defined as the foot, with rotations of the shank segment in relation to it. As such, if the FJC is in a proximal position (closer to the knee) the motion of the shank contains, in general, more rotation than if it were located more distally. Whereas, if the FJC is located distally, movement of the shank can be described as more translational over the foot (Figure 33).

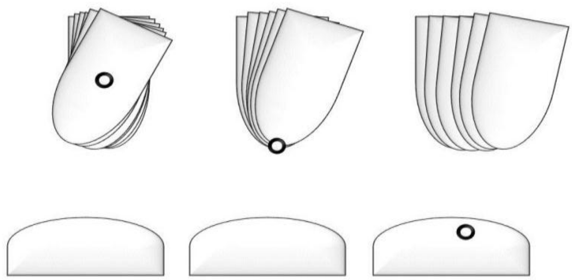


Figure 33 - The position of the FJC can be used to describe the *average* motion of the shank segment in relation to the foot. From left to right, a FJC which begins proximally and moves more distally is presented. On the left the shank rotates about a point contained within itself. Whereas to the right, the shank rotates about a point located in the foot (indicating the shank remains more horizontal and makes a linear shift over the foot).

Implications

In the outset of *Study II* the primary interest was to investigate how a prosthetic foot moves, in an effort to establish the best place to position reflective markers for motion analysis involving a prosthetic limb. The results suggested that although the reliability of the data gathered was adequate, the validity of the results was questionable. For this reason, further use of ankle kinematics of the prosthetic limb was ruled out as a dependent variable in *Study III* and *IV*. Recently other researchers have begun to explore this area in more detail [114] and the results tend to support the decision to discontinue the use of describing prosthetic foot kinematics based on rigid-body assumptions until further understanding has been gathered.

Study III

The aim of *Study III* was to investigate if vibratory feedback from under the prosthetic foot could be used to improve postural stability in transtibial prosthesis users. The results suggest that, in the methods used in this study, vibratory feedback can improve certain aspects of postural stability, whilst having a potentially negative effect on other aspects.

Mediolateral results and potential causes

The results suggest the use of vibratory feedback as utilized in the system tested does have some very encouraging effects, particularly in the reaction times of the LoS test. Though, there are some effects of the system which warrant caution and further research is necessary before a similar system is evaluated outside of a laboratory setting. The fact that the participants had decreased mean reaction times in the LoS test is an encouraging result; but that they had increased mean mediolateral excursion is potentially problematic. The postural control mechanisms for mediolateral and anteroposterior directions are different [45, 46]. In quiet stance anteroposterior movement is mainly controlled by the ankle, whilst mediolateral movement is controlled by the hip and trunk. It is possible that the addition of mediolateral feedback, something not mediated by the ankle, made control in the mediolateral plane more complicated for the participants. Huffman et al. [116] was able to show that feedback in the mediolateral direction decreased the amplitude of mediolateral trunk angle excursion, while increasing the velocity of trunk angle excursion. This is in contrast to the results in *Study III*, and may be due to the feedback modality. In the previously mentioned study, the feedback received indicated the trunk angle of the participant [116]. In *Study III* the participants received information about mediolateral pressure distribution under the prosthetic foot, but corrections came from more proximal structures at the hip and trunk. It is possible that this was actually somewhat of a perturbation, or distraction for the participants as the control mechanism (hip and pelvis motion) and feedback modality were not matched appropriately.

Relative risk of the increased mediolateral shift

The results of the mean of the AMP_{ML} variable in *Study III* was 2.04 cm with vibration and 1.62 cm without vibration. These are smaller in amplitude to the results of Norris et al. [42] for older “high-risk” fallers (2.26 cm) though greater than the “low-risk” fallers (1.08 cm). The mean difference between vibration and no-vibration for this study was 0.42 cm for eyes-open/stable condition (directly comparable to Norris et al.) and the mean of all conditions was 1.01 cm. The eyes-open/stable condition is less than half the difference between the low-risk and high-risk groups in the previously mentioned study (1.18 cm). One must therefore weigh the results of the statistical analysis against the potential clinical significance of the results. Persons with TTA have, at baseline, an increased risk of falling. That they show results on the high-end of the spectrum is not surprising. They have increased measures of many variables related to postural stability. The question in this instance is: does the addition of vibration place them at increased risk of falling? I think it could be argued based on the amplitude results, and the small differences observed between the vibration and no vibration conditions, the answer is ‘not likely’.

Mediolateral results not the same between Standing Balance (SB) and Limits of Stability (LoS)

That there was improvement in the LoS reaction time is not entirely consistent with the results seen in the AMP_{ML} variable. The reaction time composite scores used for analysis in *Study III* are themselves the combined results of multiple directions. When separated by direction they show a direction specific result indicating improvements anteriorly and towards the prosthetic side (Table 7). If the participants have improvements in the lateral direction (towards the prosthesis) this means they either benefit from the mediolateral feedback, or are able to make use of the anteroposterior feedback in executing a mediolateral shift of the CoM. In order to answer this question future research should separate the mediolateral feedback and anteroposterior feedback to see if there is an interaction effect between feedback direction and CoP excursion.

Directional control of postural shifts

Though not included in the primary analysis, following completion of the study some exploratory analysis was conducted on the direction specific reaction times. The reaction times were separated into component directions for analysis. Anterior and posterior remain the same in the presented data (Table 7). Though, right and left are reorganized to view shifts towards the prosthesis and towards the intact limb. The additional results show the prosthesis users respond slower in all directions when compared to similarly aged individuals, though the largest improvements with vibration come in the anterior direction, and in the direction of the prosthetic limb (Table 7).

	Anterior	Posterior	Intact	Prosthesis
RT with vibration (s)	0.881 (0.299)	0.846 (0.306)	0.899 (0.301)	0.849 (0.280)
RT without vibration (s)	1.046 (0.281)	0.897 (0.366)	0.961 (0.295)	1.023 (0.351)
Mean difference (s) (VIB-NOVIB)	*-0.165 (0.318)*	-0.050 (0.407)	-0.062 (0.251)	*-0.174 (0.397)*

Table 7 - Mean and (SD) for the reaction times split by direction. All times in seconds (s). The Anterior and Prosthesis mean differences and (SD) are italicized for emphasis only as no further statistical analysis was conducted on the data.

Ability to coordinate rapid postural shifts delayed

Few studies investigating LoS have reported reaction time composite data. Two sources of comparative data that exist are the normative data from the Neurocom System [40] and a publication by Nolan et al. [117]. Taking the normative data from the Neurocom system it is clear that regardless of the addition of vibration the sample of prosthesis users in *Study III* responded slower than a sample of similar age. Using the composite scores for comparison the sample in *Study III* responded with similar times to a sample of healthy individuals between 60-69 years (0.90 s [SD 0.36]) with vibration, and similar to individuals between 70-79 (1.05 s [SD 0.37]) years without vibration (*Study III, Table 3*). Using the data from Nolan et al. it is clear the experimental group in *Study III* had reaction times of the magnitude of individuals between 70-79 years (0.87 s [SD 0.30]), regardless of vibration condition.

The importance of directional resolution of feedback

A possible explanation why the participants demonstrated significant improvement could be linked to the results of Isakov et al. [67]. These authors suggested that the ability of a prosthetic user to elicit a quick shift of their CoM was mediated by the sensation from the bottom of the foot giving an indication of where the CoM was, and in turn, needed to be at the end of the postural shift. This is a similar type of motion as executed in the LoS test in *Study III*. It is possible that the feedback from the vibrating tactors provided just enough information to the participant about where their CoM was in relation to the BoS. This may have provided the advantage that made the reduced RTs evident in the results. Sensation from the bottom of the foot has been shown to be linked to postural stability [118] and to be able to respond to dynamic situations [97, 119]. The ability of transtibial prosthesis users to utilize this sensory feedback in rapid voluntary responses is encouraging. As no collection of real-time data regarding the operation of the device was done, it is not possible to link the characteristics of the vibratory signal to the improvements. Future research should focus on identifying how the signal characteristics could further influence each variable.

Implications

The use of vibratory feedback has a significant effect on how quickly a person with a transtibial prosthesis can coordinate a rapid postural shift. This postural shift is the result of appropriate sensorimotor coordination to execute the movement in the intended direction. An understanding of how participants respond with EMG would help to further explain the contribution of the sensory component in this sensorimotor coordination for transtibial prosthesis users. *Study IV* aimed to investigate this by exposing a group of transtibial prosthesis users to perturbations which elicited rapid muscular response.

Study IV

The aim of *Study IV* was to investigate the role of limb-position and weight-bearing on EMG response latency following support surface rotations in the prosthetic limb and the intact limb of persons with TTA. The results show the TTA-group had delayed responses both in the intact limb and the prosthetic limb (Figure 27). These delays are in both the toes-up direction and the toes-down direction.

EMG response latency delayed in a number of muscles

For the intact limb there were delays in the EMG latency of the gastrocnemius muscle in the TTA-group in the toes-up direction irrespective of weight-bearing or limb-position (TTA-group=182 ms, Control-group=116 ms). This is of interest as the intact ankle is anatomically the same in both the TTA-group and Control-group, indicating some external influence. Postural adaptations [65] and known bilateral sensory changes as a result of being a person with an amputation [64, 120] could help to explain the results of *Study IV*.

In the toes-down direction there was a significant interaction effect between group and limb-position which indicated the TTA-group responded slower than the Control-group in the vastus lateralis muscle in the toes-down direction when the support surface rotation was only received through the prosthesis (TTA-group=195 ms; Control-group=126 ms). This clearly indicates that there is an effect of the prosthesis on the EMG latency, but the cause of this could be multifactorial. The delayed response could be the result of local joint stimuli [97, 100, 102] due to the known mechanical constraints of the prosthesis [121]. Alternatively it could be due to delayed motion of the CoM [95]. Future research should address the question of whether it is a local joint feedback interaction or related to the task-variable of maintenance of the CoM in a position of stability that causes the delayed reaction.

The TTA-group also had delayed EMG response latency on the side with a prosthesis with the biceps femoris muscle responding slower in the toes-up perturbation (TTA-group=180 ms, Control-group=129 ms). As with many transtibial prosthesis users it is customary to 'pre-flex' the prosthetic socket as part of a standard prosthetic alignment procedure. This is done to increase the loading area within the socket and to pre-load the knee-extensors to increase their effectiveness in eccentrically controlling knee flexion during loading response in gait [95, 122, 123] (Figure 34). Knee-flexion has been shown to have an influence on automatic postural responses, including increased latency times of some muscles, in toes-up support surface rotations [107]. This fits the results of *Study IV* and future studies could exclude this confounding factor by producing individual prostheses for each participant that could be aligned without the previously mentioned pre-flexion. This would not necessary reflect a real-life situation, but would allow investigation of the effect of pre-flexion.

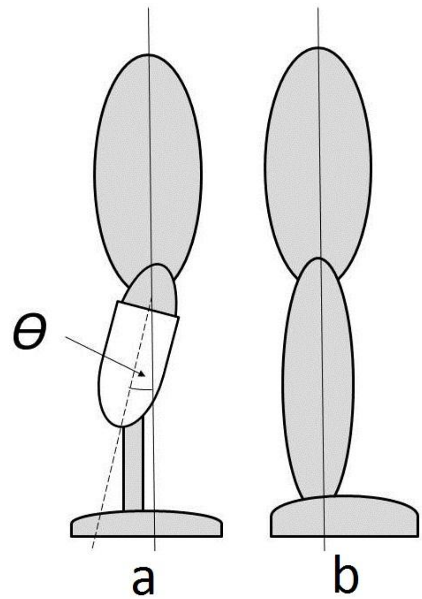


Figure 34 – Difference in the *neutral* position between the side with a prosthesis (a) and the intact limb and control limbs (b) in the sagittal plane. The angle of pre-flexion (*ϴ*) of the prosthetic limb means the neutral flexion angle is greater than the intact limb of the TTA-group and Control-group. Solid line represents vertical reference line, dashed line represents long-axis of prosthetic socket.

The first EMG response (if present generally in the ankle musculature) after a perturbation is the stretch reflex (SR). This can occur as early as 40 ms, but is of little functional significance in postural stability as the SR's main purpose is maintenance of stiffness of associate joints [124]. There were very few EMG responses that came prior to what is considered the borderline for automatic postural response ($< \approx 100$ ms) [91]. Most occurred in the Control-group during increased weight-bearing. However, the TTA-group did have some which came close to this threshold. The responses in the vastus lateralis and tibialis anterior muscles in the toes-down direction (130 ms and 137 ms, respectively) indicated quite rapid responses to the perturbation. One must consider why the gastrocnemius muscle did not have a similar response. As the pooled-group analysis includes the trials in which both feet are on the platform, it is possible the prosthesis caused a delay in the response from the intact limb. Similar to the discussion earlier regarding the mechanical characteristics of the prosthetic foot and local joint stimuli, it is possible the prosthetic limb causes sufficient dampening of the rotational perturbation that the stimulus is not transferred to the intact limb at the same rate as in the Control-group. The prosthetic foot causes a coupling delay between the support surface and the residual limb of the prosthetic user, causing a delayed physiological response. Nashner and Cordo [94] stated that automatic postural responses typically occur up till ≈120 ms in the ankle, followed by muscles of the thigh ≈10-30 ms after those at the ankle. The *voluntary* response to the perturbations occur from ≈180 ms at the ankle, and up. This means that for the TTA-group there is a possibility that the automatic postural response is largely absent and their responses were, in general, more voluntary. Regardless of the mechanism, the presence of a delay in the EMG response latency to perturbation is indicative of an individual's ability to respond to a balance threat and prevent a fall [96]. This may well indicate that under situations where pitch plane perturbations are elicited, individuals with a TTA are at increased risk of falling when compared to able-bodied individuals.

Temporal organization of lower-limb response

Although not part of the original analysis, it is useful for visualization to plot the EMG latency times. In figure 35 the intact and prosthetic limb of the TTA-group and the Control-group are plotted on their own lines. Best-fit lines have been added to help with visualization. It is clear that the TTA-group have on average slower reaction times as the best-fit line is shifted upwards by 15-20 ms. Yet, this move is almost completely parallel to the Control-group. This would seem to fit the temporal pattern of response evident in what is called the “ankle” strategy [91]. Both groups show a slight increase in the EMG latency times in the proximal muscles indicating a similar pattern of latency times, just greater in magnitude for the TTA-group. This would suggest that, in the intact limb, the TTA-group have a similar *pattern* of temporal response but one that is universally delayed across the muscles of the lower-extremity. A similar pattern was seen in the muscles on the prosthetic side.

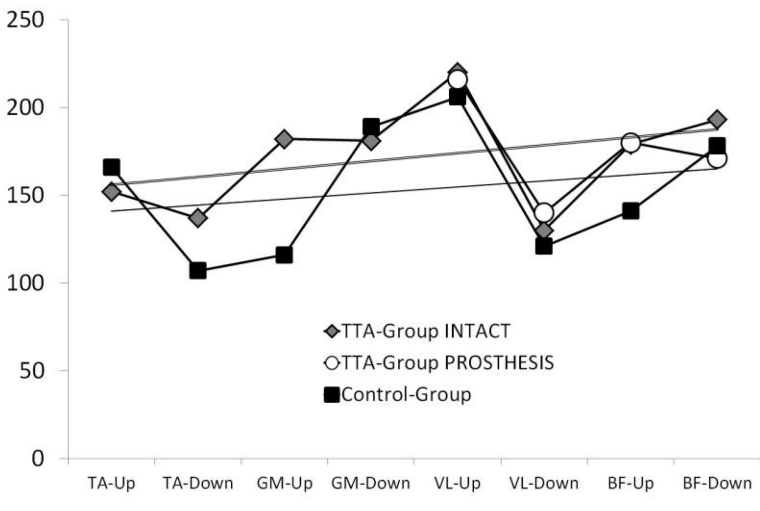


Figure 35 - Individual EMG response latencies for each muscle from distal to proximal (left to right) for TTA-group and Control-group. Best-fit line only for reference between intact limb of TTA-group and Control-group. TTA-group responded slower in all cases but the slope is the same for both groups.

Incongruence between perturbation and actual movement

A link between *Study I* and *Study IV* of interest is the incongruence between the mechanical axis of the force platform and the FJC of the prosthetic feet (Figure 36). This incongruence means that rather than a purely rotational perturbation centered about the ankle there is an element of linear translation in the perturbation when received on the prosthetic side. This is not the case on the intact side. This poses the question of whether this could have influenced the results. As the FJC across all feet was positioned anterior and inferior to the mechanical axis of the platform it is possible to describe the *real* perturbation. In the toes-down perturbation, the TTA-group would have experienced an element of linear translation in the sagittal plane, but not horizontal to the support surface. In fact they would have experienced a rapid drop and backward shift of the support surface, in combination with the rotation of the support surface. Conversely, in the toes-up rotation, the TTA-group would have experienced a rapid rise and anterior shift of the support surface in combination with the rotation.

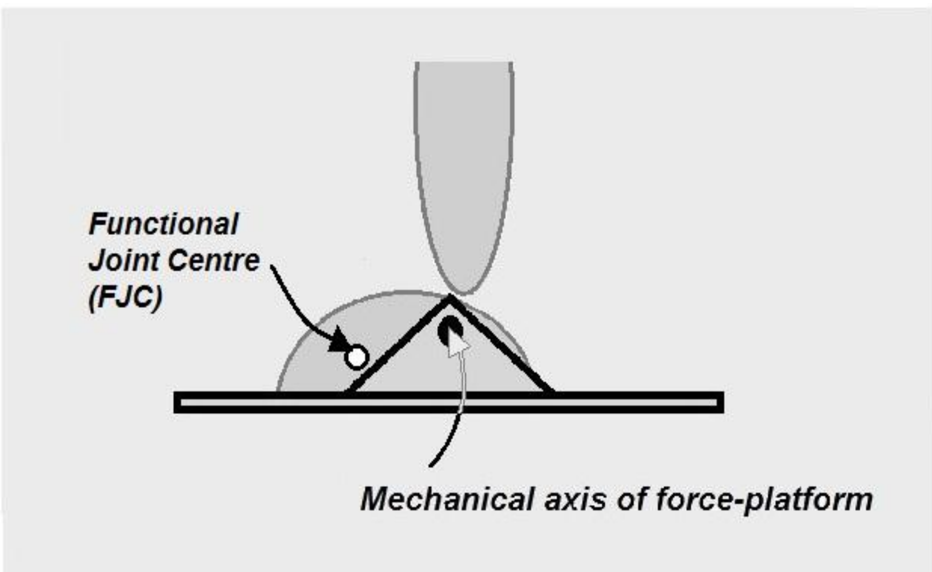


Figure 36 – Graphical description of the incongruence between the intended mechanical rotation of the support surface and the calculated FJC shown in *Study II.*

Method of platform perturbation

For safety reasons, no fastening of the foot to the platform was done in the assessments in *Study III* and *IV*. If this was done the knee on the prosthetic side would be subjected to 16 degrees of hyperextension in ~140 ms (assuming otherwise rigid segments) (Figure 37). Fastening the foot to the support surface would have created a potentially dangerous situation for the participants. Even without anchoring the foot, one participant requested to stop the data collection during the increased weight-bearing condition when only the prosthetic limb was on the forceplatform. Another participant commented the dorsiflexion movement was very difficult to overcome and likened it to "going for a ride".

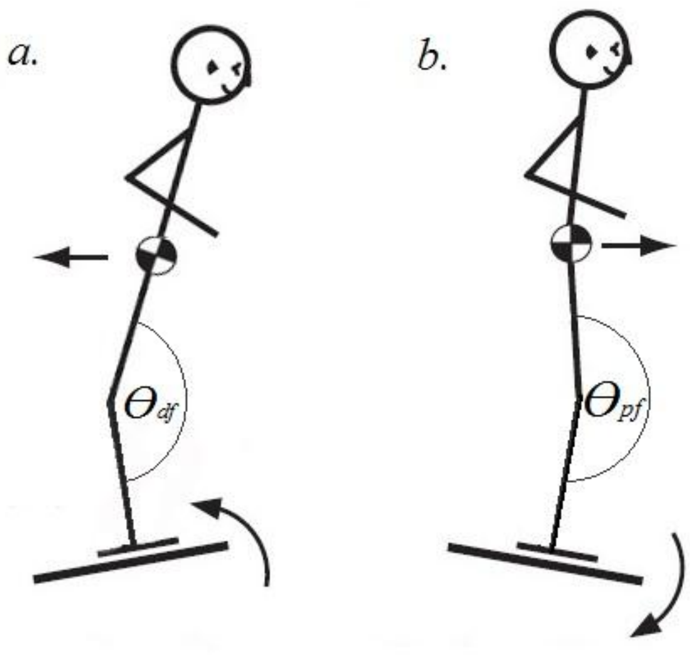


Figure 37 - Maximum angles of angular excursion of the knee following perturbation in the toes-up (Θ_{df}) and toes-down (Θ_{pf}). During the toes-up condition Θ_{df} could be reached by ≈ 140 ms. Image modified from [94].

Weight-bearing distributions

Utilizing the 25-50-75% weight-bearing distribution is very specific and may not have been an accurate reflection of a *real-life* situation. Although asking the TTA-group to assume their self-selected postural distribution may have increased the external validity of the results, this would have increased the complexity for the Control-group. Using the matched distributions made execution of the study and comparison of the two groups easier to accomplish. The methods could have documented the self-selected distribution of each participant in the TTA-group and then dictated this as the goal of the Control-group. In this way the external validity of the results would have been maximized for the sample in this study. Self-selected weight-bearing distribution could be a useful addition to the methods in the future.

Cutoff frequency used in filtering

The decision to use a low-pass filter with 100 ms as the cutoff frequency was based on previous works involving postural perturbations [88, 93, 107] including one study involving prosthesis users [60]. This may have resulted in a portion of the high frequency component of the EMG signal being removed due to the filtering. Winter [125] proposed a cutoff frequency at 200 Hz. Post-processing analysis was conducted to produce two data sets from the same individual in order to investigate the potential influence of this on the results of this study. One data set was digitally low-pass filtered using a fourth-order Butterworth filter with a cutoff frequency of 100 Hz, and a second data set with 200 Hz. The mean difference was then calculated for the paired data sets. The mean difference between the data sets was a 14 millisecond delay in the data set with a cutoff frequency of 100 Hz. Had the methods used 200 Hz as per Winter [125] the overall latency times most likely would have been 14 ms smaller for both the TTA-group and the Control-group, representing a systematic shift in the data, not an error affecting the significance of the statistical analysis. The SD of the 200 Hz data set (87 ms) was smaller than the 100 Hz data (95 ms), suggesting the statistical interpretation based on the 100 Hz data set was in fact the more conservative of the two. If one considers the results of *Study IV*, it is possible that in the toes-down direction both the vastus lateralis muscle of the limb with the prosthesis (*Tables 2 and 3, Study IV*), as well as the tibialis anterior of the intact limb would have been found significantly different as they could have been considered *borderline* variables. This is of course only conjecture, but to base the conclusions on more conservative statistical analysis could be argued more prudent than the alternative.

Platform Movement Onset

Identification of platform motion was accomplished using the raw data of one marker located at a distance of 30 cm from the mechanical axis of the force-platform. Choosing not to filter the data was done so as to include any background noise present in the marker position signal in the overall movement. Automated methods for determination of threshold crossing are used extensively in EMG analysis [126] and were utilized in this investigation. This method was further utilized in the determination of the platform movement onset. Given the known effect of length of the moving window, filtering frequency, and threshold (typically some multiple of the SD of background activity) in EMG studies [126], it was necessary to understand how this may affect the results of onset using coordinate data. The validation of this method was accomplished by first considering the likely component of noise in the signal. As the platform was completely stable, it is unlikely there was a large component of low-frequency noise in the signal. This leaves the high-frequency component to be removed [4]. In a preliminary validation, the data was processed into two data sets. One in which the onset was identified using raw data and one in which it was identified using data that had been low-pass filtered using a fourth-order Butterworth filter at a cutoff frequency of 150 Hz. The results showed for 99 trials (one participant, all trials) the mean difference to be 0.000 ms. At a resolution of 2 ms, there was not one value in the 99 trials that showed a single-digit difference.

Postural coordination and automatic postural response

Study III and *Study IV* clearly show a relationship between being a prosthetic user and having delayed rapid postural and neuromuscular response. It is interesting to consider the link between the two studies. The mean difference between the vibration condition and no-vibration condition in *Study III* was 113 ms. How much of this is the EMG response latency reduction (which itself is ≈ 150 ms) and how much is the coordination and execution of movement is something that is currently unknown. As in *Study III* there was not simultaneous collection of EMG signals so it is not possible to answer this question. If the LoS test was conducted whilst simultaneously collecting EMG data it would have helped to answer if this improvement was due to quicker muscular response or more effective postural response. Future research should address this question.

Implications

The results of *Study IV* suggest that transtibial prosthesis users have delayed EMG response latencies of multiple muscles of the lower-extremity following support surface rotations. This may place them at increased risk of falling as a result of inadequate ability to respond to external balance threats. Future research should identify if there are properties of the prosthetic limb that can decrease these latencies, thus allowing prosthesis users to respond faster to external balance threats.

13 LIMITATIONS

Study I

As part of the literature search EMBASE database was not included in the literature search. This may have prevented relevant articles from being found.

Study II

A limitation with the study, as it was presented, was the definition of the sagittal plane. In the published study it is not clearly defined and due to this oversight it is possible a reader could misinterpret the orientation of the y-direction. A clarification of this is provided (Figure 38) in which the axis-ϕ designates the calibration axis used to position the y-direction frame in the inferosuperior orientation. This would also have a systematic effect on the magnitude of the results. As the participant served as his own control it is also possible there were confounding effects of the contralateral limb.

The study design was effective in testing a method, but case study design limits the generalizability of the results.

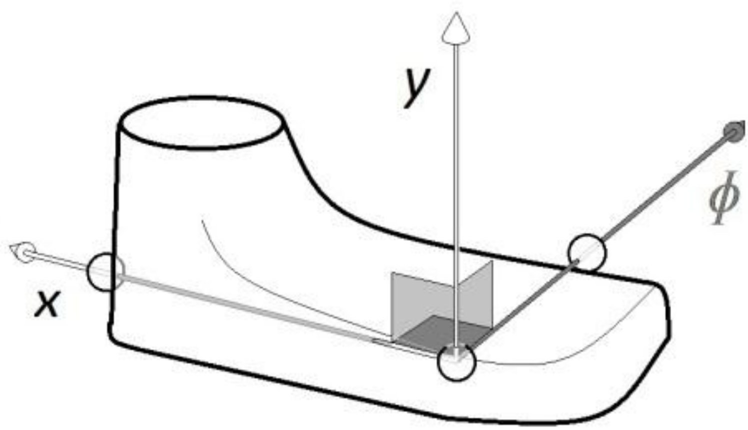


Figure 38 - Corrected three-dimensional frame of reference for the limb-based coordinate system, where x and y designate the two orthogonal planes extracted for analysis, and phi (ϕ) designates the calibration axis used for the other two planes, but not used in further analysis.

Study III

A relevant and potentially informative piece of information would have been the description regarding operation of the vibration device. If simultaneous data collection documented when the device was on, and its amplitude, it would have been possible to explore if there was a link between device operation and the documented improvements. Future research should include this variable in the data collection. Additionally, the participants were otherwise healthy prosthesis users, mostly male, making the generalizability of the results to the wider population of transtibial prosthesis users questionable.

Study IV

The choice of 25-50-75% weight-bearing may not directly reflect *real-life* situation, affecting the external validity of the results. As with *study III*, due to the sample, generalizability of the results to the wider population of individuals with TTA due to vascular disease is questionable.

Due to the focus on the lower extremity musculature there may have been reactions in the upper extremity which were not recorded, such as potential reactions in the muscles of the spine, abdomen, neck and arms. The conclusions can only be applied to the reactions of the lower extremity.

General Limitations

For *Study III* and Study *IV* there are characteristics within the experimental group that one may consider influential on the results. In neither study were the participants separated by residual limb length, or classified by age. The experimental group in both studies was allowed to wear their currently functioning prosthesis which eliminates some potentially confounding factors, such as accommodation time, but raises others, such as the influence of the different prosthetic feet and form of prosthetic suspension.

14 CONCLUSIONS

Study I

The level of evidence within the reviewed studies was generally low, though increasing in more recent publications. Quality of the studies is not increasing at the same rate as the level of evidence. Due to methodological differences comparison of studies is difficult. A number of methodological problems in the reviewed studies can be addressed by making a small number of methodological changes in future studies involving kinematics of transtibial prosthesis users such as the inclusion of a control group when appropriate.

Study II

Prosthetic feet have different FJCs from each other, in addition to that of an anatomical ankle. Reliability of the FJC method is adequate to justify continued use.

Study III

The use of vibratory feedback as provided by the tested system caused increased mediolateral CoP excursion during standing balance, but reduced reaction times in the limits of stability test. The results suggest the system evaluated may have both beneficial and negative effects on different measures of postural stability.

Study IV

Transtibial prosthesis users have delayed EMG response latency times in muscles of both the intact and prosthetic limb. These delays were in the intact limb for both toes-up and toes-down direction, whereas the prosthetic limb was only delayed in toes-down direction. Limb-position influenced latency times in the intact limb, but not for the prosthetic limb, indicating unilateral compensation when the perturbation was received through the prosthesis.

15 FUTURE PERSPECTIVES

At various points in time during the completion of this thesis additional questions came to the author's attention. These questions were often the result of contemplating the importance of the results and how they were related to previous research. A summary of some of the questions and future research related to each study are:

- How sensitive is the FJC position to factors not related to the prosthetic foot (external validity)? For instance the individual's weight or the velocity of the individual during testing?
- What is the effect of the FJC position on further derived variables such as joint torques and powers?
- Is there a clinical relevance with the FJC? Can it be subjectively assessed by prosthesis users in relation to comfort or ease of ambulation, or is it simply a laboratory based variable with application in quantitative research?
- What is the quantifiable difference between the use of various kinematic models when investigating prosthesis users?
- Are the reaction time improvements seen in *Study III* a result of reduced muscular response times or more efficient biomechanical response?
- Can alterations be made to the feedback system characteristics to reduce the negative effect on the AMP_{ML} variable, whilst maintaining the positive effect on reaction times?
- Do the reaction time improvements due to vibratory feedback found in *Study III*, have the potential to reduce EMG response latencies to perturbations such as those in *Study IV*?

16 IN SUMMARY

One could summarize the aims of this thesis in two parts with the following statement: The aims of this thesis were to evaluate how persons with a transtibial amputation *move* and how they *sense*. These two statements encompass the two major subjects covered, motion analysis and postural stability.

In the context of *moving,* three-dimensional motion analysis was used as an evaluation tool. The results from the systematic review in *Study I* suggest researchers cannot be sure of how they *move* because the methods vary so much between studies. The results of *Study II* suggest that if the same biomechanical models are used as in able-bodied individuals, a prosthetic foot and ankle *moves* very different to an intact ankle.

Various tests of postural stability were used to investigate how transtibial prosthesis users *sense* with a prosthetic limb. The results of *Study III* suggest they may utilize vibratory feedback in an open-loop response to fast movements of the body. *Study IV* suggests they have sensory deficits as a result of the prosthesis, and that this not only affects the side with an amputation but also the intact limb.

This thesis has clearly shown that if researchers want to use rigid-body assumptions on foot and ankle segments as a whole, they must be aware that a prosthetic foot behaves differently to an intact foot. These differences are of sufficient magnitude to warrant discontinued use of ankle kinematics as a variable in research within this group. There also must be an effort to propose a consensus on methods for collection and reporting of kinematic data of transtibial prosthesis users, as current methods restrict valid comparison between researchers.

This thesis has also shown that transtibial prosthesis users are able to use vibratory feedback to improve reaction times for rapid shifts of the centre of gravity. Additionally, the thesis has shown that individuals with a unilateral amputation have delayed EMG response latency not only in the prosthetic limb, but also in the intact limb. This delay may indicate a bilateral functional consequence as a result of being a person with a unilateral transtibial amputation.

SAMMANFATTNING PÅ SVENSKA

SYFTET med avhandlingsarbetet var att kritiskt granska och utvärdera de rörelseanalysmetoder som används vid studier av personer som använder transtibial protes samt att ge förslag till metodförbättringar. Ytterligare syfte var att utvärdera om vibratorisk feedback kan förbättra den postural stabiliteten hos individer som använder transtibial protes och att undersöka hur muskulära respons vid en plötslig förändring av underlaget påverkas av att vara protesanvändare .
MATERIAL OCH METOD I *Studie I* utfördes en systematisk granskning av 68 vetenskapliga artiklar innefattande kinematisk analys av försökspersoner med transtibial protes. *Studie II* undersökte rörelse hos protesfötter enligt en metod för att hitta positionen för funktionellt ledcentrum. *Studie III* utvärderade påverkan av vibratorisk feedback på postural kontroll hos 24 försökspersoner med transtibial protes. *Studie IV* undersökte hur protesen påverkade tiden för den muskulära responsen, mätt med EMG, i det intakta benet respektive det amputerade benet hos 23 försökspersoner med transtibial protes och jämfört med en matchad frisk kontrollgrupp (n=23).
RESULTAT *Studie I* påvisade generellt låg evidens- och kvalitetsnivå i de inkluderade studierna samt att metodologiska problem försvårar jämförelser mellan studier. *Studie II* visade att positionen av det funktionella ledcentrumet skiljer sig åt både mellan olika protesfötter och jämfört med en intakt fot. *Studie III* visade att vibratorisk feedback, baserat på belastning av protesfoten mot underlaget, medförde ökad avvikelse av tryckcentrum i mediolateral riktning och minskad reaktionstid vid snabba förflyttningar av tyngdpunkten. *Studie IV* påvisade fördröjd muskulär respons i såväl det intakta som det amputerade benet hos individer med transtibial protes. Fördröjd muskelrespons påvisades också i det intakta benet när rörelsen av underlaget mottogs via protesbenet.
SLUTSATS Metodologiska problem gör det svårt att tolka kinematisk data för individer som använder transtibial protes och rörelsemönstret skiljer sig åt både mellan olika protesfötter och jämfört med en intakt fot. Vibratorisk feedback kan förbättra vissa aspekter av postural stabilitet och den automatiska posturala muskelresponsen är långsammare hos individer med transtibial protes jämfört med frisk kontrollgrupp. Dessa fynd kan bidra till ökad förståelse av hur studier av rörelseanalys hos individer med transtibial protes kan utformas samt hur denna grupp upprätthåller postural stabilitet med protes.

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