Gait stability and Balance Strategies of both Acquired and Congenital Lower Limb Prosthetic Users in Response to Perturbations

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ABSTRACT

Falls are a health care problem for lower limb prosthetic users. The study of gait stability in lower limb prosthetic users facilitates improved insight and knowledge in different adaptation strategies of the human body in order to walk as functionally as possible with a prosthesis. The aim of this thesis was to determine how prosthetic users cope with unbalanced situations during walking and how these coping strategies may differ from able-bodied individuals. Improved understanding of such mechanisms may help reduce fall incidence. A number of prosthetic factors were considered including the use of a prosthetic foot incorporating an ankle joint, compared to a conventional prosthetic foot. Additionally, the effect of different alignments and the aetiology of the amputation or absence (congenital vs acquired amputation) was also considered.

The study was conducted using an advanced dual-belt instrumented treadmill (CAREN). The protocol of perturbations in the study was adopted from a previous work by a group of researchers in University of Strathclyde (Roeles et al., 2018). Interventions used were anteroposterior (AP) perturbations by means of sudden changes in the walking speed to mimic a slip that can be faced in real-life situations. Main Outcome Measurements measured were AP and ML margins of stability (MoS) Hof et al. (2005). Step length, width and time were also measured to investigate the coping strategy following perturbation.

Prosthetic users were less stable than able-bodied individuals. The involvement of the prosthetic side to recover stability was limited therefore, during rehabilitation stability training tasks for the intact side may help the prosthetic users enhance their overall stability and may reduce the fall incidence rate. Energy storing and return prosthetic feet may provide a sufficient level of stability compared to the feet which incorporate a moving ankle mechanism. The Ossur

Pro-Flex foot demonstrated enhanced stability in the AP direction. Alignment changes from the optimal alignment may impose extra challenge to the stability. A short prosthesis was found to be the most challenging alignment change in response to perturbation. The prosthetic user with congenital related limb anomaly was found to be more stable than the prosthetic users with other lower limb loss.

The outcomes of this study are novel and have potential to improve the understanding of how prosthetic users (acquired and congenital) react in when stability is compromised and the variables which may affect this further (foot design and alignment). It is envisaged that greater understanding of different adaptation strategies of the human body may help influence future prosthetic treatment, prescription, alignment and potentially component design.

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AP	Anterior-posterior
BoS	Base of Support
CAREN	Computer Assisted Rehabilitation ENvironment
CoM	Centre of Mass
CoP	Centre of Pressure
CWS	Comfortable walking speed
HBM2	Human Body Model 2
ML	Medio-lateral
MoS	Margins of stability
RoM	Range of Motion
XCoM	Extrapolated Centre of Mass

1 Introduction

1.1 Study Overview

Rehabilitation following an acquired lower limb amputation or congenital absence is important so that a person has the best opportunity to adapt following an amputation (Ephraim et al., 2003). In the case where prosthetic rehabilitation is indicated, the aim of the prosthesis is to restore functional loss and quality of life as far as possible (KOVAČ et al., 2015). Socket design, prosthetic component selection and alignment, all contribute towards optimal gait pattern as well as reduced socket interface pressures. The use of a prosthesis may aim to promote independence and activities of daily living and also the ability to cope when unbalanced situations occur without a fall (KOVAČ et al., 2015). One of the important aspects to help in improving the quality of life for prosthetic users during rehabilitation is fall prevention (Hunter et al., 2020). This thesis therefore discusses gait stability for the prosthetic users.

Study of the gait stability of prosthetic users facilitates better insight and knowledge to different adaptation strategies of the human body to walk as functionally as possible with a prosthesis (Gard, 2006, Hak et al., 2013c). It is essential to understand of how individuals will use the provided prosthesis and what factors may affect this use such as aetiology of amputation, alignment and component design. Understanding of how such factors may contribute to stability may reduce the incidence of falls and thus help improve the quality of life for the prosthetic user whilst maximising the potential of the prosthesis to meet the needs of user.

Insight may be used to develop better rehabilitation training programmes for lower limb prosthetic users and provide the required understanding which may help when developing new components. This study will focus on prosthetic users who have amputation or absence below the level of the knee joint. To help to understand how humans maintain balance and recovery from sudden unbalanced situations, the dynamic stability of an able-bodied control group will initially be investigated. The dynamic stability of the prosthetic users will then be examined and compared to the control group in order to investigate the potential differences in recovering mechanisms.

Decision making during the rehabilitation process involves the assessment of whether an individual is suitable for a prosthesis and if so, what components to provide (Devinuwara et al., 2018).

For transtibial prosthetic users, the type of prosthetic foot may considerably affect the outcome performance depending upon activity (Houdijk et al., 2018b). Currently several prosthetic feet designs are available for all activity levels (K levels). The range of motion provided by the foot and ankle complex differs depending on design and may not exactly replicate that of the normal foot and ankle during the normal gait cycle. Prosthetic foot designs vary in function, weight, and affordability. The current study will investigate what additional function prosthetic foot used in conjunction with an ankle joint may add in comparison to more conventional designs in relation to the dynamic stability and recovering from unbalanced situations. The investigation was focused towards the potential benefits of the feet with ankle joint in reference with dynamic stability compared to 'conventional' prosthetic foot without a moving ankle joint.

In addition to prosthetic foot design, it may be argued that a further important element in the prosthetic intervention is the alignment of components. Transtibial alignment refers to the special relationship between the prosthetic socket and the foot. Incorrect prosthetic alignment may lead to several issues that can impact the dynamic stability of the prosthetic user. These include decreased gait pattern symmetry (Fridman et al., 2003); increased functional demands on the contralateral side and imposed a less comfortable walking condition for the prosthetic users (Isakov et al., 1994). In addition , the increased loading on the intact side may play a great role in developing osteoarthritis (Morgenroth et al., 2011) and finally increased pressures residual limb/ socket interface pressures (Seelen et al., 2003). All factors will be examined to investigate if they influence stability that may affect the ability of the prosthetic users to recover from unbalance situations that they may face. Therefore, this study also investigates the effect of the alignment on the gait stability. It is hoped that results from this preliminary study will help design further studies to examine specific aspects of the prosthetic prescription with respect to balance and stability.

One of the commonly promoted goals during rehabilitation is to obtain optimal intact side and prosthetic side gait symmetry (Yang et al., 2012). Several researchers have promoted this idea in their work (Sadeghi et al., 2000), whilst other have established quite the opposite (Hak et al., 2014, Hof et al., 2007). Gait symmetry for intact and prosthetic sides will be investigated in the current study particularly after the event of unbalanced situation.

While some papers have investigated the dynamic stability and recovery mechanisms for the acquired related limb loss (for example limb amputation as results of diabetes or trauma), no previous work investigated the stability and walking patterns for the congenital related lower limb absences and anomalies. This may be understandable as the rate of lower limb loss as a result of congenital cases is far lower than any other amputation cause in several developed countries. For example, in Scotland based on the Scottish Physiotherapy Amputee Research Group Annual Report the incidence of congenital deformity, 11 cases out of 714 in 2017 and there were no significant rates changes between 1 January 2007 to 31 December 2017 (Smith et al., 2019). However, in other countries the congenital limb loss still seen as one of the major causes. The results in a study of Banza et al. (2009) that has been conducted in Malawi showed that (60%) of the limb loss in children aged under 18 were congenital limb deformities. Similar results were observed in a study of Brazil (de Godoy and de Godoy, 2016) as well as in Jordan (Al-Worikat and Dameh, 2008). These results promote the need of further investigation for this group. This thesis discusses this group, including definition, classification, prosthetic management, rehabilitation and dynamic stability.

The outcomes of this study are novel and have potential to enhance the quality of life for lower limb amputation or absence (acquired and congenital). It may also help therapists and prosthetists to better understand the difference between this group and more able-bodied. It is envisaged that greater understanding of different adaptation strategies of the human body may help influence future prosthetic treatment, prescription, alignment and potentially component design.

The purpose of chapter one is to familiarise the reader with the state of knowledge regarding the concept of the dynamic stability and its importance. The literature outlined in this chapter examines the methods that have been previously adopted to study the dynamic stability, the methods used to create unbalance situations, and stability status. This review aimed to provide the justification for the current thesis. The specific aims and objectives of each chapter are presented in the end of this chapter.

1.2 Lower Limb Functional Anatomy

Understanding the functional anatomy of the lower limb is not only important to understand how the musculoskeletal system of lower limb interact with each other but also to understand gait. This includes how to describe the abnormal gait pathologies; and to understand the impact of external factors (such as weight, different types of sports, amputation, etc) on the lower limb joints (Arráez-Aybar et al., 2010, LaMont, 1986). In addition, the functional anatomy of the lower limb is a fundamental aspect in order to understand the joints' movements and biomechanics (Agur and Dalley, 2017, LaMont, 1986).

Each lower limb has 30 bones: femur, patella, tibia, fibula, seven tarsal bones, five metatarsal bones, and 14 phalanges (Agur and Dalley, 2017, Anderson, 1983, Saizar, 1981, Singleton and LeVeau, 1975). Additionally, the musculoskeletal system of lower limb includes ligaments, joints, muscles, nerves, and tendons (Agur and Dalley, 2017, Anderson, 1983, Saizar, 1981).

1.2.1 The Hip Joint

This joint can be described as a synovial ball and socket joint that include the head of the femur and pelvic acetabulum (Singleton and LeVeau, 1975). Because it is a ball and socket joint, a wide range of movements are allowed at this joint such as flexion, extension, adduction, abduction, rotation (internal and external), and circumduction. The range of motion is limited and controlled by the capsular ligaments, tibiofemoral ligaments, pubofemoral ligament, the ischiofemoral ligament, and the opposite group of muscles (Dee, 1969, Ito et al., 2009, Singleton and LeVeau, 1975). Nevertheless, when the hip joint is flexed those structures will be relaxed. Thus, hip dislocation is more likely to occur when the hip joint is flexed. In contrast, the hip joint is very strong during extension due to ligaments' tightness during extension position (Singleton and LeVeau, 1975).

The main two functions of the hip joint are body weight bearing and provide stability during activities (Singh, 2014). Even though, this joint has a wide range of motion, it is a very stable joint (LaMont, 1986). The stability of this joint comes from various aspects such as the shape of the joint (two third of spheroid), deep articular cartilage, fibrous articular capsule, structure of the femur (angle between the neck and shaft of the hip joint), tight ligaments, and muscles (Figure 1.1) (Singleton and LeVeau, 1975). All of those aspects increase the connection between the head of the femur and acetabulum in order to keep the head of the femur inside the socket tightly. Besides, those structures support the hip joint to tolerate the stress and strain during standing and walking (Dee, 1969, Singleton and LeVeau, 1975).

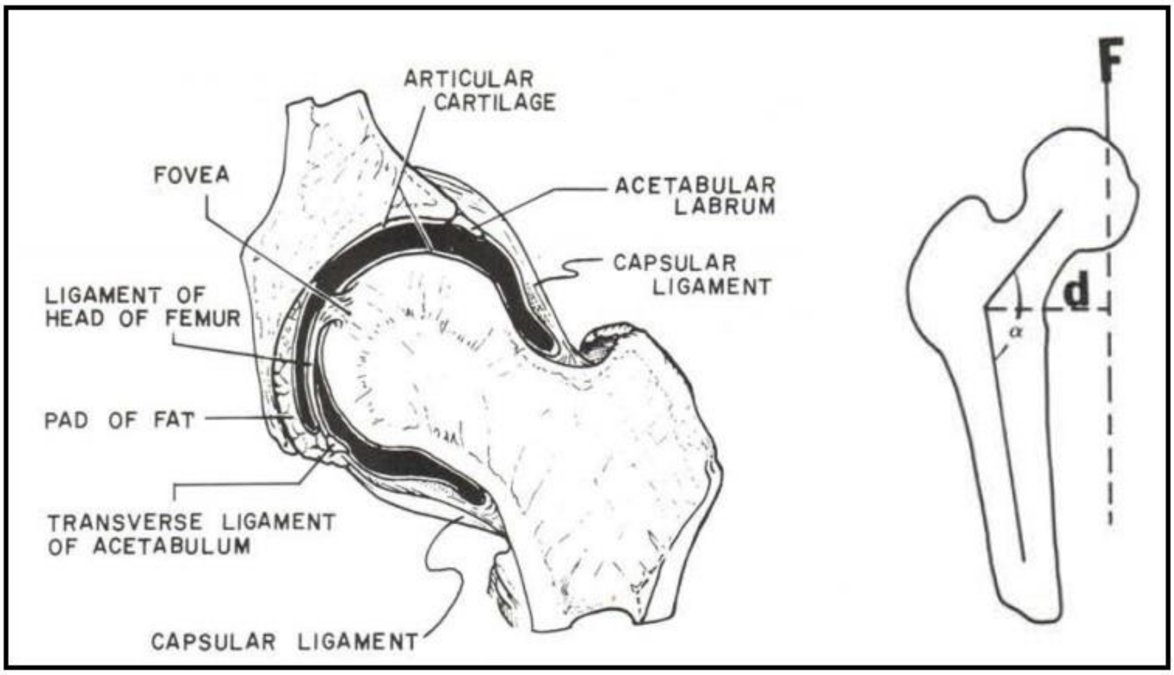

Figure 1.1 The sources of the hip joint stability (Singleton and LeVeau, 1975).

1.2.2 The Knee Joint

The knee joint is considered as a complex modified hinge joint that consists of the three joints: patellofemoral joint, medial tibiofemoral joint, and lateral tibiofemoral (Andriacchi, 1988, Andriacchi et al., 1986, Cantrell and Varacallo, 2020). The patellofemoral joint is diarthrodial joint, while the tibiofemoral joint is a hinge synovial joint. Because it is a modified hinged joint, flexion, extension, valgus, varus, and internal rotation are facilitated. Nevertheless, the knee flexion and extension movements are the primary movements of this joint (Andriacchi, 1988, Andriacchi et al., 1986)

The stability of the knee joint is critical because it is a weight bearing joint; its stability coming from muscles (quadriceps, hamstring, and popliteal muscles), ligaments, tendons, the joint capsule, and cartilages (Amis and Dawkins, 1991, Recondo et al., 2000, Watanabe et al., 1993). These components act within three bundles to resist anteromedial, intermediate, and posterolateral dislocations or injuries (Amis and Dawkins, 1991). Nevertheless, the medial side of the knee joint is the more commonly injured than the lateral side because it faces more force during walking than the lateral side (Recondo et al., 2000, Watanabe et al., 1993).

1.2.3 The Foot and Ankle Joint

The foot is a very complex structure that has 23 bones with 33 joints. During gait, the foot has a crucial function in transferring the force between the body and the ground. Therefore, all joints of the foot align together to provide the ability to walk and achieve daily activities (Bozkurt and Doral, 2006, Brockett and Chapman, 2016, Cantrell and Varacallo, 2020, Castro, 2002, De Ridder et al., 2015).

The complex ankle joint is a hinged synovial joint that includes three bones: tibia, fibula, and talus. Those three bones are articulated with each other and with other bones within three joints: the talocalcaneal (subtalar), tibiotalar (talocrural), and transverse-tarsal (talocalcaneonavicular) joint (Bozkurt and Doral, 2006, Brockett and Chapman, 2016, Cantrell and Varacallo, 2020, Castro, 2002, De Ridder et al., 2015).

The talocrural joint is responsible to provide dorsi-flexion and plantarflexion movements. The talocrural joint is a diarthrosis joint that is covered by capsule and ligaments to provide stability during walking. While, the transverse-tarsal joint is also called Chopart's joint is responsible to support inversion and eversion movements of the foot (Bozkurt and Doral, 2006, Brockett and Chapman, 2016, Leardini et al., 2014, Snedeker et al., 2012).

As a result, the stability of this joint is due to shape of the articulated bones (bracket shape or mortise), strong ligaments, muscles (anterolateral, posteromedial and lateral compartment), and capsule (Bozkurt and Doral, 2006, Brockett and Chapman, 2016, Cantrell and Varacallo, 2020, Castro, 2002, De Ridder et al., 2015) .

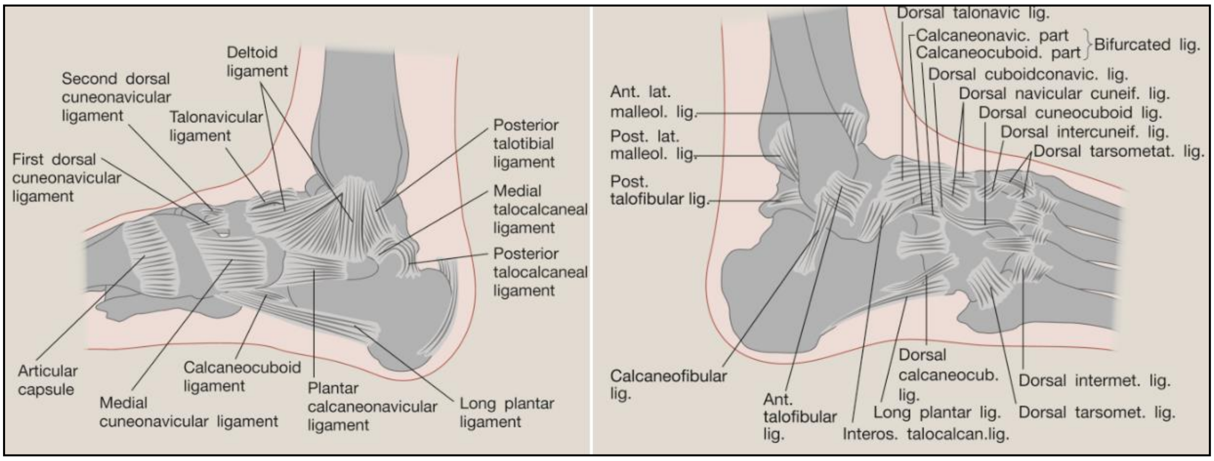

Figure 1.2 The stability factors of the complex ankle joint Brockett and Chapman, 2016

The normal ankle-foot movement during gait is reported as follow (Jacquelin Perry 2010); at the initial contact (0-2 % of the gait cycle), the heel bone which is in line with leg and perpendicular to the ground. The plane of the metatarsal heads is perpendicular to the tibia and parallel to the ground. The ankle joint is in ten degrees of dorsiflexion. No forces from the leg are on the foot in any of the three body planes that might cause inversion or eversion, abduction or adduction, dorsiflexion or plantarflexion. During the first 10 % of the gait cycle which is usually called loading response or weight acceptance the first plantarflexion arc is taking place. On average, this reach a maximum of ten degrees of plantarflexion at 8% of the gait cycle. The foot support is provided by the pretibial muscles. Less precise control will allow the ankle to drop into plantarflexion. During the loading response, the initial impact vector is vertical, so the forces are directed into the floor which provides the stability for the limb. The plantarflexion torque happens when the body vector is posterior to the ankle, this torque drives the foot towards the floor. This ankle plantarflexion action again is decelerated by the pretibial muscles.

When the forefoot contacts the ground (foot flat) the ankle changes towards dorsiflexion. The foot now is stationary, and the tibia becomes the moving part. At the 20% of the gait cycle the neutral alignment is reached. Throughout mid-stance and the first half of terminal stance, the ankle motion dorsiflexion continues to reach the maximum of approximately ten degrees angle by 48% of the gait cycle. Both the soleus muscle assisted by the gastrocnemius muscles play role in restraining the rapid rate of the ankle dorsiflexion following the forefoot floor contact. Soleus eccentric activity continues until the end of mid-stance. Hence, the soleus contributes in the ankle's progression and stability.

Following the second double support (contralateral initial contact), the ankle goes into rapid plantarflexion to reach the maximum of thirty degrees angle at the end of stance phase. During the terminal stance, there is a strong demand for the soleus and gastrocnemius action to support the body weight and to control dorsiflexion torque. The second and final dorsiflexion is initiated by the toe-off. Last task during stance is push-off which is the primary function of the gastro-soleus muscle group. In addition to the muscles, the Achilles tendon plays a role in the push-off due to the elastic recoil that returns the stored energy by tendon stretch. This will resist ankle dorsiflexion. The importance of the push-off is that firstly, it contributes to limb swing acceleration and secondly, it also contributes to centre of mass (CoM) acceleration (Zelik and Adamczyk, 2016). The aspect is important concept for the dynamic stability. Push-off and ankle plantarflexors muscles were reported to be as means to smooth the CoM trajectory during the step-to-step transition according to Saunders et al. (1953).

The neutral ankle position (0°) usually occurs around mid-swing and maintained throughout the rest of the swing phase. The entire ankle range of motion used during walking is reported to be between 20° to 40° (Jacquelin Perry 2010).

1.3 The lower limb prostheses

The term of prosthetics is widely used to describe the process of prescription, designing, fabrication, and finishing the artificial substitution of any missing part of the body (Agarwal 2013, Shurr et al., 2002). Lower limb prostheses are described as the implemented devices that are fabricated by specialists (certified prosthetists) to replace the missing part of the lower limb due to amputation/ deficiency (Agarwal 2013, Shurr et al., 2002).

The core function of the lower limb prostheses is restoring the normal function of the missing parts including posture, balance, and comfort during ambulation (Agarwal 2013, Girijala and Bush, 2018, Greitemann, 2017, Heinemann et al., 2014, Kerstein et al., 1975, Keszler et al., 2019). This section will focus on the lower limb including amputation prevalence, rehabilitation, and prosthetics.

Lower limb amputations may be conducted at different levels. Each level has different type of prostheses, prosthetics components, gait training, and rehabilitation programme (Girijala and Bush, 2018, Greitemann, 2017, Heinemann et al., 2014, Kerstein et al., 1975, Keszler et al., 2019, Knapp, 1968). The major levels of the lower limb amputation are: foot, through ankle, below knee, through knee, above knee, and hip amputation (Ebskov, 1992, Pinzur, 1999, Vu et al., 2019, McGee and Dalsey, 1992). Nevertheless, the above knee (transfemoral) and below knee (transtibial) amputations are the most common levels of lower limb amputation (Agarwal 2013, Shurr et al., 2002,(Gebreslassie et al., 2018).

Different risk factors may lead to lower limb amputation. The major common reasons of amputations are diabetes, peripheral vascular disease, tumors (such as bone cancer), trauma, arthrosclerosis disease (Ebskov, 1992, Pinzur, 1999, Vu et al., 2019, McGee and Dalsey, 1992). Nevertheless, diabetes and trauma are considered as global risk factors of the lower limb amputation (Ahmad et al., 2014, McGee and Dalsey, 1992). Diabetic patients have less blood flow to their extremities specially the lower limb extremity due to narrowing of the blood vessels. That leads to nerve injuries (peripheral vascular disease), ischemia, unhealed ulcer, and then amputation as a final treatment to remove the ischemic parts (Levin, 1995, Bild et al., 1989).

In the UK, up to 100 people have lower limb amputation every week due to diabetes (Diabetes UK and NHS Diabetes, 2009). Between 2003 and 2013, diabetic patients aged between 50-84 years were six times more likely to have a lower limb amputation than non- diabetic patients in England (Ahmad et al., 2016). Moreover, around 6000 people have toe or foot amputation per year due to diabetes in the UK and this number, unfortunately, is increasing yearly (McInnes, 2012).

Following amputation, the person have to adapt their life to align with the new situation and be prepared for prostheses such as changing their normal lifestyle and working environment (Furtado et al., 2017). It has been reported that most prosthetic users face considerable challenges to return to their life and work after the amputation. Therefore, the rehabilitation, that includes physician, surgeon, physiotherapist, occupational therapist, and prosthetist should provide an intensive approach to facilitate optimum prosthetic rehabilitation (Kristen, 1973, Ülger et al., 2018). This pre-prosthetics stage is essential to enhance the muscles power, sensation, and weight bearing ability of the residual limb through intensive rehabilitation programmes (Agarwal 2013, Shurr et al., 2002).

The rehabilitation programme after lower limb amputation must include muscle strength, flexibility, balance, cardiovascular training, and gait training (Christiansen et al., 2015, Clavagnier, 2019, Cochrane et al., 2001, Czerniecki et al., 2012, Esquenazi and DiGiacomo, 2001). It has been suggested that the rehabilitation programme that includes mobility protocol is able to increase the residual limb recovery after the surgery, prepare the residual limb for the prostheses and increase functional mobility among those with lower limb amputation (Marzen-Groller et al., 2008). It is important to state that this stage may last for six to eight weeks until the person is ready to be fitted with a lower limb prosthesis (Agarwal 2013, Shurr et al., 2002).

When a person is ready for prosthetic fitting, they will be referred to the prosthetic clinic to be fitted with prostheses. Different types of lower limb prosthetics with different components may be prescribed based on specific criteria (Agarwal 2013, Shurr et al., 2002). Those criteria may be related to level of amputation and the prosthetic users themselves such as their age, weight, health status, activity level, the requirements, goals, working environment, living area, personal preference, and financial situation (Agarwal 2013, Shurr et al., 2002).

1.3.1 The prosthetic foot

Prosthetic foot function is designed to mimic as much as possible the normal foot function during walking as described previously in section (1.2.3). Ideally, prosthetic foot design should incorporate the normal dorsiflexion and plantarflexion RoM as well as push-off capability.

The ‘push-off’ aspect of prosthetic foot design has been investigated in a number of publications. Morgenroth et al. (2011) who investigated the effects of prosthetic foot push-off on mechanical loading in transtibial prosthetic users (n=7) showed that reduced push-off ability is associated with knee osteoarthritis in lower limb prosthetic users. Additionally, increased prosthetic push-off may reduce the burden on the intact side. Adamczyk and Kuo (2014) examined the push-off effect on gait asymmetry in transtibial prosthetic users (n=11). The researchers concluded that reduced push-off capability the prosthetic feet would result in increased gait asymmetry. Other gait abnormalities were also correlated with the reduced push-off including slower forward velocity of the body centre of mass (CoM), greater energy consumption and more work.

To select the appropriate foot design for an individual, clinical information about each foot is required. In this section a description of the feet fitted and used by the study’s participants is provided based on the manufacturers’ descriptions along with the previously published work that aimed to assess these feet. The outcome of this part of the investigation may be helpful in interpretation of results in future chapters.

Prosthetic feet can be made from wood, rubber, carbon fibre and titanium as well as other materials. Structurally, the prosthetic foot may be either non-articulated or articulated (incorporating a hinged ankle mechanism). Functionally, prosthetic feet can be grouped into the following categories; solid ankle cushioned heel known as (SACH), single-axis, multi-axis, energy store and release/return energy (ESAR) feet, hydraulic ankles and microprocessor feet (Martin et al., 2010).

In the present study, the participants were divided into two main groups; feet that did not have ankle component (those were ESAR feet) and feet that had an ankle incorporated.

Feet without ankle joints, (ESAR feet), included seven Vari-flex (Össur, Iceland) and one Soleus (College-Park, USA). Whilst those with ankle joints other category included one Elan foot (Blatchford, UK) (microprocessor controlled), two EchelonVT hydraulic ankles feet (Blatchford, UK) and two pro-flex feet (Össur, Iceland).

Energy storing and return prosthetic (ESAR) feet were introduced early in 1980s (Wezenberg et al., 2014). These feet were promoted to be able to mimic the human ankle motion more than the conventional solid-ankle cushioned feet (SACH) (Wezenberg et al., 2014). They are generally made of carbon fibre components or other spring-like materials which allow storing of mechanical energy during stance phase of gait and releasing this energy during push-off (Highsmith et al., 2016). It was reported that the ESAR feet are among the feet of preference for prosthetic users (Laferrier et al., 2018). Additionally, they have been reported to help reduce the metabolic energy needed for walking and subsequently enhance walking economy (Wezenberg et al., 2014).

One of the common ESAR feet is the Vari-Flex manufactured by (Össur, Iceland) (Figure 1.3). According to the manufacturer, this foot provides high level of confidence and security as well as facilitating natural gait with less fatigue and better ankle dynamics. Vari-Flex consists of two carbon leaf springs, both are attached by bolt at toe joint. One spring is used to keep foot stable. Whilst the other is flexed by weight in ordered absorb shock, store and then release energy. It may be used for low to high activity range.

Figure 1.3 Vari-Flex (Össur, Iceland)

Results of a recent study (Houdijk et al., 2018a) have demonstrated that the ESAR feet can play a role in enhancing the step length asymmetry that is commonly seen in prosthetic users. In addition, ESAR feet maintain the margin of stability (MoS) in backward direction. In this study, the ESAR feet were compared to SACH feet. The participants of the study were given Vari-Flex, (Össur, Iceland) as ESAR feet and a 1D10 SACH foot manufactured by Ottobock, (Germany). Authors did however highlight some study limitations. The effect of acclimation especially for the SACH feet as the participants were very familiar with the Vari-flex because they were fitted with these feet for over two years unlike the SACH that was only fitted for the study. Besides the familiarisation issues, the authors anticipated that the alignment and socket fit may have affected the overall results.

The other ESAR foot in the study was a foot called Soleus (College-Park, USA) (Figure 1.4). In concept, this foot offers same functionality as the Vari-flex foot.

Figure 1.4 Soleus (College-Park, USA)

Prosthetic feet may include ankle components that are promoted to enhance walking for the prosthetic users (McGrath et al., 2018b).

At least 37 independent scientific studies have been directed towards assessing the benefits of hydraulic ankle feet manufactured by (Blatchford, UK) compared to non-hydraulic feet for period between (2011-2020). The studies conclude that feet with hydraulic ankle unit may mimic the dynamic and the adaptive actions of muscles to provide more natural gait. Among many other advertised benefits of these feet, the ones that research group were interested to test were the ability to improve stability and reduce the risk of trips and falls (Kannenberg, 2018).

Two feet from (Blatchford, UK) were utilised by participants in this study. The first and the more advanced one was the Elan foot which is a microprocessor-controlled hydraulic foot (Figure 1.5). Using small computer and sensors, ankle plantarflexion and dorsiflexion are self-aligning during walking depending on the action required. Elan is advertised to be improving the static and dynamic stability (Bai et al., 2018). Elan is deigned to go in a dorsiflexed position at the end of stance phase and remains that position during swing, to promote better toe-clearance and reduce the risk of trips. Elan's hydraulic range of motion according to technical specification by Blatchford, is 9° (3° dorsiflexion & 6° plantarflexion). A study by (Ernst et al., 2017) demonstrated comparable results for the Elan.

Figure 1.5 Elan (Blatchford, UK)

The second foot was a foot called Echelon VT (Figure 1.6). This foot is a hydraulic control of dorsiflexion and plantarflexion. Additionally, this design offers additional rotation and vertical shock absorption, according to the manufacturer it helps to reduce the shear forces at the socket interface. This reduction may help the prosthetic user to move and adapt more freely. The company advertises this foot to be ideal for activities such as golf and hiking where the stability is an important element (Kannenberg, 2018). Similar to Elan, Echelon VT range of motion according to technical specification by Blatchford, is 9° (3° dorsiflexion & 6° plantarflexion).

Figure 1.6 Echelon VT

The final foot used by prosthetic users in the study was the Pro-flex (Össur, Iceland). The foot consists of serial layouts of flexible carbon fibre leaf springs, a bottom blade (foot board), a top blade (short J-shaped spring), and a middle blade (flat spring) as showed in (Figure 1.7). The top and middle blades are connected by a multi-centre joint construction. According to the manufacturer, this construction permits rolling motion around the main pivot during walking that simulates foot rocker in normal gait. Hence, it is claimed that this foot potentially allows a more adaptive ankle joint motion.

Figure 1.7 Pro-Flex (Össur, Iceland).

Heitzmann et al. (2018) has discussed the benefits of this novel foot. The study compared the Pro-Flex against the Vari-Flex for eleven below knee prosthetic users (n=11, mixed of 3-4 K level). The results showed that the Pro-Flex provide effective prosthetic ankle RoM (31.6°) closer to the physiologic ankle RoM than the Vari-Flex (15.2°). Besides RoM, the Pro-Flex showed greater ankle power on the prosthetic side than the Vari-Flex. Subsequently, the Pro-Flex helps to reduce the demand on the intact side. The authors of the study discussed the possible limitations that might have affected the results. Those included, the effect of the familiarization time of the Pro-flex (30-40 min) was less than the Vari-Flex (two weeks). However, the results of the study were in the favour of the Pro-Flex. This could indicate that less accumulating time is needed when fitting a prosthetic user with Pro-Flex, added value for this foot. A second limitation of this study was linked to the order of foot fitting. It was always fixed order starting with the Pro-Flex first then Vari-Flex. Another limitation was linked the walking speed, participants in their study walked slower with the Pro-Flex than the Vari-Flex. This could be a major one therefore, it was taken into consideration in the discussion part of the present study. Heitzmann et al (2018) argued that the walking speed effect over kinetics can be neglected and that these effects reflected came from fitting different feet not from the different walking speed.

Childers and Takahashi (2018) also examined the possible benefits of the Pro-Flex over the Vari-Flex for transtibial prosthetic users (n=5). The results agreed with (Heitzmann et al., 2018) results, Pro-Flex demonstrated greater RoM ($23.7 \pm 3.1°$) and returned more energy compared to Vari-Fex ($15.2 \pm 6°$). In addition to the RoM, Childers and Takahashi (2018) assessed the body CoM of these two feet. The results demonstrated that the Pro-Flex returned more energy during the during prosthetic side propulsion and this ultimately would decrease the loading on the intact limb. Finally, Pro-Flex foot according to the researchers help in reducing pressures between the residual limb and the prosthetic socket.

For the prosthetic users with ankle disarticulation, a special foot with lower build height for this type of amputation called Flex-Symes (Ossur) was used (Figure 1.8). It is made of carbon fibre full length toe lever and heel to facilitate improved forward progression and shock absorption.

Figure 1.8 Flex-Symes (Össur, Iceland).

Using a prosthetic foot which incorporates an ankle joint may seem advantageous over conventional non articulating designs. However, several factors may affect choosing feet with ankle mechanisms as they are found to be heavier, bulkier, less durable and more expensive compared to the conventional feet. Study of the performance of these advanced feet may help to argue the benefits of using them and justify (or otherwise) their increased cost. It was noticed that the stability of feet with ankle joints have not previously been assessed by means of margins of stability. Therefore, it would be beneficial to investigate the MoS differences among these different foot designs. The assumption would be that prosthetic feet with ankle joints would show improved MoS over the conventional ESAR feet without ankle joints. The results of this part may disclose important insights that could help inform prosthetists in the process of prosthetic prescription.

In addition, it was reported that prosthetic users with ankle disarticulation level of amputation would exhibit relatively enhanced functional activity level closer to the able-bodied level (Braaksma et al., 2018, Lin-Chan et al., 2003, Jeans et al., 2011). Hence, the assumption was that the participants with this level of amputation would exhibit improved gait stability and stepping strategy over the more proximal transtibial amputation.

1.4 Congenital disorders

Congenital disorders may be referred as congenital birth defects, deformities, deficiencies or anomalies, are conditions effect the body structure (named congenital anomalies), body way of operation or metabolism resulting in abnormal status (Porth, 2011). As the name states ‘congenital’ these disorders are non-acquired conditions. However, not all disorders could be noticeable at birth, they become detectable later in individual’s life (Mathews et al., 2015). Generally, birth abnormalities differ in both cause and symptoms. The causes of such conditions may be genetic agents or due to environmental effects. However, many defects are of unknown or idiopathic cause (Lowry and Bedard, 2016, Brent, 2004).

The congenital limb deficiency defects (CLDDs), are congenital anomalies which may affect one or more bony extremity. Since describing a congenital deficiency may vary and be quite confusing many have described classifications to such disorders such as Frantz and O'rahilly (1961), Burtch (1966), Henkel and Willert (1969) and some as (Kay et al., 1974, Kay et al., 1975, Day et al., 1988) have proposed an international nomenclature to describe the congenital limb deficiencies.

Thus, a plethora of terms have been used to describe CLDDs. According to (Day, 1991) the term ‘phocomelia’ was used to report all classifications and types of deficiencies up until 1973.

Historical terms	
Amelia	Deficiency of one or more limbs
Ectrodactyly	Deficiency of digit or more (commonly used with split hand or foot)
Hemimelia	Deficiency or remarkable hypoplasia of the distal section of one or more extremity bone (radial, ulnar, fibular, ortibial)
Oligodactyly	Fewer than 5 fingers or toes are presence
Peromelia	Abnormal formation of one or more limbs
Phocomelia	Proximal limbs deficiency, distal elements may presence.
Split-hand/foot	Complete or partial absence of some digit of the hand/foot, commonly associated with clefts in the hands or feet.

Table 1.1 Historical terms and their use. (Sell et al., 2012, Wilcox et al., 2015).

In 1969, a classification had been made by (Willert and Henkel, 1969) for a group of diseases under the name ‘DYSMELIA’ based on 287 patients with 136 malformed lower extremities and 557 malformed upper extremities. This classification made primarily not to include all the congenital deformities but just for dysmelia which is a group of malformations that have the same morphological patterns. The idea of this classification was that these malformations according to the writers did not fit into other classification such as (Frantz and O'rahilly, 1961) and Burtch (1966). According to the writers, the relationship of the hand and foot malformations and the limbs distal and proximal segments to each other cannot be described and expressed.

Three terms were used in this classification to describe the dysmelia subgroup: The term ‘ectromelia’ is used to describe those subgroups of dysmelia in which the radius and the tibia with their peripheral rays and the humerus or the femur are affected. The term ‘phocomelia’ is used to describe those subgroups of dysmelia in which no residues of long bones are present between the extremity girdle and the peripheral segment (hand or foot). Finally, the term ‘Amelia’ is used to describe the most severe degree of dysmelia, in which the extremity is totally absent.

As in all classifications the main reason for the classifications is to enable better communication and diagnosis; and to enhance understanding of the deformed limbs’ anatomy.

The (Willert & Henkel, 1969) classification of the patients is based on three criteria, first, the affected region of the limb and skeletal elements, second, how they affected using terms hypoplasia, partial aplasia or total aplasia and finally, if the affected skeletal elements have experienced fusion by synostosis.

Using these criteria, the teratological sequence (teratology is the study of physiological development abnormalities) of dysmelia was sub-grouped into five major classes (Henkel and Willert, 1969, Willert, 1978): Distal form of ectromelia; Axial form of ectromelia; Proximal form of ectromelia; Phocomelia; Amelia.

It is important to state that the differences in the distal, axial and proximal forms of ectromelia are in presentation appearance, they show broad variation in the skeletal abnormality severity as well as some of the cases there is synostosis and reduction.

In 1989, the International Organization for Standardization (ISO) adopted a method of description under (ISO 8548-1: 1989), which solved this issue allowing more accurate descriptions of each disorder (Day et al., 1988). Such nomenclature is beneficial in enabling communication between clinical staff. In 1990, the same standard was adopted by British Standard under (BS 7313-2:1990).

1.4.1 The international standard (ISO 8548-1: 1989)

According to the international standard there are two main types of congenital anomaly; Transverse and Longitudinal (ISO, 1989).

1.4.1.1 Transverse

In this category, the extremity has formed normally to a certain level after which no skeletal component is found. In order to describe such deficiencies first, the side should be named whether is right or left then the affected limb (upper or lower limb) and finally the level as in Figure 1.9

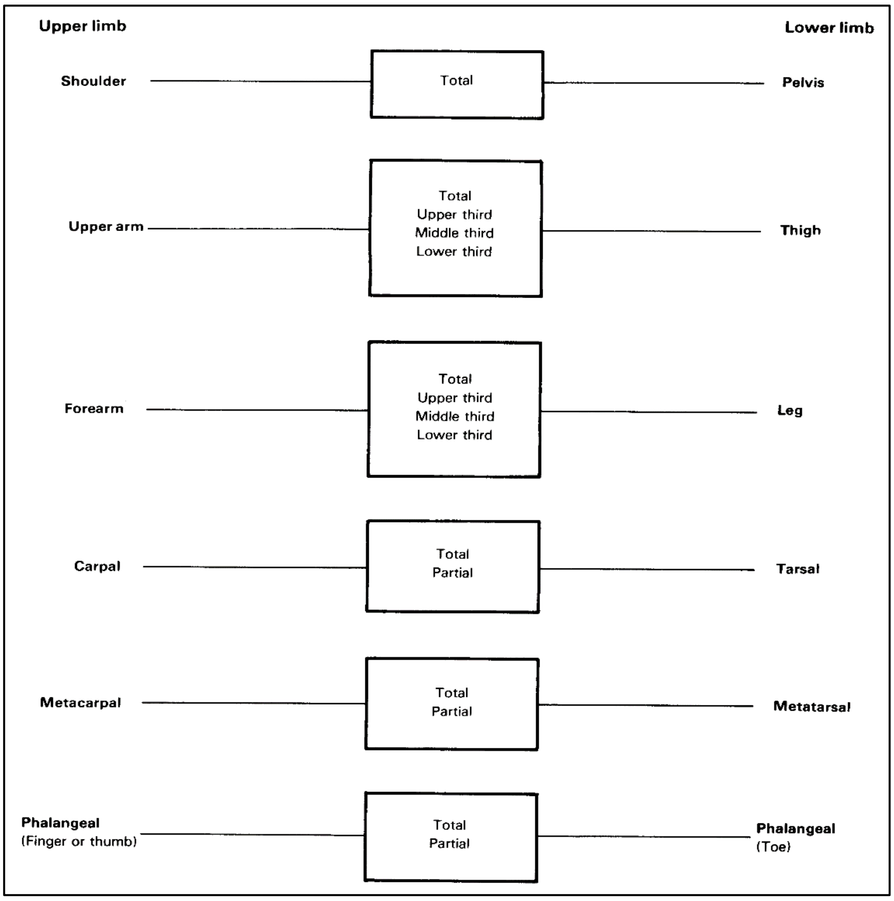

Figure 1.9 Upper and lower limbs transverse deficiencies based on the International Organization for Standardization (Day, 1991).

Based on the previous classification, it has been noticed that the total lack of the shoulder as well as hemi-pelvis including all distal elements is classified as transverse deficiency. In the case of partially absent of the shoulder or hemi-pelvis then the condition is classified as longitudinal deficiency.

1.4.1.2 Longitudinal

In this category, the extremity features partial or complete absence of one or more elements within its long axis. Unlike the transverse, normally developed skeletal elements may be founded distally to the abnormal element (Cvetkovic, 1997, Day, 1991). The following Figure 1.10 and Figure 1.11 show the describing method for the longitudinal deficiencies.

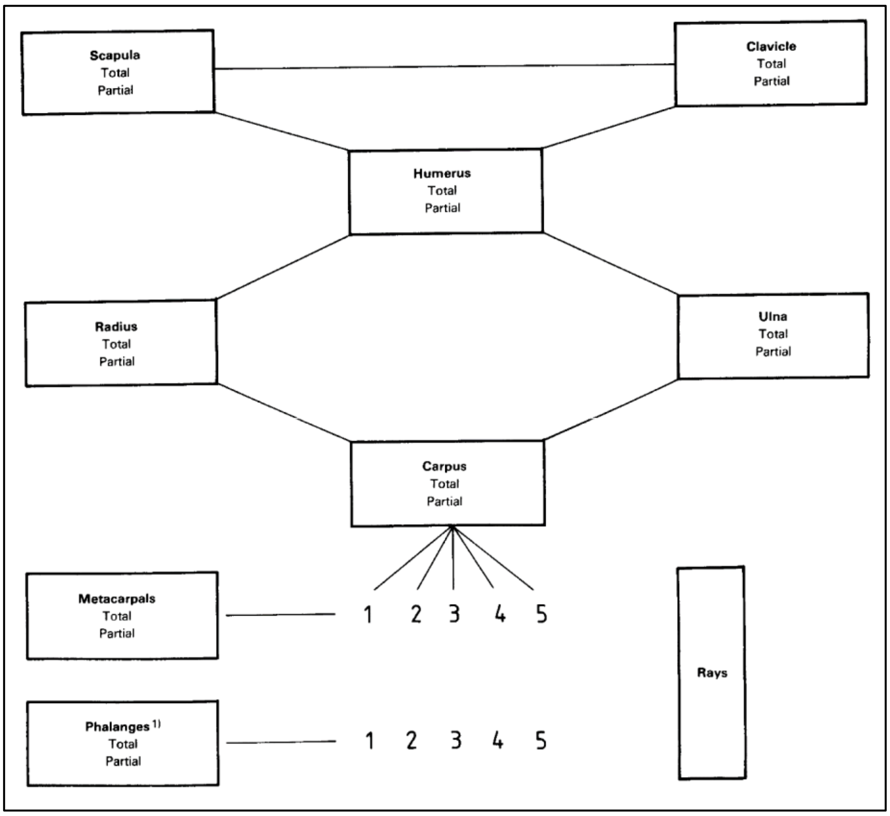

Figure 1.10 Upper limb longitudinal deficiencies (Day, 1991)

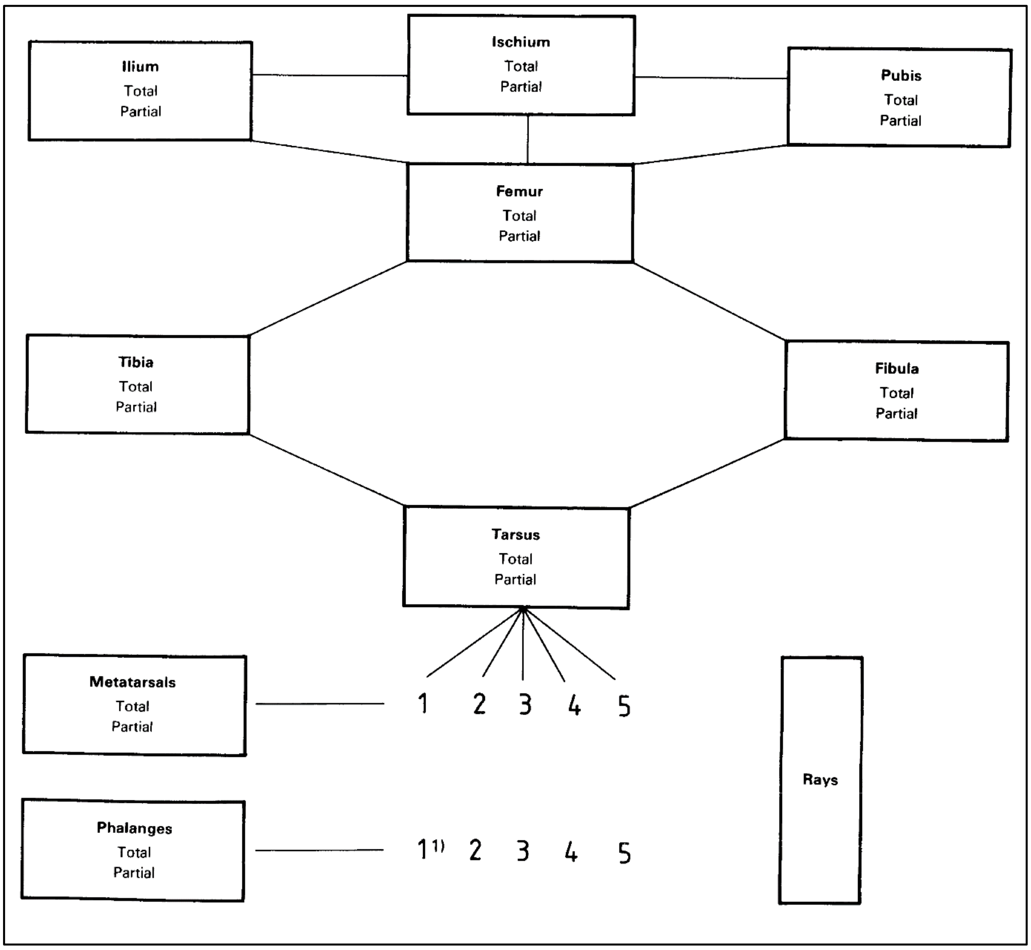

Figure 1.11 Lower limb longitudinal deficiencies. The number (1) in the figure is the great toe (Day, 1991).

As in the transverse to describe such deficiencies, the affected side should be initially named whether is right or left then the affected limb (upper or lower limb). Then for the large bones (shoulder, pelvic and limb bones) the name of the affected bone stated e.g. Fibula. Regarding the small bones (metacarpals, metatarsal and phalanges), the abnormal digit number stated along with the related phalanges from radial or tibial sides, as in the anatomical position the thumb and the great toe ‘hallux’ is considered initially. More details about this standard is presented in (Appendix I)

However, using the International Standard as a description method has some limitations. Firstly, the proposed classification is limited to skeletal deficiencies; accordingly, the segments failure of formation is the main cause of these defects. Secondly, the anomalies are described depending only on the anatomical and radiological bases without include any embryology, aetiology or epidemiology of the defects. Thirdly, terms like hemimelia, phocomelia, etc, which were used frequently, are avoided due to their insufficiency of precision and the difficulty of finding similar translation into languages that are not related to Greek. The anomalies are described as Transverse and Longitudinal. Finally, the former resembles an amputation residual limb, in which the limb has developed normally to a particular level beyond which no skeletal elements are present. All other cases are classed as longitudinal in which there is reduction or absence of an element or elements within the long axis of the limb.

1.4.2 The causes for the congenital limb deficiency defects

Regrettably, many disorders causes are undefined (Brent, 2004). The main known factors of CLDDs may be:

Genetic factor. This is the most commonly recognized and may be a single-gene, multi-factorial inheritance or chromosomal aberrations. Genetic disease is mentioned when a separate incident occurs and affects how a gene behaves in group of cells connected to each other by gene correlation. Commonly, the cause of abnormal genes due to changes in the deoxyribonucleic acid (DNA) sequence changes single-gene synthesis. Other factors may be as a consequence of chromosomal rearrangement which removes, or duplicates connected genes. Additionally, the gene factor occurs because of the errors in mitosis–a cell cycle event during which the replicated chromosomes are divided to produce two new nuclei-causing with abnormal chromosomes numbers or structure (Chen et al., 2016).

The environmental factor (Teratogenic agent), may be an external agent during the fetal development that includes maternal disease, infection or due to drug usage during the pregnancy period. The time that the environmental factor causes the defects and abnormalities is during organogenesis process, which is the process when the development and differentiation of the infant’s organs occur. Commonly, this happens within 15 to 60 days of fertilisation (Brent, 2004, Robbins and Kumar, 2010). Examples include: diabetes; infant low normal weight due to smoking as well as alcohol usage (Brent, 2004). Exposure to radiation such as X-ray may result in genetic abnormalities. The most common drug found to cause deficiencies and malformations is Thalidomide which was linked with Phocomelia (Roskies, 2019).

Other factors that may have the same impact include lack of some nutrition and vitamins (Czeizel et al., 2013) . One of the most common example is the lack of Folic Acid which has been known to be a risk factor of Neural tube defects, one of these defects is Spina Bifida (Control, 1991, van der Put et al., 1995, Czeizel et al., 2013).

1.4.3 Prenatal diagnostic testing

The process of testing the mother and the embryo before birth may enable identification of several abnormalities. Approaches are either invasive or non-invasive. The invasive approach which is a medical mechanism that involves any skin breaking or internal body cavity contacting of the individual’s body. Some of these intensive tests are now considered part of the regular prenatal care rather than for a certain reason.

Non-invasive tests include an approach called ultrasonography and blood tests. Methods are safe and may help to indicate further invasive prenatal tests. For instance, the ultrasonography is used to perform a test called Nuchal Translucency Screening Test by measuring the thickness of the fluid that fills the space at the back of the developing baby’s neck (Fetus). This test can indicate the Down syndrome, Trisomy 18 (Edward's syndrome) and heart problems (Hyett et al., 1999, Pandya et al., 1995). In case that this area is increased more than normal, the chance of the above abnormalities is increased.

Another test implemented mostly on 20-week is anomaly scan (may be conducted between 18-20 weeks), also may be known as a mid-pregnancy scan. It is carried on by ultrasound, the main purpose of this scan is not just to find the gender of the baby but to examine that the baby is developing normally and detects if there is any defect.

Paperwork by (Carvalho et al., 2002) discussed the importance of conducting the test to detect the fetal structural malformations by 11-14 week which is almost one month earlier than the mid-pregnancy scan, according to Carvalho’s work approximately 22.3% of the cases of the fetal structural abnormalities were detected at the 11–14 week scan which indicated the importance of a second-trimester anomaly scan to be included in the antenatal care routine to help increase the detection of fetal defects. On the other hand, some other conditions, such as bowel obstructions, may not be detected until later in the pregnancy.

During the scan, the breakdown of the picture in ultrasound screen is as the bones will look white meanwhile the soft tissue of the fetus will look grey and spotted. The amniotic fluid (the fluid that surrounding the fetus inside the uterus) will look black (NHS, 2013).

The sonographer (the one who performs this test) will have a list of conditions to examine the fetus for. These conditions can be very serious in which the fetus may not survive, or they can be treatable conditions, which can be treated when the baby is born. In the case that the condition is treatable, this scan will help the care team to set the proper treatment programme in advance which subsequently will increase the rate of the conditions to be successfully managed once the fetus is born (Richter et al., 2020).

The sonographer will be able to calculate the fetus's heartbeat. Furthermore, he will be able to check the fetuses' organs this includes the head's shape and structure, face if there a cleft lip and the bones of the spine to check its alignment. The abdominal wall of the fetus will be checked to make sure that it is completed and covers the internal organs. The heart chambers, the valves as well as the major veins and arteries will be examined. Also the fetuses' stomach and kidneys (NHS, 2013).

Regarding the extremities, the sonographer will check the arms, hands legs and feet, as well as the fingers and toes but without being able to count them.

In order to check the growing of the fetus's parts, some measurements will be recorded these include, the head circumference, abdominal circumference and thigh (femur) bone. These measurements should match the expected measurement of the fetus.

1.4.4 The anomalies that the scan may detect

Some abnormalities are easier to be detected than others meanwhile some are quite hard to be detected at all. The following (Table 1.2) shows the list of some abnormalities and the possibility percentage of the being detected by the examiner (Rossi and Prefumo, 2013).

The name of the condition	The chance of seeing it in the scan (%)
Anencephaly (the top of the head absence)	98 %
Cleft lip	75 %
Exomphalos (protrusion of the abdominal wall contents mainly liver and bowel)	80 %
Extremities' absence or shortening	90 %
Spina Bifida	90 %
Kidneys absence or abnormalities	84 %
Diaphragmatic hernia (abnormal hole in the diaphragm)	60 %
Hydrocephalus (increased fluid within the brain)	60%
Edwards' syndrome (chromosomal abnormalities, Trisomy 18)	95 %
Heart defects (chambers, valves or vessels)	50 %

Table 1.2 Some abnormalities and the possibility percentage of the being detected by the examiner (NHS, 2013, Rossi and Prefumo, 2013).

A study by Kudla et al. (2016) showed the importance of using the 3D/4D prenatal Ultrasound in diagnosis of a condition called Proximal Femoral Focal Deficiency (PFFD) in the 12th week of pregnancy. In the study, the difference in the extremity's length was displayed using trans-vaginal 2D, 3D and 4D projection.

Another study by Biko et al. (2012) the Magnetic Resonance Imaging (MRI) have been used to evaluate the PFFD patient and the results were that the MRI can be used in defining the anatomy. The MRI showed findings of the PFFD that not seen with the use of radiographs.

1.4.5 Common congenital anomalies

This section discusses some common congenital anomalies in terms of definition, aetiology management and rehabilitation.

1.4.5.1 **Phocomelia**

1.4.5.1.1 Definition

The word phocomelia is from the Greek: φώκη— fo´ke— ‘‘seal,’’ plus μέλος—melos— ‘‘limb,’’ and it describes the shape of the patient’s extremity which is similar to the flipper on a seal (Bermejo-Sánchez et al., 2011, Zimmer, 2012).

Phocomelia depending on a study of Bermejo-Sánchez et al. (2011) is used to mention a rare congenital malformation in which the extremity proximal part is missing or significantly hypoplastic (incomplete development) with intact or nearly normal hands or feet. It can be humerus or femur, radius or tibia, ulna or fibula, furthermore the true phocomelia is distinguished when the extremity intermediate segments are totally missing, and the hands or feet being attached directly to the trunk.

Others like (Willert and Henkel, 1969) say that the term ‘phocomelia’ is used to describe those subgroups of dysmelia in which no residues of long bones are present between the extremity girdle and the peripheral segment (hand or foot).

1.4.5.1.2 Causes

Phocomelia is strongly linked to a drug called Thalidomide (Miller, 1991, Speirs, 1962, Taussig, 1962). Others like (Maier, 1965) said that the drug also led to 'dysmelia syndrome' rather than just Phocomelia. These defects happened when a pregnant woman take the drug (Lenz and Knapp, 1962, Miller and Strömland, 2011).

1.4.5.1.3 Thalidomide: Background information

Thalidomide (a[N-phthalmido] - glutarimide) the trending medicine name is Immunoprin. In 1957 the drug was launched in the West Germany market and its first marked name was Contergan. At first was administered as a sedative. Moreover, it was claimed that this medicine has a useful effect treating on insomnia, gastritis, anxiety, as well as tension. The manufacturer company advertised that the medicine is safe and harmless for the pregnant women and can be used for nausea and morning sickness. No teratogenic effects were found when the drug was tested on rodents as well as the conducted routine screening tests had not showed that the drug has teratogenic effects on humans or mammals (Taussig, 1962, Lenz, 1988).

After 4 years of the drug launch, the number of congenital limbs anomalies had occurred in the West German population. The first to draw attention to this phenomenon was Lenz(Lenz and Knapp, 1962). Thereafter, Lenz said that there was a correlation that mothers of some of the affected children had been consuming thalidomide during pregnancy time. Following reported observation, trading Thalidomide was stopped in the market in Europe, after 9 months the drug was withdrawn in Japan’s market as well (Kida, 1987).

Study of consuming intervals in regard with outcome congenital malformation found that the thalidomide teratogenic effects took place between 20 and 35 days after conception. In this time, thalidomide use caused many abnormalities, commonly malformations that includes the cranium and face structures as well as extremities (Miller, 1991).

According to (Miller, 1991) the malformations as a result of thalidomide are:

Dysmelia which divided into three different degrees depending on the severity as

1. Mild dysmelia includes triphalangeal thumbs –a condition when the thumb has an extra joint-, thenar muscle aplasia and shoulder weakness.
2. Moderate dysmelia includes missing or hypoplastic thumb, middle and index fingers flexion contractures, radial and ulnar hypoplasia, missing phalanges, shoulder and hip hypoplasia.
3. Severe dysmelia includes missing radius, phocomelia, amelia (complete absence of upper or lower limb), severe anomalies of shoulder and hip.

Facial abnormalities this includes

1. Wide range of internal and external ear malformation starting from anotia (absence of the external ear) to mild ear malformations as well as sensorineural hearing loss which there is an abnormality in the inner ear, cochlea or hearing nerve with or without external ear.
2. Abnormalities in ocular motility that will influence the horizontal movement such as limit abduction and Duane syndrome (the capability to move the eye).
3. Facial nerve palsy or Bell's palsy (damage to the seventh cranial nerve) this may happen in different degree of severity and either unilateral or unilateral or bilateral.
4. Irregular lacrimation (the flow of tears) commonly tearing during eating or sucking as well as insufficient emotional tearing.
5. Other anomalies in ocular, like uveal coloboma (when the uvea is the middle layer of the eye), glaucoma, microphthalmos, refractive error, ptosis, and cataract.

Systemic Anomalies this includes

1- Kidney malformation which can be hypoplasia or positional.

2- cardiac anomalies.

3- Anal atresia or imperforate anus.

4- Spinal anomalies

5- Chest abnormalities

6- Central nervous system (CNS) problems.

Miller (1991) suggested that since the reports had showed that the heart abnormalities were higher in new-borns than the infants, the cardiac condition may be the reason for the early deaths and for the sudden abortions in some women who consumed thalidomide during the pregnancy.

Genetic factors have been also linked with phocomelia. A condition called Roberts syndrome sometimes may refer to by Roberts-SC Phocomelia Syndrome, SC Phocomelia Syndrome, is a genetic related disorder has similar symptoms as the phocomelia. The syndrome affected individual has abnormal number of chromosomes either reduced or extra resulting in abnormal cells development which subsequently will lead to abnormal development of extremity, organ such as ear or other structure (Ismail et al., 2016).

Studies have reported that abnormal blood supply to be increasing the risk of phocomelia. A study of (van der Horst, 1971) suggested that phocomelia can be a result of a condition called Anomalous Origin of the Subclavian in which there is insufficient blood supply, another paper of (Weaver, 1998) linked the reduced blood supply to malformation of limbs' intermediate segments.

1.4.5.1.4 Epidemiology

Epidemiologic data about phocomelia are inadequate, a study of (Bermejo-Sánchez et al., 2011) focused on phocomelia in a large scale depending on databases from many countries. 19 surveillance programmes of birth defect were used in a period between 1968 and 2006. The results of the study showed that there were 141 phocomelia cases out of 22,740,933 total births which produces an overall total prevalence of 0.62 per 100,000 births (95% confidence interval: 0.52–0.73).

According to (Mustapha, 1990) that there were approximately 349 infants were born in a period between 1959 and 1962 in the UK with congenital malformations as a result of thalidomide consumption.

1.4.5.1.5 Classification of phocomelia

According to a paper of (Frantz and O'rahilly, 1961) , phocomelia has been classified under class called transverse intercalary failure of formation. They divided the phocomelia into three types as:

- Complete Phocomelia, (Type I) in which the hands/feet or digits are attached directly to trunk (Figure 1.12)
- Proximal Phocomelia. (Type II) in which the forearm/leg bones are present between hand/foot and trunk (Figure 1.13)
- Distal Phocomelia. (Type III) in which the hands/feet are attached directly to arm/thigh (Figure 1.14).

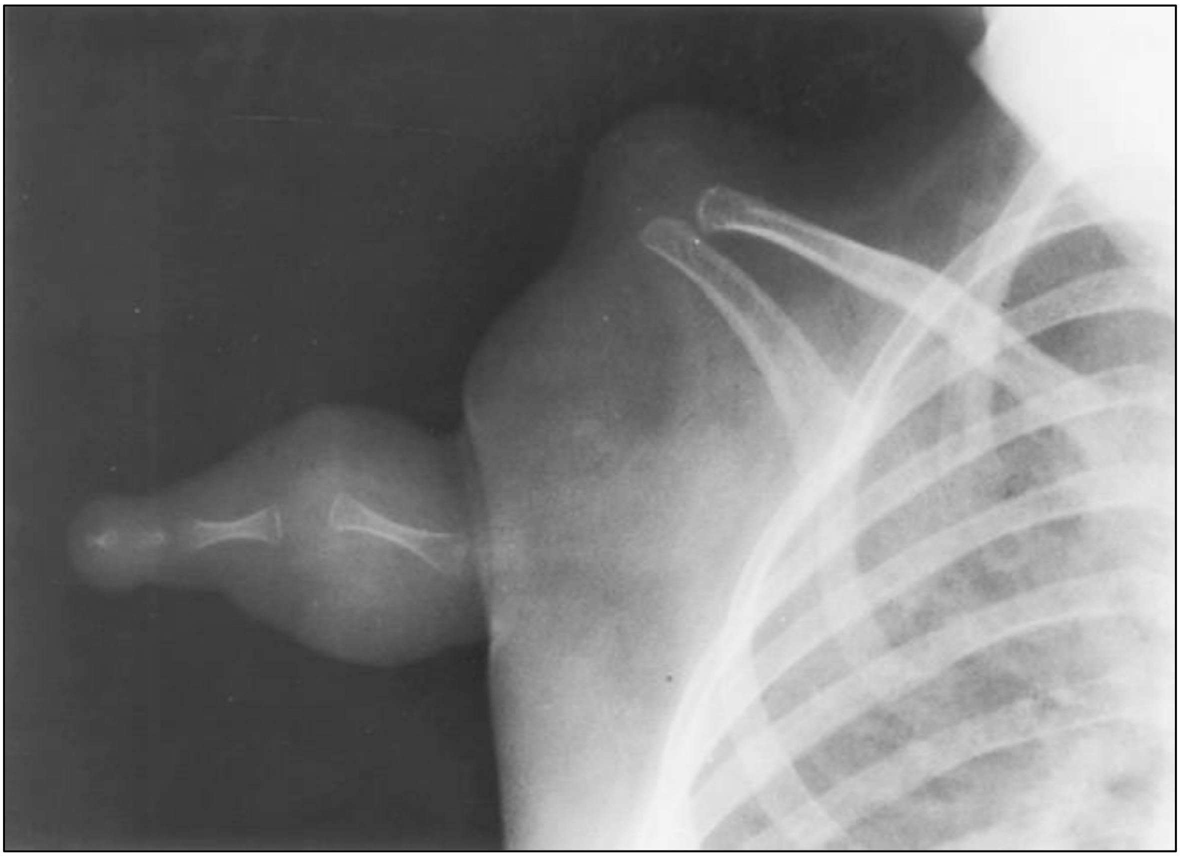

Figure 1.12 Phocomelia Type I according to Frantz & O'rahilly, 1961 classification.

Figure 1.13 Phocomelia Type II according to Frantz & O'rahilly, 1961 classification.

Figure 1.14 Phocomelia Type III according to Frantz & O'rahilly, 1961 classification.

Although this classification is simple and clear, it has not been validated in practice according to a study of (Tytherleigh-Strong and Hooper, 2003). In addition, in the paper they argued that it can be hard to distinguish phocomelia from other congenital anomalies that effect upper limbs. The methods that they used is based on X-rays of 24 patients; 19 of them was as result of thalidomide.

1.4.5.2 Congenital longitudinal deficiency of the fibula (Fibular Hemimelia)

1.4.5.2.1 Definition

Congenital longitudinal deficiency of the fibula or as also known as Fibular hemimelia (FH), longitudinal fibular deficiency, is one of birth and congenital long bones disorder where the fibula in completely or partially absent. The congenital defect with complete fibula absence is higher prevalence than fibula partially absent incident (Rodriguez-Ramirez et al., 2010). Among the long bones congenital deficiencies the FH found to be the most common (Caskey and Lester, 2002). The record has shown that between 7 to 20 per million live births could be affected with fibular hemimelia (Coventry and Johnson jr, 1952). Another investigation found that around 80% of incidents have unilateral defect and more often the right side is the most affected side (Birch et al., 2011). Furthermore, this pathology could be associated with other limb (lower limb, upper limb, or spine) defects or only with mild fibular shortening (Coventry and Johnson jr, 1952, Bohne and Root, 1977).

As a result, for this congenital defect, verities of clinical and radiological features are associated with this pathology. The clinical features consist of knee instabilities, equinovalgus (club foot) with lateral ray foot absence, tibia bowing, leg length discrepancy, and knee valgus (Bohne and Root, 1977). While the radiographic features could show the radiograph image the proximal cartilaginous epiphysis is absent for five years old children , the superior margin of the proximal fibula is lower than the tibial physis, with hypoplastic lateral condyle (Bedoya et al., 2015). Also, magnetic resonance imaging (MRI) plays a core role in the FH early examination. The MRI provides useful information about the soft tissue deformities associated with FH, joint instability, and the size and course of the fibrous (Maffulli and Fixsen, 1991).

1.4.5.2.2 Causes

The main reasons behind congenital longitudinal deficiency of the fibula is unknown; however, some risk factors could increase the incidents of having lower limb defects including: viral infections, thermal injury, bacterial toxins, intrinsic reduction in the size of mesenchymal condensation with abnormal Chondrification process (process of cartilage formation), or early compromise of blood supply (Maffulli and Fixsen, 1991).

1.4.5.2.3 Classifications and Scales

According to Achterman-Kalamchi classifications, fibular hemimelia has two main types: Type I (including A and B types) and type II based on the extend of the deficiency (Figure 1.15). In type IA, with minimal hypoplasia of the fibula, the proximal fibular epiphysis is distal to the level of the tibial growth plate and it is smaller than on the normal side, while the distal fibular growth plate is proximal to the dome of the talus. In Type IB, there is partial absence of the fibula, proximally the fibula is also absent for 30 to 50 per cent of its length, while distally it is present but does not support the ankle. In contrast, Type II, with complete fibula absence, included all limbs where there is complete absence of the fibula or where only a distal, vestigial fragment is present (Achterman and Kalamchi, 1979).

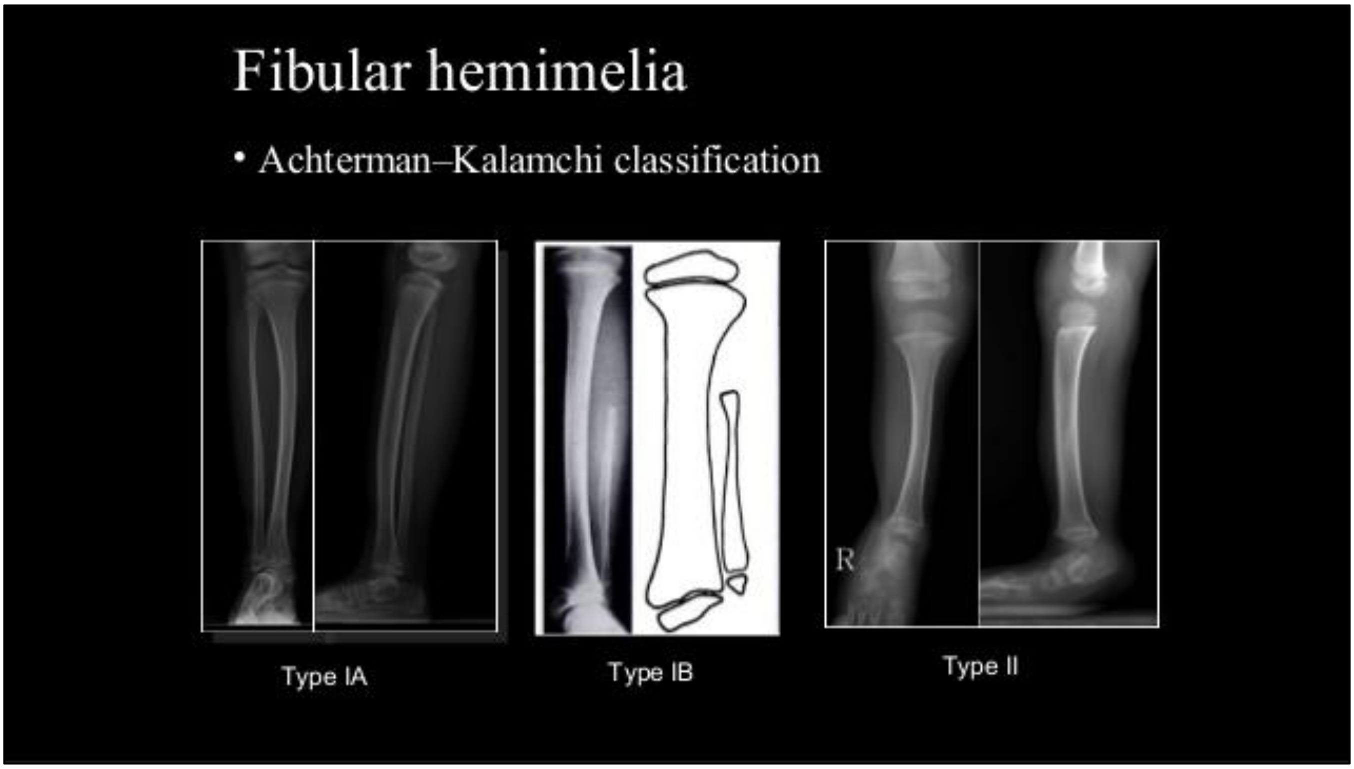

Figure 1.15 X-rays of Achterman-Kalamchi classifications (Ludhiana, 2016).

More results based on the previous study the femur deficiencies such as congenital shortening or proximal femoral focal deficiency were seen within both groups. Also, both groups have bowing of the tibia (Achterman and Kalamchi, 1979). For foot and ankle deformity, the study found that Type I had ball and socket ankle joint deformity since the fibula extended to the ankle but did not participate in ankle joint articulation. While type II had Tibiotalar ankle joints, joint instability, dysplastic of the distal tibial epiphysis. But both groups had lateral rays' absence in forefoot and tarsal coalitions in hind foot. Besides, the results showed that type II has increase in the percentage of tibial shortening than type I. This shortening is correlated with femoral shortening and number of missing lateral foot rays (Achterman and Kalamchi, 1979).

It has been noticed that the Achterman-Kalamchi classifications describes the scale based on fibula without considering any other changes at the ankle and foot.

Another used classification scale is called Coventry and Johnson classification which is a developed scale shows the severity of fibular hemimelia considering the ankle and foot defects. According to this classification, Fibular hemimelia consists three scales: Type I includes Partial unilateral absence of the fibula, with minimal leg length discrepancy, and without any additional limb deformities; Type II presents unilateral or almost complete fibula absence, leg length discrepancy, with tibial bowing and foot deficiency; finally, type III is a bilateral deformities and associated with spine and upper limb abnormalities such as ulnar hemimelia or amelia and syndactyly (Coventry and Johnson jr, 1952).

Many others suggested other new versions of classifications system based on their clinical experiences such as Stamatakis (Stanitski and Stanitski, 2003). The classification depends on fibular and ankle (structure) morphology, hind-foot alignment, and foot rays' abnormities.

1.4.5.2.4 Managements and interventions

The current interventions for FH could be either surgical or non-surgical indications depending on the severity of the case, leg length discrepancy, and functional defects of the other joints. Based on the Achterman and Kalamchi classification scale, the nonsurgical intervention such as lengthening using the Ilizarov technique (Figure 1.16) is recommended if the patients with type IA or 1B, have a predicted shortening less than two cm after skeletal maturity with a functional foot with more than three rays. While the surgical amputation interventions are indicated when the patients have shortening more than five cm at birth, or more than 25 cm of predicted shortening at skeletal maturity (Coley, 2013).

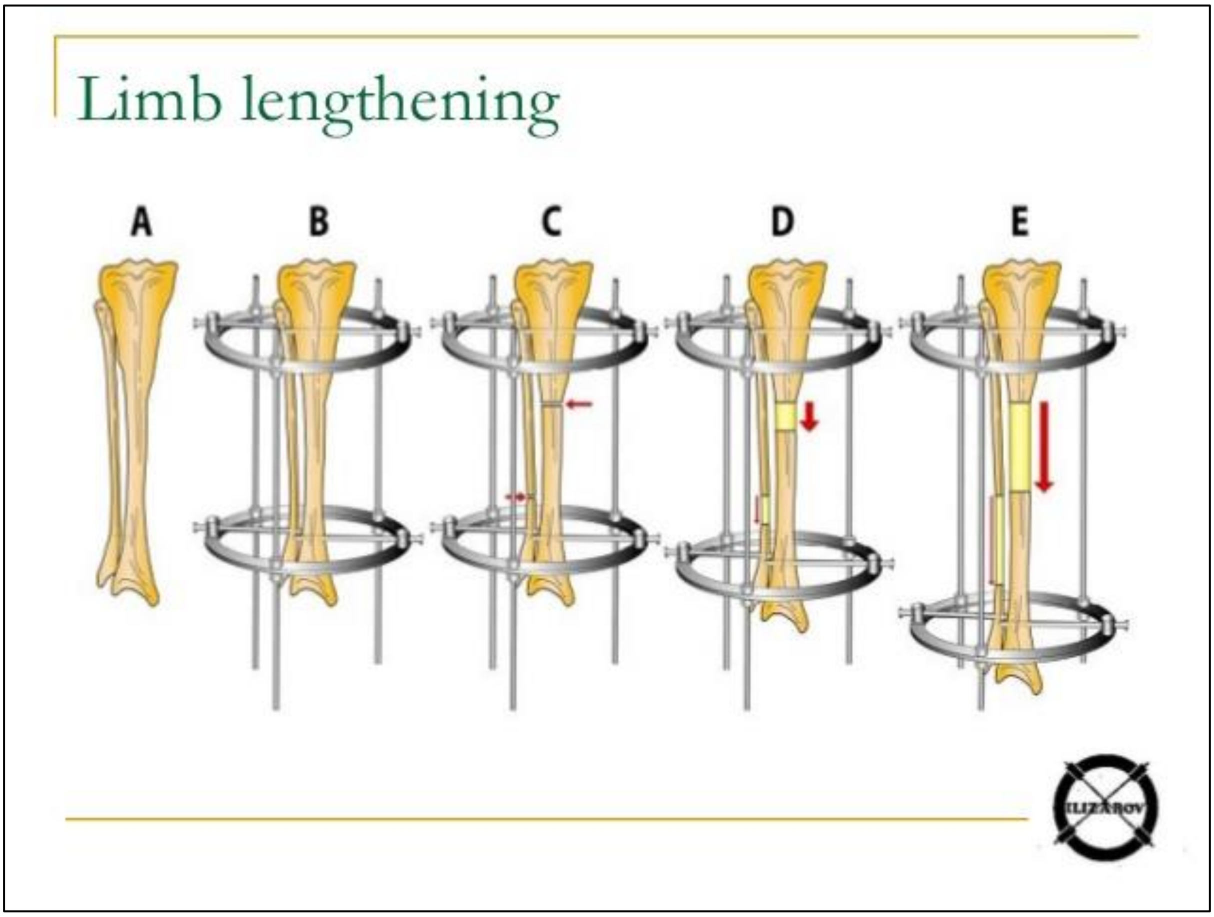

Figure 1.16 The Ilizarov technique (apparatus), external fixation used to lengthen the bone (Mamun, 2013).

The early amputation option, which are mainly ankle disarticulation (Syme's or Boyd's amputation, provides quick recovery and adapt to the new life with short stay at the hospital; however, the most important disadvantages for amputation are losing limb sensation and it is an irreversible method. In contrast, tibial lengthening technique preserves the limb with the sensation and proprioception, but it requires long stay in hospital, with psychological impacts, and the need for multiple operations (Naudie et al., 1997, Gibbons and Bradish, 1996).

Naudie et.al (1997) study found that well children who fitted prostheses after amputation have better function improvement and quick recovery than tibial lengthening. Also, some cases had need further corrective surgery or amputation after tibial lengthening (Naudie et al., 1997).

Another study evaluated three patients had Achterman Kalamchi type II fibular hemimelia with leg length discrepancy of over eight cm who had had Ilizarov lengthening. The result revealed that the three patients had early complications such as infections and knee contracture. In addition, the three patients had also late complications appeared after lengthening by up to six years such as: knee valgus, ankle instability and subluxation, and gait deformity. These subsequent deformities required additional surgery to correct the functional problems (Cheng et al., 1998, Sharma et al., 2014).

Ankle disarticulation amputation for this defect looks less bulky than use to be due to lateral malleolus absent which provides better suspension, distal weight bearing, and better cosmetic appearance. Also, the length discrepancy which is normally associated with fibular hemimelia keeps space or room for the prosthetics' foot and ankle. However, the knee valgum will be the common problem combined with this defect, the medial placement for the foot could be a temporary solution. If the knee valgum reduces the functional level, then tibial osteotomy might be required (Michael and Bowker, 2004) .

A study by Marquardt (1981) stated that in all severe cases when there is more than one ray deficiency of the foot, with severe tibia shortening and bowing the preferable methods are ankle disarticulation or modified Boyd amputation stump combined with tibial corrective osteotomy.

Marquardt (1981b) also addressed contraindications to perform amputation or disarticulation, when the child suffers from severe malformation of the upper extremities. The reason that in such cases after performing the amputation or disarticulation, the toes are essential for grasping specially for self-care. In such cases, ankle arthrolysis laterally will be performed, as well as, arthrolysis of the talo-calcaneo joint in case if it present, furthermore valgus contracture disconnection as well as if necessary, posterior transposition of the peroneal tendon(s), to be done after three-dimensional correction osteotomy of the tibia.

According to Marquardt (1981b), the good results after performing these operations made them more careful with partial foot amputation or ankle disarticulation since it has been showed that it's not recommended to perform amputation operation of children between 3 years and puberty for psychological reasons.

1.4.5.3 **Proximal Femoral Focal Deficiency (PFFD)**

Many congenital anomalies can affect the femur, include congenital short femur, coxa vara, as well as partial or complete femur absence which also known as proximal femoral focal deficiency (PFFD) (Shetty and Khubchandani, 1998). PFFD is a congenital disorder that effects the hip bone and the proximal part of the femur also it may or may not include ilio-femoral joint, resulting in abnormal hip and shorting as well as altered function of the affected limb, the condition can be unilateral or bilateral.

According to the ISO classification the PFFD consider to be longitudinal deficiency of the lower limb. The PFFD commonly associated with the presence of other anomalies such as in 50-80% of the cases there is a fibular hemimelia (Aitken, 1969), the lack of kneecap, as well as it may be accompanied with shortening in tibia or fibula along with foot deformities.

A subject with PFFD usually characterized by partially absence of the femur proximally with overall short limb. The biomechanical abnormalities of a PFFD individual are limb-length discrepancies, malrotation (abnormal rotation), instability of the proximal joint, as well as proximal musculature inadequacy (Kudla et al., 2016, Aitken, 1969).

Vascular changes associated with PFFD have been found in a study by Chomiak et al. (2009) using computed tomographic angiography. In addition, a knee arthroscopy showed that there are ligamentous changes connected with the PFFD particularly to the cruciate ligaments (Chomiak et al., 2012).

1.4.5.3.1 Causes

The cause of the PFFD is unknown (Kudla et al., 2016); many assumptions has been linked to cause PFFD, including ischemia, irradiation, mechanical or thermal injury as well as chemicals and hormones agents (Aitken, 1969, Epps Jr, 1983, Panting and Williams, 1978).

A study by Epps Jr (1983) introduced a theory that may result in PFFD, saying that damage to the cells of neural crest that form the precursors of the peripheral sensory nerves of Lumbar 4 and 5 lead to PFFD. Another investigation has (Boden et al., 1989) suggested another theory for the cause of PFFD, arguing that a defect in proliferation (a process in which results in an increase number of the cells) and maturation of the chondrocytes (the only cells that found in cartilage connective tissue) in the proximal growth plate may result in PFFD. However, according to Panting and Williams (1978), the thalidomide is the only factor that has showed to be the definitive cause.

No proof showed that the PFFD can be a result of genetic factor (Koman et al., 1982, Oppenheim et al., 1998)

1.4.5.3.2 Epidemiology

PFFD incidence range is between 1 to 2 per 100,000 live births (Oppenheim et al., 1998, Kudla et al., 2016). The PFFD consider to be the most common lower limb longitudinal deficiency. Depending on a study of Biko et al. (2012), the PFFD mainly happened in unilateral form approximately 85-90% unilateral and 10-15 % of the cases are bilateral.

1.4.5.3.3 Classification

Based on the X-ray images, many have introduced a classification for the PFFD such as (Frantz and O'rahilly, 1961), (Amstutz and Wilson, 1962) , (Aitken, 1969) , (Hamanishi, 1980) and (Gillespie and Torode, 1983, Torode and Gillespie, 1991).

Aitken classification of the PFFD consists of four types based on X-ray;

Type A

- Consider to be the least in severity degree of all 4 types.
- Short femur (as in all types) with coxa vara as well as lateral bowing in upper third of the femur.
- Adequate acetabulum that contains the femoral head.
- Pseudarthrosis development at the Sub-trochanteric region.
- In most cases, there is pseud-oarthrosisossification at the skeletal maturity, however the varus angulation may be very severe.

Type B

- Delayed ossification in the capital femoral epiphysis as well as and a mild dysplastic acetabulum.
- The upper end of the femoral shaft placed above the head of femur.
- At skeletal maturity, there is osseous connection (usually by defective cartilage) is seen between the femoral head and shaft (fail to ossify).

Type C

- Severe dysplastic acetabulum.
- Absence or very short femoral head with no attaches to the femoral shaft.
- There is tapered (reduced in thickness) at the proximal end of the shortened femur.

Type D

- The most severe type of PFFD, absence in both acetabulum and proximal femur.
- No proximal tuft is present; the femur is only represented by the femoral condyles.

A description of the other PFFD classification methods is presented in (Appendix II)

1.4.5.3.4 Management of PFFD

It is important to bear in mind that there is no single approach of intervention that can be applied to all patients with PFFD, each has to be assessed individually. For instance, Panting and Williams (1978) who used Amstutz's classification in his study, have set the basic difficulties when dealing with PFFD patients; hip instability, mal-rotation, inadequate proximal musculature, and inequality of leg length. The hip instability; the reason that cause this is related to the abnormal development of ilio-femoral.

(Doig, 1970) suggested arthrodesis (joint ossification) between the distal femur and the pelvis, which make the knee to be operated as a hip.(Amstutz, 1969) advised femoral head and neck excision with transplantation of the fibula.(Westin and Gunderson, 1969) offered a proximal segment excision and a femoral shaft placement in the acetabulum.

According to Cristini (1973) the intervention include many arthroplasties including reconstruction of the acetabulum and interposition of the soft tissue as well as femoral excision and placement of the tibia in the acetabulum.

The muscles, in most cases gluteal muscles are present in a good condition, the quadriceps commonly are in a hypoplastic condition. A demonstration of active extension of the knee and flexion of the hip and the Sartorius muscle is a posture called ‘sitting tailors’. This posture included a bulky and strongly muscle contraction. Furthermore, action of the hamstrings and iliopsoas in the presence of a pseudarthrosis of the upper femoral shaft may also play a part.

According to Panting and Williams (1978), surgical treatment is not required since the function is good despite the deformed posture. However, Westin et al. (1976) suggested a procedure in order to reduce the fixed flexion deformity by leg conversion to be a single lever by knee fusion.

Leg length discrepancy considered as the major disability with all PFFD types ranging from mild degree in type 1 to be severe in type 5. Generally, the femur represents approximately 20 to 40 % the length in relation to the normal.

Amstutz (1969) had done a follow-up to PFFD patients and suggested that the difference in femoral length continued to be constant during growth or increased in case that proximal migration happened at the pseudarthrosis.

In case that the inequality in length is relatively small, leg-lengthening procedures or epiphysiodeses of the normal leg may solve the issue. the combination of formal shorting, the flexion deformity at the hip joint and the mild hypoplasia of the leg will lead the foot of the abnormal side to be at the same level of the normal knee at the skeletal maturity. To manage these various options can be offered.

Shoe raise (Patten), this method can be used when the child has relatively short limbs and the discrepancy is not so remarkable. However, this may not functional and cosmetically acceptable when inequality becomes more noticeable.

Rotation-plasty is a surgical operation wherein a limb segment is removed and the remaining limb below the removed segment is rotated 180 deg. and reattached. In the case of PFFD, ankle functions as knee joint. It was first by described by (Borggreve, 1930) then popularised by (Van Nes, 1950).

The reason behind performing such a procedure is to provide the advantages of a below knee amputation in a situation where the other options can be an above the knee, hip disarticulation or a fixed-knee extension prosthesis, the indication for rotation-plasty, severe PFFD, cancer and sometimes infection (Wick and Alexander, 2006).

Despite that this method offers the design and function of a below-knee prosthesis, two concerns can be faced, first the difficulties in prosthetic fitting as well as poor cosmetic appearance (Panting and Williams, 1978).

Ankle disarticulation option (Syme's), Aitken (1959) discussed the benefits of Syme's amputation in providing an excellent end-bearing stump allowing ready fitting of a prosthesis.

1.4.6 Prosthetic management

Once the individual diagnosed with congenital disorder, rehabilitation is vital. The management and treatment of congenital disorder requires ongoing input from all members of the multi-disciplinary team to achieve the best possible results (Boonstra et al., 2000).

The rehabilitation team may consist of a physician, prosthetic and orthotic specialist, physiotherapist, occupational therapist, others and the patient family and friends.

Following the thalidomide crisis in the UK, a society called 'Thalidomide Society and Trust' was established which aimed to manage the financial settlements that the drug manufacturers and the British Government offered (Mustapha, 1990). During that time the prosthetics providers found many difficulties managing these deformities. According to (Mustapha, 1990) in the UK, four units committed to develop the needed prosthetic and/or orthotic designs and fitting techniques. The study said that the prescriptions of the patients were altered and the new methods that were available that time were taken into consideration including thermoplastic, light-weighted modular limb components, new feet and electrically powered hand and elbows.

Despite that, the research did not include any information about the specific changes and how these changes affected the outcome of the patients. It mentioned that the change of prescriptions may include the use of new feet for example, but it did not say anything about how these feet effect the gait of the patient for instant.

Regarding prosthetic management, there are issues to consider when fitting a patient with prosthetics including, the functionality of prosthetics the weight of the prosthesis, the cosmetic appearance, the alignment and choosing the optimal components that meet the patient's needs.

A research focused on the effectiveness of prosthetic rehabilitation of children with limb deficiencies present at birth done by Sener et al. (1999) showed that 84.7% of the lower limb group became independent walkers without requiring walking aids. Meanwhile, in the upper limb group 41% were completely independent in self-care, feeding and hand skills. These statistics show how the rehab plays a critical role in patient's life and much care should be given to this group. Similar results were found in a study by Boonstra et al. (2000).

The management of a congenital patients can be a challenging task since there is no single protocol of treatment to apply for all patients. Obviously, children are not adults. They need interdisciplinary care involving their parents' help. Particularly, sensitive introduction to the idea of limb replacement.

The initial lower limb prostheses are advised in the case of lower limb defects as soon as the child can pull him/herself up which can be performed for example by holding onto furniture. Usually, this is possible at the age of 8^{th} and 9^{th} months (Wenz et al., 1998). Delays in providing the prosthesis in cases that no complicating factors are found may minimize the child's acceptance to the prosthetic and therefore the wheelchair would be the preferable intervention (Boonstra et al., 2000).

The prostheses aim to facilitate the locomotion and also to approximate the outward body as much as possible. For children, prostheses can be manufactured with light materials for the joints and adapters.

Usually for the ankle-foot mechanisms, infants were given SACH foot to provide a standing stable base as well as flat initial contact by the age of walking. This foot offers durability is affordable and lightweight. When the child reached the school-age and adolescence, energy storing releasing feet (ESAR) are prescribed, which offer flexible keels that bend in loading stage in early stance to store energy and recoil in late stance to release energy. This gives the individual more spring when walking, running and jumping. For the alignment of the feet in relation to the socket, at first the foot should be aligned anteriorly which gives more stability according to the paper. The alignment afterwards can be changed to favour mobility as the wearer gains confidence (Eshraghi et al., 2018, Krebs et al., 1991).

Several case studies have introduced a special prosthetic intervention for congenital cases. A study by (Hall and Bochmann, 1969) discussed the prosthetic options that a PFFD patient can get, the study divided the intervention for bilateral and unilateral. In the paper, they discussed the advantages and disadvantages of the rotation-plasty, saying that the effective gait can be granted by extension prosthetics but still awkward device that does not offer knee bending either during walking or sitting. The advantage of Van Nes prosthesis that the child can obtain some control of the prosthetic knee unit and can bend his or her knee on sitting, kneeling, and bicycle riding.

Roux and Pieters (2007) discussed the prosthetic management of a PFFD case 56 years after rotation-plasty. The case showed that complex prosthetic fitting problems many years after rotation-plasty in PFFD can be managed leading to a satisfactory level for the patient. Finally, that, the key for a satisfactory prosthetic management is a proper evaluation of such unique cases.

Other papers have compared the functional outcome between having an amputation or unique prosthesis for congenital cases. Recently, Calder et al. (2017) have examined the benefits of each option for prosthetic users that have congenital absence of the fibula. The study included thirty-two prosthetic users which can be considered as a good sample size for this group. The study concluded that individuals who have had amputation in childhood showed significant enhanced short-term functional outcome with prosthetic intervention. Whilst prostheses interventions offer reasonable long-term function however, outcome scores were lower.

It was noticed that studies that focused on the clinical prescription and use of prosthetic foot and ankle mechanisms usually did not include prosthetic users with congenital related limb loss. Hafner (2005) in a review of the literature showed that the main population of the studies that aimed to evaluate the feet and ankle, were as a result of trauma followed by vascular and only two as a result of congenital disorders. This shows the insufficient information about the prosthetic foot that can be fitted for the congenital related patients. In addition, lack of knowledge for the clinical factors to consider for prescription in case that the patient is congenital related.

1.5 Why dynamic stability is important

This chapter will explain the importance of dynamic stability, examine the currently used methods to assess this and provide reasons for using the *margin of stability* (MoS) as a method to evaluate human balance. The dynamic stability of able-bodied walkers and prosthetic users will be described. Potential gaps in the literature will then be highlighted to obtain the critical hypothesis and research questions of this study.

1.5.1 Falls are a health care problem

Falls are one of the leading causes of injuries including bone fractures, head injuries, and even deaths (Bergen et al., 2016). According to the NHS, falls are considered as the most common cause of injuries leading to death in people over the age of 75 in the UK (NHS, 2020). Approximately 1 in 3 adults aged 65 and over will experience one fall a year, with half of that number facing more frequent falls.

Populations such as those with lower limb loss/absence have an even higher incidence of falls. Many studies such as Miller et al., 2001, Miller and Deathe, 2004 and Pauley et al., 2006 have reported that the up to 50 per cent of subjects with lower limb absence encounter a fall each year. It was found that most falls take place during walking (Niino et al., 2000, Tinetti et al., 1995).

In addition to the physical impacts, falls also influence quality of life. For instance, an increase in the number of falls may lead to an increase in the fear of falling, thus resulting in the limitation of a persons' activities and social engagements. Such factors may also have an effect on the mental health of an individual, such effects could include: depression, social isolation, and feelings of helplessness (Pin and Spini, 2016).

The total cost of injuries caused by falls in the UK in 2013 was estimated to be around £2 billion (Tian et al., 2013). Whilst in other countries it can be even higher, according to the U.S. Centre for Disease Control and Prevention, in 2015, the total medical costs for falls was $50 billion in the USA, these costs included treatment and intervention strategies.

Fall prevention training programmes are therefore crucial as they emphasize practical strategies that may be able to reduce fear of falling and increase activity levels. Hence, such strategies are essential in order to prevent falls and understand how humans maintain balance.

1.5.2 Causes of falls

Several risk factors have been related to increase the chances of having a fall (Masud et al., 2001, DeCarlo and Bradley, 2020). There are over 400 risk factors for a fall. These falls can be categorised into three groups; environmental, task related and personal.

Environmental factors are considered as the external factors and include wet floors; poor lighting; obstacles like carpets and rugs and external perturbations (Ahmad et al., 2017). Task related factors are mainly due to the complexity and speed of a task, such as tiredness or fatigue and load lifting (Ahmad et al., 2017). While the personal risk factors are considered as natural factors that reflect personal differences among individuals. One example of a personal risk factor is age; older people have a higher risk of falling due to balance problems; muscle weakness; poor vision and long-term health conditions such as dementia and hypotension. Additionally, gender; ethnicity; drugs and medications usage; living alone; psychological status; impaired cognition and foot problems may also affect balance (Deandrea et al., 2010, Hamacher et al., 2011, Society et al., 2001).

A history of falls, balance and gait impairments may also indicate a higher risk of falls and are commonly observed in the elderly and prosthetic groups (Rubenstein and Josephson, 2006, Hamacher et al., 2011, Knight, 2018, Steinberg et al., 2018).

Falls most commonly occur during walking, and may result from trips or slips (Davis, 1983, Robinovitch et al., 2013). In everyday activities, individuals must respond to unbalanced situations or perturbations to maintain stability and prevent falling. Perturbations may be internal such as neuromuscular noise; expected perturbations as in walking over uneven surfaces or un-expected falls such as slips (Bruijn et al., 2013).

1.6 The concept of stability

The term ‘stability’ describes a system behaviour when the body state is either in equilibrium (static) or changing with time (dynamic).To define a system as ‘stable’, it should remain in or go back to its state after being disturbed (Beatty, 2005). If this concept is applied to the human walking pattern, the word ‘gait stability’ includes biomechanical aspects of stability during walking. The dynamic stability may then be defined as the capacity of the human neuromuscular system to successfully bring back or/and keep working in spite of disturbances that come about during activities of daily living (Bruijn et al., 2013). As a result, study of gait stability is correlated with two critical aspects: the body centre of mass and base of support (Bruijn et al., 2013, Hof et al., 2005, Winter, 1995, Ivanenko et al., 1997, Pai and Patton, 1997)

1.6.1 Centre of mass (CoM) Concept

The body mass is defined as 'the quantity of matter composing a body' (Jammer, 2009). The ‘centre’, by definition, is the average position of all points, which form an object. In every object, there is a point at which the object can be balanced. That point is where the mass is equally distributed in all directions. This hypothetical point is called centre of mass (CoM). It is a useful reference point to allow calculations in mechanics that provides a simplified way to visualise an object's motion. In biomechanics, the CoM is used to understand the human locomotion system (Minetti et al., 2011).

1.6.2 Base of Support (BoS)

The base of support (BoS) is the area under an object or individual that includes points of contact that the object or individual makes with the supporting surface (Krebs et al., 2002).

Conventionally, in order to achieve balance, the vertical projection CoM should be within the BoS (Winter, 1995, Horak, 2006). However, whilst this is the case during standing, further factors must be considered while walking. The findings of (Pai and Patton, 1997) revealed that the velocity of CoM should be included while studying the stability because subject is considered unbalanced if the CoM velocity is outwardly directed even if the CoM is above the BoS.

To include the velocity of CoM to evaluate stability, a measure called the extrapolated centre of mass (XCoM) was introduced (Pai and Patton, 1997, Hof et al., 2005). This measure provides a way to assess the CoM position and velocity in relation to the BoS. As both the CoM and BoS are in a continuous state of change, gait stability can be described as the capacity to control the XCoM whilst BoS is changing.

1.7 Methods used to study dynamic stability

To understand the dynamic stability the relationship between the CoM and BoS must be studied. This relationship can be measured and quantified providing a tool that can contribute to understanding the subject-specific fall risk. Several measures have been introduced to assess the dynamic stability; feasible-stability-region, stabilisation and destabilisation, local dynamic stability and margins of stability (Bruijn et al., 2013). Hence, in this section, the methods used to evaluate the dynamic stability will be briefly addressed and the rationale for selection of the MoS will also be included.

1.7.1 Feasible-stability-region (FSR)

The FSR was developed by (Pai and Patton, 1997). The concept was based on the inverted pendulum model, particularly the foot segment to find a range in the Anterior-posterior (AP) direction of feasible CoM velocity and position combinations for movement termination. A score of 1 is considered stable and indicates that balance is maintained within this range, (such as when the CoM motion state lies within the FSR). Conversely, a motion below the lower boundary of the FSR (such as a negative AP stability) means that the backward balance is more likely to be lost and a backward fall would be initiated. Also, a motion state exceeding the upper FSR boundary would initiate a forward fall meaning that the forward balance would be lost (Figure 1.17). Additionally, the greater distance from the FSR boundaries implies the higher the level of instability.

Initially, a two-segment sagittal model was used to study the BoS and the pendulum, with the assumption that the foot position is symmetrical and stationary for both sides (Patton et al., 1997). Later, the concept was modified and used to study the gait stability in regards to unbalanced situations (forward slips) by using seven- segment model (Pai and Patton, 1997). FSR was then also applied in the frontal plane to assess the dynamic stability in the ML direction (Yang et al., 2008, Yang et al., 2009).

FSR has been adopted to assess stability in several research studies conducted by the group of Pai; to test the stability when facing backward slips perturbations using a moveable platform as in (Yang et al., 2008), over-ground slips perturbations in (Bhatt et al., 2011) and treadmill based slips perturbations in (Liu et al., 2015, Yang et al., 2018). The results of the group projects have helped to understand how faster walking speed may be more advantageous to avoid forward slips (Yang et al., 2009) and how failing to recover from forward slips may be linked with lower gait stability (Bhatt et al., 2011, Lee et al., 2018).

FSR has not been used by other research groups (Bruijn et al., 2013). This may be as result of introduction of the margins of stability (MoS), which is derived from the same concept but uses relatively simpler biomechanical reasoning (Hof et al., 2005).

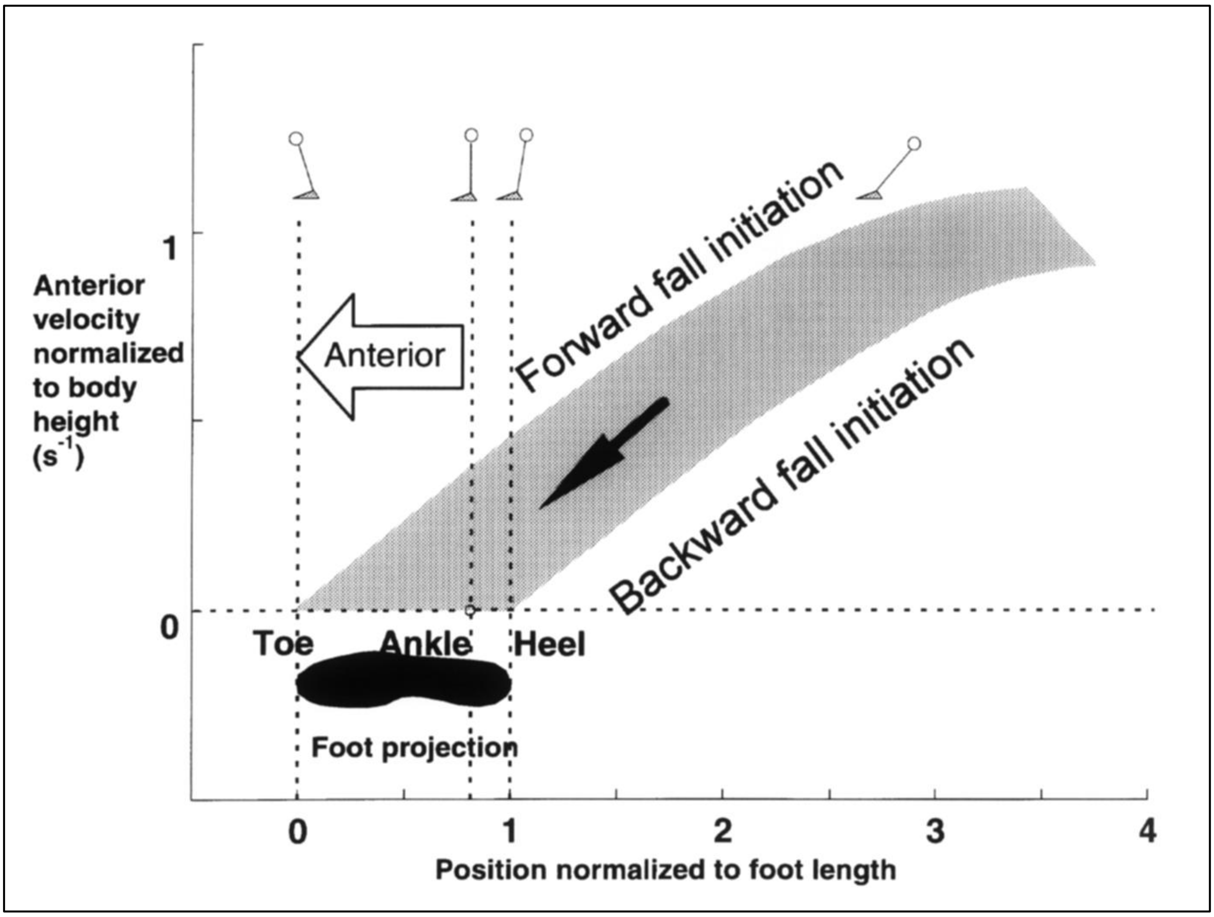

Figure 1.17 Feasible-stability-region (FSR) for the AP stability. When the combination between the foot projection (X axis) and CoM AP velocity (Y axis) is within the boundaries of FSR (shaded area) a one considered stable, whereas exceeding the upper and lower boundaries will result in forward fall and back fall initiation respectively. Adopted from (Pai and Patton, 1997).

1.7.2 Time-to-Contact (TtC)

The time to contact (TtC) is a variable measure that combines instantaneous information about the CoM trajectory. These measures were adopted by several studies to show how much time the CoM takes to cross the stability boundary which in this case is the BoS boundary (Haddad et al., 2006). The concept was first introduced by (Slobounov et al., 1998) based on the work of (Lee, 1976). The method includes the displacement, velocity and acceleration aspects of CoM movement related to BoS. Additionally, (Slobounov et al., 1998) reported that the TiC may indicate the needed time to recover from a disturbed situation before balance loss.

Although the time to contact (TiC) measure seems to provide an ability to assess the stability, several points make the measures difficult to adopt. Firstly, the measure involves an in-depth use of mathematical equations and methods of application this concept have varied between studies mainly in whether the CoM or CoP should be used. Additionally, results of a study by Lugtigheid and Welchman (2011) who examined the methods used to measure time-to-contact, concluded that there was a large variability in the accuracy of estimates of this measure. Finally, variations were also found in filtering techniques and calculations regarding how the trajectory CoM or CoP is extrapolated in regards to the BoS (Slobounov et al., 1998, Haddad et al., 2006).

1.7.3 Local dynamic stability (LDS)

The Local dynamic stability (LDS) was first described by (Dingwell and Cusumano, 2000) and measures the average logarithmic rate of a system divergence following a small perturbation (Bruijn et al., 2013). In general, the calculation of LDS quantifies if a system's current state (for example, position and velocity) is affected by the applied perturbation. One of the strong aspects of this measure is that it may be estimated from any kinematic data with no consideration to the reference frame in which the data are captured (Gates and Dingwell, 2009).

To calculate the LDS, a state space should be constructed from kinematic data of steady state walking (in which no perturbations are applied). Any kinematic time series can be used to reconstruct a state space, however, trunk kinematic data are suggested to be the most sensitive in detecting differences (Bruijn et al., 2009). The next step of the calculation is to identify the nearest neighbouring trajectories in state space for each data point and then to ascertain Euclidean distances between these points, finally finding the slope of average logarithmic rate of divergence. Lower LDS values indicate more stable gait.

The LDS indicates the system capacity to recover from small perturbations and provides a clinical measure to assess gait stability. In addition, the LDS has been adopted in many studies with several groups; for example to detect the stability difference between young and old adults as in (Roeles et al., 2018) and for adults with transtibial amputation as in (Beurskens et al., 2014).

Nevertheless, two points should be taken into account when adopting the LDS for assessing the stability. The LDS requires careful calculation; the chosen state space must contain the same number of strides for every condition and subject (Bruijn et al., 2013).

To achieve a statistically accurate estimation, relatively large datasets are required (Bruijn et al., 2009). Statistical accuracy is generally obtained when the time series is increased to include more than 150 strides. This may sometimes be difficult to achieve especially with less able-bodied participants.

1.7.4 Stabilising and destabilising forces

The concept of stabilising and destabilising forces was first presented by Duclos et al. (2009). This concept aims to assess the forces required to control the CoM motion with regard to the BoS.

Stabilising forces (F_{st}) can be defined as 'the needed force to stop the CoM from moving outside the BoS during any task', and can be calculated as follows;

$$F_s = -\frac{1}{2} . \frac{m . vCoM^2}{D}$$

m	The body mass (kg)
vCoM	The CoM velocity (m/s)
D	The minimum distance between CoP and BoS border

Greater stability force values mean that more work is needed to prevent the CoM from moving outside BoS which in turn indicates lower stability.

On the other hand, the destabilising force (F_d) may be defined as 'the needed force to move the CoM outside the BoS border and can be found as follows;

$$F_d = \left(\frac{GRF.n}{CoMz}\right) D$$

GRF	The ground reaction force
n	The unit normal vector of the contact surface
D	The horizontal distance between CoP and BoS border
CoMz	The CoM height

It is worth noting that the F_d does not take the velocity of CoM into consideration.

Finally, the index of stability may then be found by dividing the F_d by F_s this ratio provides a way to show how easy it is to make the body fall from its current position and what effort is required to do so. The F_d and F_s were only adopted in a handful of publications and mainly by the same research group. While F_s may be more interesting as it shows the necessary efforts to be stable with considering the CoM velocity, F_d shows little validity and may be considered as too simplistic by some authors (Bruijn et al., 2013).

1.7.5 Margins of stability

Several papers by Hof et al (Hof et al., 2005, Hof, 2008) have discussed the dynamic stability in detail. The work of (Pai and Patton, 1997, Iqbal and Pai, 2000) reported that the velocity of CoM should be included when studying the dynamic stability. Hof et al based on the simple inverted pendulum model (Winter, 1995), have introduced a new definition as well as a new measure, the extrapolated centre of mass position ‘XCoM’ and ‘margin of stability’.

1.7.5.1 Extrapolated centre of mass (XCoM)

The extrapolated centre of mass (XCoM) is defined as a spatial measure to describe the CoM motion state by extrapolating the CoM position in the direction of its velocity. This concept helps to extend the classical concept of static equilibrium of the inverted pendulum model, in which to achieve stability the CoM must be over the BoS by adding the velocity of CoM into the calculation.

The XCoM can calculated as;

$$XCoM = CoM + \frac{vCoM}{\sqrt{g/I}}$$

CoM	The position of CoM in the one direction
vCoM	The CoM velocity in the related direction
l	The equivalent pendulum length based on the height of CoM
g	represents the acceleration of gravity (9.81 m/s^2)

The XCoM may be used to find the spatial margin of stability (MoS, or 'b' as reported first by Hof et al 2005), and a temporal stability margin (b_t).

The margin of stability assesses the distance between the XCoM and the boundaries of BoS, accordingly, one can be considered stable when the XCoM lies within the BoS. When the XCoM exceeds the border of the BoS, a corrective step is needed to recover balance and avoid a fall, therefore one may be considered unstable, (Hof et al., 2005). The MoS can be used to assess the balance in the ML and AP directions (Figure 1.18)

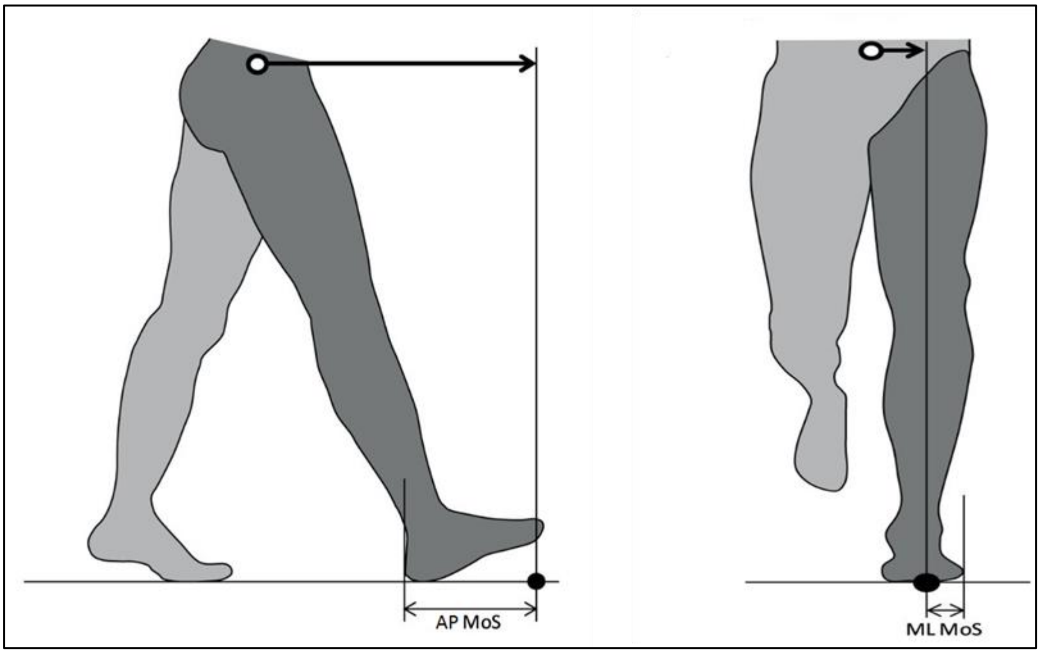

Figure 1.18 Schematic representation of margins of stability (MoS)which is defined as the minimum distance between body CoM position corrected to its velocity i.e. XCoM and BoS border, and can be estimated in both the AP and ML direction

Another measure was also presented by (Hof et al., 2005) in the same paper is the temporal stability margin. This measure shows the time in which the BoS stability borders would be reached with no intervention (e.g. a corrective step). This can be found using;

$$b_t = \frac{MoS}{vCoM}$$

1.7.6 MoS as a tool to evaluate dynamic stability

To prevent a person from falling it is essential to ensure dynamic stability during movement. Some studies have suggested that to maintain the balance during walking, the vertical projection of the body centre of mass (CoM) should be within the base of support (BoS) taking into consideration the velocity and direction of the CoM (Hof et al., 2005, McAndrew Young and Dingwell, 2012).

MoS has been used to study a variety of groups and different participants. It was adopted to study the stability of able-bodied subjects (McAndrew Young and Dingwell, 2012, McAndrew Young et al., 2012, MacLellan and Patla, 2006, Aprigliano et al., 2015, Hak et al., 2013a, Hak et al., 2012). Moreover, MoS has also been used to assess the stability of less able-bodied subjects such as in a study of participants with multiple sclerosis (Peebles et al., 2017), a history of CVA (Punt et al., 2017), Parkinson's disease (Martelli et al., 2017b), and unilateral peripheral vestibular disorder (McCrum et al., 2014).

MoS has been adopted to analyse the stability of lower limb prosthetic users for both trans-femoral (Monaco et al., 2017, Hof et al., 2007) and trans-tibial level (Curtze et al., 2010, Curtze et al., 2011, Gates et al., 2013, Beltran et al., 2014, Hak et al., 2013c). In addition, it has been selected to evaluate the differences between the stability of young and older adults (Roeles et al., 2018, Martelli et al., 2017a).

The use of MoS has proven to provide a successful model to study stability when walking in numerous conditions including continuous perturbations such as platform oscillations (Beltran et al., 2014, McAndrew Young et al., 2012, Hak et al., 2013c) and rough surfaces like loose rocks (Gates et al., 2013). The method was also used in a study that included treadmill perturbations like change of treadmill belt speed (Aprigliano et al., 2015, Punt et al., 2017, Sivakumaran et al., 2018); a rope attached to the waist to provide pulls and pushes (Hof et al., 2010, Vlutters et al., 2016); and additionally, evaluating the differences in dynamic stability during over-ground against treadmill walking (Rosenblatt and Grabiner, 2010).

Using the MoS model has several key advantages. Firstly, finding MoS is straightforward in measuring the relationship between CoM and BoS. MoS takes into consideration both velocity and the position of CoM (Hak et al., 2012). This formula eliminates the assumption of fixed ground reaction force and a fixed center of mass; thus, there is no need to find a different formula for each different position (Terry, 2014). The MoS equation therefore produces more results with time, and flexibility from which to measure MoS during walking and standing.

MoS can also be simply predicted and measured using several methods; such as through the use of kinematic data and reflective markers (Hak et al., 2012). MoS may also be measured using force-plate data alone by filtering the CoP data to obtain the CoM position, assuming that the BoS position is the same as CoP position (Hof et al., 2007). Finally, it can be found using a combination of both kinematic and force-plates as in (Hof et al., 2005). MoS may also be applied for different positions such as standing and bending forward or backward. These methods all reduce required time and effort from which to estimate MoS.

MoS therefore provides a simple applicable and a reliable tool to study the human balance. Additionally, MoS may be a helpful method in understanding how quickly subjects are able to recover the state of stability after experiencing a perturbation, by examining foot placement in terms of the extrapolated centre of mass position (XCoM).

Previous studies have supported that MoS is a useful method by which to assess the dynamic stability in different groups and situations. MoS was therefore selected as a primary measure in this study.

1.8 Perturbation

Since the main goal of gait stability studies is to understand how individuals may react to a fall/stumble and recover, it is not surprising that at least one type of perturbation was applied to mimic a fall and assess the stability. Mechanical perturbations (such as slips) and/or altering sensory input (such as visual perturbations) can influence walking stability. To achieve this, researchers have adopted and developed numerous methods of perturbations. In this section, an overview of these methods will be addressed. This review aimed to provide insight about the biomechanical effect of the perturbation on the gait stability of two groups; able-bodied adults and prosthetic users.

The term ‘slip’ is commonly used to describe disturbances that occur/ are applied during leg stance phase, whilst the term ‘trip’ or ‘stumble’ used to those that occur/are applied during leg swing phase (Pijnappels et al., 2001, Martelli et al., 2017a, Patel and Bhatt, 2018, Allin et al., 2020).

Various studies have discussed the effects of trips and slips in terms of which may be most challenging. Findings suggest that slips are linked with higher fall risks than trips (Patel and Bhatt, 2018, Bhatt et al., 2012, Grabiner et al., 2008, Bassi Luciani et al., 2012). Therefore, in the current study the methods were directed towards imposing slips in order to challenge the prosthetic users’ stability. However, it was noted that the differences in the recovery strategies when applying these different perturbations have not been fully examined at least for the prosthetic user group. The same note was reported by a group of (Patel and Bhatt, 2018). The differences between slips and trips have been discussed for the older adults aged 61-75 years as in study of (Allin et al., 2020). The results of the study support the idea that slips trigger significant differences between the perturbed trials and the baseline (steady state) whilst no significant differences were found when imposing trips. This can be one of the gaps that need to be evaluate in the future studies. For the purpose of this study and to help to keep the trial time as reasonable as possible for the prosthetic users the methods included slips only.

The perturbations are either applied continuously or individually (discrete). Continuous perturbations are imposed over long periods that result in multiple subsequent steps (McAndrew et al., 2010). In contrast, discrete perturbations are usually imposed within one step (perturbed step) followed by one or more corrective steps (Sessoms et al., 2014).

1.8.1 Perturbations adopted

1.8.1.1 Able-bodied studies

A number of studies have used platform movement in order to challenge support surface. (Shapiro and Melzer, 2010, McIntosh et al., 2017, Oliveira et al., 2012, McAndrew et al., 2010). McAndrew et al. (2010) developed a novel approach to challenge the gait stability by creating continuous perturbations taking advantage of an advanced system called CAREN. The system can provide different methods of perturbation including a platform movement and visual obstacles. The platform has a 6 degree of freedom motion base. More details of this system are described in chapter 2. The group method of perturbation was imposed as pseudo-random oscillations including both the AP and ML directions of platform movement as well as visual field. The oscillations were based on equation consisting of a sum of four sine waves of different frequencies:

$$D(t)= 0.05[1.0\sin(0.16*2\pi t) + 0.8\sin (0.21*2\pi t) +1.4\sin (0.24*2 \pi t) +0.5\sin (0.49*2\pi t)]$$

Where D(t) is the translation distance in (m), t is time in (s).

This method has been adopted in various study with various groups including able-bodied and prosthetic users (Hak et al., 2013c, Beltran et al., 2014, Beurskens et al., 2014, McAndrew et al., 2011, Franz et al., 2015).

A recent study by Roeles et al. (2018), consisted six different types of individually imposed perturbations; two platform sways, belt deceleration, belt acceleration, visual and auditory perturbations using the CAREN environment. The study perturbations' method has several advantages, the perturbations were sudden and very unpredictable. The protocol was presented in a random order. The onset of each perturbation was not the same, they were triggered randomly every 10-15 steps. This also helped in reducing the ability of the participants to proactively adapt their gait and stability to cope with the upcoming perturbations aiming to real life unbalance situation. In addition, the perturbations can be triggered separately and randomly on each side, this may be very helpful in assessing the potential differences in responses of each leg. This can be useful to answer research questions such as whether dominant leg shows relativity better stability and recovery mechanisms. The same applies to answer the question about the potential functional differences between sides for less able-bodied such as prosthetic users as a inconstancy was found among the reports that have tried to investigate this aspect (more details are addressed in section 1.6.3).

Roeles et al. (2018) study demonstrated that the visual and auditory (sensory) perturbations did not significantly challenge the gait stability. While both the acceleration and deceleration slips did provoke a response, the deceleration slips were more challenging than acceleration. Providing platform sways has affected the gait stability as well, but the one applied on the ipsilateral side was less challenging than the one applied one the contralateral side.

A group of Pai (Liu et al., 2015, Bhatt et al., 2012, Bhatt et al., 2005, Bhatt and Pai, 2009, Yang and Pai, 2007), has been working on a novel over-ground slip. Their experimental approach consists of a pair of low-friction movable platforms fixed side by side in an over-ground walkway as in (Figure 1.19). Platforms were computer-controlled and the release mechanism occurred when the participant's right side contacted the right platform. The protocol aimed to assess the immediate effect of slip training on stability. Although the participants were not told the location or timing of a slip or how it would happen, the slip can be only unpredictable at the first trial. If for example the test to be repeated with the same participant, the participant may be aware of what would happen. This method of perturbations would lower friction of the support surface rather than the friction between the BoS and the support surface. It noted that this method of perturbation along with the outcome measure (Feasible-stability-region FSR) has not been adopted by any other research group.

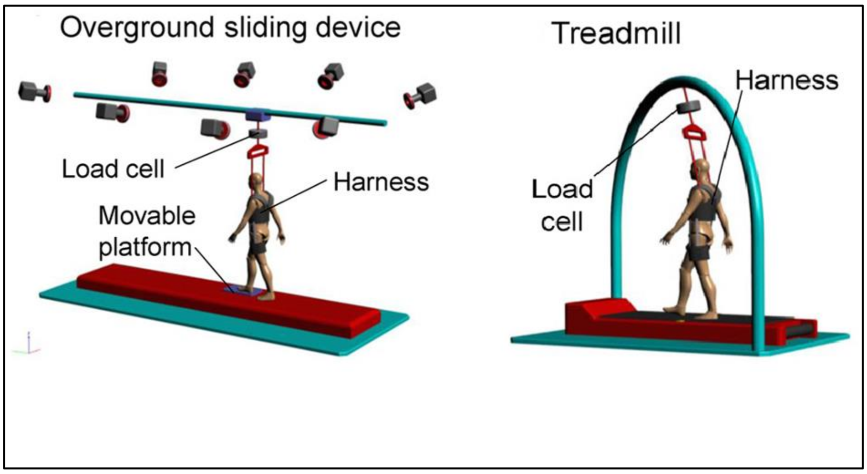

Figure 1.19 Experimental setups to impose slip perturbations during over-ground and treadmill walking.Adopted from (Yang et al., 2018).

More recently, Pai et al. used the treadmill to impose slips by applying sudden changes in belt speed (Yang et al., 2018). The treadmill in the method is called ActiveStep (Simbex, NH), and consists of a single belt treadmill that delivers a combination of forward and backward acceleration perturbations whether at initial contacts or toe off. It was noted that the perturbations profiles were integrated within the software that comes with the machine, hence the only thing that can be edited is the perturbations onset time which may add some restrictions. Their results showed that imposing slip perturbations using a regular treadmill (a limited version as they have referred to in their study) can improve dynamic stability control consequently reduce falls. This can be helpful in case that accessing to more advanced equipment is not possible for reasons such as costs and availability of such equipment. The authors, on the other hand, have addressed some limitations to the study. The perturbations in the study could be deemed as anticipatory slips and that did not create real-life slip. This may reflect on transferring the study's findings to a real-life situation. They have also mentioned that the design of the study was a blocked with no involvement of any randomised training. They suggested that if the protocol was a combination of a blocked and randomised one, the training would be more effective.

The treadmill also used to provide perturbations in many studies, Shapiro and Melzer (2010) developed a prototype to provide multidirectional perturbations (BaMPer system). The system consists of treadmill (single belt 140 cm length and 60 cm wide) mounted on a moving platform (160 cm wide and 200 cm). The system can provide longitudinally, laterally or combination of both surface perturbations. The system is controlled by a computer. According to the developers, the advantage of this system was; relatively cheaper than other systems (cost $17,000). The system was adopted recently in a study by (Madehkhaksar et al., 2018) to investigate the effects of the perturbations on gait stability and parameters of able-bodied people (n=10). Results indicated that able-bodied people changed their gait parameter in order to regulate the MoS and decrease the probability of falling. However, the researchers have reported some challenges related to the system, due to technical issues in the treadmill, some of the expected perturbations were not delivered. The treadmill did not have attached reflective markers, it was not clear if attaching markers to the treadmill was not possible at all or it was a protocol error. Nonetheless, absence of treadmill marker led to difficulties in finding the exact frame where the perturbations where provided. There was also a delay in triggering the perturbations due to the treadmill setup limitations, this resulted in perturbations not being provided in the desired timing which meant to be at the moment of mid-stance. Regarding the experiment, the trials were not conduced in randomised order which could lead participants to anticipate what would happen next thus reflected on the overall results. Finally, the authors have mentioned that the sample size is relatively small hence the need of a larger sample studies in the future.

Slips perturbation were adopted by a group of (Martelli et al., 2017a, Aprigliano et al., 2015). The group designed and developed system called SENLY, whereby the slips were mimicked be means of movable treadmill belts. The Full details of this system were described in (Bassi Luciani et al., 2012). The equipment setup consists of two belts that can be controlled independently in AP and ML directions. However, slips were only provided as a sudden and unexpected forward movement of the right belt (AP only). The slips were triggered when the right initial contact was detected by the force sensors. It was not clear why researchers only applied slips to the right side. Also whether it was possible to change slips side so that they would be applied on the left side as well or not. The results of the study showed the direct relationship between intensity of the perturbation and effect on the MoS. The study results also supported the statement that the elderly people exhibited reduced ability to recover from the slips when compared to younger group. The MoS of the elderly group was less than the younger group which indicated that the XCoM was located in a more backward position with reference to the leading foot compared to the younger subjects The group addressed a few limits of the study, the experimental protocol was constrained and not fully generalizable. Providing many perturbations at the same side each time could have affected the participants' reaction (proactive adaption effect) as the perturbations would at some point become predictable.

Finally, the sample size was small and the elderly participants in the study considered to be relatively highly active in comparison to their age group.

Other researchers have aimed to provide waist-level perturbations during walking. Techniques used include application on a treadmill as in (Martelli et al., 2017b, Hof et al., 2010). These studies used rods or cables attached to the waist that can provide pull or pushes at a certain time during walking as in (Figure 1.200). The rationale behind choosing this method according to the authors (Martelli et al., 2017b) was to study the effect of the uncommon perturbations rather than those commonly linked with causes of falls (tripping and slipping). They concluded that the imposed perturbations may not reflect to other forms of stability challenges that people face in everyday situations.

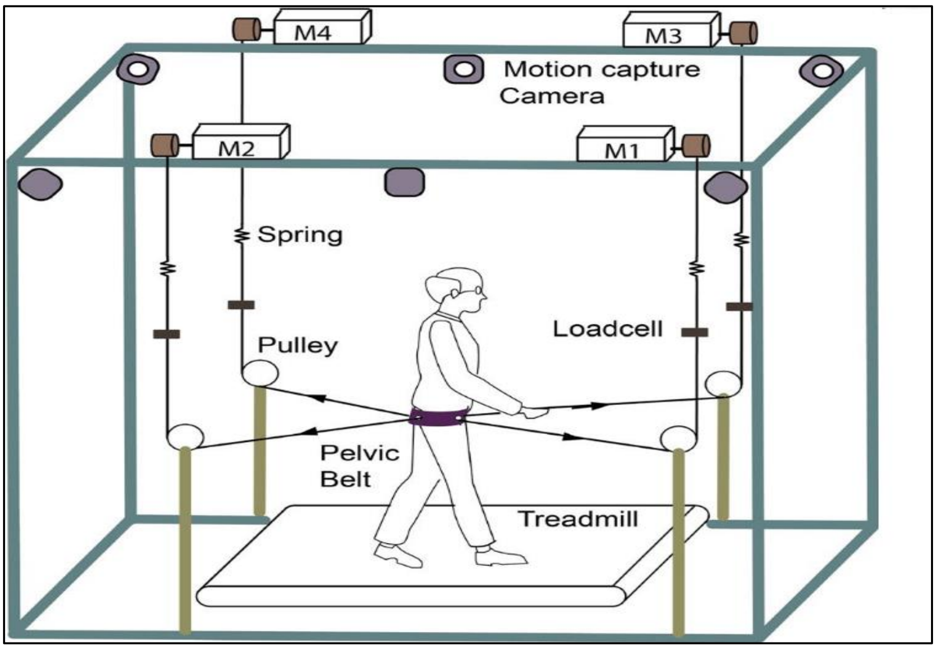

Figure 1.20 Experimental setup to impose waist perturbations in the AP and ML direction. Adapted from (Martelli et al., 2017a).

Hof et al. (2010) investigated the balance of able-bodied people after providing lateral waist perturbations. Participants experienced a push to the right controlled by a computer at every 10% of the gait cycle (i.e. at 10%, 20%, ... 100%) in the first protocol and at 40% or 90% of the gait cycle in the second protocol. Each participant was given 400 pushes over 2 hours. Similarly to (Martelli et al., 2017b), the perturbations in this study were not similar in a real time situation. The researchers concluded that providing slips or trips would trigger greater effects over the gait stability.

A further study developed a novel robotic waist perturbations for over-ground walking (Olenšek et al., 2016). Despite the several limitations, the pilot study results were promising, and the further work was suggested to optimize the method. Limitations included the inability to measure the ground reaction forces hence the studies chose the treadmill. Because perturbations were always imposed at the same side and time (following left initial contact), this may have allowed participants to anticipate the upcoming perturbations.

1.8.1.2 **Perturbation in prosthetic users' studies**

While various types of perturbations were presented to challenge the gait stability of able-bodied subjects, fewer studies have assessed stability in prosthetics users.

Most studies incorporating prosthetic users used the CAREN system as the study equipment to challenge the balance. Hak et al. (2013c) adopted the method of continuous platform movement by McAndrew et al. (2010) to provide ML platform translation as perturbation protocol for prosthetic users subjects. The same method was also adopted by Beltran et al. (2014) besides platform movement, their protocol included visual continuous perturbation.

Vrieling et al. (2008) also used the CAREN to impose perturbation to lower limb prosthetic users (mixed 3 transfemoral and 5 transtibial) by providing an AP platform sway, and visual perturbations by means of blindfolding. This study also used the Stroop test to create dual task challenge. The Stroop test is a neuropsychological color and word test involves showing words in different fonts and asking participants to read the words. For example, showing the word blue in a red font. Their results showed that the prosthetic users relied more on the contralateral side and recommended that rehabilitation should be directed towards improving control in the intact side. Also, blindfolding and a dual task activity considerably affect the outcome measures. The limitations of this study however included mix of different amputation levels. It is very challenging to assume that a person with transfemoral amputation is functionally similar to one with transtibial and well established that the knee joint plays great role in walking and overall stability (Abulhasan and Grey, 2017). The sample size was limited and participants' activity levels relatively higher, which mean that results may not be generalizable to other prosthetic users. The researchers also highlighted that the perturbations were not very unexpected, and the participants might have developed a pattern to anticipate the upcoming perturbations.

Work by (Sturk et al., 2018, Sturk et al., 2017) disused the gait stability of transfemoral prosthetic users (n=10). The study took the advantage of the CAREN 6- degree of freedom to test the stability in eight different walking conditions. In the study, continuous perturbations were imposed similar in concept to (McAndrew et al., 2011) method but different in the axes. The method included level walking was used as baseline, downhill slope, uphill slope, top-cross-slope, bottom-cross-slope, medial-lateral translations, rolling-hills and simulated rocky using rumble module. The method of perturbations appears to be inclusive and cover most of the real-world scenarios. However, the average time for each trial was 5 minutes, in addition participants walked for 10-15 minutes as a warm-up session. This might be a downside factor somehow affecting the performance outcome of the participants. Requesting a prosthetic user especially an above knee one to walk actively for around 40 minutes is rather long even when breaks are provided. This issue, however, may be solved by breaking these conditions into more than one session. The authors have also described another limitation that was associated with the various prosthetic components. In the study, the participants wore relatively varied prosthetic components which might have affected the overall walking performance across the conditions. This issue is not unpopular across the studies that assess the prosthetic user. The authors have argued that despite this variability, all participants were fitted with the components that designed for variable walking speeds and conditions. In the following study, the same group was subdivided into two groups K3 (lowest) and K4 (highest) activity level. Although the study sample was small, significant differences between these two group were found.

Sessoms et al. (2014) introduced a method to mimic a trip experience when walking by applying sudden changes in treadmill speed applied in the CAREN as well (both belts were moving at the same speed). The perturbation profile consisted of treadmill decelerations (15m/s^2) from the normalised speed for 50 ms then, acceleration (-15m/s^2) for 270 ms. The protocol was executed on 12 transtibial subjects. The study method to challenge the stability seemed promising, the authors however, reported some limitations with the study. The authors argued that simulating a trip using a treadmill is challenging as it is difficult to replicate the sensory and motor condition that occur when challenging the swing phase by external obstacles. Meanwhile, they suggested that treadmills can create whole body motion pattern that might be similar to the one following an actual trip. Finally, they suggested that their method can be optimized by increasing the clinical and laboratory tests using the instrumented treadmill like the CAREN.

Other studies did not provide any mechanical perturbations. Whereas only different walking surfaces were involved. Two relatively similar adopted this method. In Gates et al. (2013) which evaluated the stability of transtibial prosthetic users (n=12) when walking over flat ground vs. walking over a loose rock surface (Figure 1.21 a).

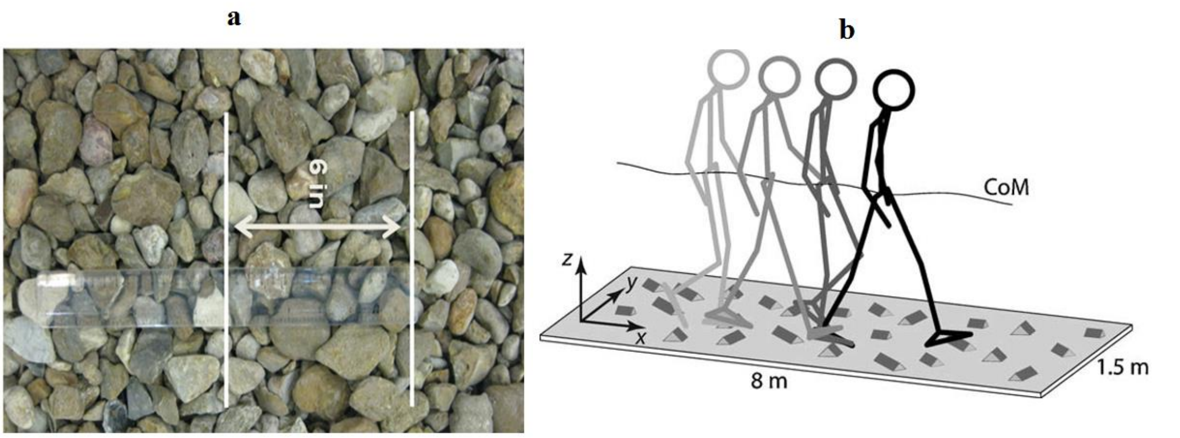

Figure 1.21 Walking surfaces used in two studies to test the stability of prosthetic users subjects while walking over them. a) loose rock surface used in Gates et al. (2013), b) custom made surface made from strips and wooden prisms used in Curtze et al. (2011).

Walking over different surfaces was also adopted in an older study of Curtze et al. (2011) who has investigated the stability of transtibial prosthetic users (n=18). This study sample is relatively large one when compared to other prosthetic users' stability studies. The stability was challenged by asking participants to walk over a custom made irregular surface with strips and wooden prisms (Figure 1.21 b).

The results of both Gates et al. (2013) and Curtze et al. (2011) studies were compared to the results when participants walked over smooth/flat surface. The methods of these study appear to be straightforward as minimal equipment is required. On the other hand, several limitations have been identified for these studies. Firstly, the methods of perturbation do not include a way to separate the reaction of each side which is needed for the current study in order to evaluate whether prosthetic users tend to exhibit asymmetry in their gait or not. This limitation has been observed in Curtze et al. (2011) study results as the margins of stability for the prosthetic side did not differ from the intact side. However, the MoS of the prosthetic side was found significantly different than the intact side in Gates et al (2013). It was noticed that in both studies the MoS was defined using the same concept in the ML direction only.

Secondly, the imposed methods may have not been strong enough to trigger a reaction as well as the participants still could anticipate the provided perturbations, this could explain why there were no significant differences in terms of MoS as well as major adaptions in the gait parameters when walking over irregular surfaces compared to walking over a flat ground. These results were also seen in a study of MacLellan and Patla (2006) who assessed the adaptions to regulate dynamic stability. Whereas no mechanical perturbations were imposed as well. They used a compliant surface made of foam as a walkway to assess the dynamic stability of able-bodied adults (n=8).

The authors of both prosthetic studies have addressed the same limitation regarding the CoP position. As they could not measure the exact position of the CoP when participant walked over the irregular surface. In Gates et al (2013) study the position of 5th metatarsal marker was used meanwhile in Curtze et al. (2011) study, the AP axis of the foot was used to find the CoP position. They have addressed that CoP positions in their studies were necessary to estimate the base of support. However, using the foot markers' positions were noticed to be also adopted in the MoS studies.

Kendell et al (2010) investigated the dynamic stability of prosthetic useres, the study included a relatively larger sample size (n=20). In addition, the researchers used portable pressure sensors that can be fitted into the participants' shoes. They are called F-Scan pressure-sensor (Tekscan Inc., Boston, USA). The participants walked over four different conditions; level ground, uneven ground (foam mats) , ramp (7-degree incline) and stairs (12-step stairwell). The method seemed to be not complex and the sensor can be a helpful research tool especially when more expensive and complex equipment are not available. However, the research group reported the outcome measure the maximum lateral force placement (MaxLat), was not sensitive to differences between limbs, conditions, and groups. They suggested further research towards the use of outcome measure for the prosthetic users. Another possible limitation might be related to the fact that the walking conditions may not strong enough to challenge the participants stability Kendell et al. (2010).

In a study of Hof et al. (2007), three different walking speeds were used to test transfemoral subjects (n=6) while walking over treadmill. No mechanical perturbations were imposed. However, participants have been asked to perform a Stroop test while walking. This test is known to be used to assess the cognitive capability (color and word test), it was included to challenge the participants while walking over the treadmill at three speeds of 0.75, 1.00 and 1.25 m/s. During the study, the prosthetic users individuals have used the side bars of the treadmill, this may somehow have affected the outcome of the study. This might have affected CoP estimation. Also, it was noted that in the study there were no details regarding the prosthetic components' types or/ and mechanisms that the participants have been fitted with.

In case of above the knee prosthetic users, the prosthetic knee type/ mechanism can play a major role in gait pattern and stability, the same but with a less degree applies for the prosthetic foot.

Another method of evoking a fall developed by Curtze et al. (2010) was by sudden release of prosthetic users subjects (n=17) from fixed forward- inclined orientation of 10% as showed in Figure 1.22.

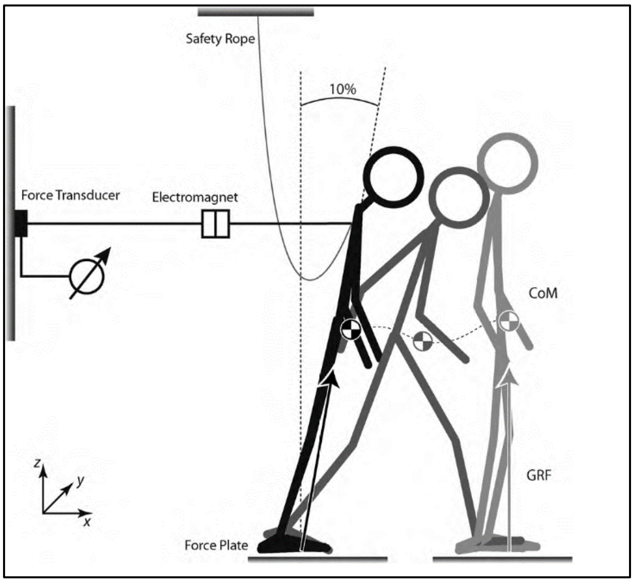

Figure 1.22 A method of evoking a fall by sudden release of the prosthetic users subjects. Adopted from (Curtze et al., 2010)

This method does not include participants walking, and arguably only allows users to exhibit one step. Although this can help in understanding the initial reaction in forward falling, one step may not be enough to fully understand the mechanisms that prosthetic users use to cope with the challenge to restore balance.

Participants movement may also have been limited by the attached cable. Using this method may not help in distinguish the potential differences between the intact and prosthetic sides. This was reflected on the results as no differences were found in MoS for both sides as well as when comparing the control able-bodied group to the prosthetic users' group. This has led the authors to assume that prosthetic users were equally efficient in recovering from the fall as controls. Another limitation of this method is that the fall itself was expected which may have given the prosthetic users to prepare to the way of recovery. This may not reflect a real-life situation whereas falls happen unexpectedly in fractions of seconds.

1.9 The MoS for able-bodied and the prosthetic users (PU)

In daily activities, static balance assessments may not be enough to challenge the basic strategies for maintaining balance as controlling the balance is often needed during walking (Buckley et al., 2002).

In order to improve gait stability during the rehabilitation of prosthetic users, it is essential to understand the strategies adopted by prosthetic users to optimize these aspects of walking, together with how these strategies may vary from strategies used by able-bodied people. This section provides an overview of the gait stability status for able-bodied.

1.9.1 Able-bodied

A study of Hak et al. (2012) found that able-bodied participants tended to increase their step width, shorten the step length and take faster steps in respond to continuous ML perturbations. This resulted in increasing the ML MoS to cope with the applied perturbations and resulted in an increase of the AP MoS. The core question of their study was regarding the walking speed and whether or not participants would speed up or slow down. Results indicated no significant changes in the walking speed to cope with the perturbations. Authors concluded that the walking speed is not a factor in a recovery mechanism. Although, the participants in the study showed clear adaptations to cope with perturbations, authors argued that these may be only used when continuous ML perturbations are applied (as in this study). This may limit the generalisation of such findings and suggests need of further investigation of discrete perturbations.

Similar results were reported by (McAndrew Young et al., 2012) in ML direction in able-bodied participants who showed decreased MoS in AP direction. Both studies used similar methods of perturbation.

Able-bodied subjects' ML MoS were reported as being the same on uneven vs. smooth surface (MacLellan and Patla, 2006) but differed in the AP MoS.

Rosenblatt and Grabiner (2010) assessed the MoS in the frontal plan only, demonstrated that the ML MoS of the able-bodied participants did not change significantly when they walked overground or on a treadmill, but that step width varied. In their study, the research group discussed the possible effect of the familiarization time for the treadmill as a potential limitation. In addition, researchers addressed that potentially there was an error in the calculated MoS as they were found to be (2-3 times) larger than other studies' results although in the study they only adopted the minimum MoS during stance phase.

In a recent investigation of stability using mechanical perturbation, Madehkhaksar et al. (2018) reported that the AP MoS were decreased as an effect of ML perturbation application whilst the MoS in ML were increased. Additionally, in their study, the AP MoS variability increased during the forward perturbations.

In response to varied means of perturbation, a study of Roeles et al. (2018), found that there were no significant differences between young and older adults in MoS after applying the perturbations, and the participants expressed decreased ML MoS and increased AP in platform movement perturbations. Another study of Martelli et al. (2017a) that investigated the differences in stability among young and elderly adults using MoS. The results supported typical assumption that the elderly individuals are less stable than the young group, and the latter group was significantly less effected by the applied perturbations. These results were in line with older studies' findings as in (Bosse et al., 2012, Carty et al., 2011, Karamanidis et al., 2008).

1.9.2 Prosthetics Users

Due to the limited joint mobility, the lack of distal muscles as well as the lack of feedback from the affected lower limb, persons with lower limb absences/amputations experience difficulty in maintaining dynamic stability during walking (Winter et al., 1996, Lamoth et al., 2010).

These factors also lead to reduced ability of control of the CoP position while standing on the prosthetic side. As a consequence, prosthetics users may tend change their stance control methods and develop alternative control strategies while walking (Viton et al., 2000, Buckley et al., 2002, Aruin et al., 1997). In addition, prosthetics users are reported to have reduced response to mechanical perturbations and ability to modify their walking pattern to comprehend environmental changes (Houdijk et al., 2012, Hofstad et al., 2006).

Therefore, prosthetics users are assumed to be less stable than able-bodied subjects and experience more difficulty when facing disturbances. Prosthetic users are also assumed to be more reliant on the intact foot to achieve balance. However, many researches who have attempted to demonstrate these assumptions reported conflicting and sometimes counterintuitive results (Curtze et al., 2010, Curtze et al., 2011, Gates et al., 2013, Beltran et al., 2014, Hak et al., 2014, Hak et al., 2013c, Kendell et al., 2010, Vrieling et al., 2008, Major et al., 2020, Kline et al., 2020).

In a study of Curtze et al. (2011), prosthetics users found to be equally stable when walking over rough and smooth ground. Conversely, prosthetic users showed increased stability when walking over uneven ground in a study of Lamoth et al. (2010).

Similarly, prosthetic users were found to be equally stable as able-bodied when recovering from perturbation according Curtze et al. (2010), or less stable than able-bodied as reported by Lamoth et al., 2010 , Beltran et al., 2014 studies or more stable than the able-bodied as found by (Kendell et al., 2010, Gates et al., 2013).

1.9.3 Between limbs differences

The reported results regarding differences in side to side were also found to be inconsistent. According to (Hof et al., 2007), prosthetics users with above-knee amputation showed greater mean MoS in the prosthetic limb than in the intact one. Also, Kendell et al. (2010) reported that the below knee subjects were more stable on the prosthetic side as well. While Gates et al. (2013) found that prosthetics users tend to be less stable on the prosthetic side and the control subjects showed no differences in stability between limbs.

In contrast, McAndrew Young et al. (2012) concluded that the able-bodied subjects showed significant differences between right and left sides.

Studies by (Curtze et al., 2010, Curtze et al., 2011) showed that the prosthetics users were equally stable on both limbs. These findings were also supported by Beltran et al. (2014), not only for the prosthetics users group but also for the control able-bodied group.

Vrieling et al. (2008) suggested that the intact limb should be given more attention including complex balance tasks during the rehabilitation programmes. This was also suggested by (Hof et al., 2007, Schoppen et al., 2003)

By contrast, Curtze et al. (2010) suggested that both limbs should be trained equally as the prosthetic users were equally efficient in both feet when recovering from falls.

1.9.4 Recovery Strategies (compensatory)

To maintain balance during dynamic tasks, corrections to BoS relative to CoM are achieved through proper foot placement, therefore, such foot trajectories during continuous motion, provide the essential mode of error correction in order to allow stability during gait. Study of foot trajectories may provide vital insight to help to understand gait stability. It is therefore not surprising that most research papers study the recovery strategies by assessing the gait spatiotemporal parameters, especially when MoS is adopted to quantify the stability. Gait parameters may help in understanding how subjects adapt their gait to prevent falling. Commonly, the adopted gait spatiotemporal parameters include step-length, width, time and frequency (Hamacher et al., 2011, Hak et al., 2013a, Houdijk et al., 2018b). Additionally, it was reported by several studies that increased variability of spatiotemporal parameters during walking is linked to increased risk of falls (Sivakumaran et al., 2018, Callisaya et al., 2011, Hausdorff et al., 2001).

The above measurements are relativity straightforward to compute and can be captured with relatively affordable devices for instance, floor mats and footswitches, all of which are able to provide foot kinematics and differences in foot trajectories.

In this section, reported recovery strategies adopted by able-bodied as well as prosthetics users to cope with changes in the balance status will be described.

1.9.4.1 Strategies adopted by able-bodied to maintain balance

A recent study by a group of Madehkhaksar et al. (2018), found that during perturbed trials, subjects exhibited shorter and wider steps as well as higher cadence.

Subjects showed increased step length and width when walking over compliant surface in a study of MacLellan and Patla (2006).

Vlutters et al. (2016) reported that subjects walked with a shorter single support duration when faced external pelvic perturbation. This suggests that timing of events (stance and swing phase) could be a form of recovery from external perturbations.

When comparing recovery strategies among young adults (25.1± 3.4 years) and older adults (70± 7.6 years) as in Roeles et al. (2018), no variations were found in subjects' mechanisms to increase their stability in the ML direction. Their steps were wider, shorter and faster than the steady state gait ones; the AP MoS was also increased

Similar strategies were also adopted to cope with platform continuous movement by able-bodied young adult (32±7.5 years) as reported by Hak et al. (2012). Further, results from

Martelli et al. (2017a), demonstrated similar findings, that there were no differences in compensatory mechanisms between young (24±2.7 years) and elderly (65±4.8 years).

1.9.4.2 Strategies adopted by prosthetics users to maintain balance

Results by Curtze et al. (2010) showed that prosthetics users exhibited significant longer steps on the prosthetic side when recovering. In addition, this study reported that the prosthetics users were equally efficient as the able-bodied participants in recovering. They reported that there were no statistically significant differences in prosthetic users as a group in terms of gait parameters and dynamic stability when walking on different surfaces over rough and smooth. However, there were some differences between controls and prosthetics users Curtze et al. (2011). These were mainly in step length and width; the prosthetic users walked in shorter and wider steps compared to the controls. Same strategy was adopted by the prosthetics users to cope with the destabilising environment is a wider and shorter steps as reported in a study of Beurskens et al. (2014).

Steps length of prosthetic users were also found short in results that were reported by Hak et al. (2014), although the study group walked on self-paced mode where no perturbations were imposed.

In a study of Parker et al. (2013), results showed that the prosthetic users showed a significant reduction in walking velocity as well as a significant reduction in step length on both sides.

Gates et al. (2013) reported that both prosthetic users and controls subjects’ stability were not affected by changing the walking surface.

1.10 Discussion

Studies that focused on the dynamic stability of prosthetic users are limited and counterintuitive. This indicates that more attention should be given to this group in order to enhance gait stability and adaptability during the rehabilitation. This can be achieved by understanding the strategies adopted by prosthetic users to optimize these aspects of walking, as well as how these strategies differ from strategies selected by able-bodied people. Creation of unpredicted unbalanced situations (also known as perturbations) that are faced during day-to-day activities may help in providing needed knowledge.

Several aspects were noticed regarding the prosthetic users stability studies; firstly, the dynamic margins of stability direction; most often, the presented direction was only in ML direction as in (Beltran et al., 2014, Sturk et al., 2017, Gates et al., 2013) without description of the stability status in AP.

In terms of which type of perturbations that would help to reveal insights about dynamic stability. The use of the popular McAndrew et al. (2010) method of perturbation appears to be revolutionary. However, application of continuous perturbations may not reflect a real-life situation (ecologically valid). Additionally, as the aim was to study the recovery response after a sudden event, imposing continuous perturbation may not help this present study to do so as subjects will be continuously perturbed at each initial contact. This method along with (Sturk et al., 2017) may be used as an anti-falling rehabilitation training in case the rehabilitation facility does have the CAREN system. Where participants can benefit from the continuous perturbations to practice and perform muscle strengthening exercises.

It was noted that mostly in the perturbation studies two main limitations have been reported, those were the possibility of anticipating the upcoming applied perturbations, and the randomizing of the trial.

This present study aimed to impose slip perturbations that can be experienced in real-life situation hence this study protocol focused on developing discrete slip perturbations.

To the research team's knowledge, the effect of acceleration and deceleration slips on dynamic stability in both AP and ML directions of the prosthetic users has not been investigated before.

In addition, to the research team's knowledge, none of the prosthetic users' studies imposed slips perturbations with multiple amplitudes. Applying different intensities of perturbation would help in indicating the minimum magnitude that should be exceeded in order to challenge dynamic stability. In other words, the threshold where any changes below it may be coped with no or minimal effort.

The perturbation protocol developed by Roeles et al. (2018) appeared to be overcoming these issues by randomising the applied perturbations as well as in the protocol the onset of the perturbations was not fixed and randomly generated every (10-15 step) which make the anticipation of the upcoming slips challenging for participants. Therefore, the present study perturbation protocol was based on their method.

Lastly, an inconsistency in applying the MoS concept was found. A clear example of this inconsistency was found when comparing the following two studies. The study by Beltran et al. (2014) assumed that greater ML MoS mean values of the prosthetic subjects indicated less ML stability. Whilst the opposite was reported by Hak et al. (2013c) who assumed that greater ML MoS mean values of the prosthetic subjects indicated greater ML stability. Despite this both studies have used the same methods and equipment.

1.11 Aims and objectives

The overall aim of this thesis was to study the dynamic stability of prosthetic users. This aim required the development of a method to provide destabilising yet safe situations to prosthetic users.

Based on the literature review, the status of dynamic stability for the prosthetic users is not clear and further investigation is required.

The hypothesis of this study was that the prosthetic users are less stable (lower MoS values) than the able-bodied. In addition, the prosthetic users would show less stability on the prosthetic side when comparing to the intact one.

The outline of chapters is as follows:

Chapter two: *General methods*: the study protocol and methods are described. An overview of equipment used to provide, and record perturbations is provided.

Chapter three: *Controls participants' margins of stability*: The initial study of the control group (able-bodied) is described. The aim of this chapter is to investigate the dynamic stability of the control group to help understand how humans maintain balance. Also, to investigate whether increased MoS indicates better stability or indicates the status instability. A further aim of the chapter is to investigate if there was a trend in the recovery mechanism when able-body persons faced unexpected destabilising events, and whether the applied perturbation affected gait spatio-temporal parameters along with gait stability measures. This chapter is essential to provide a criterion in dynamic stability that may help in comparing and understanding the stability of less able-bodied populations, (in this case, the prosthetic users).

Chapter four: *Prosthetic users' margins of stability*: The main study of prosthetic users' dynamic stability. This chapter aims to investigate the status of the dynamic stability of this population (within group study). Then, to compare the status of the prosthetic users to the control group in ***chapter three***. In addition, the chapter discusses potential benefits of incorporating a hydraulic prosthetic ankle mechanism over a conventional prosthetic foot and ankle.

Chapter five: *The prosthetic alignment effects on margins of stability*: The first case study is presented. In this chapter the effect of the prosthetic alignment is discussed. The main purpose is to investigate how alteration of optimal alignment affects the dynamic stability of a prosthetic user. This may help clinicians to understand how the prosthetic alignment might affect the gait stability.

Chapter six: *Lower limb congenital absence & anomaly:* The second case study. The aim of this study is to examine the dynamic stability on a person with congenital below knee anomaly. Then, to compare the status of the prosthetic user to the groups in chapter three & four.

Chapter seven: *General study conclusions clinical implications, limitations, and future work*: Provides summary conclusions, and clinical implications of the results. Limitations and direction for future work are described and suggested.

2 General Methods

2.1 Introduction

The aim of this study protocol was to create similar unbalanced situations that a person would likely experience in the real world. This may help to assess the dynamic stability and gait patterns in a safe fall-free environment.

An advanced system called CAREN was used in this study to impose sudden slips in a safe environment. These slips were caused by accelerations and decelerations from the fixed walking speed.

The MoS was adopted to evaluate the dynamic stability. To study the mechanisms of balance recovery, the gait spatiotemporal parameters were measured.

In this chapter, a description of the study's methods including equipment and techniques is provided. Additionally, the steps of how the data was prepared and analysed are described.

2.2 Identifying the relevant work

Table 2.1 shows the searched databases and individual journals along with the key words that were used in order to identify relevant published work for inclusion in this study literature. All works were read, assessed and integrated into the final review if they were considered relevant for inclusion.

Searched databases and individual journals	Key words
Medline, ScienceDirect, Prosthetics and Orthotics International, Journal of Prosthetics and Orthotics, SAGE journals online, SpringerLink, and Google Scholar.	Falls, centre of mass, amputation, prosthesis, lower prosthetic users, dynamic stability, margins of stability, rehabilitation, perturbations

Table 2.1 Literature review key words and relevant databases and individual journals to identify the relevant published work.

2.3 Equipment

2.3.1.1 **The Computer Assisted Rehabilitation ENvironment (CAREN)**

The use of instrumented treadmills is increasing in gait research since they may offer some advantages over the typical gait labs. Here in this study, a system called (CAREN) was adopted. The Computer Assisted Rehabilitation ENvironment Extended system (Motek, Amsterdam, Netherlands) is based at the National Centre for Prosthetics & Orthotics in the University of Strathclyde (Figure 2.1).

In 1998, the company developed the first prototype of CAREN and the first CAREN system was installed in 2000 at the University of Groningen in the Netherlands (MOTEK, 2016, MOTEK, CAREN Capture System).

The CAREN is an adaptable, multi-sensory system used for the purposes of clinical analysis, rehabilitation, and for the evaluation of the human balance system. The CAREN system in this study uses a dual-belt instrumented treadmill with integrated forceplates recording at 1000 Hz as a walkway. Furthermore, it includes a virtual reality environment as well as a surround sound system. .

For the detection of reflective markers, CAREN uses twelve infrared Vicon Bonita cameras within its Motion Capture System (Vicon Nexus : Oxford Metrics, Oxford, UK) (Figure 2.1), meaning that it can record at various sampling frequencies. In this study, data was recorded at 100 Hz. A safety harness was integrated within the system, in order to protect the participants in case of a fall.

2.3.1.2 The benefits of using the CAREN as study equipment

The CAREN offers numerous advantages for gait assessment. Firstly, using the treadmill as a walking base allows trial sessions, whether training or research, to be done in small area. This means that an unlimited number of steps can be performed. Experimental sessions can be very short, which is advantageous for the evaluation of children or less able-bodied adults. Importantly, the walking speed can be controlled which will increase the reliability of any measures (Lee and Hidler, 2008, Tesio and Rota, 2019).

In terms of the effect of using a treadmill on the walking pattern (kinematics and kinetics), there is a debate whether there are significant differences between walking on a treadmill vs walking over-ground or they both relatively the same.

Some studies have reported results in favour of the treadmill use. Results in a study of Van Caekenberghe et al. (2013), showed that walking over a high-quality treadmill as the case for the CAREN is dynamically equivalent to walking over-ground. Results from (Lee and Hidler, 2008) showed that in general, very few differences were observed in temporal gait parameters or kinematics between treadmill and over-ground walking. The authors of the same study concluded that the use of treadmill for training especially for those who are with neurological injuries appears to be justified. The split-belt treadmill walking has been reported to be a useful rehabilitation tool for people who have transtibial amputation in a recent research by Kline et al. (2020). This idea was also supported by Reisman et al. (2013) who assessed the gait pattern of poststroke individuals.

On the other hand, other studies have raised some objections against the use of the treadmill. Rosenblatt and Grabiner (2010) found that the spontaneous average velocity reported on the treadmill is lower by approximately 30% than the one reported on firm ground. Similar results were reported in a study of Marsh et al. (2006). Supported by experimental findings, a study by Mignardot et al. (2017) argues against the use of the treadmill. The authors see that treadmill walking may limit the learning of adaptive capacities to real-life environments.

Zeni Jr and Higginson (2010) showed that stride-to-stride variability may be seen when using a split-belt treadmill and a familiarizing session of 5 minutes prior to data collection may overcome this issue.

For the purpose of this study, the use of a treadmill can be justified based on the advantages that such an advanced tool can provide. The aim was to have a perturbation protocol that one might face during daily activity.

The CAREN can help in evaluating balance. These challenges should be applied in a safe environment; the CAREN can provide such an environment. Several methods can be applied in the CAREN to create unbalancing situations. These can be auditory, visual and mechanical perturbations. The use of mechanical perturbations includes belt speed changing and multiple degree-of-freedom of platform translations.

Even though perturbations can be also given when walking over ground (for example obstacles placed on the walking base), these perturbations are still predictable to the participants. Using treadmill-based perturbations gives the advantage of providing unpredictable perturbations, thus enabling the creation of situations that are comparable to real world experiences. (Sessoms et al., 2014, Lee et al., 2018).

Figure 2.1 The Computer Assisted Rehabilitation ENvironment (CAREN), University of Strathclyde, the CAREN consists of dual-belt instrumented treadmill, VR, and sound system. 12-Vicon Bonita cameras as a motion capture system.

2.4 Data integration

In this study both D-Flow and Vicon Nexus (Vicon, Oxford, UK) software programmes were used to obtain the subjects' data.

2.4.1 D-Flow

Data Flow (D-Flow) is a software system developed by Motek, (Amsterdam, The Netherlands), which allows the operator to control several devices such as a virtual reality, a treadmill, a motion platform, and any infrared motion capture cameras.

In this study, the treadmill belts speeds including the magnitude, any changes (acceleration and deceleration) and the perturbation onsets were all controlled by D-flow (version 3.28.0) using the Lua programming language. In addition, the virtual reality was set to allow optical flow synchronizing with the treadmill speed.

Besides the controlling features, D-flow provides access to real time data from the input devices. D-flow consists of several modules with many functions; an example is presented in Figure 2.2. The main one in this study is called Motion Capture module (MoCap).

2.4.1.1 **MoCap**

This module performs several functions. These include receiving motion capture data from any hardware. The module can also calculate the human body model (HBM2) which is integrated within the system. Finally, the module can detect gait events. Gait event timing is needed to control the application of perturbations during the trials and record data.

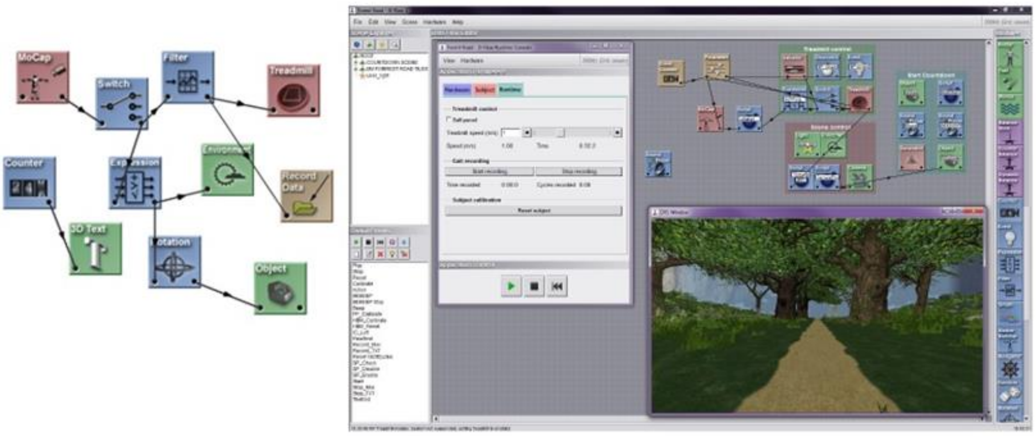

Figure 2.2 D-Flow and its modules, the software used to control the treadmill speed, by applying a written script that changes the speed of the treadmill suddenly (increasing and decreasing) to create the unbalanced situations (slips and trips). Besides that, the d-flow was used to calculate the model outputs (kinetics and kinematics) using MoCap module

2.4.2 Vicon Nexus

The Vicon Nexus is a data capture software for clinical gait and biomechanics developed by Vicon, (Oxford, UK.) The software offers several features. These features include data capture such as kinematics data that involves 3D motion capture obtained from the cameras, and kinetics data that involves forces, moments and powers. Figure 2.3 shows the Vicon Nexus interface.

In addition, the software can be used to post-process the captured data. This feature includes the ability to filter the raw data using several filtering options.

Prior to capturing the required movement, several steps should be conducted to prepare the system. Firstly, system calibration; the Vicon active wand was used to both calibrate the infrared cameras and to set an origin (global reference) which was placed in the middle of treadmill in this study. The final step of calibration is to zero level sensors.

The next step is usually to conduct a markers labelling: this step is required for the system to define each of the attached markers. In this study, the marker-labeling step was executed in Nexus.

Markers labelling usually takes place following the static trial capturing. The recorded static trial was opened, and the subject model was reconstructed in the 3D perspective view.

In some cases, the infrared cameras could not see the markers during certain times of the trial session. This will result in frames with breaks in trajectories these are called gaps in these markers' data. This problem can be resolved by using the post-processing in Nexus which allows gap filling for the missing markers as well as labelling of the unlabeled markers. Gaps were filled using two gap filling options: 1) the rigid body filling was used for pelvis markers as this option are recommended when a rigid or semi-rigid relationship exists between markers.

2) pattern filling was used for single markers such as ankle markers this option is suitable when marker with a trajectory similar to the one whose gap is needed to be filled. This is typically the case when the trajectories originated from markers attached to the same segment.

Nexus can export the data in several formats such as c3d, text, and mox extensions which add flexibility to use different software to conduct data analyzing. In this study the c3d files was used.

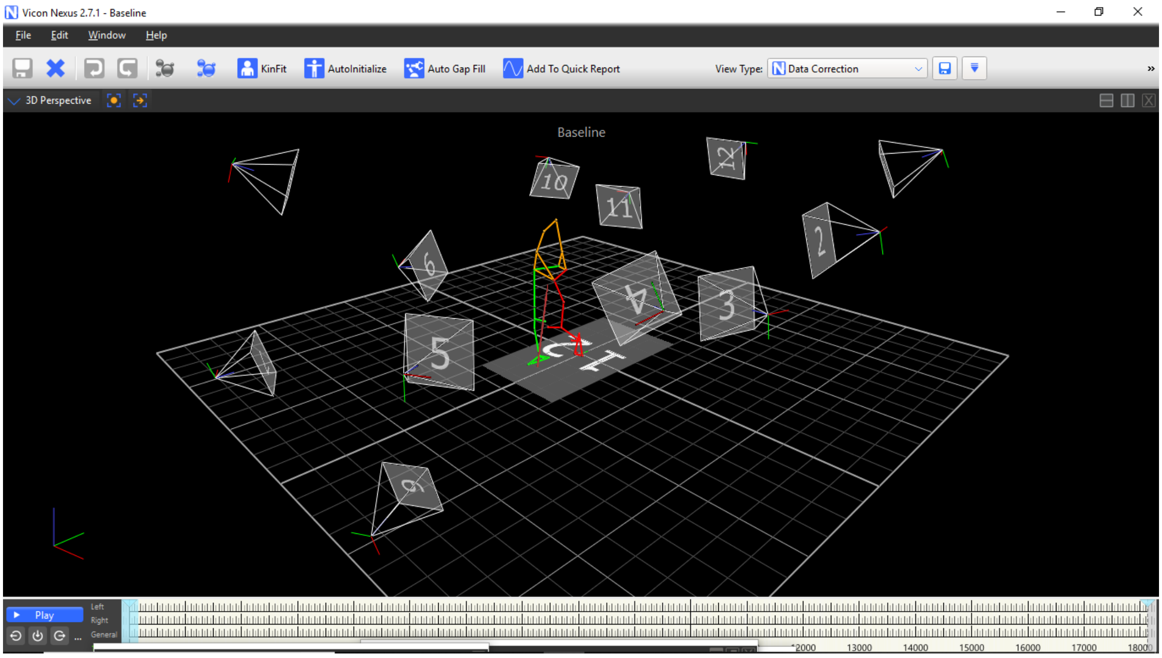

Figure 2.3 Vicon Nexus interface showing a skeleton model,12 Vicon Bonita cameras setup and force sensors (1, 2). The global reference frame was located at the centre of the treadmill.

2.5 The study's protocol of perturbation

The aim was to challenge a participant's stability in forward / backward direction. This can be challenged by imposing perturbation in the AP to compromise the relation between the BoS and CoM. As discussed in chapter 1 the stability in forward and backward direction is lost when the XCoM passes the anterior and posterior boundaries of the BoS, respectively.

In addition, in present study, the aim was to challenge the stability of each leg during the stance phase by providing an unexpected slip –like experience of different intensities for both lower limb prosthetic users and able-bodied groups. This may allow investigation of the differences in gait stability as well as side-to-side asymmetry.

In order to achieve these aims, a previously developed Lua code was adopted. The code was firstly introduced by Motek (MM Gait 2.1 - Perturbations), then a group of researchers in the university Strathclyde developed this code (Roeles et al., 2018, Roeles et al., 2017). The code contained two types of sudden mechanical perturbations; acceleration and deceleration from the pre-selected speed triggered during the stance phase.

Stance leg perturbations, which are referred to as slips, are normally applied by sudden changes in belt speed around initial contact or toe-off. Applying belt accelerations during the stance phase will act as a backward slip, subsequently making the CoM move toward the anterior border of the BoS. This perturbation will challenge the participants' forward balance. Whilst imposing belt deceleration will act as forward slips challenging the backward balance by making the CoM moves toward to the posterior border of the BoS. This does not necessarily signify that the balance in the ML direction will not be affected by applying these perturbations. Therefore, the study investigated the stability in both AP and ML directions and evaluated the recovery stages that should be performed to maintain a dynamic balanced system after perturbations.

When running the Lua code, a total of 12 mixed randomised acceleration and deceleration perturbations per trial, 6 on each foot, 3 of each type were applied. In the protocol, the perturbations were provided separately on each leg i.e. two perturbations would not be applied at the same time for both belts.

During each block application, the code generated both perturbation types in different times and orders (Figure 2.4).

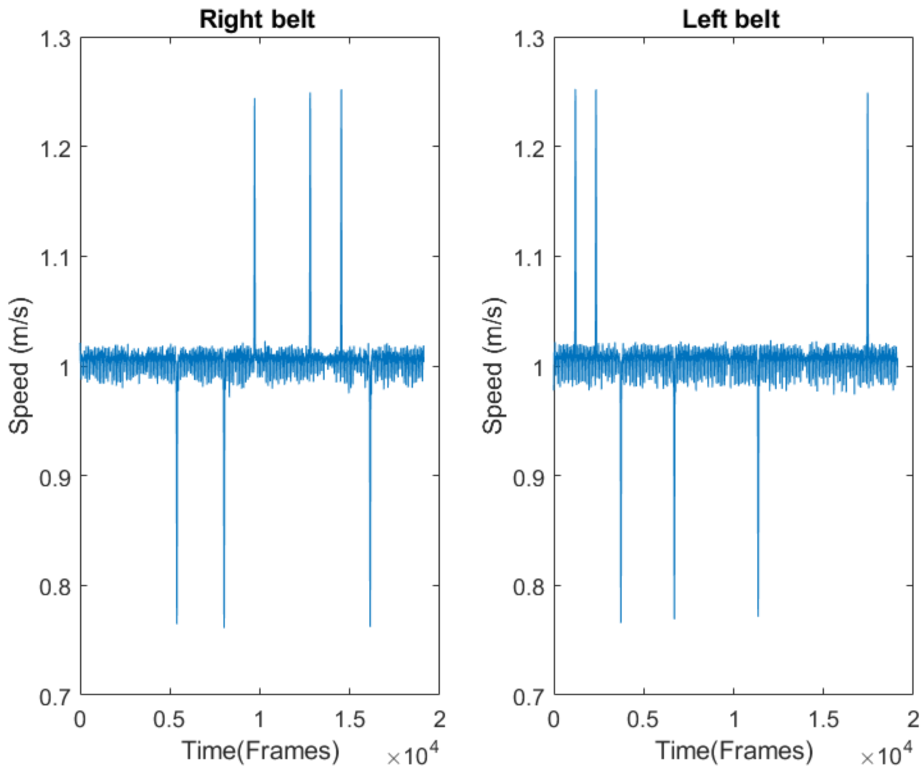

Figure 2.4 Example of Right and Left belts speed changes during one perturbed trial (Low intensity block applied) while walking with fixed speed 1m/s. The protocol contains mixed perturbations; three of each accelerations and decelerations from the fixed walking speed. The perturbations were applied separately on each leg as well as the order and time of them changed randomly for each application. Each frame of time equals 0.01 s

The perturbation protocol included three blocks of different intensities; Low, Medium and High. During the Low intensity block, the belt speed changed by ±25% of the walking speed, during the Medium one by ±45% and during the High ±65%.

The perturbation intensities did not aim to create a certain fall for several reason; One can imagine that this is not useful in clinical practice considering the costs, size and complexity of the required systems. As full-body safety harnesses with an instrumented system that minimizes the effect of a fall is needed to provide a maximum safe environment. Besides, falling even when it does not include any physical impact, can have a psychological impact on the participant leading to an increase the fear of falling which will result in limiting the activities especially for a group such as prosthetic users (Pin and Spini, 2016, Miller and Deathe, 2004). Therefore, the aim was to provide the perturbation that would challenge the gait stability of those and yet would not cause a fall. This enables studying the adopted strategies to prevent falling and the status of their stability.

In addition to the mixed set of perturbations during the same trial, the perturbations' intensities were applied randomly which all lead to prevent the anticipation of the speed changing as well as further reduce of the proactive gait adaptations. Figure 2.5 below provides an example of running the three intensities blocks.

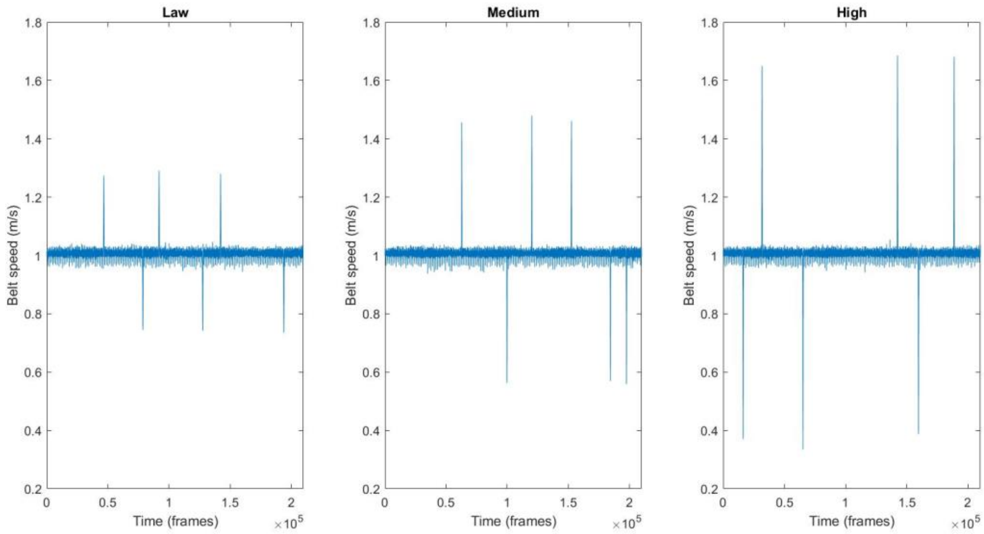

Figure 2.5 Schematic representation of 3 perturbed gait trials for left leg of a subject walking at fixed speed of 1m/s. Perturbation blocks resulted in six different belt speed changings, three accelerations (+25, +45 and +65%) and three decelerations (-25, -45 and -65%). The perturbations' blocks and types were all presented randomly. During each application, the code generated the perturbations differently in time and ordered. The Low intensities (±25%), Medium (± 45) and High (±65). Each frame represents 0.001 s of time.

2.5.1 Perturbation Profiles

Acceleration and deceleration perturbations followed the same perturbation profile for the three intensities, as showed in Figure 2.6. The profile included three phases; increasing, constant and decreasing. During first phase the value of perturbation increased until the preset amplitude was reached. Then, the perturbation value stayed constant for a given time (~0.1 s). Finally, during decreasing, the value returned to the initial value, the total time of perturbation is approximately 0.55s. To make sure that participants had recovered from the perturbation before applying the next one, the perturbations were applied every 10 to 15 steps.

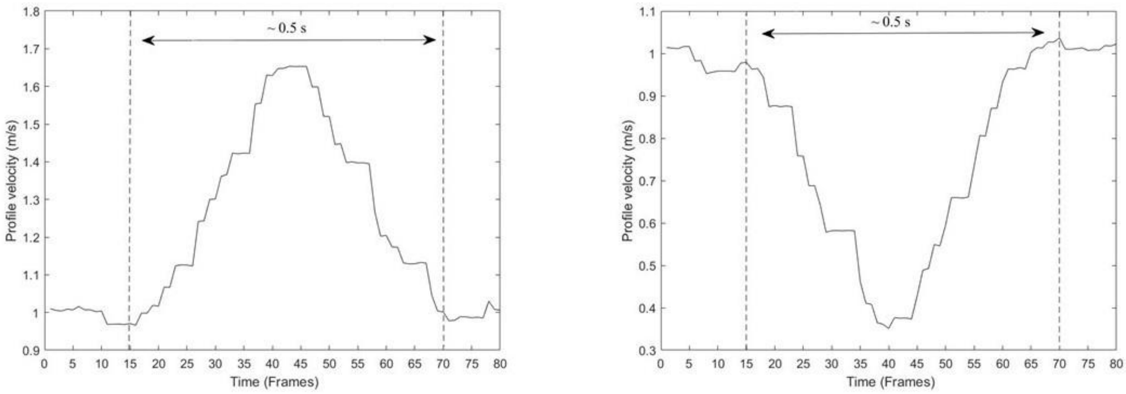

Figure 2.6 Perturbation Profiles , the same profile were applied for both the acceleration and deceleration perturbations. The time from the perturbation start until it returned to the fixed speed, was approximately 55 frames that equal to 0.55 s. The profile consists of 1) Increasing phase 2) constant phase (~0.1 s) 3) decreasing phase.

2.5.2 The perturbation onset

To provoke slip like action, the perturbations were triggered after the initial contact; during the stance phase of the tested leading leg, the peak of the perturbation was during the mid-stance. Figure 2.7 shows the perturbation onset in relation to the gait cycle. The MoCap model can detect the gait events through several approaches, here in this protocol the gait initial contacts were found using the (Zeni et al., 2008) velocity algorithm.

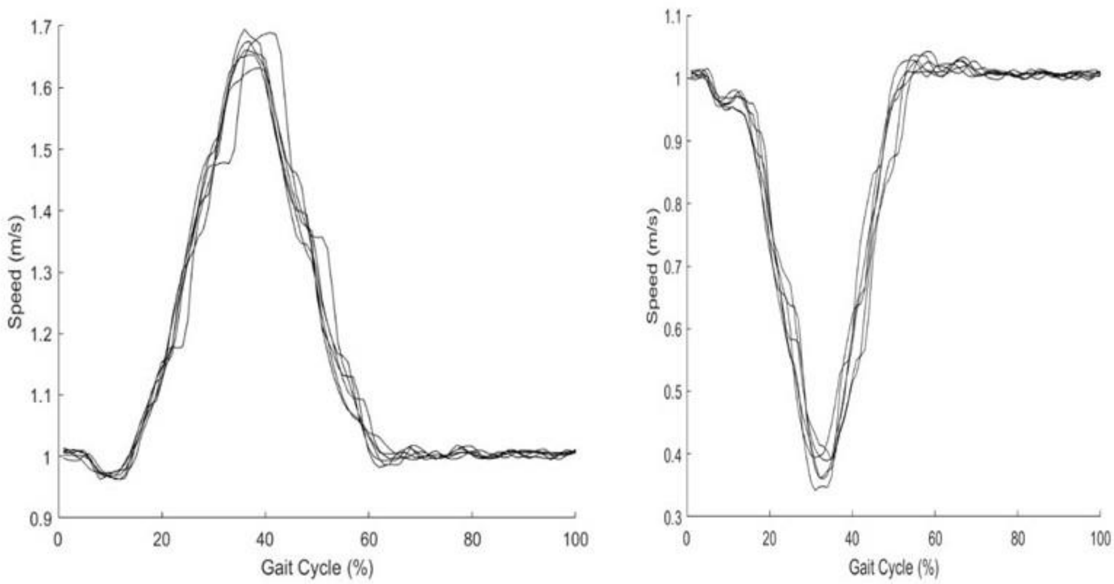

Figure 2.7 Perturbation in regard to Gait Cycle. The perturbations were trigged during the stance phase, after initial contact around 15% of the Gait Cycle to act like a sudden slip, the peak of the perturbation was during the mid-stance.

2.6 The Self-Paced Mode

CAREN as a feedback-controlled treadmill allows subjects to walk at their preferred speed, i.e. walking in a self-paced (SP) mode. The self-paced mode offers new clinical possibilities for instance measurement of long-term gait variability or fatiguing. There are several ways of applying this mode (Sloot et al., 2014). In this study, this was implemented by a real-time algorithm that considered the pelvis position in the AP direction, as found by the markers attached to the pelvis (LASIS, RASIS, LPSIS and RPSIS) in relation to an origin position on the midline treadmill (Figure 2.8). To ensure the safety of the participants, the treadmill has sensors attached at the end of it. In case a subject passed these, the treadmill would automatically stop.

The reason for including this mode in the protocol is to study the gait stability as well as gait spatiotemporal parameters for prosthetic users while walking in a more natural way (close to the over-ground walking) allowing to record a large number of steps within the trial, and compare if the results remarkably differ from the pre-selected fix speed.

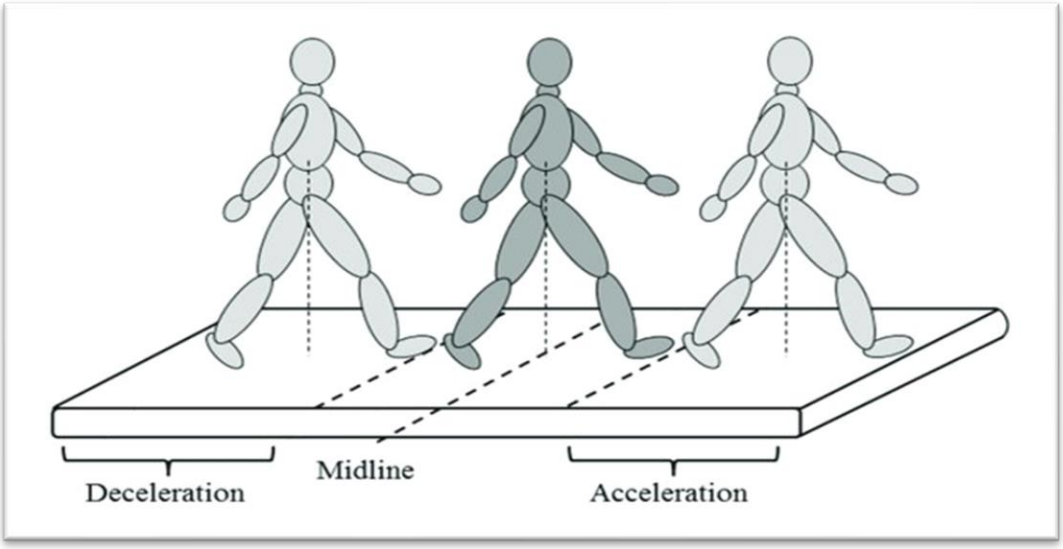

Figure 2.8 Self-Pace mode (SP) in CAREN, the CAREN takes into account the AP positions of the pelvis markers, to change the speed of the treadmill accordingly, forward position from the middle acceleration while backward leads to deceleration.

2.7 Walking speed

Initially, the aim was to provide a standard perturbation in terms of intensities and onset time regarding the gait events, so the walking speed was set to a fixed value of 1 m/s. However, the research team was aware that this speed may or may not be suitable for the prosthetic users. In case the subject did not manage this speed, the protocol included another approach to find the walking speed that suited this group.

Firstly, the treadmill speed was gradually increased until the subject could walk with minimal effort. Then, the speed was increased until the subject reported feeling uncomfortable. From there, the speed was gradually decreased until the subject reported the comfortable speed again. The treadmill speed was then set to the average of the two reported comfortable speed points (Hak et al., 2013a). The same perturbation protocol was applied for both walking speeds CWS and 1m/s.

2.8 Finding Gait Events

To detect gait events, data from heel markers and foot markers were used in an AP Velocity-Based Treadmill Algorithm by (Zeni et al., 2008). Briefly, the velocity of the markers was found by differentiating the AP marker positions and by taking into account the sampling frequency as in the following:

Then, the frame where the velocity of the heel markers turned from positive to negative was defined as heel contact, whilst frame where the velocity of foot markers turned from negative to positive was defined as foot toe-off. Finally, to assure correct identification, the gait events were verified by visual inspection.

2.9 The centre of mass position estimation

There are many ways in order to estimate the position of CoM, using both kinematics or/and kinetics (Lafond et al., 2004). In the study of Lafond et al. (2004) three methods to estimate the CoM were compared to each other. These were the kinematic method; the zero-point-to-zero-point double integration which also known as gravity line projection (GLP) and the COP low-pass filter method (LPF). The results of the study showed that GLP method provides similar CoM trajectories compared to the kinematics method and that LPF method underestimates the CoM displacement. A study of (Tesio and Rota, 2019) on the other hand provided some limitations of the GLP method these included the failure to provide CoM displacements within the body system or with respect to the ground frame. As well as that the researcher needs to be fully familiar with the concept of this method which add complexity element to this method.

Generally, the kinematics method is considered as the 'golden standard' (Hof, 2005) maybe because of its overall simplicity and reliable results (Cotton et al., 2011). The kinematics method included body tracking of the participants. When going through studies that focused on stability in response to gait perturbations, two main technique of finding the CoM have been adopted. These are full body model that includes tracking of the participants. In addition, a modified model which assumes that the CoM can be represented by the centre point of the pelvis which also known as reduced kinematic model (Beurskens et al., 2014, Madehkhaksar et al., 2018).

In this study, the trajectory of the CoM was estimated by using pelvis markers to define a pelvis segment and finding the middle point between the pelvis four markers (right and left ASIS, right and left PSIS) relative to the global reference frame (Whittle, 1997). The reasons of using adopting this method were firstly, the simplicity of applying the concept as simple math was needed. When compared to full body tracking, the method requires less time and expertise as only 4 reflective markers are needed whilst the full body may need up to 30 reflective markers. A study by (Forsyth et al., 2018) who evaluated the efficacy of using the pelvic method to estimate CoM position in response to gait perturbations, showed that pelvic method can be reliable and appropriate for clinical use without losing accuracy and precision. Havens et al. (2018) also supported this model and suggested that this simple CoM approximation is comparable to more complex multi-segment models. Finally, this method was adopted to find the CoM for MoS in several studies (Peebles et al., 2017, Peebles et al., 2016, Hak et al., 2012, Hak et al., 2013b, Hak et al., 2014, McAndrew et al., 2011, Major et al., 2020).

2.10 Finding the velocity of CoM

CoM velocity was found by first computing the first derivative of the CoM position divided by 0.01 as in (McAndrew Young et al., 2012). The instantaneous velocity of the treadmill was then added to the calculation (Hak et al., 2014).

2.11 Evaluating the Gait Stability

Inconsistency in interpretation the value of the MoS were found among the reported studies. Whether greater MoS value indicates greater stability or it indicates the opposite way. The initial rational behind the development if the MoS, was that increased MoS may be interpreted as being more stable (Hof et al., 2005). Based on this original paper, the present study's hypothesis was aligned with this interpretation.

The following equation was used to find the MoS in both AP and ML direction: $MoS = BoS - XCoM$

(XCoM), represented the position which can be in (x,y,z) of the centre of mass plus its velocity in that direction times the pendulum natural frequency used in model calculated as

$$\sqrt{\frac{\boldsymbol{g}}{\boldsymbol{I}}}$$

Where l is the equivalent pendulum length (m) based on the height of CoM. it was calculated as the distance from the ankle markers (right and left LM) of the foot to the CoM (vertical projections positions) at the time the of heel strike (Peebles et al., 2017, McAndrew et al., 2011) (~leg length). The letter g represented the acceleration of gravity (9.81 m/s^2).

The equation of the MoS was adjusted to account for the direction of the lab coordinate system. The greater MoS values indicated more stable status while negative values represented unstable condition meaning that the XCoM exceeded BoS and a corrective step should be taken to avoid falling, (Bruijn et al., 2013, Hof et al., 2005).

For the BoS boundaries, the ML 5th metatarsal marker position of the foot was used to determine the ML border of the BoS (as it is the most lateral edge of the BoS) (Gates et al., 2013, Beltran et al., 2014). Whilst the AP position of heel marker was used to quantified the AP BoS border (as it is the most posterior edge of the BoS) (Roeles et al., 2018) , Figure 2.9 shows the BoS and the MoS in regard to the leading foot.

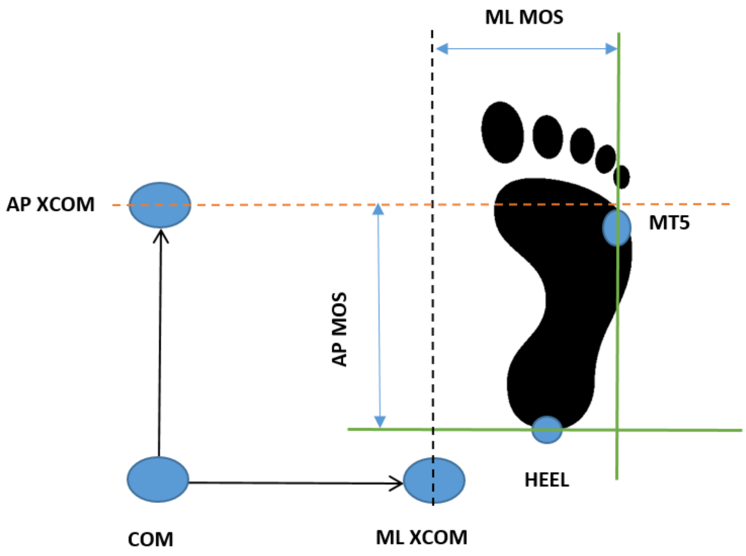

Figure 2.9 Schematic representation of margins of stability (MoS) for the right side in the Medio-lateral (ML) and Antero-posterior (AP) direction

2.12 Gait Spatiotemporal parameters

The following spatiotemporal gait parameters were calculated to study the recovery mechanisms:

- Step length was defined as the AP distance between ipsilateral and contralateral heel strike events (Supakkul, 2017).
- Step width was defined as the ML distance between ankle markers.
- Step time was defined as the elapsed time between ipsilateral and contralateral heel strike events.

For all data, the gait cycle was defined as the frames between two consecutive initial heel contacts and the stance phase was defined as the frames between heel contact and toe-off of the lead foot. These frames were time-series normalised to 101 samples per stride using linear interpolation to represent gait cycle percentage.

2.13 Procedure

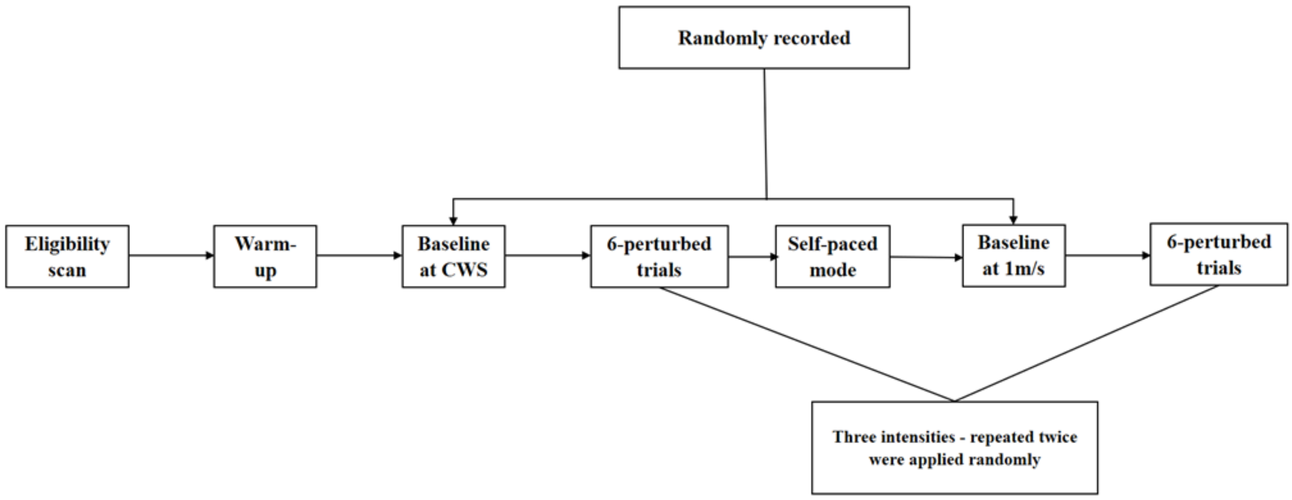

Figure 2.10 The study's procedure for both control and prosthetic users groups

First, an eligibility screening was conducted to make sure that the participant met the study's criteria. If the participant was found eligible for the study, a consent form was handed to obtain the participant's approval.

Full information about the study protocol was given for each participant prior to the recording sessions.

In order to make the reflective markers as close to the landmarks as possible, participants were asked to change into appropriate tightly fitted clothing. Then, 10 reflective markers were applied to the following anatomical landmarks:

- Right and left ASIS
- Right and left PSIS
- Right and left lateral malleolus
- Right and left heels
- Right and left 5th metatarsals

The able-bodied participants walked with their own comfortable shoes, whilst the prosthetic users walked with their own prosthesis with no changes were made to their prosthesis.

After that, the participants were asked to step in the platform, and immediately they were attached to the safety harness.

Two calibration trials were recorded; firstly, the static calibration trial was recorded in which the participants stood still and the markers were labelled and saved. The static trial is needed for the MoCap to estimate the position of joint centres, bone lengths, and joint axes. Then, the functional calibration was done by asking the subject to walk for a few steps. This is needed to initiate the model and to help the system to map the markers. This process is essential for the MoCap for two purposes. Firstly, the Lua code needs to identify the heel markers for gait event detection. Also, this process is vital for the self-paced mode to know the position of pelvic markers.

To become familiar with the machine, participants performed a 6-minute warm-up trial, Figure 2.10 summaries the trial sessions. During the warm-up the CWS were found (see section 2.7).

Following that, the treadmill belts speed was set to a fixed speed, participants then walked for 3 minutes with no perturbations (baseline trial) which was used as the reference to compare the stability and gait patterns to the perturbed trials.

After the baseline, subjects completed 6 randomly mixed perturbed trials where each intensity was repeated twice.

Participants were aware that they were going to experience speed changes (either increasing or decreasing) that would be applied separately on each side, but the perturbation onset was not known. They were asked to walk as they walk usually and to try to recover from the perturbations in the best possible way as well as to keep walking after the perturbation. In addition, they were given breaks between trials and whenever they asked for.

A SP trial was recorded; initially, participants were given some time to be familiar with the mode, the collected data was taken for the final 3 minutes of the trial.

For participants that managed to walk at 1 m/s, the perturbation protocol was repeated with the 3 different intensities and 3 of unperturbed walking at 1m/s.

Once the participant completed the walking trials, the participant was assisted from the motion platform immediately after being unclipped from the harness. The reflective markers were carefully removed.

The following data was collected during the trials:

1. Forceplate data
2. Marker 3D positions
3. Treadmill speed

The treadmill speed was recorded at 1000 Hz in Nexus using an analogue 4-output Phidget (Phidgets Inc, Calgary, Canada) Figure 2.11 . All of the data, including forceplate, markers and treadmill speeds were synchronized enabling comprehensive post-processing. .

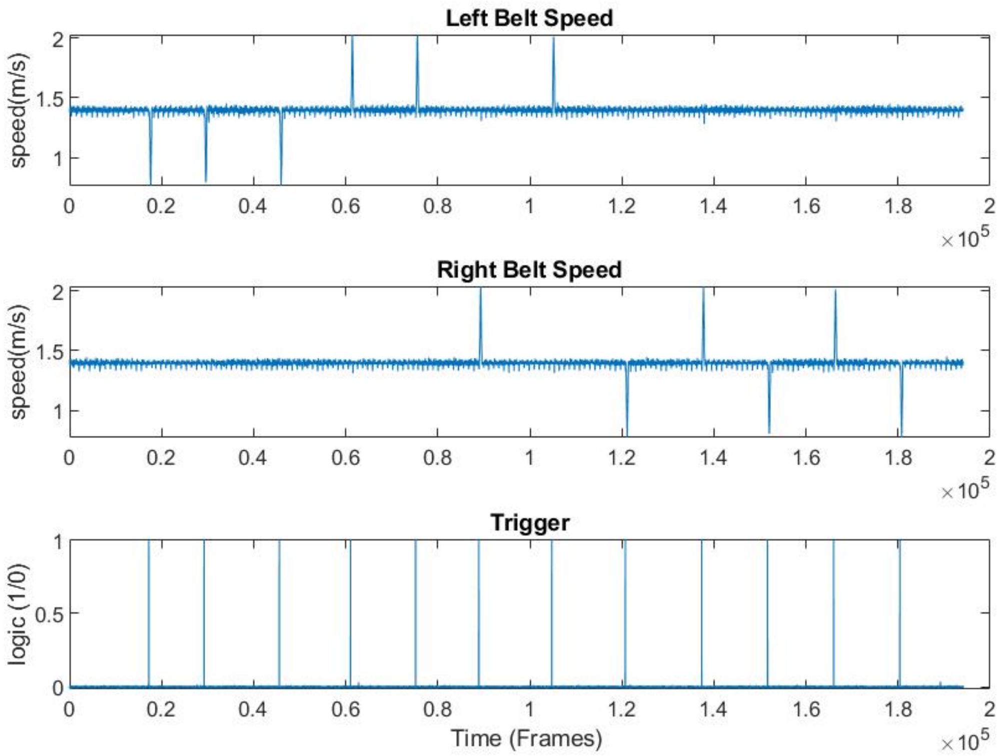

Figure 2.11 treadmill speed data with trigger.

2.14 Data analysing

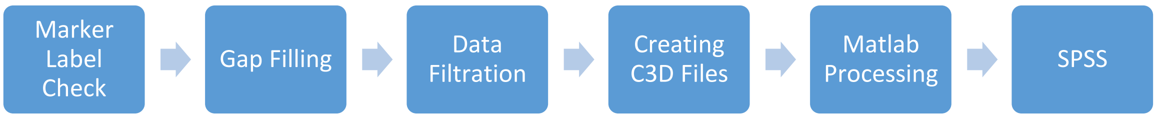

Figure 2.12 Summary of data post processing for calculating MoS and Gait Spatiotemporal.

2.14.1 Preparing the data

Figure 2.12 provides an overview of how the data was processed. In details, the makers were initially checked to see if any were mislabeled or if any were overlapping. Then the markers' gaps were filled using gap filling features in the Vicon Nexus. The rigid body filling was used for pelvis markers while the pattern fill was used for single markers.

After that, the trajectory data was filtered in Vicon Nexus 2.7 using second-order Butterworth filter with a cut-off frequency of 6 Hz. As according to (Winter et al., 1974) this cut-off frequency was found to be the highest frequency in kinematics related to gait. All of the forceplate and treadmill speed data was also filtered using a second-order Butterworth filter with a cut-off frequency of a 20 Hz. Lastly, the Forceplate and treadmill speed data was downsampled to match the 100 Hz markers' data, so the frames that represent the initial contacts and toe-offs are the same for all data.

2.14.2 C3D files

The data was exported from Vicon Nexus in format of c3d file. The toolbox provided by Barre and Armand (2014) (BTK tool) was used to read the c3d files into the Matlab (Mathworks, Inc., Natic, USA).

2.14.3 MoS and spatiotemporal for steady state trials; baseline 1m/s and SP mode walking

MoS and spatiotemporal parameters were calculated continuously for the whole trial (~300 steps per 3-min trial). Mean MoS and spatiotemporal at the initial contact of the leading foot was calculated and averaged over 150 steps for each side. The variability of MoS and spatiotemporal were calculated as the standard deviation.

All processing and analyses were performed with custom MATLAB R2019b codes (Mathworks, Inc., Natic, USA).

2.14.4 Perturbed trials

For the trials with perturbations, for each ipsilateral and contralateral sides: one step before perturbation, the perturbed step and three steps after perturbation were selected to study. The timing of these steps is showed in (Figure 2.13). A Matlab algorithm was developed to compare the parameters among these steps. Where the first, second and third step after were compared to the one before to see how these steps varied. Also, they were compared to the baseline values.

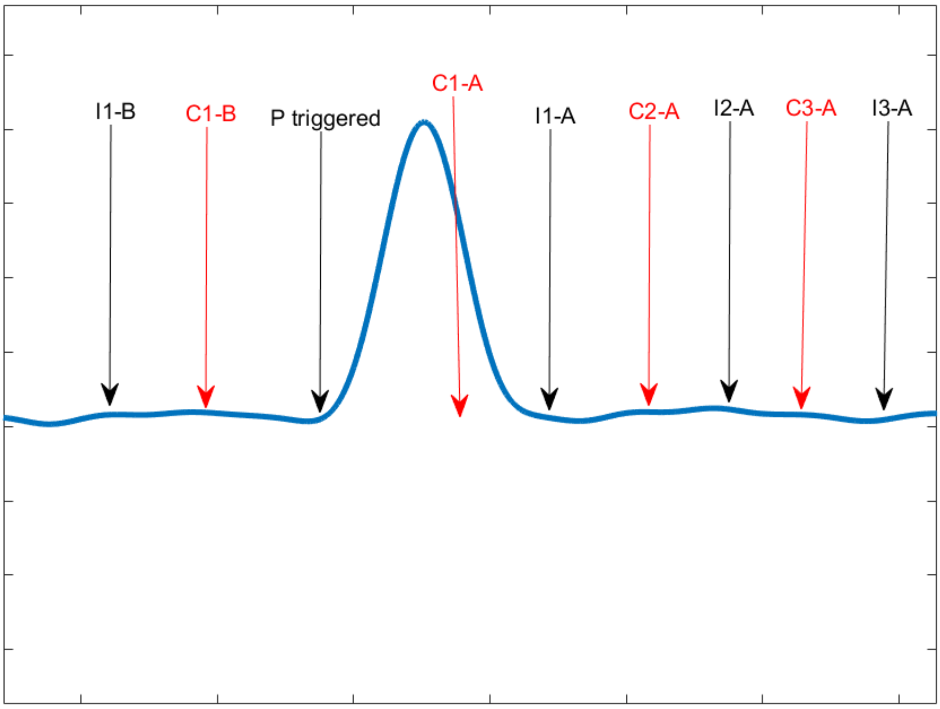

Figure 2.13 Demonstration plot of the selected steps before and after acceleration slips.The same steps were selected for both deceleration and accelerations slips. I1-B, I1-A, I2-A I3-A: represent the 1 before, 1 after, 2 after, 3 after steps for ipsilateral side. C1-B, C1-A, C2-A, C3-A: represent the 1 before, 1 after, 2 after, 3 after steps for the contralateral side. P: the step where the perturbation was triggered.

2.15 Statistical Analysis

All data was based on the average ± standard deviation. Data management and analysis were performed using SPSS v25 (SPSS Inc., Chicago, IL, USA).

The Shapiro-Wilk test was used to check data normality. The data was normally distributed therefore parametric statistical tests were used to investigate the differences. For all statistical tests, the 95% confidence interval for mean differences was calculated and the p-values represented were statistically significant when $p < 0.05$. When comparing the differences in means, confidence intervals were adjusted using Bonferroni correction.

In order to investigate the differences in MoS and spatiotemporal parameters between the right and left sides for control groups, the paired sample t-test was used. Also, the same test was performed to investigate the differences between intact and prosthetic side for the prosthetic users' group.

To assess the effect of walking at a different unperturbed walking speed, a repeated measure of analysis of variance ANOVAs was used.

To investigate whether the steps before and after the perturbations were significantly different in MoS and spatiotemporal for each side of each group, a repeated measure ANOVAs was performed. Similar test was used to see whether the steps before and after were significantly different from the baseline values.

The differences between the prosthetic users' subgroups (no prosthetic ankle and ankle) were investigated using independent t tests.

Differences between able-bodied and prosthetic users in the steady state walking were analysed using independent t tests.

To compare the difference in recovery between control and prosthetic user groups, a two-way repeated measure ANOVAs was performed. The factors were (Group × step number).

3 Control participants’ margins of stability

3.1 Introduction

This chapter discusses the dynamic stability of an able-bodied group included as participants in this study. The aim of this part of the study is to investigate how an able-bodied person would react when facing slips. This is essential to initially provide understanding of the concept of dynamic stability. Additionally, study of this group facilitates creation of reference data which will be used as a control group to provide comparison of how a less able-bodied person (such as a prosthetic user) could improve their gait stability which may lead to reduce the fall rates.

Twenty-two participants participated in this part of the study. Participant sample criteria and method of recruitment are described in this chapter.

Initially, the MoS and gait spatiotemporal measures results for this group during unperturbed walking condition are presented. This is necessary in order to provide an overview about gait stability and pattern when walking in a steadily state. Following this, results of gait stability and spatiotemporal parameters are also presented following perturbations to investigate how gait parameters have been affected.

To provide a clear picture on how the stability and the gait have been affected by imposing perturbations, step to step parameters were presented. The selected steps were ipsilateral and contralateral steps before and after perturbations. The time of these steps is showed in figure 2.13.

Finally, a conclusion regarding this group dynamic stability is provided at the end of this chapter.

3.2 Methods

3.2.1 Sample criteria

Inclusion criteria consisted of healthy participants aged 18 and above with no neurological or musculoskeletal conditions with potential to limit mobility. Participants had to have the ability to walk continuously for 5 minutes. Exclusion criteria were using walking aids, participants with vestibular dysfunction, suffering from motion sickness and pregnant subjects.

3.2.2 Method of Recruitment

A departmental email was distributed to all staff and students to advertise the research. The email contained information about the project and BME staff contact details. The participant information sheet and consent form were sent to whom contacted the research group. Participants were given a two-week period after receiving the participant information sheet to make their decision. Participants provided written informed consent prior to trials session. The project protocol was approved by the department of biomedical engineering ethics committee (DEC), University of Strathclyde.

3.2.3 Equipment

Participants walked on the CAREN (please see section 2.1 for more details). Prior to each data capturing session, the system was calibrated. The Vicon active wand was used to both calibrate the infrared cameras and to set an origin (global reference) which was placed in the middle of treadmill. In addition, the forceplates were zero level.

3.2.4 Participants

Twenty-two participants volunteered for this part of the study. Table 3.1 shows the participants' characteristics.

Characteristics (Mean ± SD)	**Able-bodied (n=22)**
Age (yr)	28.8 ± 6.8
Sex (male/female)	13 / 9
Height (cm)	173.45 ± 9.7
Body weight (kg)	69.1 ± 15.1
Found Comfortable Walking Speed (m/s)	1.2 ± 0.11
Dominant leg side (L/R)	R for all

Table 3.1 Control group characteristics.

3.3 Results

All participants completed trials without falling. In total, for each intensity of CWS perturbations, 132 acceleration slips were applied on each leg sides, as well as 132 deceleration slips were applied on each side leg sides. The same total was also applied for the 1 m/s speed.

The average number of steps per trial was 160 steps for each side, average total time of any trial was approximately 180 seconds.

3.3.1 Descriptive overview of steady state trials: baseline walking CWS, SP and 1m/s

During baseline where controls walked at CWS, control group walked at a speed of 1.2 m/s ± 0.11 m/s (mean ± SD). This speed was set according to the reported comfortable walking speed by each participant (see section 2.7.1). Whilst subjects walked at a speed of 1.3 ± 0.13 during SP mode.

Table 3.2 shows the mean and standard deviation of the MoS for both AP and ML direction and gait spatiotemporal parameters of left and right sides during unperturbed trials namely baseline, SP and 1m/s.

Subjects tended to exhibit slightly larger AP MoS values at left heel contacts than at right heel contacts for all unperturbed walking speed conditions. However, this difference did not reach statistical significance (CWS $p= 0.074$, SP $p= 0.057$, 1m/s $p=0.088$).

In addition, participants walked almost symmetrically in terms of step length and time as no significant differences were found between left and right with slightly longer and faster steps for the right side (CWS $p= 0.712$, SP $p= 0.712$, 1m/s $p=0.590$).

On the other hand, during all unperturbed walking speed conditions, subjects showed significant differences in values of ML MoS between left and right sides (CWS p= 0.002, SP p= 0.001, 1m/s p<0.001). It was noticed that participants exhibited significantly larger ML MoS at right initial contacts (CWS 0.146 ± 0.031, SP 0.146± 0.029, 1m/s 0.154± 0.030) than the left side.

Trial name / Variable	CWS Left Mean ± SD	CWS Right Mean ± SD	***p***	SP Left Mean ± SD	SP Right Mean ± SD	***p***	1m/s Left Mean ± SD	1m/s Right Mean ± SD	***p***
AP MoS (m)	0.136±0.033	0.129±0.030	*0.074*	0.151±0.044	0.144±0.040	*0.057*	0.086±0.029	0.079±0.028	*0.088*
ML MoS (m)	0.122±0.026	0.146±0.031	*0.002*	0.122±0.029	0.146±0.029	*0.001*	0.125±0.029	0.154±0.030	*<0.001*
Step length	0.721±0.056	0.723±0.050	*0.712*	0.765±0.057	0.767±0.052	*0.712*	0.658±0.037	0.661±0.028	*0.590*
Step width (m)	0.118±0.033	0.128±0.037	*0.003*	0.113±0.033	0.120±0.034	*0.004*	0.125±0.034	0.132±0.034	*0.006*
Step time (s)	0.547±0.024	0.548±0.025	*0.570*	0.536±0.028	0.538±0.034	*0.448*	0.595±0.035	0.598±0.033	*0.529*

Table 3.2 Mean ± SD of MoS in both AP and ML directions along with gait parameters for left and right side steps for control group' steady state trials (CWS, SP and 1m/s). Significant differences at p <0.05

Similarly, step width values of left and right sides were significantly different (CWS *p*= 0.003, SP *p*= 0.004, 1m/s *p*<0.006). The participants walked with wider right steps (CWS 0.128 ± 0.037, SP 0.120± 0.034, 1m/s 0.132± 0.034)

3.3.2 How did the walking speed affect the gait pattern and MoS of the group's unperturbed trials?

On average, subjects walked 10% faster during SP than the baseline (CWS). About 85 % of the group (18 participants) walked faster during SP whilst 15% walked about the same speed as the reported CWS.

Figure 3.1 shows mean MoS at left and right initial contacts (a mean of 150 steps for each side of each subject) in both AP and ML directions as well as step length and width of the three different walking speeds for the controls (n=22).

AP MoS mean and SD values at left initial contacts for the fixed trial at 1 m/s (0.079±0.028 m) were the lowest values among the three trial types and sides. Whilst, AP MoS at left initial contacts of SP trials were the largest among all unperturbed trials and sides (0.151±0.044 m). The repeated measure ANOVA showed that the AP MoS values were significantly different from each other ($p<0.001$).

The step length values of each trial were also significantly different from each other ($p<0.001$), the shorter steps were seen during walking at 1m/s while the longest ones were during SP trials. Additionally, the variations between steps were the greatest during the SP trials which were slightly more than the baseline.

There were no significant differences in mean ML MoS at initial contacts among all three unperturbed conditions ($p= 0.739$). Subjects tended to exhibit slightly larger ML MoS values when walking at a speed of 1m/s for both sides than the CWS and SP. The ML MoS mean values of CWS and SP were almost similar and slightly differed in variability.

Regarding the step width, no differences were between the values of 1 m/s and baseline (p=0.910), the same when comparing SP step width values to baseline ones. Whilst step width values when comparing the 1 m/s to SP were significantly different (p=0.008). The steps during walking at a speed of 1m/s were the widest among all trials. Similar to ML MoS the values, the step width between CWS and SP were comparable.

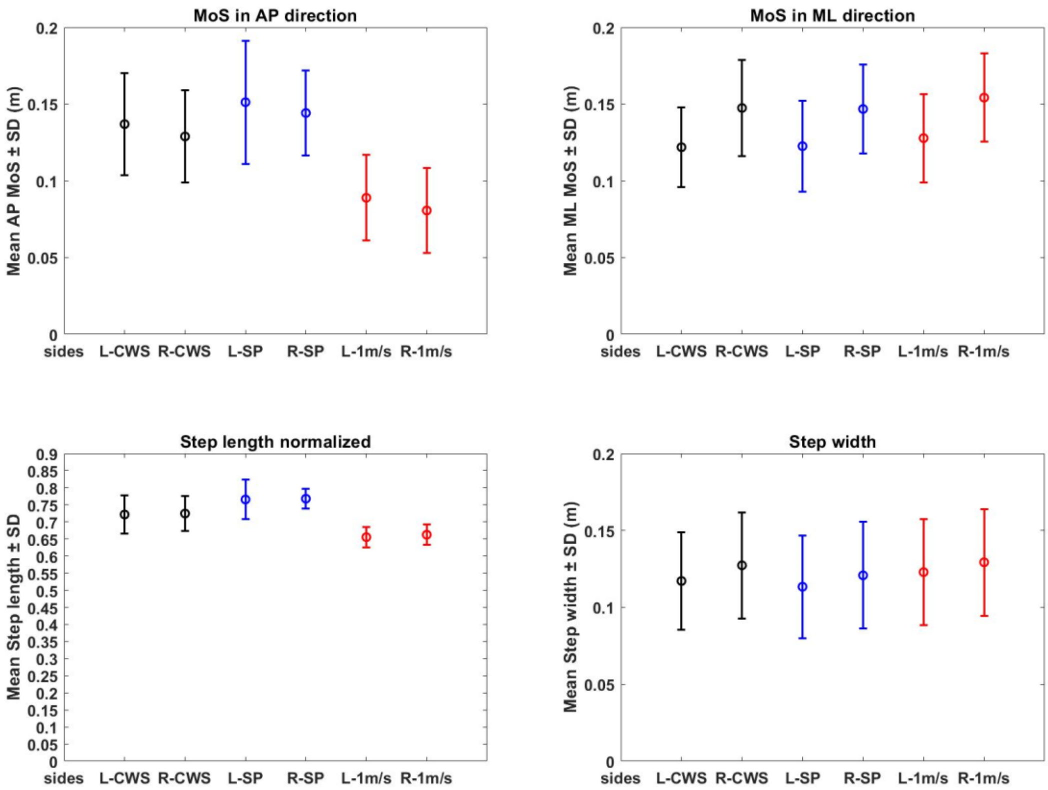

Figure 3.1 Mean MoS in both AP and ML directions as well as step length and width for the right and left sides (150 steps for each) during unperturbed trials. CWS (comfortable walking speed), SP (self-paced mode 1m/s (walking at a fixed speed of 1m/s). (o) represent mean values. Error bars indicate between-subject standard deviations.

3.3.3 The effect of the perturbations

3.3.3.1 The effect of different intensity of perturbation on gait stability and spatiotemporal

There were no significant differences among the MoS and gait spatiotemporal values of baseline, one, two and three steps after applying low intensity perturbations whether acceleration or deceleration on both left and right sides ($p>0.680$).

The medium perturbations were found to be changing the values greater than the low slips specifically following the deceleration slips. The main observation was for the AP MoS. Where the AP MoS means of left and right steps increased after imposing the deceleration compared to the baseline mean value. The left sides were affected more than the right side with a mean difference from the baseline steps of (0.018 m). The steps after were slightly greater in AP MoS than one step before, two and three steps after. Despite that, these differences did not reach statistical significance in any parameter ($p>0.088$).

The greatest variations from baseline steps and other steps were noticed after applying the highest block in the protocol. i.e. high intensity (±65% from the walking speed). Therefore, the steps after this intensity were selected to study the gait stability and recovery mechanism for this group.

As showed in (Figure 2.13), the sequence of the selected steps was as follows: ipsilateral one step before (I1-B) , contralateral one step before (C1-B) , the step where the ipsilateral perturbation was triggered (P-triggered), contralateral one step after (C1-A), ipsilateral one step after (I1-A), contralateral two steps after (C2-A), ipsilateral two steps after (I2-A), contralateral three steps after (C3-A) and finally ipsilateral three steps after (I3-A).

3.3.3.2 How did gait stability and spatiotemporal parameters differ from the steady state and steps before the perturbation?

To answer the above question, it was necessary to evaluate the mean and standard deviations of AP MoS, ML MoS, step length, step width, and step time for both ipsilateral and contralateral steps. The selected steps were one before, one after, two after and three after each type of slips (Table 3.3, Table 3.4,

Table 3.5 and Table 3.6).

3.3.3.2.1 The deceleration slips effects

Overall, the results of this study show participants recovered (i.e. returned to the steps before values) from the deceleration slips through a combined interaction of one step of each sides (one stride). Namely, ipsilateral one step after (I1-A) and contralateral two steps after (C2-A). Hence, these one steps after were called recovery steps.

It was noticed that reducing the walking speed suddenly to approximately walking at speed of 65 percent less than the reported walking speed seemed to be affecting both sides (left and right sides).

In addition, the results of this study show that when challenging the left and right sides, the mechanism of recovery from the deceleration perturbations was found to be symmetrical for both limbs. However, the degree of deviation from the one step before and CWS baseline steps was varied. Since it was found that the deviation from the means was greater when challenging the left sides in all parameters.

When the left side was challenged by deceleration slips					
Parameters	AP (m)	ML (m)	Step length	Step width (m)	Step time (s)
I1-B	0.180±0.047	0.121±0.026	0.720±0.054	0.120±0.036	0.579±0.172
C1-B	0.136±0.027	0.151±0.030	0.712±0.068	0.139±0.031	0.528±0.084
P triggered	0.177±0.046	0.121±0.027	0.723±0.055	0.126±0.033	0.579±0.187
C1-A	0.082±0.059	0.169±0.040	0.499±0.084	0.157±0.036	0.577±0.088
I1-A	0.266±0.060	0.108±0.028	0.746±0.080	0.164±0.047	0.532±0.133
C2-A	0.180±0.034	0.137±0.031	0.720±0.065	0.140±0.037	0.522±0.078
I2-A	0.186±0.049	0.120±0.028	0.701±0.073	0.138±0.041	0.570±0.132
C3-A	0.135±0.022	0.153±0.029	0.713±0.056	0.143±0.029	0.541±0.078
I3-A	0.179±0.042	0.121±0.027	0.702±0.064	0.134±0.037	0.570±0.130

Table 3.3 Mean and standard deviations of AP MoS, ML MoS, step length, step width and step time. Parameters for both ipsilateral (I) and contralateral (C) steps one before (1-B), one after (1-A), two after (2-A) and three after (3-A) deceleration slips on the left side. P triggered is the step that when the slips were triggered.

When the right side was challenged by deceleration slips					
Parameters	AP (m)	ML (m)	Step length	Step width (m)	Step time (s)
I1-B	0.131±0.031	0.150±0.030	0.728±0.055	0.133±0.034	0.543±0.063
C1-B	0.183±0.039	0.122±0.029	0.722±0.052	0.122±0.029	0.528±0.084
P triggered	0.135±0.032	0.150±0.030	0.728±0.055	0.133±0.038	0.557±0.129
C1-A	0.115±0.059	0.148±0.032	0.501±0.103	0.148±0.032	0.577±0.088
I1-A	0.208±0.046	0.139±0.035	0.739±0.080	0.174±0.047	0.500±0.096
C2-A	0.233±0.046	0.136±0.029	0.714±0.069	0.136±0.029	0.522±0.078
I2-A	0.137±0.034	0.151±0.031	0.703±0.067	0.149±0.037	0.547±0.096
C3-A	0.183±0.041	0.133±0.028	0.716±0.059	0.133±0.028	0.541±0.078
I3-A	0.131±0.030	0.152±0.030	0.716±0.055	0.141±0.035	0.547±0.096

Table 3.4 Mean and standard deviations of AP MoS, ML MoS, step length, step width and step time. Parameters for both ipsilateral (I) and contralateral (C) steps one before (1-B), one after (1-A), two after (2-A) and three after (3-A) deceleration slips on the right side. P triggered is the step that when the slips were triggered.

In the AP direction, participants exhibited larger ipsilateral MoS mean and standard deviation (I1-A: L=0.266±0.060, R=0.208±0.046 m) compared to the baseline values (L=0.136±0.030, R=0.129±0.030 m). The deviation from the means was greater for the left (0.13 m). Additionally, when compared to I1-B, participants also exhibited larger AP MoS with mean difference of (L=0.086, R=0.077 m). At C1-A, which was just after the peak of the perturbations, the mean MoS was significantly decreased and found to be the lowest MoS value of all steps (C1-A during: L=0.082±0.059, R=0.115±0.059).

Comparable to I1-A, the AP MoS means were also increased for the C2-A steps. As the left C2-A (0.13 m) , right C2-A (0.5 m) were larger the than the left and right C1-B. Left and right recovery steps (I1-A & C2-A) AP MoS means were significantly different from all other steps including baseline, one step before, two and three steps after ($p<0.001$).

In terms of normalised step length, the participants tended to show longer normalised mean and greater standard deviation I1-A steps after deceleration (L=0.746±0.080, R=0.739±0.080) compared to baseline steps. The deviations from the means were (L= 0.025, R=0.016). The same trend was found when compared to I1-B means, the recovery steps means were larger and the deviations were (L= 0.026, R= 0.011). No significant differences were found between means of I1-A, baseline and I1-B ($p>0.269$). However, the I1-A recovery steps were significantly different from I2-A $p<0.003$ and I3-A ($p<0.027$).

For the contralateral side, the C1-A was sufficiently shorter than all other steps approximately (C1-A ~ 0.5 during left and right). In addition, the C2-A were slightly longer than the steps before but almost equivalent to the baseline values. No significant differences were found ($p>0.320$)

In the ML direction, the results were opposite. Compared to CWS baseline steps and steps before participants exhibited smaller mean MoS in ML direction at I1-A (L=0.108±0.028, R=0.139±0.035 m). The mean differences from the baseline values were (L=-0.014, R=-0.007 m). The deviations from the I1-B mean were (L=-0.013 m, R=-0.011 m).

For the contralateral side, when challenging left and right the C1-A steps ML means were notably increased from the steps before (when L =0.169 m, when R=0.148 m). Different pattern was found for the contralateral one after steps when the ipsilateral sides were challenged. Compared to the C1-B and the C2-A when the left side was challenged the C2-A ML MoS was reduced (0.137±0.031). Whilst there was an increase in ML MoS for the C2-A when right side was challenged (0.136±0.029). Moreover, the standard deviations were slightly different. For both sides, means ML MoS of the I1-A and C2-A were significantly different from all other steps (p<0.001).

In terms of step width, participants took wider I1-A left and right steps as well as greater standard deviations (L=0.164±0.047 m, R=0.174±0.047 m) to cope with the deceleration slips. The differences from the baseline were equal for left and right sides (0.046 m) whilst differences in mean from I1-B were (L=0.044 m, R =0.041 m).

For the contralateral sides, the C2-A and C3-A were very similar to the C1-B steps when the left side was challenged. But when the right side was challenged, the C2-A steps were wider than the C1-B steps. However, the mean step width of C1-A steps was greater than the C1-B when both sides were challenged (when L=0.157±0.036 m, when R=0.148±0.032 m).

The I1-A for left and right recovery steps width means were significantly different from all other steps including baseline, one step before, two and three steps after (*p*<0.001). At the same time no significant different were found among the C2-A, C1-B and C3-A.

The step time of both I1-A and C2-A was less i.e. faster steps compared to the step time of steps before and baseline steps. However, the deviations from the steps before was more obvious for the I1-A steps (L=0.532±0.133 s, R=0.500±0.096 s). The same trend was found when challenging both sides. The C1-A of both sides' steps were found to be the longest in terms of step time (0.577s). The I1-A steps of both left and right time statistically different than other steps (*p*<0.013).

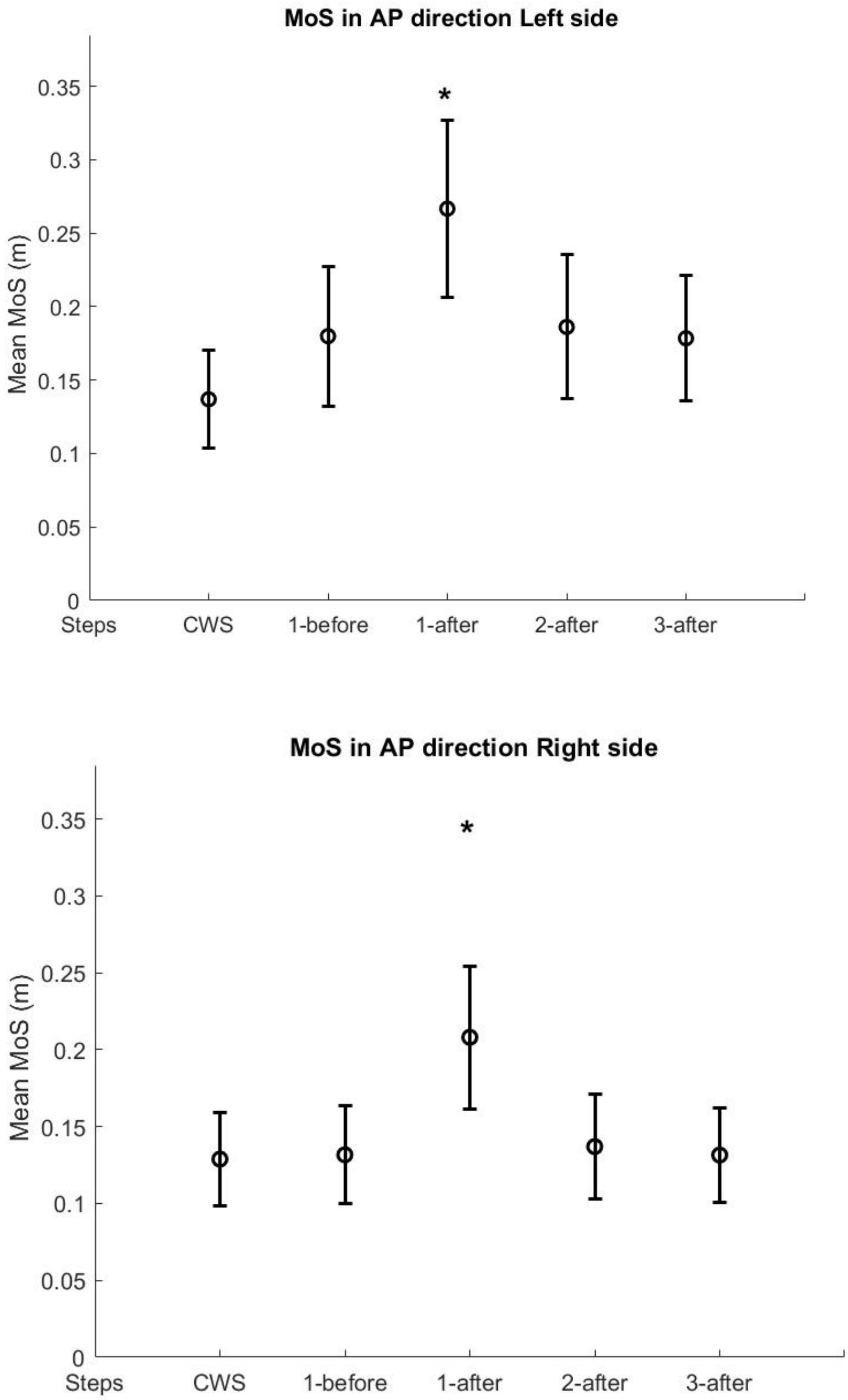

Figure 3.2 Mean MoS in AP direction of CWS (reported by the participant), one step before the deceleration (1-before). One step after deceleration (1-after), two steps after deceleration (2-After) and three steps after deceleration (3-after) for the left and right sides. (o) represent mean values. Error bars represent between-subject standard deviation. (*) indicate significant differences between steps.

3.3.3.2.2 The acceleration slips effects

The effect of acceleration slips contrasted with deceleration method of recovery from perturbations. Between-limb differences were found with a different degree of deviation from the one step before and CWS baseline steps as well (

Table 3.5 and Table 3.6).

In AP direction, compared to the baseline values (L=0.136±0.030, R=0.129±0.030 m), participants exhibited larger MoS mean and larger standard deviation on the left side I1-A step (0.168±0.068). Whilst on the right side participants decreased AP MoS (0.120±0.057 m). The deviation from the means was greater for the left I1-A (0.032 m) than the right (-0.009 m). In addition to, when compared to I1-B, participants exhibited smaller AP MoS on both sides with mean differences of (L=-0.014, R=-0.011 m). ANOVA tests revealed no significant differences between left and right recovery steps (I1-A & C2-A) in AP MoS from all other steps including baseline, one step before, two and three steps after (p>0.069).

The mean normalised step length of I1-A were smaller (i.e. shorter steps) compared to both baseline steps and I1-B. Participants tended to show matching I1-A length on both sides (L=0.608±0.099, R=0.608±0.090). The deviations from the means were (L= -0.113, R=-0.115). Compared to I1-B means, the steps deviated by (L=-0.113, R=-0.118). Left I1-A step length means were significantly shorter from baseline means (p<0.001). But left I1-A was not different from I1-B and steps after (p>0.072). While the right I1-A were found significantly shorter than all other steps (p<0.007).

For the contralateral side, when left and right sides were challenged the C1-A step length mean were the greatest of all steps (when L= 0.803±0.063, when R=0.803±0.064). The C2-A steps were found to be shorter than the C1-B for the left and right. ANOVA tests showed no significant differences when comparing the C2-A to other steps (*p*>0.471).

When the left side was challenged by acceleration slips					
Parameters	AP (m)	ML (m)	Step length	Step width (m)	Step time (s)
I1-B	0.182±0.047	0.120±0.027	0.722±0.054	0.126±0.041	0.579±0.172
C1-B	0.137±0.029	0.150±0.031	0.727±0.053	0.135±0.034	0.551±0.107
P triggered	0.181±0.045	0.120±0.027	0.721±0.057	0.124±0.035	0.542±0.179
C1-A	0.255±0.050	0.133±0.026	0.803±0.063	0.110±0.030	0.479±0.114
I1-A	0.168±0.068	0.125±0.030	0.608±0.099	0.145±0.040	0.531±0.133
C2-A	0.128±0.034	0.161±0.034	0.717±0.064	0.171±0.041	0.560±0.110
I2-A	0.184±0.045	0.120±0.026	0.730±0.060	0.127±0.039	0.570±0.132
C3-A	0.133±0.024	0.151±0.031	0.726±0.056	0.134±0.033	0.550±0.106
I3-A	0.180±0.041	0.121±0.027	0.730±0.058	0.128±0.034	0.570±0.130

Table 3.5 Mean and standard deviations of AP MoS, ML MoS, step length, step width and step time. Parameters for both ipsilateral (I) and contralateral (C) steps one before (1-B), one after (1-A), two after (2-A) and three after (3-A) acceleration slips on the left side. P triggered is the step that when the slips were triggered.

When the right side was challenged by acceleration slips					
Parameters	AP (m)	ML (m)	Step length	Step width (m)	Step time (s)
I1-B	0.131±0.030	0.149±0.030	0.726±0.055	0.132±0.036	0.540±0.063
C1-B	0.182±0.040	0.117±0.026	0.725±0.056	0.121±0.029	0.566±0.126
P triggered	0.131±0.028	0.149±0.031	0. 729±0.055	0.129±0.034	0.523±0.133
C1-A	0.330±0.060	0.100±0.023	0.803±0.064	0.107±0.028	0.492±0.126
I1-A	0.120±0.057	0.160±0.030	0.608±0.090	0.158±0.047	0.500±0.094
C2-A	0.167±0.048	0.133±0.032	0.709±0.061	0.165±0.041	0.577±0.121
I2-A	0.142±0.038	0.146±0.030	0.730±0.063	0.137±0.035	0.543±0.096
C3-A	0.175±0.040	0.119±0.025	0.724±0.055	0.124±0.027	0.570±0.127
I3-A	0.133±0.030	0.150±0.030	0.727±0.058	0.132±0.033	0.540±0.096

Table 3.6 Mean and standard deviations of AP MoS, ML MoS, step length, step width and step time. Parameters for both ipsilateral (I) and contralateral (C) steps one before (1-B), one after (1-A), two after (2-A) and three after (3-A) acceleration slips on the right side. P triggered is the step that when the slips were triggered.

Opposed to deceleration recovery pattern, participants exhibited on average greater ML MoS mean on I1-A (L=0.125±0.030 R=0.160±0.030 m) to cope with acceleration slips compared to CWS and steps before. Whilst the standard deviations were slightly different. The mean differences from the baseline and I1-B values were greater for the right side with deviations from baseline (L=0.003 m, R=0.014 m). The deviations from I1-B were (L=0.005 m, R=0.011 m). No significant differences were found between left and right I1-A steps and other steps including baseline, I1-B, I2-A and I3-A (p>0.076).

For the contralateral steps, the ML MoS means of the steps after C1-A were decreased from the steps before by the same amount of (-0.017 m) when both left and right sides were challenged. The C2-A were greater in ML MoS than the steps before. In addition, it was wound that the C2-A steps were significantly different from the C1-B and C3-A steps (p=0.002).

As in deceleration slips, the left and right I1-A acceleration slips were found to be wider than other steps including baseline, I2-A, I3-A (L=0.145±0.040, R=0.158±0.047m). The means deviations from these steps, however, were less than the deviations from steps after deceleration slips. Where the means deviations from the baseline were (L=0.028 m, R= 0.03 m). And from I1-B the deviations were (L=0.019 m, R=0.026 m). ANOVA revealed significant differences between left and right I1-A and all other steps including baseline, I1-B, I2-A, I3-A (p>0.026).

For the contralateral, the C1-A left and right steps were found to be the narrowest steps (when L=0.110 m, when R=0.107m). The C2-A steps were also found to be wider than the steps before similar to the I1-A. Additionally, they were found to be significantly different to other steps (*p*<0.001)

The step time of I1-A was less i.e. faster steps compared to steps before and baseline steps. (L=0.531±0.133 s, R=0.500±0.094 s). The same trend was found when challenging both sides. The I1-A steps of both left and right time statistically different than other steps (p<0.013).

The contralateral sides step time pattern was different from the ipsilateral. C1-A step time for the both sides' steps were found to be the lowest (when L 0.479s, when R= 0.492s). The C2-A of both sides were found to be longer in time than the C1-B. ANOVA did not show significant difference between C2-A and baseline steps however, it showed that the C2-A steps were significantly longer in time from other steps including C1-B and C3-A (*p*<0.006).

3.3.4 Which perturbation type had the greatest effect over the gait stability and spatiotemporal parameters?

The gait stability and spatiotemporal were differently affected by both different slip types. For the deceleration, the greatest MoS deviations from the baseline values were for the steps following the deceleration slips. Particularly, AP MoS of the left I1-A with approximately 96% increase. The right I1-A was also increased by a considerable amount approximately 62 %in AP direction. The I1-A for the both left and right were also increased compared to the I1-B by approximately 50%.

In terms of contralateral, when challenging the left side, the C1-A (i.e. right side) AP MoS was decreased by 36%, then increased by almost 40% in C2-A from the baseline values. In the meantime, the same C1-A was also decreased by 40% from the C1-B. Then increased by 35% in C2-A.

When challenging the right side, the C1-A AP MoS was also decreased but with a less degree by 15%. Then increased by almost 72% in C2-A from the baseline values. In the meantime, the same C1-A was also decreased by 40% from the C1-B. Then increased by 27% in C2-A.

AP MoS in the situation of speeding up has substantially lower effect on the ipsilateral sides with a decreased in MoS by less than 10% even when comparing to both baseline values and steps before.

Despite that, it was noted that the major effect of this type of slips was on the contralateral sides. As when challenging the right sides, the C1-A deviated from the baseline means by 140% increase and 80% increase from the C1-B. The C2-A was also increased but with much less percentage of 20. Similar pattern was noticed for the C1-A when challenging the left side with an increase of 100% and 85% from the baseline and C1-B respectively. The C2-A was almost the equal to C1-B.

By percentage, the left and right I1-A mean ML MoS were not considerably different in absolute value for the steps after deceleration and acceleration slips. Whereas they were different in signals; after deceleration slips, they were decreased by almost 10% from both baseline and I1-B. Whilst they were increased by less than 5% when imposing acceleration slips.

For the contralateral side ML MoS, the C1-A and C1-2A increased by less than 20% from both baseline and C1-B when imposing deceleration on the right side. It was noted that the C1-A deviated more when the right was challenged. Meantime, the C1-A was decreased by less than 18% from the baseline and C1-B after acceleration slips.

As to gait spatiotemporal parameters, the mean step width appeared to be affected more by deceleration than acceleration slips. Compared to the baseline and I1-B, the left and right I1-A mean step width after deceleration slips were the most effected parameters, with an increase of 35%. Means step width of the left were changed more than the right sides.

The step length was more deviated after acceleration slips. Step length means of the I1-A were 20 % more than I1-B and baseline.

The step time of the I1-A steps appeared to be equally affected by the two different slip types when compared to baseline and I1-B with a decreased by an estimation of 10%.

3.4 Discussion

The aim of this part of the study was to understand how able-bodied individuals control their dynamic stability while they steadily walk and when they face sudden unbalanced situations. Understanding of this may help in prevention of falls not only for able-bodied individuals but also for less able-bodied groups such as prosthetic users.

Application of mechanical perturbations such as slips, in a safe environment can directly challenge the dynamic stability. In this way a sudden change in walking speed is imposed. A popular stability measure called margin of stability firstly developed by (Hof et al., 2005) along with gait spatiotemporal were adopted to understand the dynamic stability.

An indicator of reduced stability is the increased variability among the recovery steps. This reflects the different MoS values that subjects exhibit each time to maintain the balanced status (McAndrew Young et al., 2012). This suggestion was also supported by Gates et al. (2013)

stating that variability may reveal how well controlled this measure is under different walking condition.

The results of the AP MoS mean values for the different unperturbed walking conditions may support this interpretation. The lower MoS values and variability were found during walking at 1 m/s which was the slowest one among different unperturbed walking conditions whilst the greatest values were found during walking at the highest speed which was during SP mode. When walking at faster speed as in SP mode greater stability is needed thus the participants have increased their margin of stability to accommodate the balance requirements. Despite that the participants have been given the chance to become familiar with the SP system before recording, the system yet still new to them as more caution and extra effort were needed. Whereas when they walked at the lowest 1m/s speed which was at least 20 % less than their comfortable walking no AP increased margins were needed. This was without forgetting no extra care to be taken as the treadmill was at fixed based where they did not need to think about their position in the treadmill as in SP mode. These results correspond with previous findings (Hak et al., 2013b, Hak et al., 2013a). Walking at slower speed has been also previously linked with increased stability in a study of (England and Granata, 2007). The results of the step length support this interpretation as the participants took the longest steps while walking at SP based mode (0.765). Whilst the shortest were seen during 1m/s walking (0.658). It was noticed that in the AP direction both sides were symmetrical as no significant differences were found in both AP MoS and step length parameters when comparing left and right. With slightly greater values were found for the left side.

A consistent MoS in ML asymmetries were found between left and right initial contacts among all walking conditions (Table 3.2,Table 3.3, Table 3.4,

Table 3.5, Table 3.6 and Figure 3.1)

These observations may also support that increased MoS can be an indicator of increased stability in that direction. As all the participants have reported that their dominant side was the right side. Therefore, it is reasonable to assume that because they were relying more on the right side the ML margins of stability on the right side were higher.

Similar observations were reported by several studies. Rosenblatt and Grabiner (2010) discussed that this asymmetry may represent a functional asymmetry and these results suggest the idea of a limb preference when it comes to maintain stability in this direction. This idea was also supported by Sadeghi et al. (2000). England and Granata (2007) speculated that these asymmetries might reflect differences in stability needs of each side to be able to provide the sufficient support during walking. Similar to (McAndrew et al., 2011, McAndrew Young et al., 2012), the subjects of the present study exhibited greater ML MoS at their right initial contacts across all unperturbed walking conditions. Similar interpretation may explain the present study ML MoS results.

Increased step width was demonstrated to increase mean MoS (Hak et al., 2013b, McAndrew et al., 2010, McAndrew Young and Dingwell, 2012), this may contribute to the asymmetry in mean MoS in the left and right. As the right sides step width were greater than the left.

In terms of the perturbation results, the effect of the imposed perturbation intensity was as anticipated. The stronger was the perturbation, the greater effect on the pattern. Since during the low and medium intensities, no significant differences were found among MoS and gait spatiotemporal parameters for the recovery steps when compared to steps before and baseline. These results were also reported in previous work of Aprigliano et al. (2015) who adopted slips perturbation applied in the AP direction by the means of AP (forward) movement. However, in their research at that time the sample size was relatively small (n=5). Despite this, while later the same group has expanded their study and applied the same methods on a relatively larger study sample that contained two subgroups; elderly and young persons. The results of the study were consistent as well (Martelli et al., 2017a).

Table 3.7 below provides a summary of MoS results in both AP and ML directions after the controls participants faced the perturbations. The steps presented were the first two contralateral steps and first ipsilateral step after both deceleration and acceleration slips compared to the corresponding steps before.

Deceleration		
Step	***AP MoS***	***ML MoS***
C1-A	Decreased (varied between sides, R: ~60% *, L: ~15% *)	Increased (varied between sides R: ~15% *, L: ~20% *)
I1-A	Increased (L:50% *, R:60% *)	Decreased (~10% *)
C2-A	Increased (~35% *)	Decreased (~10% *)
Acceleration		
Step	***AP MoS***	***ML MoS***
C1-A	Increased (>90% *)	Decreased (~10%*)
I1-A	Decreased but not significantly (<10%)	No noticeable change
C2-A	Decreased but not significantly (< 10%)	No noticeable change

Table 3.7 Summary of the perturbation effect on the MoS compared to steps before. (*) represents significant difference (*$p<0.05$).*

The results of perturbations demonstrated that able-bodied individuals required a full two steps in order to go back to the balanced status as in baseline and steps before. These steps are a combined interaction of one ipsilateral step and two contralateral steps after imposing the slips. This was clearer after decelerations slips. In addition, the results showed that the stability in both AP and ML directions were challenged and changed despite that the perturbations are imposed only in the AP direction. Similar results were reported in a study of McAndrew et al. (2011), however, in their study the AP perturbations were imposed by means of AP platform oscillation.

The deceleration slips decreased the distance between the BoS border and XCoM therefore the AP MoS at C1-A was significantly decreased. The balance was recovered by taking longer, wider and faster steps as happened to I1-A and C2-A. These gait adaptions agree with several previous reports. The participants in a study of Sivakumaran et al. (2018) showed longer and wider steps to recover from AP perturbations which was provided by means of pseudo-random fluctuations in the speed of treadmill belt. Participants showed increased step width and length when walking over a compliant surface to regulate their dynamic stability in (MacLellan and Patla, 2006) study. Taking wider steps were also reported to be one of the adaptions to cope with the mechanical perturbation in several studies (Hak et al., 2013a, Hak et al., 2013b, Hak et al., 2013c, Madehkhaksar et al., 2018, McAndrew Young and Dingwell, 2012).

Following the deceleration slips, the means ML MoS of the steps I1-A and C2-A were decreased from the I1-B and C1-B steps even though that the participants at these steps took wider steps. This might be because the deceleration slips were extra challenging as the participants needed to match the forward velocity to keep up with the treadmill. Therefore, the focus was aimed towards the AP direction stability. This interpretation agrees with other reported results by (Peebles et al., 2016). The decreased in MoS ML was also reported in the result of Sivakumaran et al. (2018) research. A study of O'Connor and Kuo (2009) showed that movement strategies are thought to be independent between the AP and ML directions. It is likely that participants increased their step length to cope the AP perturbation as demonstrated by increased AP MoS at initial contacts of the I1-A and C2-A steps, the trade-off for this adaption was an increased step width to maintain ML MoS.

Although the AP MoS was increased on both I1-A and C2-A when the ipsilateral side was challenged, it was noticed that the contribution of the I1-A to recover was greater than the C2-A. This may suggest that able-bodied can recover efficiently on both sides. However, participants recovered the balance by the right side when the right side was challenged slightly more than the left side when the left side was challenged. The percentage of the right I1-A compared to the right I1-B was 60% whilst it was 50% for the left I1-A compared to left I1-B.

Participants demonstrated a range of different adaptions to cope with the acceleration slips. The step width was also increased similar to the deceleration recovery mechanism. However, this increase did indeed reflect on the ML MoS. The ML MoS of the I1-A and C2-A was increased compared to steps before and baseline. The step length on the other hand, was decreased which also affected the AP MoS and resulted in decreased AP MoS. The step time of the I1-A and C2-A was inconsistent, as the I1-A time was less meanwhile the C2-A was more than steps before and baseline. The acceleration steps may tend to shift the CoM more anteriorly which explain the significant increase of AP MoS and step length in C1-A.

The acceleration initial effect in AP MoS was greater than deceleration as seen in C1-A which were affected by nearly 90% increase. Besides the effect on MoS, the acceleration slips resulted in significant differences in step length and time of the recovery steps compared to the baseline and steps before. Nonetheless, the acceleration slips appeared to be less challenging when compared to deceleration slips. As less effort was required to go back to steady state by the steps after. These results are also supported by other studies. Ilmane et al. (2015) showed that acceleration perturbation needed one step to go back to the balanced situation whilst the deceleration perturbation took four steps. A newer study by Roeles et al. (2018) supports the same results. Another factor that may explain why the acceleration slips were less effective is that the perturbation intensities may have been too low to provoke responses for the control group. However, since the aim of this study was to study the stability in a fall free environment for the prosthetic users, the perturbation intensities were not increased.

The interpretation of increased MoS indicates higher stability is in a conflict with other reported findings of individuals who are at higher risk of falling. Firstly, in a study of Hof et al. (2007) who examined the stability of above-knee prosthetic users. The participants in that study have exhibited larger mean MoS in ML as well as wider steps. The same results were also reported in a study of McAndrew Young et al. (2012) who demonstrated increased mean MoS values for the transtibial prosthetic users. Another study by Beltran et al. (2014) who interpreted the greater mean and variability of the prosthetic users in the ML direction as a signal of less lateral stability of this group when compared to able-bodied. This idea has been also debated in elsewhere research that included arguably another less able-bodied as in the work of (Peebles et al., 2017, Peebles et al., 2016) who examined the dynamic stability of the persons with multiple sclerosis. Notwithstanding, the results of these studies in fact may also support the opposite. It can be explained as because these individuals needed a compensatory gait strategy to cope with the applied external perturbations, they tended to show increased MoS. The remarkable increase in the gait parameters for example step width, have contributed in the increased ML MoS when compared to more able-bodied. Nonetheless, the result of the next chapter may help in clarifying the meaning of MoS values.

It was noted that apart from the left side I1-B AP MoS and step time, the steps before the two types of perturbation were very much comparable to the baseline steps. These results demonstrate that the participants were capable to fully return to the steady state walking despite that they were perturbed with different intensities and types. The left side steps before however, tended to have increased mean AP MoS and step time compared to left baseline steps values. This can be explained again by the dominant vs non-dominant side idea. As the participant can cope with the changes on the right side and recover better using their dominant side whilst more preparation was needed for the left side hence the left step time mean and variability for the steps before were increased as well. Similar pattern was reported by Roeles et al. (2018), where the AP MoS for non-dominant side was significantly greater than the baseline steps.

This study's protocol including the both perturbations types and different intensities was not changed for the prosthetic users' group in the next chapter for several reasons. First, it was unclear how would a prosthetic user recover when the balanced is challenged by acceleration and deceleration, whether would the recovery mechanisms be similar to the controls or they would adopt a different method.

Secondly, in order to randomize the protocol as much as possible, preventing the prosthetic users from predicting the upcoming perturbation whether it would be increasing in walking speed slips or decreasing which may help in imposing a real time fall situation. Therefore, the protocol may be more sensitive to compare the stability of the two different groups.

Lastly, imposing different level of intensities of perturbations may have helped in revealing that to challenge an able-bodied stability, at least 60% change in walking speed is needed. The use the lowest intensity in case that the high or the medium intensities were found difficult to cope for any prosthetic user.

3.5 Conclusion

Participants showed the ability to regain balance after two steps. These were one ipsilateral step after and two contralateral steps after the imposed slips. The increased MoS may indicate increased stability. The deceleration tended to be more challenging than the acceleration. There were consistent MoS in ML asymmetries between left and right initial contacts. Even though, able-bodied persons yet can still recover efficiently on either side with slightly more advantage for the dominant side. The greater intensity of the perturbation, the greater effect on the gait. Based on these results, adaptations in step length, step width, and time can be used to adjust the MoS and regain balance. The participants took wider and faster steps to maintain the balance regardless of the applied perturbation. The step length was increased after deceleration whilst decreased after acceleration. Despite that the slips have been applied in the AP direction, both AP and ML MoS were affected. The most deviated MoS parameter was in the AP direction, particularly the AP MoS of C1-A following an acceleration. The greater recovery in AP MoS was seen by the left I1-A after deceleration. In terms of gait parameters, the step width after deceleration on the left side was the greatest recovery by 40% increased width compared to baseline.

It would be advantageous to investigate whether individuals with gait impairments are able to walk at different combinations of step length, step width and time. Additionally, how these alterations affect the MoS. Training focused on the adaptations in step length and stride width might help these individuals to better regulate gait stability, and thereby decrease the risk of falling.

3.6 Clinical implication summary

To recover from slips, individuals may benefit from training and awareness of the need to take wider and faster steps regardless of the direction of the slips. For backward slips (acceleration perturbations), shorter steps may help in maintaining forward balance (MoS) and thus potentially avoid or reduce the incidence of falls. Conversely, for forward slips (deceleration perturbations), longer steps may be beneficial in fall avoidance.

4 Prosthetic users' margins of stability

4.1 Introduction

This chapter explores an arguably less able-bodied group, the prosthetic users. As discussed in chapter one, the stability of this group is yet not clear. The aim of this chapter of the thesis was to firstly, provide an overview of the prosthetic users' dynamic stability status as group and to assess whether this group reacted differently than the able-bodied group when exposed to perturbations. The hypothesis was that prosthetic users would be less stable i.e. exhibit less MoS than the control group, increased variability, more irregularity and be more affected by the imposed perturbations.

The reported dynamic stability results of this group found to be conflicting and therefore further investigation is required. It was found that there are three different opinions when comparing prosthetic users to able-bodied individuals. Prosthetic users were found to be less stable than able-bodied subjects as in study of (Beltran et al., 2014); were reported to be equally stable to able-bodied as in (Curtze et al., 2010) and even found to be more stable as in a study of (Kendell et al., 2010). To examine the dynamic status of the prosthetic users, the same protocol was adopted as that employed in ***Chapter three*** with able-bodied subjects (section 2.4 describes the protocol).

Fifteen participants participated in this part of the study, the sample criteria and method of recruitment are addressed in this chapter. Results of gait stability and parameters are presented and discussed in this chapter. The method of perturbation was the same as for the control group in ***Chapter three***. Similar to chapter three, MoS and gait spatiotemporal measures results are initially presented for the unperturbed trials and then step to step parameters are presented. The selected steps were also similar, and the time of these steps is showed in Figure 2.13.

In addition to the overall performance of the prosthetic users as a group, the effects of the prosthetic foot on the stability and recovery mechanisms are discussed in this chapter. The aim of this part was to determine the biomechanical adaptations made by active unilateral below the knee prosthetic users when they used a prosthesis incorporating a foot with hydraulically controlled ankle compared to conventional ESAR foot. The assumption was that the prosthetic users with the more advanced feet would exhibit improved stability compared to the prosthetic users fitted with conventional feet.

Finally, a conclusion regarding this group dynamic stability is provided at the end of this chapter.

4.2 Methods

4.2.1 Sample criteria

The criteria for participation was a unilateral below the knee lower limb prosthetic user. Ambulatory without walking aids, no other known musculoskeletal problems affecting the contralateral limb, had no current problems with their prosthesis. Have been a prosthetic user walking with a prosthesis for minimum a year; and aged 18 and older.

Individuals who were unable to walk for a minimum of 1 minute, used walking aids or suffered from motion sickness were excluded. In addition, subjects who known to be pregnant or suffered from other musculoskeletal problems such as osteoarthritis, chronic back pain, knee instability affecting the contralateral limb were also excluded.

4.2.2 Method of Recruitment

The project protocol was approved by the NHS ethics committee (IRAS project ID: 244306). As well as, the university of Strathclyde ethics committee (UEC) approved the study. A poster was pinned onto notice boards, and the advert was available as a handout in the waiting area of the prosthetics and physiotherapy departments at Greater Glasgow and Clyde and Lanarkshire Health Boards. Additionally, the poster was sent to prosthetics users' charities and associations. Participants provided written informed consent prior to trials session.

4.2.3 Protocol and Equipment

Participants walked on the CAREN (please see section 2.1 for more details).

4.2.4 Participants

Fifteen prosthetic user participants volunteered for this part of the study. below shows the participants' characteristics.

Characteristics (mean ± SD)	Prosthetic users (n=15)
Age (yr)	55 ± 13.11
Sex (male/female)	10 / 5
Height (cm)	173.45 ± 9.19
Body weight (kg)	79.8 ± 15.19
Found Comfortable Walking Speed (m/s)	1.084 ± 0.254
Effected side (L/R)	(6/9)
Time since amputation (year)	19.85 ± 13.54

Table 4.1 Prosthetic users group's characteristics.

4.3 Results

All participant completed the trials without falling. In total, for each intensity of CWS perturbations, 87 acceleration slips were applied on each leg sides, as well as 87 deceleration slips were applied on each side leg sides. As anticipated not all prosthetic users managed to walk at speed of 1 m/s. Three participants walked at speed of 0.66, 0.46 and 0.92 m/s. Two participants reported the CWS as 1 m/s.

The average number of steps per trial was 160 steps for each side, average total time of any trial was approximately 180 seconds.

The level of two of participants was at the ankle (Ankle disarticulation) while the rest were classified as transtibial prosthetic users. The causes of limb loss for the sample were as follow: nine participants following a trauma, four as result of diabetes, one participant as result of Osteomyelitis and one congenital anomaly (phocomelia).

In terms of the prosthetic foot, the sample can be divided into two main group depending on the present of prosthetic ankle. Where ten participants were fitted with prosthetic foot that has no prosthetic ankle whilst five participants were fitted with prosthetic ankle foot type.

Similar to control group, the average number of steps per trial was 160 steps for each side, average total time of any trial was approximately 180 seconds.

The results are presented firstly as a group, then the prosthetic users were sub-grouped based on the prosthetic foot type and finally a comparison was made between two groups; controls (able-bodied) and prosthetic users.

4.3.1 Descriptive overview of steady state trials: baseline walking CWS, SP and 1m/s

During baseline where prosthetic users walked at CWS, prosthetic users walked at a speed of 1.084 ± 0.254 (mean ± SD). The maximum reported walking speed was 1.32 m/s. This speed was set according to the reported comfortable walking speed by each participant (see section 2.7.1). Whilst subjects walked at a speed of 1.20±0.31 m/s during SP mode. Table 4.2 below shows the mean and standard deviation of the MoS for both AP and ML direction and gait spatiotemporal parameters for the intact and prosthetic sides during unperturbed trials. Firstly, the differences between prosthetic and intact side were investigated. Then the effect of the walking speed on gait stability and pattern of this group was presented. Individuals who have reported their comfortable walking speed at 1m/s (n=2) were not repeated in CWS.

Trial	side	AP MoS (m)	ML MoS (m)	Step length	Step width (m)	Step time (s)
CWS	Intact	0.126±0.030	0.161±0.025	0.648±0.159	0.186±0.047	0.555±0.069
	prosthetic	0.099±0.048	0.157±0.037	0.657±0.170	0.185±0.046	0.585±0.105
	p	***0.003***	***0.675***	***0.410***	***0.856***	***0.032***
SP	Intact	0.146±0.043	0.160±0.027	0.726±0.142	0.179±0.049	0.537±0.046
	prosthetic	0.124±0.054	0.154±0.037	0.724±0.181	0.180±0.047	0.554±0.070
	p	***0.007***	***0.541***	***0.892***	***0.882***	***0.140***
1m/s	Intact	0.084±0.025	0.158±0.028	0.565±0.042	0.170±0.053	0.585±0.042
	prosthetic	0.070±0.035	0.150±0.041	0.570±0.061	0.174±0.054	0.595±0.022
	p	***0.047***	***0.501***	***0.715***	***0.494***	***0.410***

Table 4.2 Mean ± standard deviation of the MoS for both AP and ML direction and gait spatiotemporal parameters for the intact and prosthetic sides during unperturbed trials. ‘Baseline at CWS’, Self-paced mode (SP) and 1m/s. The step length is normalised by leg length

Prosthetic users tended to exhibit significantly larger AP MoS values at the intact initial contacts than at prosthetic side initial contacts for all unperturbed walking speed conditions (CWS p= 0.003, SP p= 0.007, 1m/s p=0.047). In addition, the AP MoS variability was greater on the prosthetic side.

Conversely, during all unperturbed walking speed conditions, subjects showed no significant differences in mean ML MoS between intact and prosthetic sides (CWS p= 0.675, SP p= 0.541, 1m/s p=0.501). Nevertheless, mean ML MoS was greater on the intact side. It was noticed that participants exhibited greater ML MoS variability at prosthetic initial contacts (CWS = ±0.037, SP = ±0.037, 1m/s= ±0.041) than the intact side.

Interestingly, prosthetic users walked almost symmetrically in terms of step length and width as no significant differences were found between the intact and prosthetic side for the unperturbed trials. Despite this, the variability was again greater on the prosthetic side. The step time on the other hand, was found to be increased for the prosthetic side among unperturbed walking speed conditions. The significant differences, however, were only found between intact and prosthetic side during the CWS (p=0.032).

4.3.2 How did the walking speed affect the gait pattern and MoS of the group's unperturbed trials?

On average, subjects walked 20% faster during SP than the baseline (CWS). About (60%) of the group (9 participants) walked faster during SP whilst 40% walked about the same speed as the reported CWS.

Table 4.2 shows mean MoS at prosthetic and intact initial contacts (a mean of 150 steps for each side of each subject) in both AP and ML directions as well as step length and width of the three different walking speeds for the controls (n=15).

AP MoS mean and SD values at prosthetic initial contacts for the fixed trial at 1 m/s (0.070±0.035 m) were the lowest values among the three trial types and sides. Whilst, AP MoS at intact initial contacts of SP trials were the largest among all unperturbed trials and sides (0.146±0.043 m). The repeated measure ANOVA showed that the AP MoS values were significantly different from each other (p<0.010).

The step length values of each trial were also significantly different from each other (p<0.001), the shorter steps were seen during walking at 1m/s while the longest ones were during SP trials. Additionally, the variations between steps were the greatest during the SP trials which were slightly more than the baseline.

There were no significant differences in mean ML MoS at initial contacts among all three unperturbed conditions (p> 0.654). The ML MoS mean values of CWS, SP and 1m/s were almost similar and slightly differed in variability.

Regarding the step width, no differences were between the values of CWS and SP (p=0.092), the same when comparing SP step width values to 1m/s. Whilst step width values when comparing the 1 m/s to CWS were significantly different (p=0.041). The steps during walking at a speed of 1m/s were the narrowest among all trials.

4.3.3 The effect of the perturbations for the prosthetic users as a group

4.3.3.1 The effect of different intensity of perturbation on gait stability and spatiotemporal

Comparable to the able-bodied group, there were no significant differences among the MoS and gait spatiotemporal values of baseline, steps before and after applying low intensity perturbations whether acceleration or deceleration on both intact and prosthetic sides ($p>0.710$).

On the other hand, the medium perturbations did challenge the stability for three prosthetic users. The major change was for the AP MoS particularly after deceleration. It was noted that these prosthetic users walked slower than other participants (CWS < 0.9 m/s).

Similar to control group the greatest variations from baseline steps were noted after imposing the highest intensity block of the protocol whereas the speed was changed by (±65%) of the CWS. Therefore, the trials of the high intensity were selected to be stability and stepping mechanisms investigation. Correspondingly, the steps before and after the slips were selected to see how the gait stability and spatiotemporal were changed. As showed in (Figure 2.13), the sequence of the selected steps was as follows: firstly, ipsilateral one step before, contralateral one step before, the step where the ipsilateral perturbation was triggered, contralateral one step after, ipsilateral one step after, contralateral two steps after, ipsilateral two steps after, contralateral three steps and finally ipsilateral three steps after.

4.3.3.2 How were the gait stability and spatiotemporal parameters differed from the steady state and steps before the perturbation?

Similar to the able-bodied group, the mean and standard deviations of AP MoS, ML MoS, step length, step width, and step time for both ipsilateral and contralateral steps. one before, one after, two after and three after each type of slips were analysed.

4.3.3.2.1 The deceleration slips effects

In the AP direction, participants exhibited larger MoS mean and standard deviation on both intact and prosthetic sides compared to baseline. In addition, both sides reacted similarly when they were challenged by deceleration. However, they are different in the amount of variation from the baseline and steps before. For the intact side, the intact I1-B steps were larger than the baseline steps, and significantly different from baseline (p=0.029). Moreover, the three intact ipsilateral steps after were significantly different from the baseline (p<0.015). Compared to the intact steps before, the first two intact ipsilateral steps after were significantly different (p<0.021). Although the intact I3-A were larger in mean values, no significant differences were found compared to step before I1-B (p=0.551). Both intact and prosthetic I1-A steps showed the most increased from the steps before I1-B. The percentage of this increase was 43 % and 53 % respectively. When challenging the intact side, the first ipsilateral step after deceleration was the largest AP MoS (I1-A intact = 0.212±0.065 m).

When the intact side was challenged by deceleration slips					
Parameters	AP (m)	ML (m)	Step length	Step width (m)	Step time (s)
I1-B	0.148±0.055	0.168±0.030	0.664±0.119	0.201±0.046	0.517±0.076
C1-B	0.124±0.063	0.156±0.033	0.648±0.154	0.193±0.044	0.586±0.145
P triggered	0.153±0.054	0.163±0.031	0.684±0.125	0.188±0.042	0.552±0.136
C1-A	0.098±0.060	0.181±0.040	0.453±0.130	0.199±0.033	0.601±0.139
I1-A	0.212±0.065	0.154±0.026	0.666±0.144	0.206±0.041	0.485±0.087
C2-A	0.169±0.076	0.149±0.037	0.641±0.164	0.193±0.052	0.547±0.156
I2-A	0.177±0.062	0.155±0.028	0.646±0.134	0.209±0.052	0.498±0.110
C3-A	0.132±0.072	0.160± 0.036	0.635±0.159	0.195±0.043	0.570±0.157
I3-A	0.161±0.053	0.159±0.029	0.636±0.119	0.210±0.045	0.508±0.093

Table 4.3 Mean and standard deviations of AP MoS, ML MoS, step length, step width and step time.Parameters for both ipsilateral (I) and contralateral (C) steps one before (1-B), one after (1-A), two after (2-A) and three after (3-A) deceleration slips on the intact side. P triggered is the step that when the slips were triggered

When the prosthetic side was challenged by deceleration slips					
Parameters	AP (m)	ML (m)	Step length	Step width (m)	Step time (s)
I1-B	0.125±0.063	0.161±0.038	0.629±0.164	0.199±0.058	0.604±0.163
C1-B	0.154±0.057	0.163±0.031	0.671±0.131	0.188±0.044	0.530±0.095
P triggered	0.131±0.064	0.156±0.040	0.658±0.144	0.187±0.045	0.575±0.141
C1-A	0.148±0.061	0.182±0.037	0.393±0.132	0.201±0.044	0.561±0.109
I1-A	0.193±0.078	0.159±0.040	0.631±0.156	0.219±0.054	0.543±0.128
C2-A	0.202±0.063	0.145±0.033	0.695±0.147	0.201±0.062	0.525±0.153
I2-A	0.146±0.066	0.183±0.093	0.606±0.162	0.218±0.047	0.591±0.173
C3-A	0.165±0.057	0.161±0.030	0.660±0.127	0.193±0.046	0.538±0.154
I3-A	0.138±0.063	0.161±0.037	0.610±0.162	0.210±0.053	0.597±0.183

Table 4.4 Mean and standard deviations of AP MoS, ML MoS, step length, step width and step time. Parameters for both ipsilateral (I) and contralateral (C) steps one before (1-B), one after (1-A), two after (2-A) and three after (3-A) deceleration slips on the prosthetic side. P triggered is the step that when the slips were triggered.

When the prosthetic side was challenged, all the prosthetic ipsilateral before and after steps were significantly different from the baseline (p<0.031). The three prosthetic ipsilateral steps after were significantly different from the steps before (p<0.012). The prosthetic I1-A step was also the largest ipsilateral step (0.193±0.078 m).

For the contralateral side when the intact was challenged, the mean MoS at the prosthetic C1-A was significantly decreased and found to be the lowest MoS value of all steps (0.098±0.060 m). The prosthetic C2-A was significantly different from the C1-B (p<0.001). When the prosthetic side was challenged the C2-A was the largest step (0.202±0.063 m). C2-A and C3-A steps were significantly different from the C1-B (p<0.011).

When the intact side was challenged, the step length of steps before and the all three ipsilateral steps was not significantly different from the baseline (p >0.821). Compared to the intact steps before, intact I2-A and I3-A were shorter and significantly different (p<0.004) whilst the I1-A was slightly longer but not significantly different (p=0.870). Meanwhile, when the prosthetic side was challenged, step length of prosthetic I1-B and I1-A were not significantly different from the baseline (p>0.242). whilst significant differences were found when compare I2-A and I3-A to baseline (p<0.020). The first step after was not significantly different from steps before (p>0.100). However, the I2-A and I3-A were significantly different from steps before (p<0.040). The prosthetic I2-A and I3-A were found to be roughly 3% shorter than steps before and was shorter by 6% shorter than baseline steps.

For the contralateral side, when the intact side was challenged the step length of the prosthetic side was seemingly changed only at C1-A. The steps C2-A and C3-A were not significantly different from the steps before (p>0.402). On the other hand, when the prosthetic side was challenged the length of intact steps have been changed. Particularly, the intact C2-A was longer than other steps (0.695±0.147) and it was significantly different from intact ipsilateral before and after steps (*p*<0.016). Only C1-A was significantly different from other contralateral steps (*p*<0.001). The shortest step was the first intact step after the prosthetic side was challenged (0.3973±0.132).

In terms of stability in the ML direction, the mean ML MoS of all three ipsilateral steps after were decreased and were smaller than ML of steps before and baseline when the intact side was challenged by deceleration. The intact before and after ipsilateral steps were not significantly different from the baseline (*p*>0.161) when the intact side was challenged. At the same time, the means ML MoS at the first two steps I1-A and I2-A were smaller and significantly different from the steps before (*p*<0.005). When the prosthetic side was challenged, the mean ML MoS was almost the same for the first and third ipsilateral after steps. despite that the mean and variability of the second ipsilateral were increased to be greater than steps before and baseline, no significant differences were found between ipsilateral steps and baseline (*p*>0.334) as well as between ipsilateral steps before and after (*p*>0.298).

When the intact side was challenged, the mean and variability ML MoS of the first prosthetic contralateral step after was increased to be the greatest ML MoS step (0.181±0.040). It is significantly different from the C1-B, C2-A and C3-A (*p*<0.001) as well as significantly different from the first intact ipsilateral step (*p*=0.026). After the increase in the C1-A, the ML MoS tended to be going back to the C1-B values for the following two contralateral steps (C2-A 0.149±0.037, and C3-A 0.160± 0.036). The C2-A was significantly different from the steps before (*p*=0.027) meanwhile the C3-A was not significantly different from the steps before (*p*=0.181). When the prosthetic side was challenged, the ML MoS of the first contralateral step was greater than steps before and baseline. In addition, it was significantly different from baseline and steps before ($p<0.001$). The second intact step after was smaller than steps before and significantly different ($p<0.003$).

Participants tended to take wider steps at the initial contacts of both sides for all steps. As the mean step width of ipsilateral and contralateral were increased when challenging both sides by deceleration. The ipsilateral after steps when the intact was challenged were slightly wider than steps before. All intact ipsilateral steps were wider than the prosthetic contralateral steps. No significant differences were found between ipsilateral steps after and before ($p>0.068$) however, all three ipsilateral steps were significantly different from the baseline ($p<0.016$). contrary, when challenging the prosthetic side, all three ipsilateral steps after were significantly wider than the ipsilateral steps before and baseline ($p<0.008$). It was noted that similar to the baseline, the prosthetic ipsilateral steps after were also not significantly different from the intact ipsilateral steps after ($p>0.122$).

The contralateral steps were wider on average than baseline steps when challenging both limbs, all prosthetic contralateral whether before or after steps were relatively the same when the intact side was challenged (~ 0.194 m), no significant differences were found ($p>0.255$). The first prosthetic contralateral step was slightly wider than other contralateral steps (0.199 m). Despite that the all three intact contralateral after steps were wider than the intact contralateral before steps when the prosthetic side was challenged, no significant differences were found ($p>0.067$). Again, the intact contralateral after steps did not significantly differed from the prosthetic ones ($p>0.162$).

When imposing deceleration slips, all intact ipsilateral steps including before and after were faster in terms of time than the baseline, however, only the first intact ipsilateral steps were significantly different from the baseline (0.485±0.087 s, $p=0.010$). Similarly, the same step was the only step that significantly faster than the step before ($p=0.013$). For the ipsilateral prosthetic steps, the results showed that the ipsilateral prosthetic steps before were longer in time than other ipsilateral steps including baseline and the three ipsilateral after. Only the first ipsilateral steps after were significantly faster than other steps ($p<0.011$).

The two and three prosthetic contralateral steps after were faster than the contralateral steps before, the first one on the hand was longer than the steps before (0.601±0.139 s). The intact contralateral steps were relatively similar to each other, but all were faster than baseline step time apart from the first intact contralateral which was slightly slower than the baseline.

4.3.3.2.2 The acceleration slips effects

The results of gait stability and parameters for both intact and prosthetic sides are presented in (Table 4.5 and Table 4.6) below.

When the intact side was challenged by acceleration slips					
Parameters	AP (m)	ML (m)	Step length	Step width (m)	Step time (s)
I1-B	0.144±0.052	0.162±0.028	0.669±0.118	0.195±0.049	0.549±0.113
C1-B	0.127±0.066	0.156±0.036	0.655±0.151	0.188±0.040	0.578±0.161
P triggered	0.154±0.057	0.163±0.032	0.680±0.117	0.192±0.045	0.552±0.137
C1-A	0.210±0.103	0.143±0.036	0.735±0.159	0.178±0.054	0.491±0.160
I1-A	0.162±0.068	0.175±0.033	0.503±0.115	0.235±0.066	0.549±0.121
C2-A	0.107±0.072	0.178±0.041	0.634±0.148	0.232±0.046	0.581±0.141
I2-A	0.156±0.055	0.158±0.030	0.691±0.144	0.185±0.041	0.527±0.126
C3-A	0.126±0.062	0.160±0.038	0.650±0.175	0.194±0.040	0.581±0.170
I3-A	0.147±0.052	0.162±0.026	0.677±0.117	0.190±0.045	0.531±0.110

Table 4.5 Mean and standard deviations of AP MoS, ML MoS, step length, step width and step time. Parameters for both ipsilateral (I) and contralateral (C) steps one before (1-B), one after (1-A), two after (2-A) and three after (3-A) acceleration slips on the intact side. P triggered is the step that when the slips were triggered.

When the prosthetic side was challenged by acceleration slips					
Parameters	AP (m)	ML (m)	Step length	Step width (m)	Step time (s)
I1-B	0.126±0.064	0.158±0.035	0.637±0.153	0.193±0.050	0.597±0.147
C1-B	0.158±0.059	0.162±0.027	0.684±0.126	0.190±0.042	0.583±0.204
P triggered	0.127±0.064	0.157±0.038	0.653±0.152	0.189±0.043	0.552±0.135
C1-A	0.227±0.096	0.145±0.029	0.774±0.130	0.169±.0480	0.515±0.201
I1-A	0.136±0.075	0.166±0.033	0.543±0.150	0.226±0.051	0.589±0.112
C2-A	0.143±0.056	0.176±0.030	0.651±0.129	0.229±0.055	0.589±0.192
I2-A	0.141±0.070	0.156±0.038	0.653±0.171	0.193±0.053	0.567±0.170
C3-A	0.157±0.055	0.164±0.026	0.680±0.127	0.207±0.045	0.584±0.205
I3-A	0.135±0.067	0.158±0.036	0.648±0.149	0.200±0.047	0.585±0.149

Table 4.6 Mean and standard deviations of AP MoS, ML MoS, step length, step width and step time. Parameters for both ipsilateral (I) and contralateral (C) steps one before (1-B), one after (1-A), two after (2-A) and three after (3-A) acceleration slips on the prosthetic side. P triggered is the step that when the slips were triggered.

When the both intact and prosthetic sides were challenged by acceleration, mean AP MoS was increased from steps before and baseline. Mean AP MoS for the ipsilateral intact steps after the acceleration were not significantly different from the steps before (p>0.205). However, before and after intact steps were significantly larger in AP mean than baseline (p<0.020). The greater deviation from the baseline for the intact side was seen in the second ipsilateral intact step with increase of approximately (30 %). Mean AP MoS for the ipsilateral prosthetic steps after were all greater than the mean AP MoS for step before. However, only the I2-A and I3-A were significantly different from steps before (p<0.018). Similar to intact side, the prosthetic before and after steps were all significantly greater than baseline values (p<0.015).

When the intact and prosthetic side were challenged by accelerations, the contralateral steps of both sides were similar in pattern. As mean AP MoS was increased greatly for the first contralateral steps compared to steps before. When the intact side was facing slips, the first prosthetic contralateral step after was the increased by almost (65%) from the corresponding steps before. Whilst the first intact contralateral step after was also increased by about (45%)

from the steps before, which was already larger than the contralateral prosthetic steps before. This increase was followed by a decrease in MoS for both side (C2-A). The contralateral prosthetic side was decreased from the steps before by (%2) comparted to (1.5%) for the intact side. C3-A steps of both sides were similar to C1-B steps.

Compared to ipsilateral steps before, the prosthetic users initially took on average shorter steps on the first ipsilateral intact and prosthetic sides when they were challenged. Then they took longer steps on the second and third. Compared to baseline, mean step length for ipsilateral steps before, I2-A and I3-A of the intact side were greater. On the other hand, the prosthetic side ipsilateral steps were all shorter than the baseline steps.

In terms of the contralateral steps of both sides, mean step normalised length were greatly increased at first contralateral step (intact 0.774±0.130, prosthetic 0.735±0.159) from the steps before. Following this increase mean step length were decreased at C2-A before going back to steps before values at C3-A.

Compared to ipsilateral steps before, mean ML MoS for the intact and prosthetic ipsilateral steps increased at I1-A (intact by 8%, prosthetic by 5%) this increase was significantly different for both sides from the steps before ($p<0.048$). Then, the ipsilateral steps were slightly decreased at I2-A (by 1%) no significant differences were found ($p>0.146$) compared to steps before. finally, mean ML MoS of I3-A were almost equivalent to steps before. Intact and prosthetic ipsilateral steps before, I2-A and I3-A were very similar to baseline. Only I1-A steps were greater in ML MoS than baseline however, no significant differences were found between these steps ($p>0.054$). Compared to steps before, mean ML MoS for contralateral steps of both sides decreased significantly ($p<0.020$) by approximately (10%) at initial contact of C1-A.

Following this decrease, the mean ML MoS increased significantly at C2-A ($p<0.001$), before going back to be close to the steps before values at C3-A.

Prosthetic users took wider steps on both ipsilateral sides in order to keep balance. Compared to steps before the step width was significantly increased ($p<0.004$) by approximately (20% intact, 17% prosthetic) at I1-A. The intact I2-A was narrower by (5%) than the steps before. This decrease in step width, however, did not reach significant level (p>0.051). The I3-A was almost equivalent to steps before. For the prosthetic side, mean step width for the I2-A and I3-A were comparable to steps before. Whilst compared to baseline, mean step width of intact and prosthetic I1-A were significantly increased ($p<0.002$) by (26% and 20%) respectively. Mean step width of I2-A and I3-A for intact and prosthetic were greater than baseline. However, no significant differences were found among these steps ($p>0.250$). mean step width for contralateral steps decreased by approximately (intact 10 %, prosthetic 5%) from the steps before at C1-A these differences were significant only for the intact side ($p=0.001$), prosthetic ($p>0.106$). Mean step width of intact and prosthetic C2-A (intact= 0.229 m, prosthetic = 0.232 m) steps were significantly greater than the corresponding steps before ($p<0.001$). Intact and prosthetic C3-A step width were greater than steps before, although these differences did not reach significant level ($p>0.070$).

When the intact side was challenged by acceleration, mean step time for I1-A were profoundly equal to I1-B (0.549 s), the variability was slightly greater (±0.121 s). Mean step time for I2-A and I3-A were less than the steps before but not significantly different ($p>0.305$). Comparing to baseline steps, mean step time of all steps were less than baseline (0.555±0.069 s). However, these differences did not reach significant level ($p>0.118$). For the prosthetic side, mean step time of I1-A, I2-A and I3-A were slightly less but not significantly different from prosthetic steps before (p>0.317). mean step time for the contralateral steps of both sides significantly decreased from corresponding steps before only at C1-A (p<0.001) as C2-A and C3-A were remarkably similar to steps before.

4.3.4 The effects of the prosthetic ankle mechanism

The results for AP and ML directions along with gait parameters for the prosthetic side steps of the ankle and no ankle groups are detailed in Table 4.6. The no ankle group included: Vari-Flex and Soleus feet whilst the ankle group included: Elan, Echelon VT, Pro-Flex. The results were represented when participant walked over two conditions: 1) Baseline where participants walked at a speed of (Ankle = 0.988±0.388 m/s, No ankle =1.193±0.168 m/s). 2) Self-paced mode where participants walked at a speed of (Ankle =1.178±0.3818 m/s, No ankle =1.303±0.0680 m/s).

As group vs group, overall, no significant differences were found between the prosthetic users fitted who were fitted with prosthetic ankle and the group with no ankle for all parameters of the two walking conditions (p>0.236). The prosthetic users with no ankle exhibited less variability among all parameter than the users with active prosthetic ankle.

During both walking conditions, the mean AP MoS at the prosthetic initial contact was greater for people with no ankle (0.124 m). Similarly, the MoS in the ML direction was greater for the same group (0.166 m).

The participants who were no fitted with prosthetic ankle took longer steps on with the prosthetic side than the people with ankle during the both walking conditions. In addition, it was noted that the variability of the step length of the ankle group was significantly greater than the no ankle group during both conditions, the differences were more clearer during the self-paced mode (0.076). During the baseline, the group with no ankle took wider steps (0.182 m). Both groups showed on average the same step width but differed in variability during the self-paced mode walking.

The prosthetic steps of the no ankle group were faster in time than the group with ankle during both walking conditions.

Trial	Prosthetic ankle (yes/no)	AP MoS Mean±SD (m)	ML MoS Mean±SD (m)	Step length Mean±SD	Step width Mean±SD (m)	Step time Mean±SD (s)
Baseline	Yes	0.092±0.057	0.157±0.055	0.614±0.208	0.178±0.057	0.614±0.157
	No	0.124±0.023	0.166±0.036	0.695±0.113	0.182±0.035	0.541±0.035
	p value	0.236	0.760	0.422	0.893	0.313
SP	Yes	0.126±0.070	0.149±0.049	0.706±0.219	0.173±0.055	0.564±0.097
	No	0.134±0.026	0.167±0.038	0.762±0.076	0.173±0.040	0.534±0.028
	p value	0.820	0.589	0.575	0.990	0.476

Table 4.7 Mean ± SD of MoS in both AP and ML directions along with gait parameters for the prosthetic side steps of the ankle and no ankle groups. The baseline was the trial where no slips were applied. SP, where participants waked at self-paced mode. Significant differences at p <0.05.

The results of individuals feet during baseline walking are presented in (Figure 4.1 and Figure 4.2)

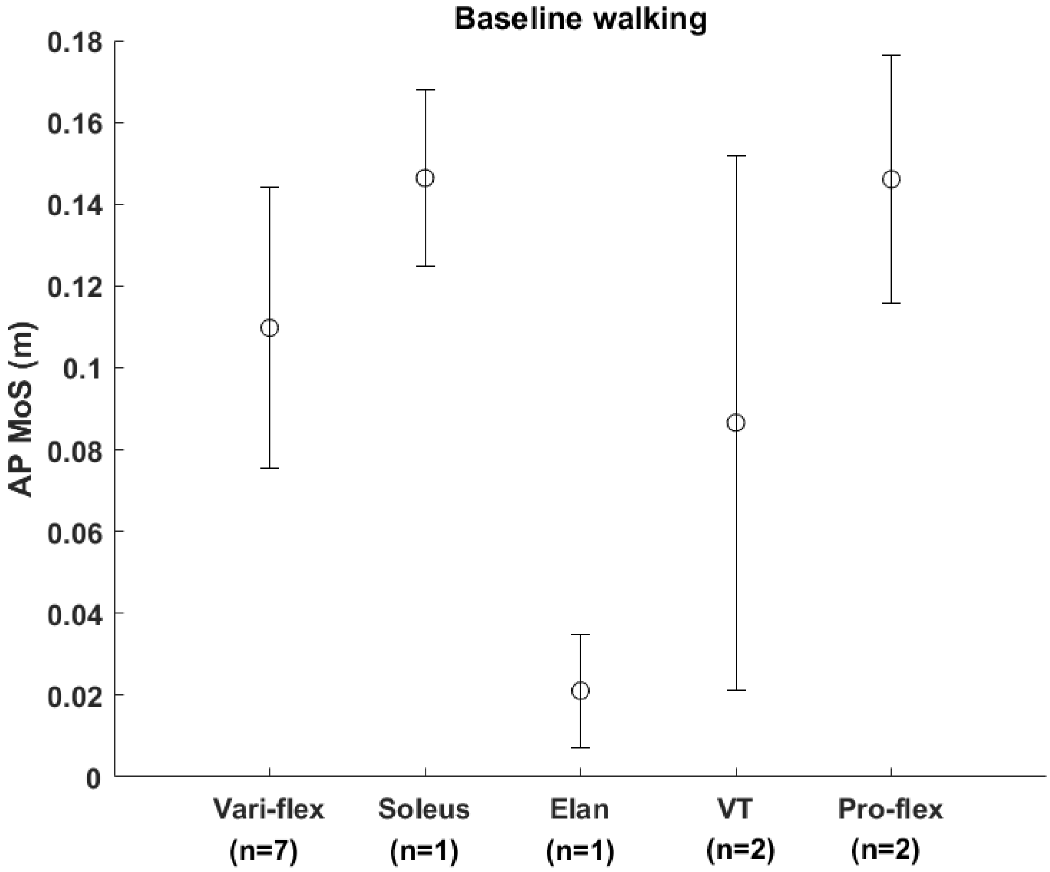

Figure 4.1 MoS in AP direction for each participant prosthetic feet when walking at CWS. (o) represent mean values. Error bars represent between-subject standard deviation.

AP MoS mean and variability at initial contacts of Elan were the smallest among all other feet (0.021±0.014 m). Whilst AP MoS mean for Soleus and Pro-Flex were the greatest (0.15 m). between-subject variability of Echelon VT was the greatest (±0.06 m) among the prosthetic feet. Vari-Flex feet showed relatively improved AP MoS mean.

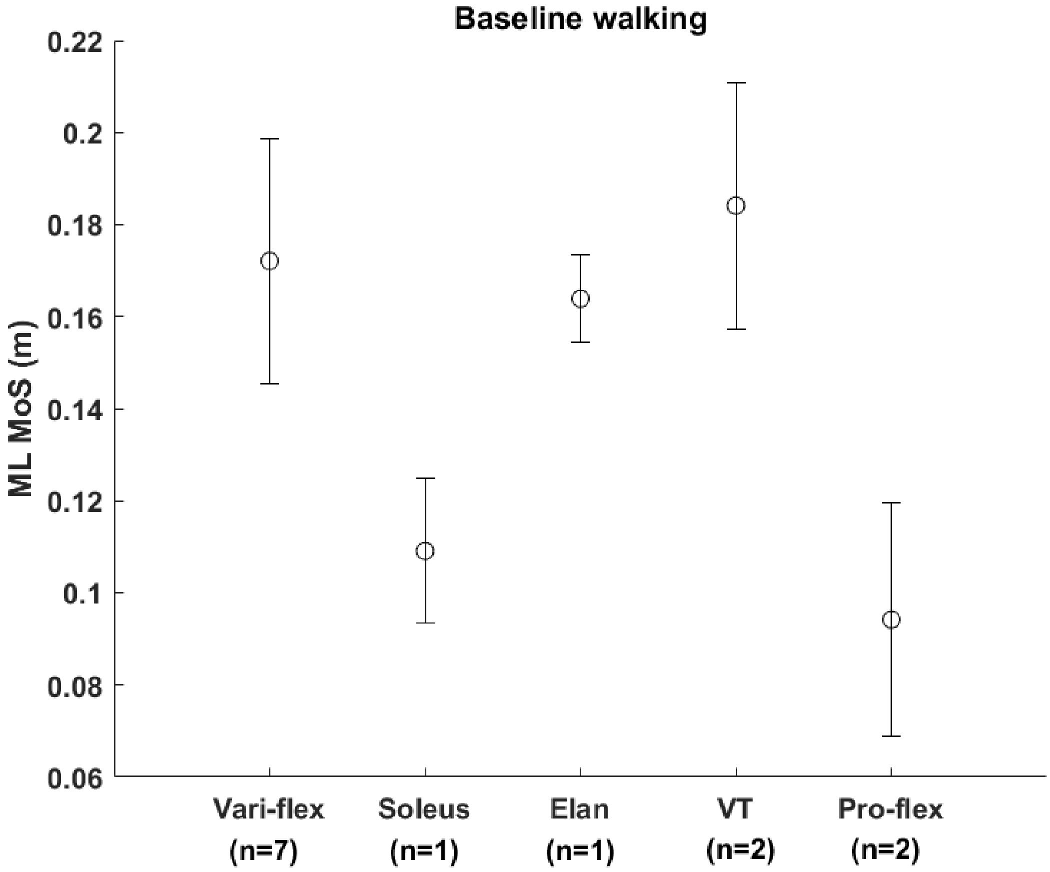

Figure 4.2 MoS in ML direction for each participant prosthetic feet when walking at CWS. (o) represent mean values. Error bars represent between-subject standard deviation.

Unlike the MoS in the AP direction, mean ML MoS for Elan (0.16 m) and Echelon VT (0.18) were improved compared to other feet. Mean ML MoS for Soleus was the second smallest among the prosthetic feet (0.10 m). The results of the Vari-Flex were also comparable to Elan and VT. Mean ML MoS for the Pro-Flex were the smallest (0.095 m).

4.3.4.1 **The effect of the prosthetic foot type in response to slips**

(Figure 4.3 and Figure 4.4) below provide overview results of each prosthetic foot type in response to deceleration slips.

In general, the prosthetic three ipsilateral steps after of all feet were relatively greater than ipsilateral steps before as well as baseline. In addition, compared to baseline values, prosthetic ipsilateral before were at least (20%) relatively greater except Soleus whereas the baseline mean value (0.146 m) were greater than the prosthetic ipsilateral steps before (0.100 m). Overall, mean AP MoS for Pro-Flex steps foot were greater than other steps. Whilst Elan foot mean AP MoS were the smallest. Mean AP MoS for prosthetic intact for C1-A steps at which the perturbation yet was not finished, were decreased compared to steps before for all feet except Vari-Flex and Echelon VT intact C1-A in which the mean was equal to steps before.

When Vari-Flex was challenged, AP MoS variability of both prosthetic and intact sides were utterly maintained indicating the ability of miniating balance thought the slips. Unlike Soleus feet in which the variability has remarkably increased especially for the intact side. The results of Vari-Flex showed that the prosthetic side mean AP MoS were always less than the intact side.

Mean AP MoS results for Echelon VT feet were relatively similar for the prosthetic and intact sides except the C1-A at which the MoS was decreased significantly.

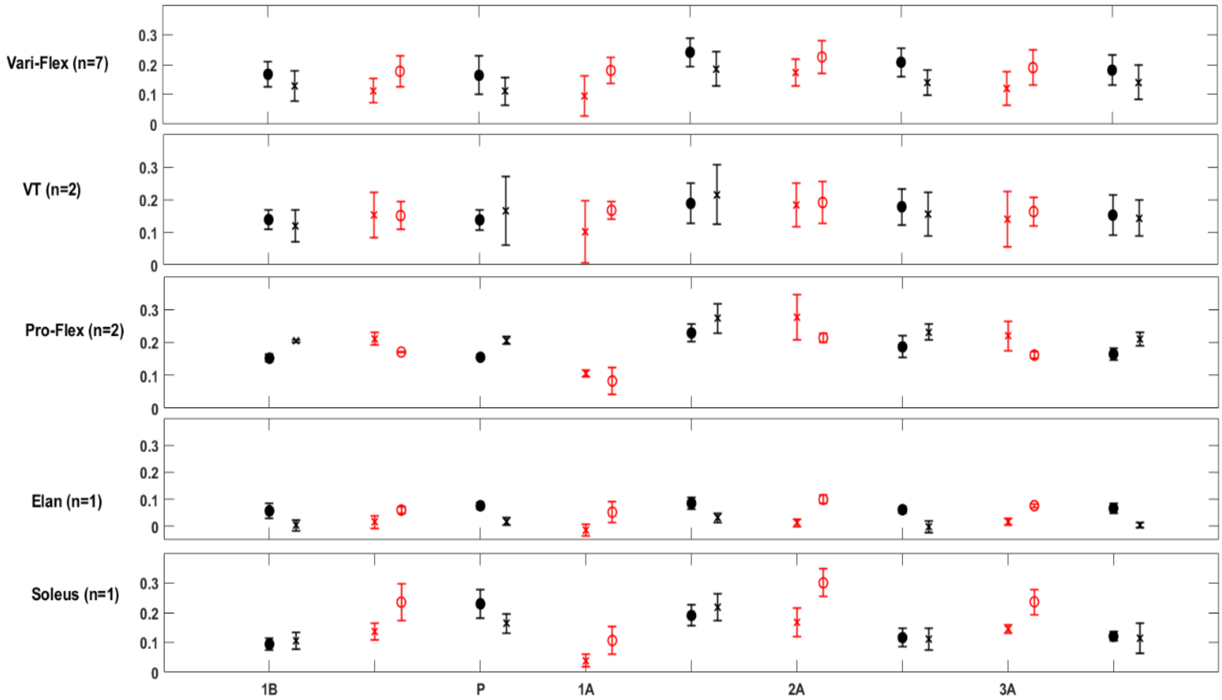

Figure 4.3 Mean and standard deviations of AP MoS at initial contact for each foot when imposing the deceleration slips. (●) represent mean values of intact sides. (x) represent mean values of prosthetic sides. Error bars represent between-subject standard deviation. The black error bars represent ipsilateral steps whilst the red error bars represent contralateral steps. (B): steps before, (**P**) triggered steps and (A) steps after.

Pro-Flex mean AP for the prosthetic I1-B were the greatest among all feet (0.200 m), in addition the variability was the smallest among other feet (± 0.003 m). It was noted that mean AP MoS of all prosthetic steps before and after were greater than the comparable intact sides. Mean MoS for prosthetic I1-A was increased by approximately (35%) from the corresponding steps before. the prosthetic I2-A were also increased from the steps before (20%) after this the I3-A returned to the steps before status. The intact sides steps followed the same pattern.

Mean AP MoS results for Elan feet were the smallest among the feet particularly for the prosthetic side. There was less involvement from the prosthetic side to recover the AP MoS, as when comparing the prosthetic ipsilateral steps before to steps after no remarkable change in the mean AP MoS was seen. On the other hand, mean AP MoS for the intact side steps after increased from steps before. In addition to changes in mean, an increase in AP MoS variability for the intact side was also seen. Another observation was that during the deceleration slips, mean AP MoS for prosthetic C1-A was significantly reduced to be (- 0.013 m) which was an indicator of instability.

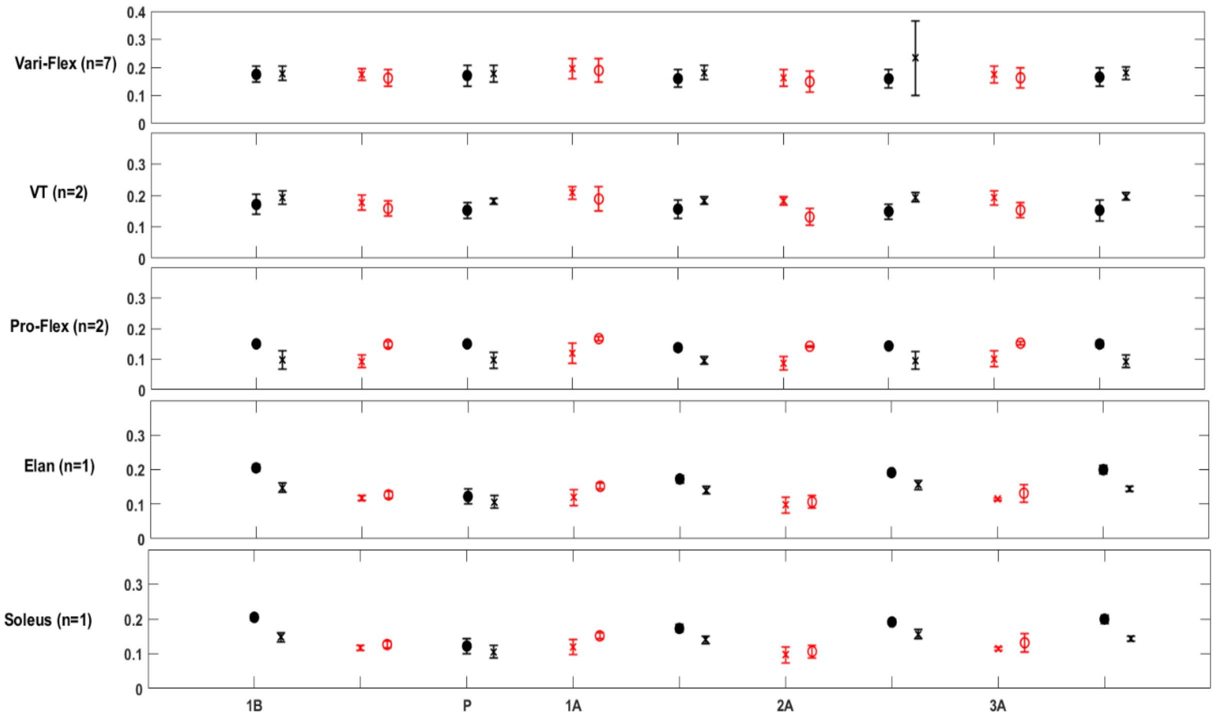

Figure 4.4 Mean and standard deviations of ML MoS at initial contact for each foot when imposing the deceleration slips. (●) represent mean values of intact sides. (x) represent mean values of prosthetic sides. Error bars represent between-subject standard deviation. The black error bars represent ipsilateral steps whilst the red error bars represent contralateral steps. (B): steps before, (**P**) triggered steps and (A) steps after.

In terms of the MoS in the ML direction when the prosthetic users were challenged by deceleration, mean ML MoS for Echelon VT feet were the greatest among all feet (approximately 0.19 m) which were similar to baseline walking. In addition, mean ML MoS for the prosthetic sides were greater than the intact sides. This was also seen for Elan prosthetic side steps. Comparable to baseline values, mean ML MoS for Pro-Flex prosthetic steps were the smallest among all other feet (<0.01 m).

When comparing the prosthetic ipsilateral before and after steps to baseline results, no remarkable change in mean ML MoS were noticed among all feet apart from Soleus foot. However, the MoS variability in this direction has been altered.

Same as in AP direction, MoS variability in ML for the Vari-Flex was maintained throughout the steps apart from the second prosthetic ipsilateral step after which the variability was increased remarkably from the steps before. Additionally, mean ML MoS for Vari-Flex were very comparable to the performance of Echelon VT as the overall mean was (approximately 0.175 m).

It was noticed that ML MoS variability of the intact side for the Pro-Flex was equal throughout the steps (approximately ±0.008 m). Compared to baseline results, mean MoS for the steps were greatly equal (~ 0.09 m).

Elan results showed that the ML MoS variability of the intact side was increased for all steps after compared to steps before. The opposite was seen for the prosthetic side which the variability was decreased. Mean steps for prosthetic steps were around (0.16 m).

ML MoS variability for the intact side were also increased for the steps after compared to steps before when Soleus foot was challenged. Mean ML MoS for prosthetic ipsilateral steps were increased by at least (45%) from the baseline steps.

The results of MoS for each foot in both AP and ML directions when the participants were challenged by acceleration are presented in (Figure 4.5 and Figure 4.6).

Unlike the deceleration, in response to acceleration slips inconsistent mechanism among the feet were found; both Elan and Soleus reacted the same as both ipsilateral prosthetic steps AP MoS were deceased compared to baseline. Elan ipsilateral steps were greatly decreased approximately (90-100%) less than the baseline values. Whilst the decrease in Soleus from the baseline ranged between (%35 and 40%). For the other feet, the same steps were increased from the baseline. Vari-Flex steps increased by (20-30%), Echelon VT mean AP MoS increased between (20-40%), and Pro-Flex steps were the greatest increase from the baseline by (60-70%).

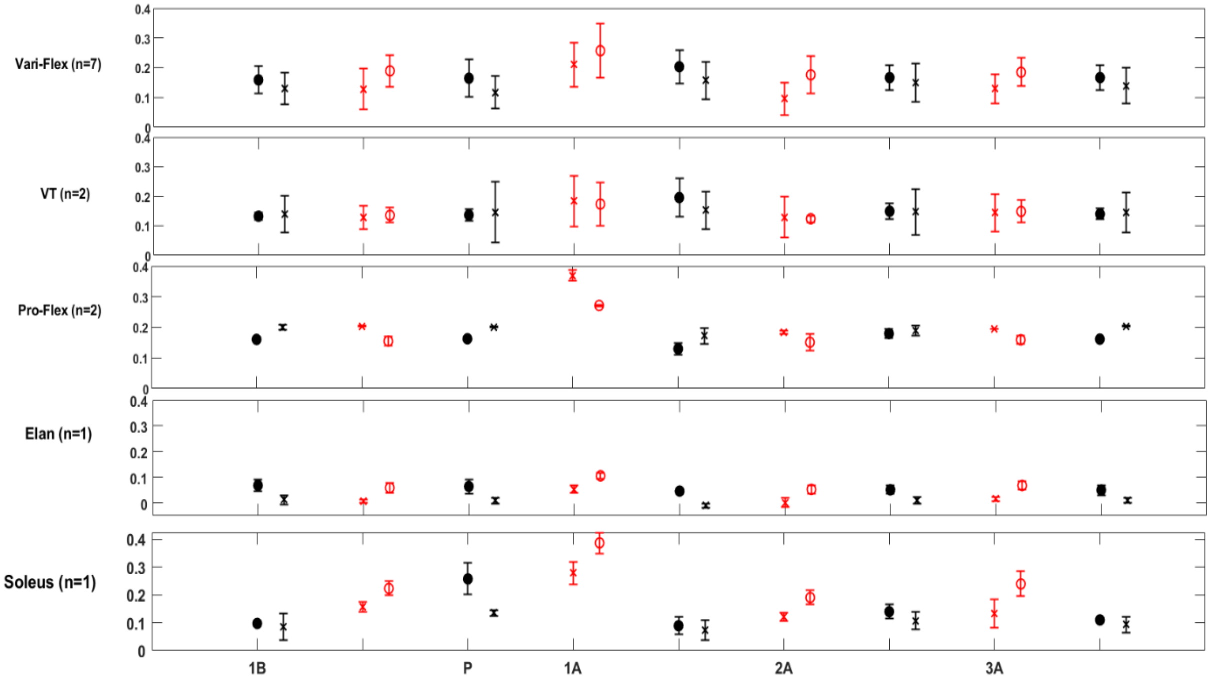

Figure 4.5 Mean and standard deviations of AP MoS at initial contact for each foot when imposing the acceleration slips. (●) represent mean values of intact sides. (x) represent mean values of prosthetic sides. Error bars represent between-subject standard deviation. The black error bars represent ipsilateral steps whilst the red error bars represent contralateral steps. (B): steps before, (**P**) triggered steps and (A) steps after.

Overall ML MoS results in response to acceleration were very similar to the results in response to deceleration (Figure 4.6), the results were slightly more affected by deceleration than acceleration slips. ML MoS mean and variability for Vari-Flex feet were also maintained throughout the steps approximately (0.16±0.03 m). Mean ML MoS for the prosthetic side were greater than intact sides for both Echelon VT and Elan feet. Echelon VT results were also the greatest among the feet. Meanwhile the lowest results were seen for Pro-Flex feet.

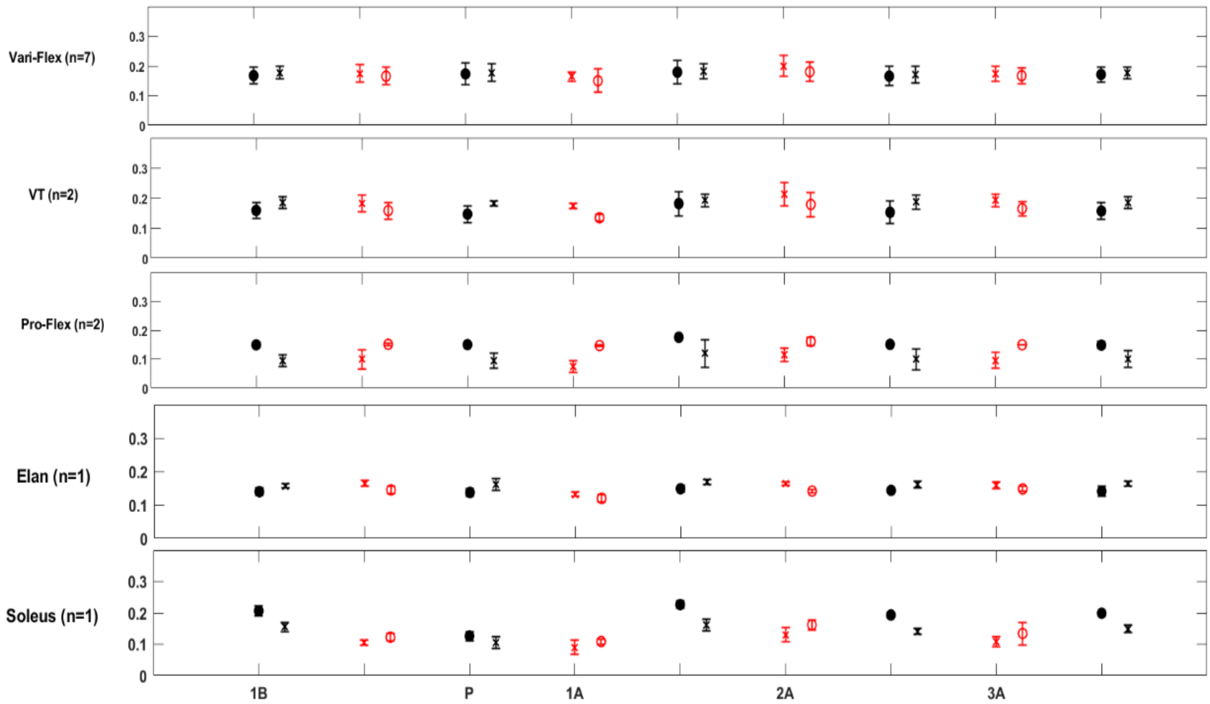

Figure 4.6 Mean and standard deviations of ML MoS at initial contact for each foot when imposing the acceleration slips. (●) represent mean values of intact sides. (x) represent mean values of prosthetic sides. Error bars represent between-subject standard deviation. The black error bars represent ipsilateral steps whilst the red error bars represent contralateral steps. (B): steps before, (P) triggered steps and (A) steps after.

The results of the Ankle disarticulation (Syme's) prosthetic uses (n=2) are presented in (Figure 4.7, Figure 4.8 and Figure 4.9).

Ankle disarticulation prosthetic users showed comparable AP and ML MoS results for the prosthetic and intact sides during bassline (Figure 4.7). AP MoS results for intact and prosthetic sides were (0.120 ±0.03 m) and (0.11 ± 0.45 m) respectively which were similar to the results of participants with Vari-Flex feet (Figure 4.1). Meanwhile, ML MoS results for intact and prosthetic sides were (0.180±0.02 m) and (0.150 ± 0.04 m). ML results were only greater than Soleus and Pro-Flex.

The results of the ankle disarticulation prosthetic users when they were challenged by deceleration followed the same trend similar to other feet. For the AP direction the balance was maintained by increasing the MoS of the first ipsilateral steps and second contralateral steps after deceleration (Figure 4.8). The first prosthetic ipsilateral step after were greater than the matched intact steps which was similar to the pattern of users fitted with Pro-Flex and Soleus (Figure 4.3). AP MoS of both sides were greater than steps before even after three steps. The variability of first contralateral intact steps seemed to be greater than the matched prosthetic steps, indicating that the intact sides were affected by the deceleration more than the prosthetic side.

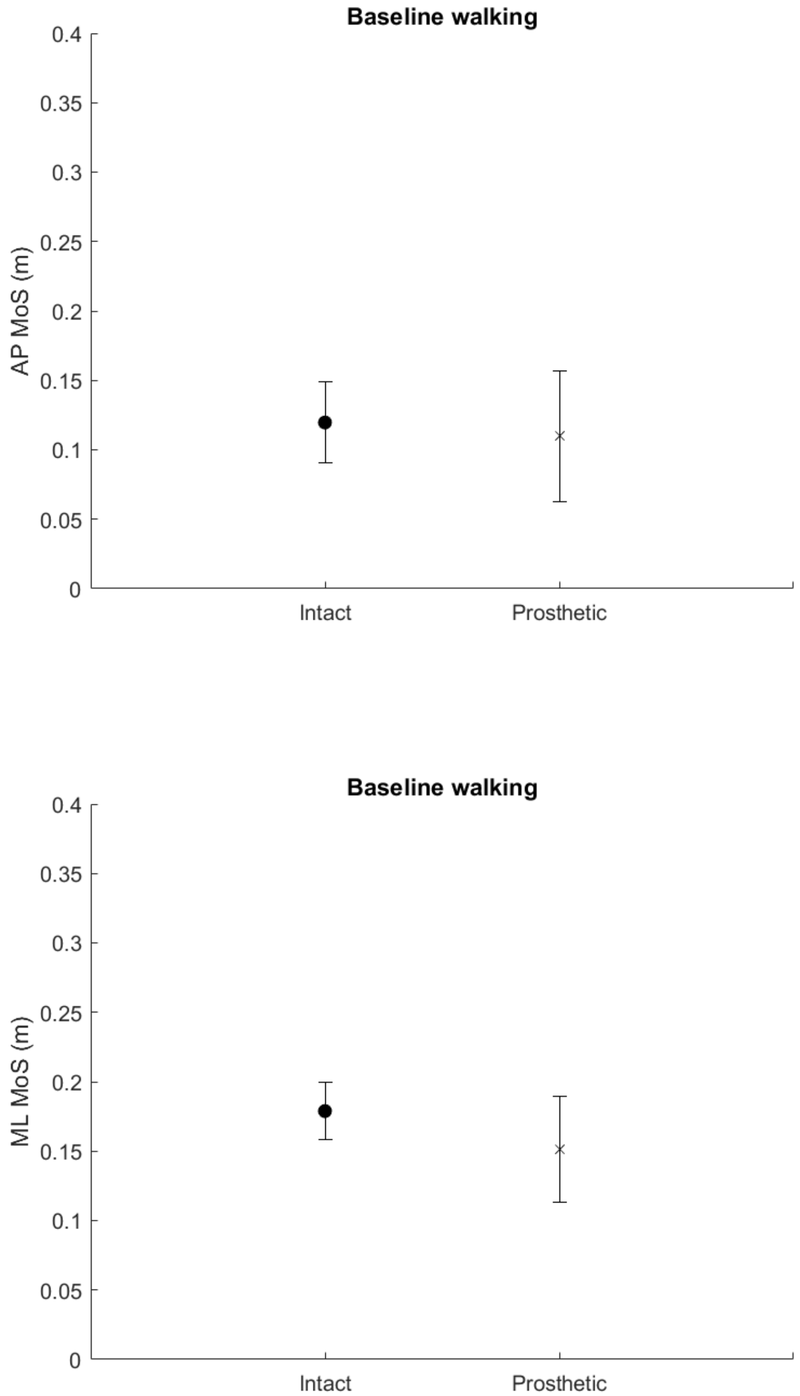

Figure 4.7 MoS in AP and ML directions for Ankle disarticulation prosthetic users (n=2) during baseline walking. (●) represent mean values for intact side whilst (x) represent mean values for prosthetic side. Error bars represent between-subject standard deviation.

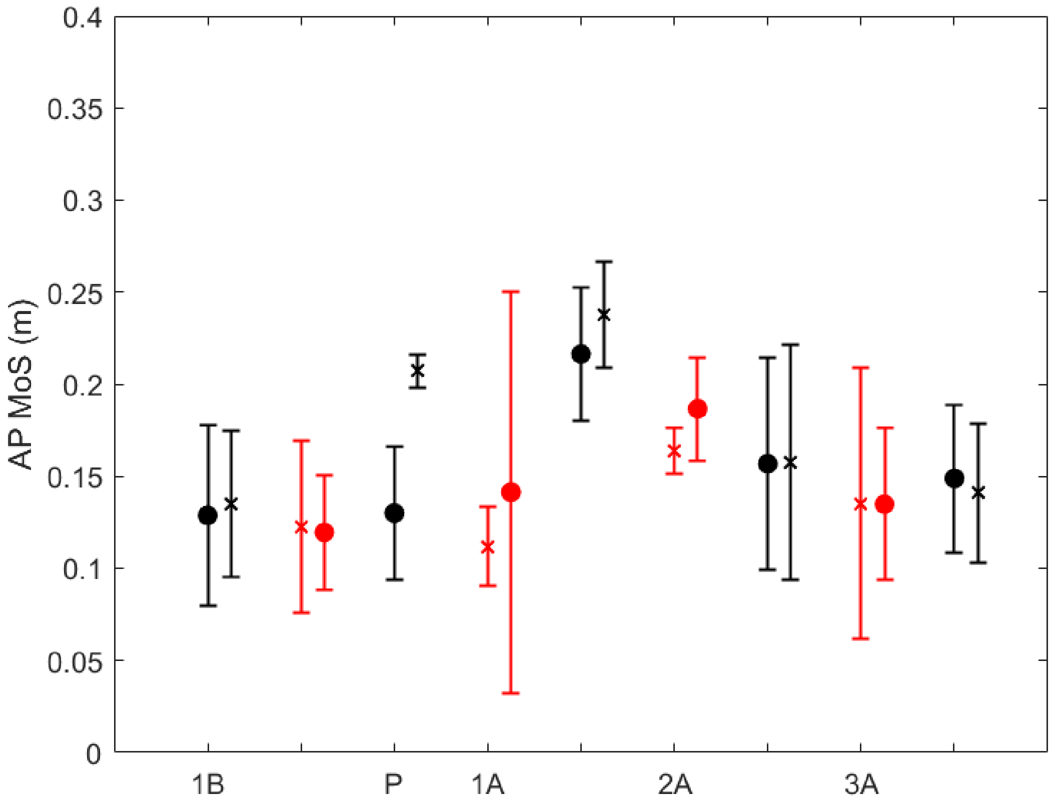

Figure 4.8 Mean and standard deviations of AP MoS at initial contact for ankle disarticulation prosthetic users (n=2) when imposing the deceleration slips. (●) represent mean values of intact sides. (x) represent mean values of prosthetic sides. Error bars represent between-subject standard deviation. The black error bars represent ipsilateral steps whilst the red error bars represent contralateral steps. (B): steps before, (**P**) triggered steps and (A) steps after.

For the ML direction, mean and variability of the MoS for intact side were always greater than prosthetic side. Compared to intact steps before (0.19 ± 0.02 m), the ML MoS for the intact ipsilateral steps after decreased, the most deviation from the steps before was at the first intact ipsilateral steps (0.16±0.18 m). Similar pattern was noted but with less degree for the prosthetic side, as prosthetic ipsilateral steps before (0.15±0.007 m) where slightly greater than ipsilateral steps after approximately (0.145±0.003 m). The ML MoS results for ankle disarticulation prosthetic users were comparable to the results of Echelon VT. However, here the prosthetic sides results were always less than the intact which opposite to those feet where the prosthetic side results were greater than intact (Figure 4.4).

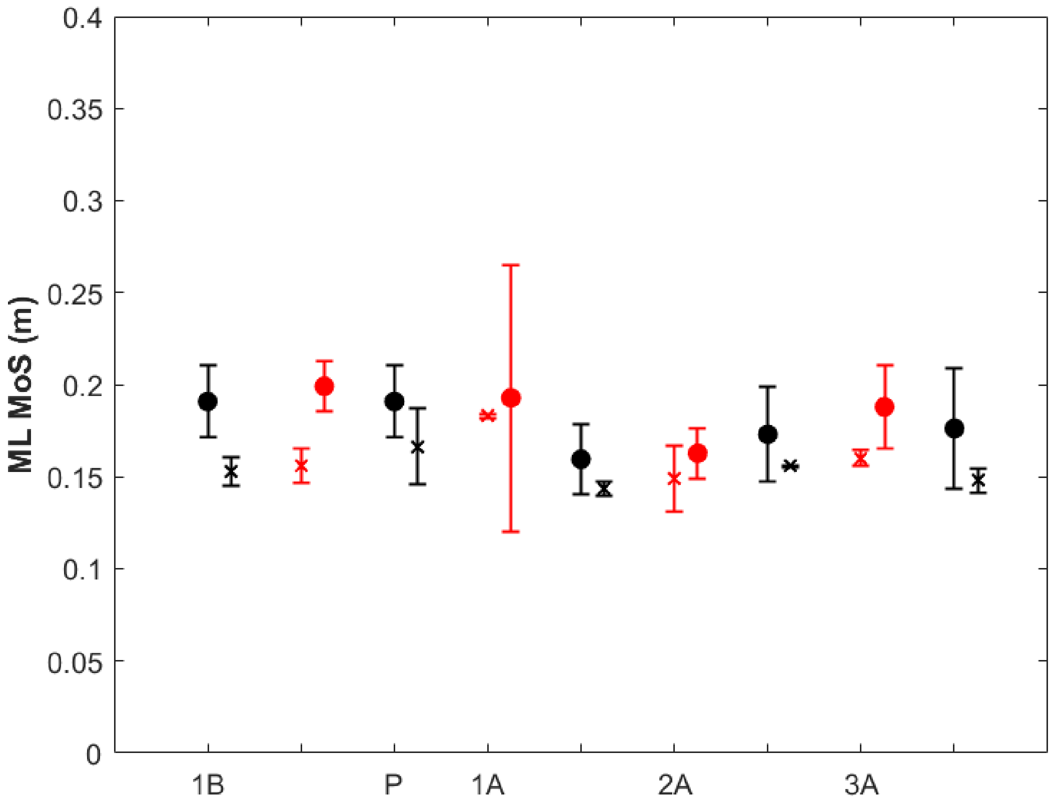

Figure 4.9 Mean and standard deviations of ML MoS at initial contact for ankle disarticulation prosthetic users (n=2) when imposing the deceleration slips. (●) represent mean values of intact sides. (x) represent mean values of prosthetic sides. Error bars represent between-subject standard deviation. The black error bars represent ipsilateral steps whilst the red error bars represent contralateral steps. (B): steps before, (**P**) triggered steps and (A) steps after.

4.3.5 The differences between able-bodied and prosthetic users

The differences in gait stability and stepping mechanisms were investigated between the prosthetic users as a group vs the able-bodied subjects.

4.3.5.1 Baseline and Self-pace mode walking

On average, able-bodied participants walked faster than the prosthetic users for the two walking conditions baseline and self-paced mode. The control group during baseline walked at (1.2±0.11 m/s) and walked at (1.3±0.13 m/s) during self-paced mode. Whilst the prosthetic users walked at (1.08±0.254 m/s) during baseline and walked at (1.20± 0.31) during SP mode. However, no statistically significant differences were found in walking speed between control and prosthetic users groups for the both walking conditions (baseline: $p = 0.103$, SP: $p = 0.138$). Nonetheless, the between-subjects' differences were greater for the prosthetics users group. Figure 4.5 below shows mean and standard deviations of MoS and gait parameters of able-bodied and prosthetic users groups during baseline walking.

During the baseline, on average prosthetic users exhibited smaller AP MoS on both intact and prosthetic side than the able-bodied. Mean AP MoS at initial contact for prosthetic side were significantly lower than the able-bodied left and right sides ($p<0.036$), the AP MoS variability were greater on the prosthetic sides. The intact side, however, did not significantly differ in means and variability from the left and right sides in AP direction ($p>0.480$). The minimum AP mean MoS among all sides was (0.02 m) and it was seen for the prosthetic side. The maximum AP mean MoS was (0.24 m) and it was seen for the left, right and intact sides. Whilst the maximum AP MoS for the intact side was (0.17 m).

For the ML MoS, the prosthetic users group showed greater mean and variability on both sides compared to able-bodied. Regardless, mean ML MoS of the prosthetic and intact sides were only significantly greater than the left side ($p<0.002$).

In terms of the gait spatiotemporal parameters, the prosthetic users took shorter steps than the able-bodied. Also, the variability of step length for both prosthetic and intact side were significantly greater than the right and left side of the able-bodied ($p<0.020$). Nonetheless, no significant differences were found in means between the prosthetic users and the able-bodied ($p>0.215$).

The prosthetic users exhibited wider steps at the initial contacts on both sides compared to able-bodied. The mean step width for prosthetic and intact sides were significantly greater than left and right sides ($p<0.001$).

Overall, for the step time, the prosthetic users walked slower than able-bodied. The prosthetic side was the longest in time than other steps, however no statistical differences were found in mean and variability when comparing the prosthetic and intact sides to the right and left ($p>0.200$).

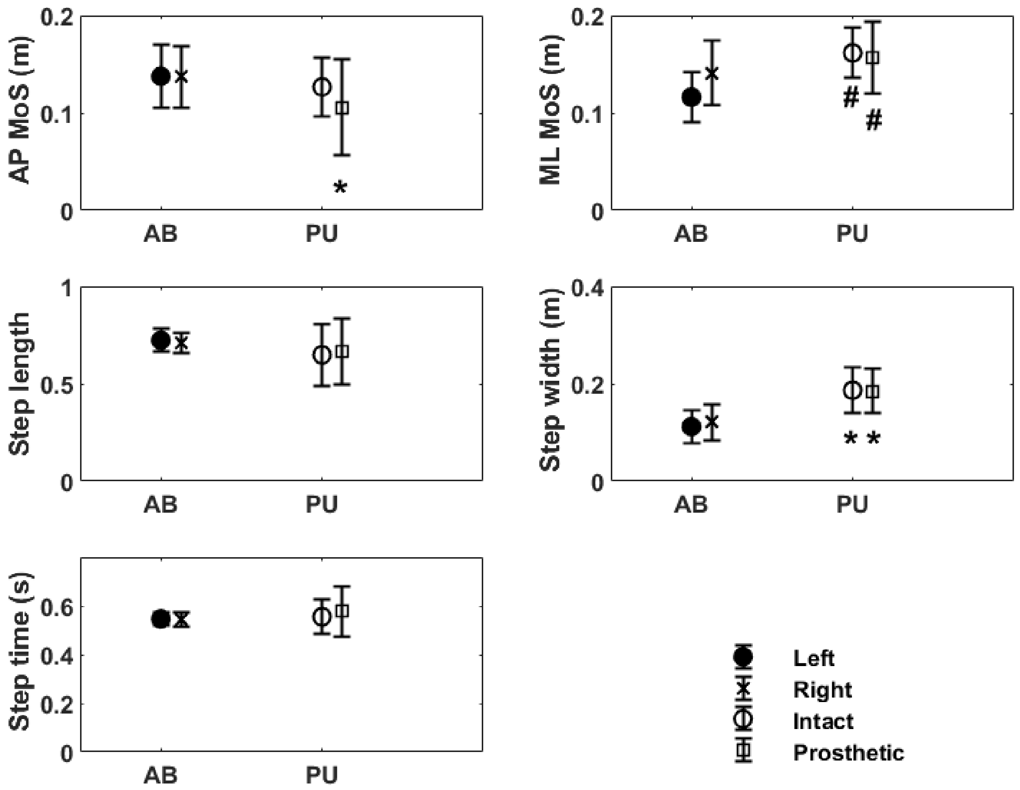

Figure 4.10 Mean and standard deviations of MoS and gait parameters of able-bodied (**AB**) and prosthetic users (**PU**) groups during baseline walking. (●) represent mean values of left sides. (**x**) represent mean values of right sides. (**o**) represent mean values of intact sides and (□) represent mean values of prosthetic sides. Error bars represent between-subject standard deviation. (*) indicate significant differences in mean from control left and right sides. (#) indicate significant differences in mean from control left side only. Significant differences at $p < 0.05$

Figure 4.11 below shows mean and standard deviations of MoS and gait parameters of able-bodied and prosthetic users groups during self-paced mode walking. The MoS and gait parameters were similar to the baseline walking. Mean AP MoS of prosthetic and intact sides were smaller than the left and right. But, unlike the baseline, mean AP MoS of prosthetic side were also not significantly different from the left and right sides ($p>0.091$).

Mean ML MoS for prosthetic and intact sides were also greater than the left and right side and were only significantly different from the left side ($p<0.004$).

Mean normalised step length at initial contact of prosthetic and intact sides were smaller (i.e., shorter steps) than the left and right side. No significant mean differences were found ($p>0.318$), however, the variances of the two groups were not equal ($p<0.006$).

Mean step width of both prosthetic and intact sides were significantly greater than the left and right sides ($p<0.001$). Mean step time for prosthetic users were smaller than able-bodied group ($p>0.281$).

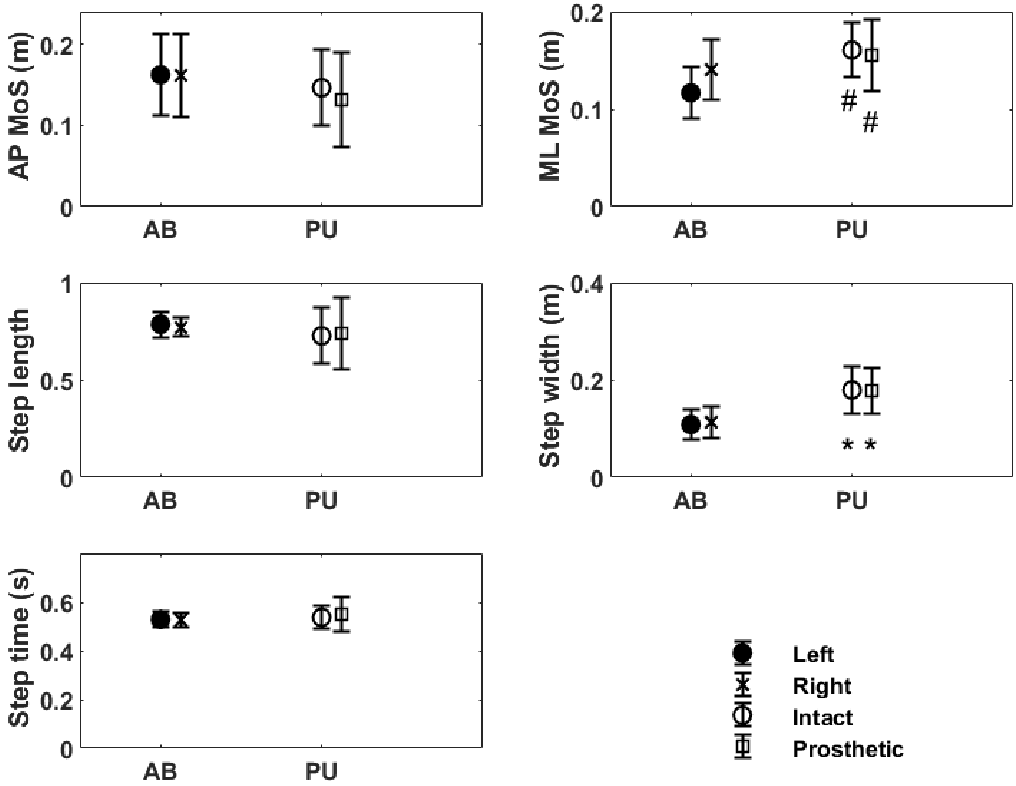

Figure 4.11 Mean and standard deviations of MoS and gait parameters of able-bodied (**AB**) and prosthetic users (**PU**) groups during self-paced walking. (●) represent mean values of left sides. (**x**) represent mean values of right sides. (**o**) represent mean values of intact sides and (□) represent mean values of prosthetic sides. Error bars represent between-subject standard deviation. (*) indicate significant differences from control left and right sides. (#) indicate significant differences from control left side only. Significant differences at p <0.05.

4.3.5.2 **The effect of deceleration slips**

The AP results showed that the response to the deceleration slips was similar in pattern for both groups. It was noted for the all steps, the variability of the prosthetic steps was the greatest among all steps (>0.06 m). Figure 4.12 below shows the mean and standard deviations of AP MoS for able-bodied and prosthetic users when facing deceleration slips.

The ipsilateral prosthetic steps before were the smallest mean MoS among all other corresponding steps (0.125 m). However, these steps were only significantly smaller in mean than the left side (p=0.014). When the deceleration slips were imposed on any given side, it was noted that the contralateral prosthetic steps mean AP MoS was also the smallest mean (0.124 m). Similarly, these were only significantly smaller in mean than the left side (p=0.006). The mean AP MoS for first ipsilateral steps after for all sides followed the same pattern, all mean AP MoS were increased from the baseline and before steps. The prosthetic and left I1-A were the most deviated from the baseline mean AP MoS by approximately 95%. The intact and right sides were deviated by 70% and 60% respectively. When comparing to the steps before, the prosthetic I1-A was the most deviated steps by 55%. The intact side on the other hand, was the least deviated steps by 40%. Mean AP MoS for the prosthetic and intact I1-A steps were significantly smaller only than left side (p<0.023), no significant differences in mean were found when the prosthetic and intact I1-A were compared to mean AP MoS for the right I1-A (p>0.325). The first contralateral prosthetic and intact steps after the slips were similar in behaviour to the corresponding left and right steps. But the prosthetic and intact steps were less deviated from the steps before than the left and right corresponding steps. The control group was able to return to the mean AP MoS before the slips by second ipsilateral steps, they I2-A and I3-A were similar to the steps before. On the contrary, the prosthetic users were not able to return to the step before even after three ipsilateral steps. Mean AP MoS for the prosthetic ipsilateral steps were elevated. The prosthetic and intact steps after (I2-A and I3-A) were also smaller in mean than the left and right, however, these differences did not reach a significant level (p>0.069). The same pattern was also seen for the intact contralateral two and three steps after (p>0.114). the prosthetic contralateral two and three steps after were significantly smaller from only the left contralateral steps (p<0.025).

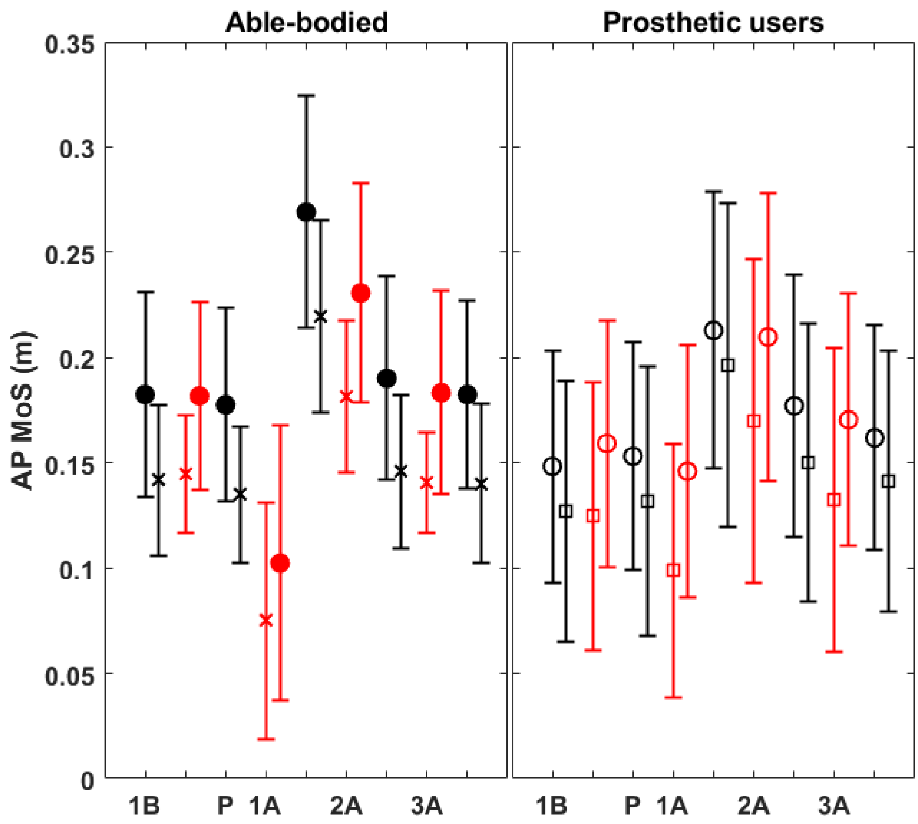

Figure 4.12 Mean and standard deviations of AP MoS at initial contact for able-bodied and prosthetic users groups when imposing the deceleration slips. (●) represent mean values of left sides. (x) represent mean values of right sides. (o) represent mean values of intact sides and (□) represent mean values of prosthetic sides. Error bars represent between-subject standard deviation. The **black error bars** represent ipsilateral steps (sides the slips were applied on) whilst the **red error bars** represent contralateral steps (when slips were applied on the other side). (**B**): steps before deceleration, (**P**) triggered steps when the slips were triggered and (**A**) steps after deceleration.

On average, prosthetic users exhibited greater mean ML MoS at initial contact than the able-bodied for ipsilateral and contralateral steps before and after. Figure 4.13 shows the mean ML MoS for both groups in respond to the deceleration slips.

Similar to the baseline and SP walking, the intact and prosthetic ipsilateral steps including before and after steps were only significantly greater than the left side ($p<0.003$). Both groups showed approximately similar ipsilateral steps before mean ML MoS compared to the baseline steps mean. The first ipsilateral steps mean MoS of both were nearly significantly greater than the right corresponding steps ($p=0.070$). In addition, no significant differences were found between prosthetic users first contralateral mean ML MoS and similar steps of the able-bodied group ($p>0.060$). The second and third intact and prosthetic contralateral steps, however, were only significantly greater than left side ($p<0.001$). It was noted that the variability of the prosthetic I2-A was the greatest among all other steps (0.094 m).

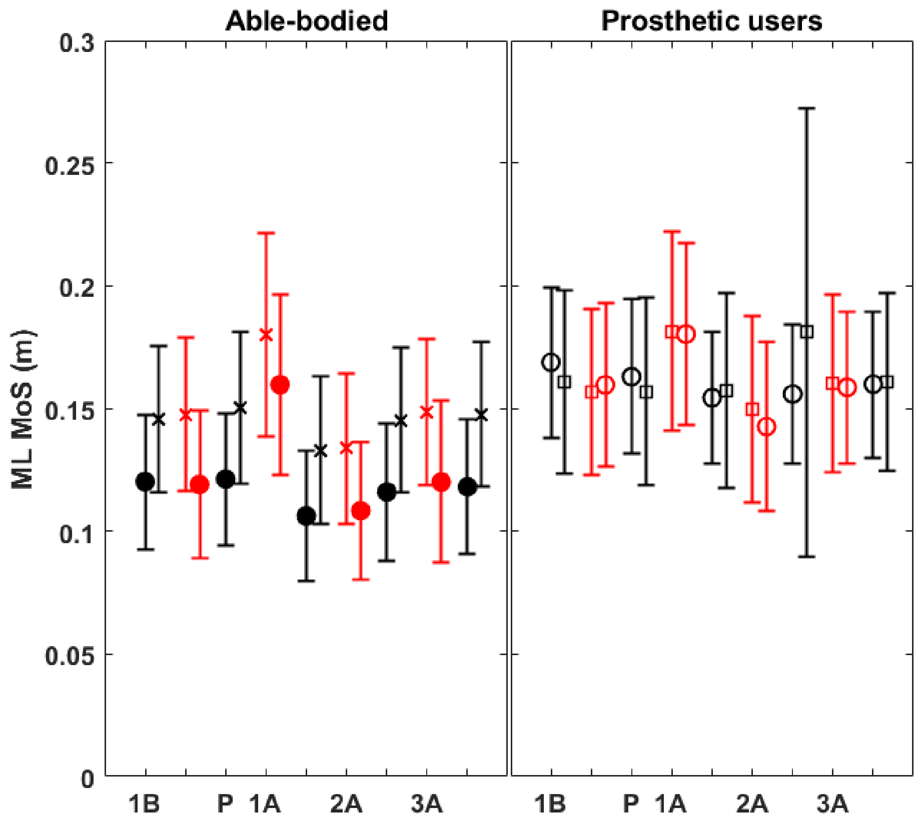

Figure 4.13 Mean and standard deviations of ML MoS at initial contact for able-bodied and prosthetic users groups when imposing the deceleration slips. (●) represent mean values of left sides. (x) represent mean values of right sides. (o) represent mean values of intact sides and (□) represent mean values of prosthetic sides. Error bars represent between-subject standard deviation. The **black error bars** represent ipsilateral steps (sides the slips were applied on) whilst the **red error bars** represent contralateral steps (when slips were applied on the other side). (**B**): steps before deceleration, (**P**) triggered steps when the slips were triggered and (**A**) steps after deceleration.

In terms of the step length, the prosthetic users overall took shorter steps than the able-bodied across all steps. Figure 4.8 provides the mean and standard deviations of normalised step length at initial contact for able-bodied and prosthetic users groups when imposing the deceleration slips.

The shortest steps were mostly seen for the prosthetic side. The shortest step, however, was the first contralateral intact step. In addition, the prosthetic users showed greater step length variability; the prosthetic step length variability was the greatest among all other steps. The able-bodied tended to show less step length asymmetry. There were no significant differences in mean step length between prosthetics users and able-bodied, for baseline, SP and steps before perturbations.

Mean step length for the first three ipsilateral steps were all significantly longer for the able-bodied than the prosthetic side ($p<0.043$). When comparing prosthetic users' steps with the perturbations applied to the intact side, only the second intact steps were significantly shorter than for able-bodied ($p=0.042$). Despite the fact that the contralateral prosthetic steps were shorter, they were not significantly different to the able-bodied contralateral steps ($p>0.090$). However, when the perturbations were applied to the prosthetic side, there were significant differences between the first step on the intact side and the able-bodied first step ($p< 0.009$).

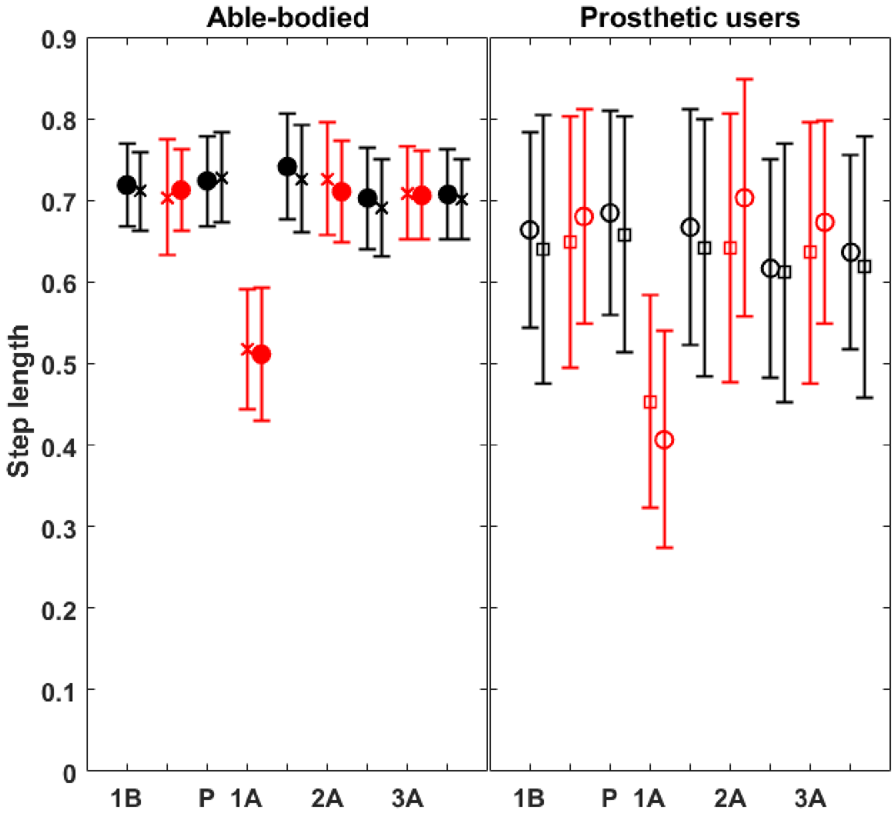

Figure 4.14 Mean and standard deviations of normalised step length at initial contact for able-bodied and prosthetic users groups when imposing the deceleration slips. (●) represent mean values of left sides. (x) represent mean values of right sides. (o) represent mean values of intact sides and (□) represent mean values of prosthetic sides. Error bars represent between-subject standard deviation. The **black error bars** represent ipsilateral steps (sides the slips were applied on) whilst the **red error bars** represent contralateral steps (when slips were applied on the other side). (**B**): steps before deceleration, (**P**) triggered steps when the slips were triggered and (**A**) steps after deceleration.

The prosthetic users showed the same pattern as in baseline and SP regarding the step width (Figure 4.15). The intact and prosthetic steps including ipsilateral and contralateral were clearly wider than the left and right steps. The prosthetic users' mean step width for all ipsilateral and contralateral steps were significantly greater than the left and right (p<0.001). Besides the mean, the prosthetic users showed increased variability particularly on the prosthetic side.

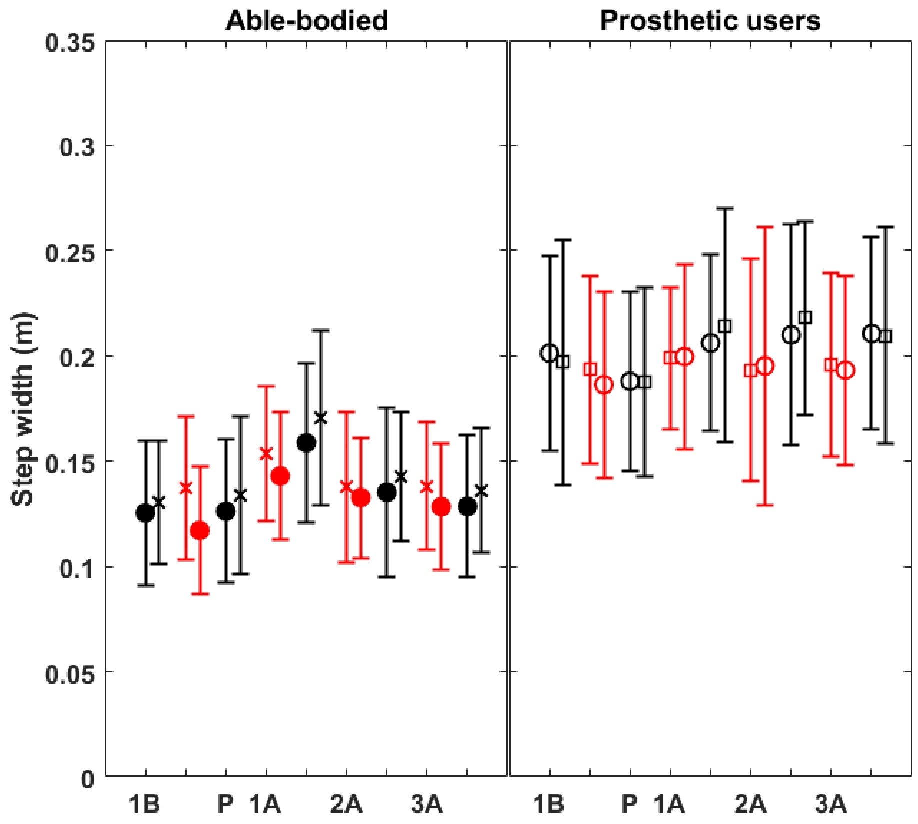

Figure 4.15 Mean and standard deviations of step width at initial contact for able-bodied and prosthetic users groups when imposing the deceleration slips. (●) represent mean values of left sides. (x) represent mean values of right sides. (o) represent mean values of intact sides and (□) represent mean values of prosthetic sides. Error bars represent between-subject standard deviation. The **black error bars** represent ipsilateral steps (sides the slips were applied on) whilst the **red error bars** represent contralateral steps (when slips were applied on the other side). (**B**): steps before deceleration, (**P**) triggered steps when the slips were triggered and (**A**) steps after deceleration.

Figure 4.16 shows mean and standard deviations of step time at initial contact for both groups when facing deceleration slips. Among all steps, the mean step time at initial contact for left side steps and prosthetic side steps were the greatest (i.e. longest in time). In addition to the mean, the variability of these steps was also the greatest. Mean step time and variability for the right and intact steps were the lowest. However, no significant differences were found between able-bodied and prosthetic users for all corresponding ipsilateral and contralateral steps (p>0.130).

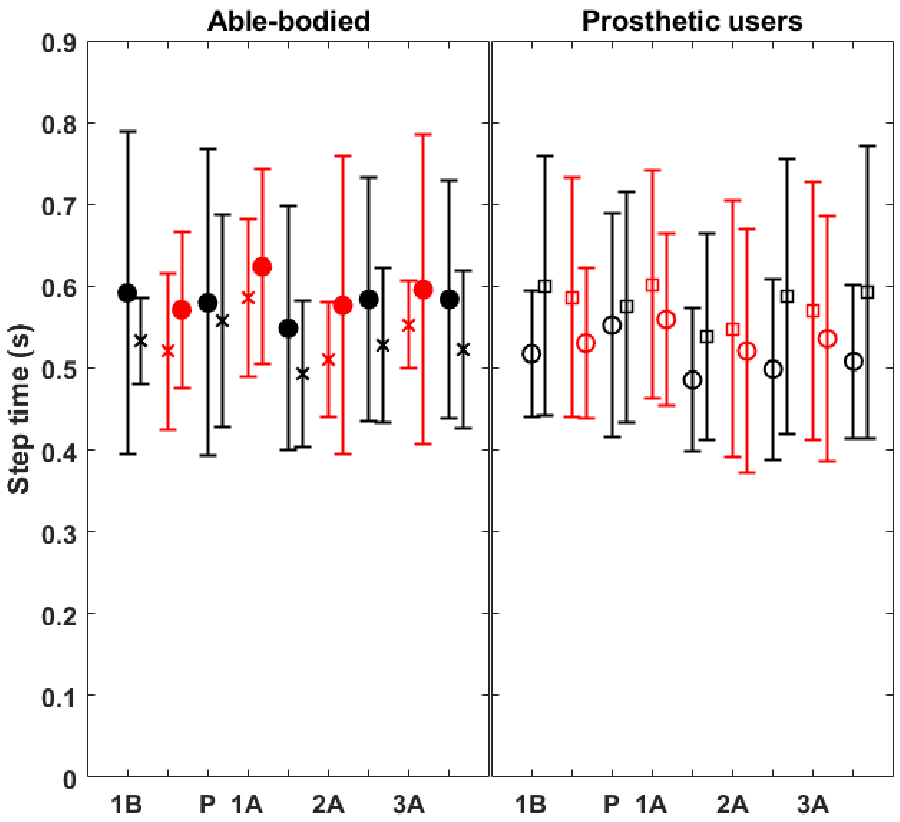

Figure 4.16 Mean and standard deviations of step time at initial contact for able-bodied and prosthetic users groups when imposing the deceleration slips. (●) represent mean values of left sides. (x) represent mean values of right sides. (o) represent mean values of intact sides and (□) represent mean values of prosthetic sides. Error bars represent between-subject standard deviation. The **black error bars** represent ipsilateral steps (sides the slips were applied on) whilst the **red error bars** represent contralateral steps (when slips were applied on the other side). (**B**): steps before deceleration, (**P**) triggered steps when the slips were triggered and (**A**) steps after deceleration.

4.3.5.3 The effect of acceleration slips

The results of the gait stability and stepping mechanisms are presented below in (Figure 4.17 , Figure 4.18 , Figure 4.19 , Figure 4.20 and Figure 4.21)

Mean AP MoS of the intact side and prosthetic sides for steps before accelerations were significantly less than left sides (p<0.040), whilst they were fairly similar to right sides. However, mean AP for the prosthetic sides were the smallest among all corresponding steps (0.126 m). No significant differences were found when comparing mean AP of the first ipsilateral steps for both prosthetic and intact sides to left and right sides (p>0.060). Among all steps mean AP for right I1-A steps were the smallest (0.118 m) followed by the prosthetic side (0.135 m). The intact and left sides were relatively similar in MoS value (0.160 m). As in the first ipsilateral, no significant differences were found when comparing the second and the third ipsilateral steps after acceleration were found (p>0.060).

in terms of the contralateral steps, mean AP MoS for contralateral steps of prosthetic side were the smallest among the corresponding steps (0.125 m). The prosthetic C1-B were only significant different from the left (p=0.010). The intact C1-B were relativity comparable to left and right C1-B and no significant differences were found among these steps (p>0.220). Mean AP MoS prosthetic and intact first contralateral steps after were significantly smaller than the corresponding left steps (p<0.002). For the following contralateral steps, mean AP MoS for the prosthetic C2-A and C3-A were also only significant different from the left steps (p<0.017).

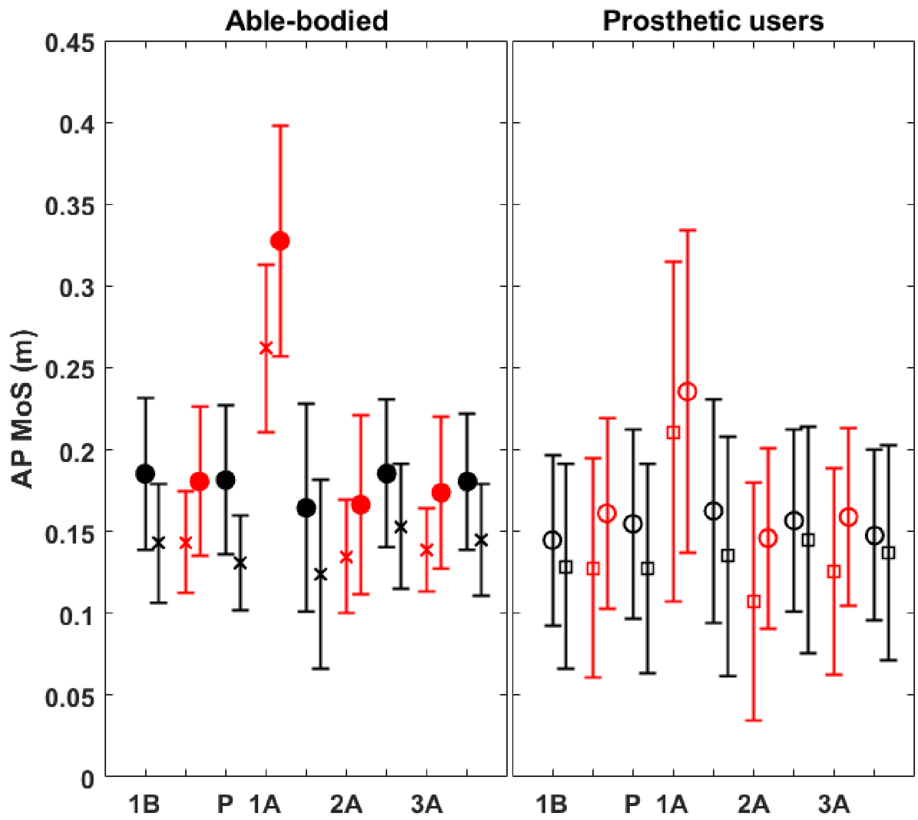

Figure 4.17 Mean and standard deviations of AP MoS at initial contact for able-bodied and prosthetic users groups when imposing the acceleration slips. (●) represent mean values of left sides. (x) represent mean values of right sides. (o) represent mean values of intact sides and (□) represent mean values of prosthetic sides. Error bars represent between-subject standard deviation. The **black error bars** represent ipsilateral steps (sides the slips were applied on) whilst the **red error bars** represent contralateral steps (when slips were applied on the other side). (**B**): steps before acceleration, (**P**) triggered steps when the slips were triggered and (**A**) steps after acceleration.

Generally, the intact and prosthetic sides mean ML MoS were comparable unlike the right and left sides where mean ML for MoS right side were greater than the left (Figure 4.18). As in the baseline, mean ML MoS for prosthetic and intact for all steps were only significantly greater than the left sides (p<0.001). Overall, both groups exhibited the same pattern when comparing the steps before to after.

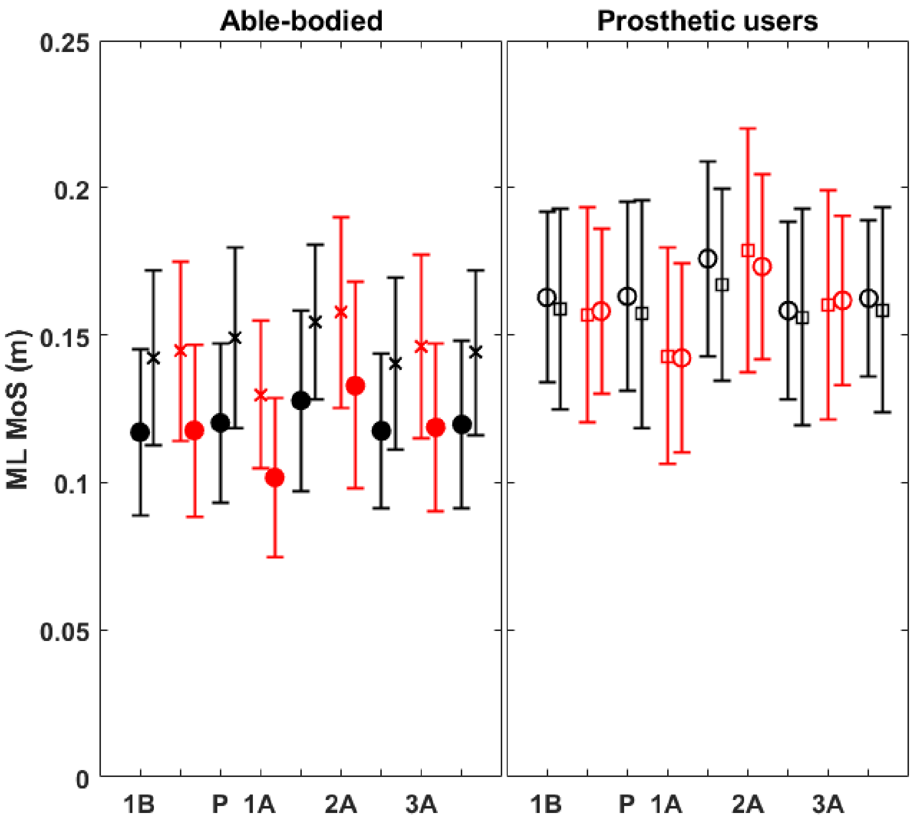

Figure 4.18 Mean and standard deviations of ML MoS at initial contact for able-bodied and prosthetic users groups when imposing the acceleration slips. (●) represent mean values of left sides. (x) represent mean values of right sides. (o) represent mean values of intact sides and (□) represent mean values of prosthetic sides. Error bars represent between-subject standard deviation. The **black error bars** represent ipsilateral steps (sides the slips were applied on) whilst the **red error bars** represent contralateral steps (when slips were applied on the other side). (**B**): steps before acceleration, (**P**) triggered steps when the slips were triggered and (**A**) steps after acceleration

In terms of step length in response to acceleration (Figure 4.19), it was noticed that the variability of normalised step length of for the prosthetic users was significantly greater than able-bodied ($p<0.008$). The variability in step length was the greatest for the prosthetic sides. In general, the prosthetic group reacted similarly to the able-bodied as they took shorter steps on the first ipsilateral steps compared to the steps before then substantially, they returned to the steps before status on the second and the third ipsilateral steps. On average, the intact I1-A steps (0.5) were the shortest among all other steps followed by prosthetic I1-A steps (0.55). both steps were significantly shorter than the corresponding I1-A left and right steps (p<0.003). no other significant differences in mean were found among other steps (p>0.106). However, on average, the prosthetic users took shorter steps compared to able-bodied.

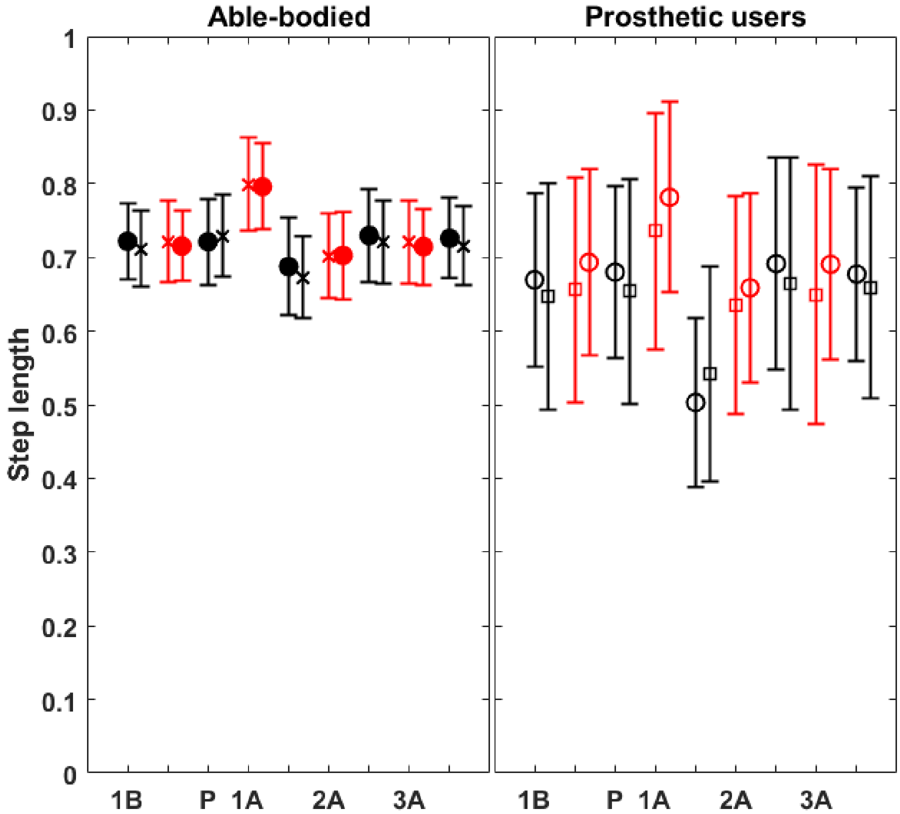

Figure 4.19 Mean and standard deviations of normalised step length at initial contact for able-bodied and prosthetic users groups when imposing the acceleration slips. (●) represent mean values of left sides. (x) represent mean values of right sides. (o) represent mean values of intact sides and (□) represent mean values of prosthetic sides. Error bars represent between-subject standard deviation. The **black error bars** represent ipsilateral steps (sides the slips were applied on) whilst the **red error bars** represent contralateral steps (when slips were applied on the other side). (**B**): steps before acceleration, (**P**) triggered steps when the slips were triggered and (**A**) steps after acceleration

Mean step width of all prosthetic and intact including steps before and after steps were significantly greater than the corresponding left and right steps (p<0.001). In addition, the same pattern was seen for both groups. Where the first ipsilateral steps after acceleration were wider than the steps before whilst the second and third ipsilateral steps after were roundly equivalent to ipsilateral steps before. For the contralateral side steps, the intact and prosthetic side contralateral steps were utterly similar to each other meanwhile the right contralateral steps were wider than the left sides.

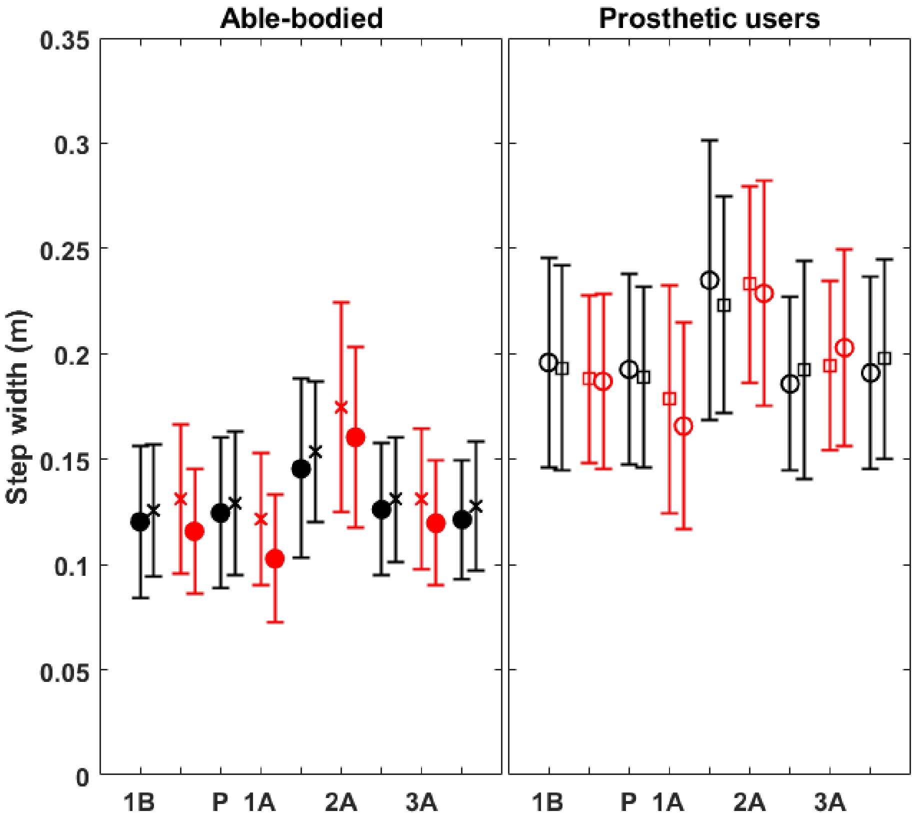

Figure 4.20 Mean and standard deviations of step width at initial contact for able-bodied and prosthetic users groups when imposing the acceleration slips. (●) represent mean values of left sides. (x) represent mean values of right sides. (o) represent mean values of intact sides and (□) represent mean values of prosthetic sides. Error bars represent between-subject standard deviation. The **black error bars** represent ipsilateral steps (sides the slips were applied on) whilst the **red error bars** represent contralateral steps (when slips were applied on the other

side). (**B**): steps before acceleration, (**P**) triggered steps when the slips were triggered and (**A**) steps after acceleration

For the step time, the differences in the variability between left and right step time followed similar pattern in which the left variability was always greater than the right side. Whereas, the differences variability was inconsistent. For example, the step time variability of the prosthetic side was greater than intact step time for steps before meanwhile it was less than intact step time for contralateral steps. It was noted that on average, the intact ipsilateral mean step time of before and after steps were less than the corresponding intact sides. No significant differences were found when comparing the step time of the two groups (p>0.125).

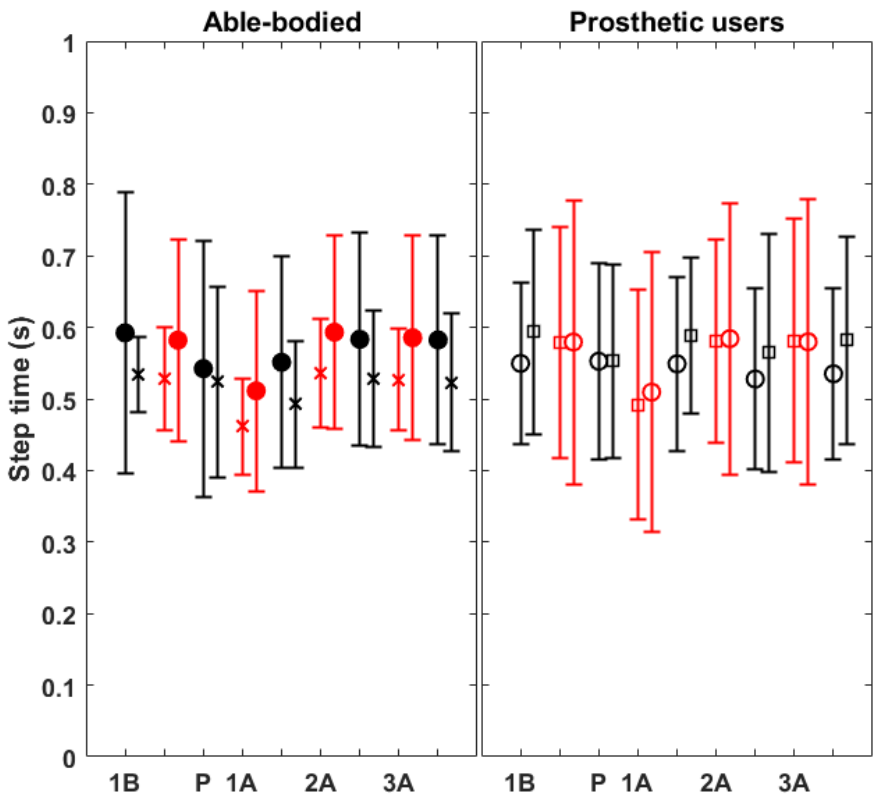

Figure 4.21 Mean and standard deviations of step time at initial contact for able-bodied and prosthetic users groups when imposing the acceleration slips. (●) represent mean values of left sides. (x) represent mean values of right sides. (o) represent mean values of intact sides and (□) represent mean values of prosthetic sides. Error bars represent between-subject standard

deviation. The **black error bars** represent ipsilateral steps (sides the slips were applied on) whilst the **red error bars** represent contralateral steps (when slips were applied on the other side). (**B**): steps before acceleration, (**P**) triggered steps when the slips were triggered and (**A**) steps after acceleration.

4.4 Discussion

The aim of this chapter was to provide an in-depth study of prosthetic users' stability and stepping strategies, and how these stepping strategies might differ from the ones that the able-bodied would use. Results were initially presented for the whole group to see how this group coped with the imposed slips. Individuals were then grouped based on the prosthetic foot type to examine the effect of prosthetic ankle over the stability. Finally, the prosthetic users group was compared to the able-bodied to investigate the differences in stability and recovery mechanism between the two groups.

4.4.1 Prosthetic users as a group

Comparable to the results in chapter three, the results of prosthetic users' MoS especially in the AP direction support the original interpretation that the greater MoS value indicates greater stability (Hof et al., 2005). This is indicated by the fact that AP results on the prosthetic sides were consistently less than intact side (Table 4.2, Table 4.3, Table 4.4, Table 4.5 and Table 4.6). Similar findings were reported previously in other studies. Previous work in the field of prosthetic user's stability by a team of Hak et al reported that MoS AP at initial contact of the prosthetic sides were significantly lower than the non-prosthetic side (Hak et al., 2014, Houdijk et al., 2018a, Hak et al., 2013c). Despite the fact that Hak et al., methods of perturbation were different from the present study's methods (description of the method is addressed in section (1.8).

For the frontal plane results, although the mean ML MoS on the intact side was greater than prosthetic side during steady state walking conditions, these differences did not reach a significant level (p>0.500). However, the MoS variability of on the prosthetic side was always greater than the intact indicating that intact side' stability was greater. The ML MoS results generally agree with results from (Beltran et al., 2014) study.

During unperturbed trials (Table 4.2), the prosthetic users showed no differences in mean step length and width between intact and prosthetic sides. The differences were in variability, the greater variability mostly was more on the prosthetic sides. Usually, prosthetic users show asymmetry for these parameters; the intact step length found to be commonly shorter than the prosthetic step length (Isakov et al., 1997, Robinson et al., 1977, Hak et al., 2014). In addition, the step width is found to be greater for prosthetic side (Hak et al., 2013c). These studies concluded that these asymmetries seen as compensatory mechanisms in order to maintain balance and improve stability.

A group of Roerdink et al. (2012) have discussed the step length asymmetry phenomenon and linked it to trunk progression. The authors showed that the longer prosthetic step was a result of asymmetry in trunk progression along with symmetric forward foot placement relative to the trunk, in addition to the limited muscular action. During prosthetic single limb support, the trunk does not forwardly displace as far as during intact single limb support resulting in shorter intact step. Whilst the symmetric forward foot placement relative to trunk movement would maintain this observation.

Nonetheless, similar to the results of this study, other studies reported the lack of systematic difference between intact and prosthetic step lengths within a sample (Dingwell et al., 1996, Vanicek et al., 2009). These observations may be attributable to firstly, the sample size in which the size was not big enough to show significant differences. Secondly, variations among the sample, as some participants were relativity more active than the other participants. The more active participants may have the ability to walk with symmetrical pattern.

In both cases, the results of this study may indicate that asymmetries for step length and width variables are not necessarily related with stability as all participants have completed the trials with no falls. In addition, the results suggest that not in all cases symmetrical pattern should be an aim of rehabilitation as well as further investigations are needed to be able to identify those measures for which achieving a more symmetrical pattern is most beneficial.

The mean and variability of step time for the prosthetic sides were greater than the intact sides. These results indicate that on the prosthetic side more time was needed to make sure that the placement of the foot is ideal since no feedback was provided from the prosthetic sides. Unlike the intact sides which was faster in time. The increased step time has been linked with decreased stability in a previous work by (Hof et al., 2007).

Table 4.8 provides a summary of MoS results in both AP and ML directions after the prosthetic users participants faced the perturbations. The steps presented were the first three steps for contralateral and ipsilateral sides after both deceleration and acceleration slips compared to the corresponding steps before.

Deceleration		
Step	***AP MoS***	***ML MoS***
C1-A	Decreased (intact~3%, prosthetic~20% *)	Increased (intact~12%, prosthetic~16% *)
I1-A	Increased (intact~45% *, prosthetic~55% *)	Decreased (intact~8% *, prosthetic no change)
C2-A	Increased (intact~35% *, prosthetic~37% *)	Decreased (intact~12%, prosthetic~3%)
I2-A	Increased (intact~20% *, prosthetic~16% *)	Decreased (intact~8% *, prosthetic~13%*)
C3-A	Increased (intact~8%, prosthetic~6% *)	No noticeable change
I3-A	Increased (intact~8%, prosthetic~10% *)	No noticeable change
Acceleration		
Step	***AP MoS***	***ML MoS***
C1-A	Increased (intact~45% *, prosthetic~65%*)	Decreased (~10% intact * & prosthetic *)
I1-A	Increased (intact~15%*, prosthetic~10%*)	Increased (~10% intact * & prosthetic *)
C2-A	Decreased (intact~15%*, prosthetic~10%*)	Increased (~10% intact * & prosthetic *)
I2-A	Increased (intact~8% *, prosthetic~12%*)	No noticeable change
C3-A	No noticeable change	No noticeable change
I3-A	Intact no change, prosthetic increased~10%	No noticeable change

Table 4.8 Summary of the perturbation effect on the MoS compared to steps before. (*) represents significant difference ($p<0.05$).

In response to deceleration slips (Table 4.4, Table 4.5 and Table 4.8), mean AP MoS for ipsilateral steps of both sides were increased compared to baseline and steps before perturbation. The prosthetic side three steps after were all significantly larger than prosthetic side baseline and steps before perturbation. The three ipsilateral steps after intact side being challenged were also significantly larger than baseline but only the first two steps after were statistically significantly larger than steps before. These results indicate that prosthetic users as a group could not return to baseline values and the situation before the slips even after three steps.

In the ML direction, mean ML MoS for the three intact ipsilateral steps after were less than steps before and baseline. Whilst the prosthetic ML MoS for steps after seemed to be comparable to steps before and baseline. When the intact side was challenged by deceleration, the balance was recovered by mainly the intact side by taking firstly a longer step at I1-A. The steps then appeared to be shorter and shorter for the I2-A and I3-A indicating that the prosthetic users were trying to walk in caution. Besides to the shorter steps, the steps were wider and faster. At the same time, the AP MoS for the prosthetic C1-A was significantly decreased to be lowest among all steps (0.098±0.060 m). The involvement of the prosthetic side to retrieve the balance when the intact was being challenged was limited. The prosthetic C2-A and C3-A steps practically the same as steps before. It seemed that the main goal was to maintain the current situation and minimise the drifting as much as possible as no great step length and width were observed for those steps. On the other hand, when the prosthetic side was challenged, the intact contralateral particularly C2-A steps were longer, wider and faster compared to steps before. Prosthetic I1-A steps were similar in length to steps before. But then similar to intact side, the prosthetic steps were shorter for both I2-A and I3-A. All three prosthetic ipsilateral steps after were wider and faster than steps before.

After the acceleration slips, the prosthetic users exhibited increased AP MoS at ipsilateral initial contacts of both sides (Table 4.5 , Table 4.6 and Table 4.8). Compared to baseline, mean AP MoS of all before and after steps were significantly greater than baseline values indicating that the acceleration slips were also challenging to the prosthetic users. The MoS in ML was also challenged after acceleration however, unlike the AP MoS, the ML MoS were only significantly greater than steps before the first ipsilateral step. The acceleration initial effect was greater than deceleration as the first contralateral step after (C1-A) were the longest and the fastest among the steps. To recover balance after acceleration, prosthetic users took shorter and wider first ipsilateral steps. But then, prosthetic users tended to increase their step length and maintain the step width for the following ipsilateral steps. The involvement of the prosthetic side to cover balance when the intact was being challenged was more than the one in deceleration. This may indicate that deceleration slips were more challenging than acceleration slips.

4.4.2 Comparison between prosthetic users and able-bodied

In terms of the question regarding which group was more stable, the hypothesis was that the prosthetic users would show reduced MoS (i.e. reduced stability), increased variability, more irregularity and be more affected by the imposed slips. The results supported this hypothesis except in the ML direction.

As a group, the prosthetic users exhibited significantly less AP MoS on the prosthetic sides among all trials, unlike the able-bodied participants who showed relatively similar AP MoS on both right and left sides (Table 3.2). The prosthetic side AP MoS results were found to be the lowest among all sides including the intact, able-bodied left and right sides results. As mean AP MoS for the prosthetic side were significantly less than able-bodied left and right sides (Figure 4.10 , Figure 4.10 and Figure 4.17). The intact sides were fairly comparable to right and left sides of the controls participants. Besides mean values, the variability in AP MoS was always greater on the prosthetic side. For example, AP MoS for intact side during baseline walking were (0.126±0.030 m) and for the prosthetic side were (0.099±0.048 m). Meanwhile AP MoS for the left and right sides were (0.136±0.033 and 0.129±0.030 m) respectively. These results support the assumption that the prosthetic users would show decreased stability compared to the able-bodied in AP direction. Similar results were reported in one of Hak et al studies (Hak et al., 2014). In which the prosthetic users exhibited reduced AP MoS compared to able-bodied.

In terms of the ML MoS, at the first look, the results conflicted with the interpretation. The MoS in ML for the prosthetic users were always greater than able-bodied. As discussed in section 3.4), this observation was previously reported in number of studies. Firstly, in one of MoS concept founder's paper Hof et al. (2007) showed increased MoS in the ML for the above knee prosthetic users. A population is well-known to be significantly less stable not only than able-bodied but also than the below knee participants due to many factors including the absence of anatomical knee and muscles (Penn-Barwell, 2011, Taheri and Karimi, 2012). Hof et al. (2007) has discussed this observation in a logical approach. Firstly, that the final foot position must be planned beforehand, the act of stepping provides the gross control and then the ankle plays delicate role by providing minor corrections after the foot placed in the final position all depend on the active feedback and distal sensation. Since the prosthetic users lack this feedback, they compensate by taken wider steps making sure that the CoM and its XCoM projection always within their base of support. The disadvantage of this is that the steps would be always wider. Beltran et al. (2014) has reported similar results for below knee prosthetic users who have showed increased ML MoS in all walking conditions including baseline, platform continues translation and continues visual perturbation. Unlike Hof et. al (2007), no description of the step width was presented in Beltran et al. (2014) study. Instead, they have measured the distance between base of support and the location of CoM at the initial contact (BoS-CoM). The BoS-CoM results showed that the distance between CoM and BoS at the initial contact of the intact sides were the greatest among all other steps, the right is greater than the left whilst the prosthetic and left steps were lowest. Beltran et al. (2014) concluded that because the prosthetic users showed increased ML MoS they were less stable than able-bodied. Prosthetic users in the present study as a group took consistently wider steps than the able-bodied in all walking conditions. These results are commonly presented in the prosthetic user's studies. Prosthetic users showed significantly wider steps compared to able-bodied when walking over irregular surface in study (Curtze et al., 2011). Similarly, prosthetic users took wider steps in respond to continuous perturbations in a study of Hak et al. (2013c). Hak et al. (2013c) additionally reported that prosthetic users during unperturbed walking, did walk with wider steps order to increase the ML MoS, which was concluded to as compensatory mechanism for the lower local dynamic stability (LDS).

Similarly, below knee prosthetic users took wider steps to maintain balance when walking over rock surface in (Gates et al., 2013). Not only the below knee prosthetic users but also the above knee showed these results as discussed in Hof et al. (2007) study. These results suggest that prosthetic users always walk in wider base than able-bodied regardless of the walking condition which support Hof et al. (2007) interpretation. In addition, these results explain the increased MoS in ML for the prosthetic users compared to able-bodied. On the other hand, Gates et al. (2013) adopted contrasting interpretation of the increased ML in the MoS arguing that the increased in the MoS indicated increased stability even for the prosthetic group. Kendell et al. (2010) has also supported the same idea that the prosthetic users were more stable than the controls.

Additional observations may support that the prosthetic users were less stable than controls. Firstly, prosthetic users showed increased irregularity among steps more than the controls. Also, the results showed that the prosthetic users could not manage to fully return to the baseline status as the controls could. More corrective steps were needed to recover from slips. The acceleration was more challenging to the prosthetic users than the controls.

Despite differences in the outcome measures, both groups responded in a similar trend to the manipulations of stability. For example, when deceleration slips imposed on the right side and left side, the mean AP MoS for the first ipsilateral step after increased compared to steps before. Similar fashion was found when the intact and prosthetic sides were challenged. Regarding the stepping mechanisms, comparable trends were found as well. Both groups increased their step width, length and decreased their step time to cope with deceleration slips. Prosthetic users, therefore, have the capacity to adopt the same mechanisms to handle challenges of gait stability as able-bodied individuals. These results agree with previously reported work by (Hak et al., 2013c).

However, the groups did vary in how much change from the before status. When the right sides were challenged for example, the first ipsilateral steps were longer by 36%, wider by %37 and faster by 10% from steps before. Whilst when the intact and prosthetic sides were challenged, the corresponding steps were longer, wider by less than 1% and faster by 6%.

4.4.3 Comparison between limbs

The hypothesis was that prosthetic users as a group would show significant limb to limb differences regardless of the walking condition. Contrary to expectations, the results support this hypothesis only for the AP MoS measure. Since no significant differences between limbs were found apart from AP MoS.

The prosthetic users' ability to recover from dis-balancing situations is limited on the prosthetic side. Hence during rehabilitation, imposing complex stability tasks in challenging environments on both sides but more for the intact side would help the prosthetic users enhance their overall stability and reduce the falls rate.

4.4.4 Comparison among the prosthetic feet

The hypothesis was that individuals who were fitted with advanced prosthetic feet that included ankle mechanisms would show enhanced performance compared to individuals who were fitted with basic prosthetic feet. To provide knowledge on each foot stability performance, MoS parameters were evaluated and compared with each other (Table 4.7, Figure 4.3, Figure 4.4, Figure 4.5 and Figure 4.6). Outcomes from this section provide clinical professionals with important decision-making insights.

When investigating the results for the two sub-groups as showed in Table 4.7, no significant differences were found between the two sub-groups. This may indicate that having a high-end prosthetic foot did not add any significant enhancement in stability's performance. MoS in both directions were less by approximately (35% and 5%) in AP and ML respectively in individuals with advanced feet. Study of the spatiotemporal measures, however, may help explain the MoS results. Individuals with advanced feet exhibited shorter and narrower steps during baseline. This may indicate that individuals with advanced feet did not need to increase their step length or walk with wider base in order to contain the CoM. Their prosthetic ankles especially those with hydraulic mechanism provided the fine role that Hof et. al (2007) discussed. But the prosthetic users with advanced ankle showed increased time which according to Hof et. al (2007) this might be another compensation that prosthetic users would exhibit to maintain balance.

In terms of each foot performance, in the AP direction, the Pro-Flex (n=2) showed increased MoS with reduced variability compared to other feet. The results of the prosthetic side were interestingly better than the intact side. The results indicated that Pro-Flex prompted enhanced stability and reduced the demand on the intact side as well. Pro-Flex performance was previously investigated by means of prosthetic ankle angle and power during level, incline and decline walking using CAREN system in a study of Tomkin et al. (2018). Although the study included only three participants, the results were promising. Pro-Flex did provide greater ankle range of motion during level and slope SP walking, which helped in reduce compensatory gait strategies. Comparable results were reported as well in a bigger study (n=11) by Heitzmann et al. (2018). Heitzmann et al., 2018 study was discussed in detail in section (1.3.1). A doctoral dissertation by Gunnarsson (2019) showed that Pro-Flex has combination of very high ankle power and range of motion. Also, that Pro-Flex did help in reducing the load on the intact side by over 10%. The researchers reported that since launch, Pro-Flex is perceived as the most advanced mechanical foot available in the market. Childers and Takahashi (2018) concluded that the high amounts of range of motion and high energy return ability of the Pro-Flex foot to is due to the structure of this foot that being deformed throughout its range of motion.

Conversely, Pro-Flex showed the lowest MoS in the ML direction among all feet (intact ~ < 0.15 m, prosthetic ~ < .0.1 m).

Both ESAR feet; Vari-Flex (n=7) and Soleus (n=1) showed enhanced AP and ML MoS results compared to the high-end Elan (n=1) and Echelon VT (n=2) feet. Additionally, they showed comparable results to the more advanced Pro-Flex. These results again suggested that ESAR feet might be as good as the advanced feet. No comparison by the means of MoS between ESAR feet and more advanced feet was reported previously. However, Houdijk et al. (2018a) have compared the ESAR feet to SACH feet and the results were that the ESAR feet showed increased MoS compared to SACH.

It was noted that Elan foot and Echelon VT were fitted to relatively slower walkers. This might raise the question about the foot selection during rehabilitation. Obviously, every patient wants to be fitted with the latest and the most advanced components despite that they may not use the fitted components to their full potential due to the limited activity level. This may cause to unsuitably consume the materials and public funds (if the prosthetic services are covered). The underuse of the prosthesis and optimizing the prosthetic prescription have been previously discussed in doctoral thesis by (Schaffalitzky, 2010). Their work showed that physical factors were clearly important for prosthetic rehabilitation. In addition, prosthetic prescription depends on the environment characteristics and the situations in which the selected technology is to be used. Individual's personality, preferences and acceptance are factors can play a role in decision-making and the use of the technology.

Nonetheless, an important note should be considered regarding the Echelon VT and Elan feet. The manner in which these feet are set up may play a role in their function. The preferred hydraulic resistance level is set based on the participant preference but primarily selected by the prosthetist and normally these feet are set up for the normal level ground walking. Since no changes were made by the research group within participants this might have affected the overall performance of the feet. McGrath et al. (2018a) supported this idea and the work in their study was presented in a case series study design.

4.4.5 The performance of the ankle disarticulation prosthetic users

The assumption was that prosthetic users with ankle disarticulation showed enhanced performance, less steps to step irregularity and asymmetry than the prosthetic users with more proximal level of amputation. The results supported this assumption.

Ankle disarticulation prosthetic users walked relatively faster (1.3 m/s) than other prosthetic users. during the SP mode they managed to walk as fast as (1.6 m/s). The MoS on both prosthetic and intact sides were to some extend comparable (Figure 4.7, Figure 4.8 and Figure 4.9). When they were challenged by deceleration, they relatively managed to return to the balance status after one stride (I1-A and C2-A) which was similar to control reaction. The prosthetic side involvement in retrieving balance was relatively more than the average of the prosthetic users a group. The results are in line with previous work (Braaksma et al., 2018, Lin-Chan et al., 2003, Jeans et al., 2011). However, further longitudinal studies comparing walking stability at different levels of amputation/anomalies are needed, as the number of ankle disarticulation prosthetic was only two and they were relatively more active.

4.4.6 Limitations and future work

The prosthetic users and able-bodied in the study were not age matched. The able-bodied individuals were substantially younger than the prosthetic users. Therefore, some of the differences in stability and stepping mechanism may potentially have been due to age-related differences in physical capacity or general health status. The effect of age over the stability has been well investigated for the able-bodied. Results of a large recent study that included (105) participants showed that the age had significant impact on the ML MoS (Herssens et al., 2020). It was noticed that for the prosthetic users the age aspect has not been investigated. In addition, the assumption that elderly individuals would exhibit reduced performance was not in the line with the results of the participants of the present study. As there were three participants aged over sixty years old have showed enhanced results, they were relatively fast walkers (CWS >1.3 m/s) which was faster than the average walking speed of able-bodied.

The sample size was small which might have affected the power of the study especially when the participants were sub-grouped based on the prosthetic foot type. The results of two sub-groups may have been affected by the limited number in each group and by within group differences in the activity level. Therefore, testing the two different feet on the same participants in a larger study would be more appropriate. In addition, it may help in answering the question whether fitting advanced prosthetic foot would enhance the stability and the therefore the high price tag would be justified. Altogether would be helpful to identify individuals who might benefit from greater ankle range of motion. The initial assumption would be that optimum use of the advanced feet is greatly linked with relatively higher-level of activity individuals.

Another potential limitation was the variability among the prosthetic users. Nonetheless, this variability in fact increases the chances of study sample of being more representative to this group of walkers. Relatively high overall walking ability of the study sample was commonly reported as a potential limitation (Hak et al., 2013c, Beurskens et al., 2014).

Other factors besides the foot type may have also effects on the results like socket type, alignment, suspension mechanism, stump length and the cause of amputation. Providentially, some of these aspects were not varied among the prosthetic users. Firstly, all the prosthetic users were fitted with the same socket (Total surface bearing TSB). A reviewed paper by (Safari and Meier, 2015) discussed the effects of current prosthetic socket designs using qualitative outcomes concluded that TSB sockets lead to greater activity levels and satisfaction. These results were also reported in an older study by Hachisuka et al. (1998).

Regarding the mechanism of suspension, the prosthetic users were all fitted with silicon liner and most of them used pin and lock. This part was also previously investigated in a study of Ali et al. (2012). The study examined the different mechanisms of suspension, the results were in favour of the use of the silicone liners whether with or without shuttle lock. However, the study highlighted two problems associated with these liners, the increased stump sweating which requires careful personal hygiene. Secondly, some prosthetic users found the processes of donning and doffing the prosthesis to be sometimes harder.

In terms of the stump length, apart from the prosthetic users with ankle disarticulation, other prosthetic users' stump length was medium compared to the intact side. No short or long stumps were found. A study of Subbarao and Bajoria (1995) examined the stump length for prosthetic fitting and concluded that longer stump length may cause some issues in the process of the rehabilitation including the high risk of re-amputation and prosthetic fitting difficulties.

Performance and cause of amputation have been correlated previously in the prosthesis field. Usually prosthetic users with diabetes have been reported to be showing reduced performance compared to individuals with other aetiologies. The results of one prosthetic user with diabetes were in line with these results in terms of the reduced overall performance (Kuhlmann et al., 2020, SARAF, 2015). However, the other one has showed improved results very comparable to other aetiologies if not better. The prosthetic users in Beltran et al. (2014) study were all traumatic limb loss related. The authors have highlighted that the sample was relatively high active and therefore the results may not be extrapolated to other sub-groups. Similarly, the sample was in the study Hak et al. (2013c) in which all participants had traumatic amputation. Here, in the present the sample contained individuals with amputation of different causes which provides more representative sample that may reveal insights about each aetiology performance. On the other hand, sub-grouping prosthetic users based on the aetiologies and comparing performance would be beneficial. Especially in revealing important information that may help in developing and providing prosthetic services that fit the patient needs the most.

All prosthetics users walked with their comfortable prosthetic alignment they use daily. However, the prosthetic alignment can play a critical role in walking and stability, it was unknown to what extend different alignment can impact individual's stability therefore this aspect should be explored in future investigations. The next chapter is a preliminary study aimed to discuss this aspect.

4.5 Conclusion

Prosthetic users walked with a wider base of support than able-bodied controls regardless of the walking conditions. In response to the perturbations, both able-bodied participants and prosthetic users took wider and faster steps regardless of the imposed perturbations. In addition, for the both groups the step length decreased after acceleration. However, the step length was greatly increased after deceleration for the able-bodied participants whilst was nearly the same as steps before for the prosthetic users. Despite that the slips were imposed in the AP direction, both AP and ML MoS were affected for both groups.

The deceleration perturbations were more challenging for the prosthetic users similar to the able-bodied participants. The acceleration perturbations, however, were also challenging for the prosthetic users. Hence, including the both types of perturbations to study the stability of prosthetic users is beneficial. Whilst including the acceleration perturbations would be helpful to the able-bodied individuals in the randomising the upcoming perturbations so they are less anticipated.

Most adjustments strategies were made by the intact side whilst the involvement of the prosthetic side to retrieve stability was limited. Therefore, complex training tasks that improve muscles strength and control for the intact side during rehabilitation could be beneficial to enable prosthetic users to manage real life perturbations and avoid falls.

Future longitudinal studies directed towards investigating falls in prosthetic users may be able to explain the relationship between falls and gait disorders in prosthetic user.

ESAR feet (Vari-Flex and Soleus) allowed an improvement of recovering from perturbations while preserving the margin of stability for prosthetic users walking at day to day walking speed. Since they both were found to be reliable prosthetic feet in facing slips that may be faced in real time situations. This may possibly explain the high subjective preference for ESAR feet in people with a lower limb amputation. Pro-Flex foot showed significantly promising results in the AP direction in perturbations recovering, further study with a larger group fitted with this foot is needed. Surprisingly, the results for the high-end feet (Elan and Echelon VT) with hydraulic ankle movement were not significantly different to the less advanced feet. However, this possibly related to the subjects' activity level in a way that those participants were not using these feet to their potential high capability. Without forgetting the small sample size, hence, further investigations with an appropriate sample size are required.

The results supported that the able-bodied individuals are more stable than prosthetic users. Nonetheless, the results also demonstrated that prosthetic users may have the ability to recover from slips using the same mechanisms and stepping strategies.

4.6 Clinical implications summary

Results indicate that walking with wider and faster steps with no significant change in step length, may help prosthetic users to recover from slips and thus avoid falling. Therefore, it may be feasible to provide training and awareness to prosthetic users which may help them react in a similar way to promote recovery, maintain stability and avoid falls. Such training and awareness may be promoted during rehabilitation, prosthesis fitting and delivery and could be valuable throughout entire prosthetic use.

In prosthetic clinics where 'high-end' prosthetic feet may not be readily available, more traditional designs of ESAR feet may yet be reliable enough to achieve satisfactory gait stability. On the other hand, before fitting top-end prosthetic components, the individual's potential to benefit from those components should examined. Fitting an advanced prosthetic foot such as Pro-Flex for a highly active individual who may utilize it to its full functional capacity, may also have benefits especially in terms of the users' ability to recover from slips and stability. Such considerations in foot prescription are currently not considered but may be of use to prescribers and prosthetic component designers in the future.

If a gait analysis system including a controllable treadmill (even up to the complexity of the CAREN system) was available in a prosthetic clinic, this study protocol may be use in evaluating and provision of 'anti-falling' training.

5 The prosthetic alignment effects on margins of stability

5.1 Introduction

The prosthetic alignment is defined as the 3D position and orientation of the prosthetic components in relation to each other (Isakov et al., 1994). Commonly the prosthetic socket is the reference point for the other components to set up the optimal prosthetic alignment (Seelen et al., 2003). The optimal alignment, which is according to manufacturers' instructions, is the essential factor for the successful rehabilitation of the prosthetic users because a comfortable alignment generally allows ambulation with minimal metabolic cost and without impacting or damaging the residual limb (stump) (Schmalz et al., 2002). Besides, the optimal alignment was reported as essential factor in helping to keep the static and the dynamic balances during standing and walking (Vrieling et al., 2008).

The alignment of prosthetic users especially below-knee level includes six parameters. These are 1) AP translation (shift) of the socket with reference to the foot; 2) AP tilt; 3) ML translation of the socket with reference to the foot; 4) ML tilt; 5) length of the prosthesis 6) foot angle (Berme et al., 1978, Zahedi et al., 1986).

Initially, it is important to note that investigating each type of these alignments and to what extent they might affect the gait of prosthetic users could be revealing vital insights for the prosthetists and the rehabilitation team.

Previous work has investigated the effect of these alignments. Chow et al. (2006) examined the effect of AP translational and tilt changes on gait symmetry for persons with below knee prosthesis (n=7). The study results showed that truly symmetric gait was not found between intact and prosthetic side. However, the applied changes may have played role in decreasing the asymmetry in some gait parameters. Namely the vertical ground reaction forces, stance duration, step length and time for maximum swing phase knee flexion.

Boone et al. (2013) have studied the effect of four socket's alignment changes in both AP and ML directions for below knee prosthetic users (n=11). The applied changes were socket flexion, extension, abduction and adduction. The degree varied between 3 and 6. In their work, the reaction moment of the socket was adopted as a primary outcome measure. The findings suggested that changes in both directions have significantly affected the socket reaction moments.

Friberg (1984) argued that in order to prevent lateral balance loss during standing, walking and running, prosthesis length should be matched with the length of the contralateral (intact) limb. Besides, it was found that the incorrect prosthesis length has led to chronic pain symptoms for low back, hip and knee. Same results were supported by (Lee and Turner-Smith, 2003).

A previous work by Schmalz et al. (2002) showed that changing the angular attitude of the prosthetic foot resulted in pronounced effects on prosthetic users' gait. The changes were ten degrees increased dorsiflexion and plantarflexion from the optimal. The dorsiflexion resulted in displacing the load line more posteriorly whilst the plantarflexion has led to displace the load line more anteriorly. In addition, both changes have significantly increased oxygen consumption when compared to the optimal alignment.

Failing to provide a satisfactory alignment could result in serious issues that may affect the quality of life for the prosthetic users. Those may include walking difficulties and reducing the overall stability increasing the chance of falling, stump pain and tissue breakdown (Fridman et al., 2003, Isakov et al., 1994, Kolarova et al., 2013). In addition, prosthetic poor alignment can also lead to increased demands on the intact limb, pressure changes inside the prosthetic socket, and increased asymmetry of walking all in all may lead to decreased activity level and falling (Fridman et al., 2003).

Nevertheless, it is important to state that finding the optimal alignment is commonly known to be based on two things, the prosthetist judgment and more importantly the feedback from the patient (Berme et al., 1978). Sometimes, from the prosthetist's prospective the alignment would be optimal, but the same alignment would not be preferable for the patient. Some prosthetic users feel more comfort with shorter or longer prostheses or with slight dorsiflexion or plantarflexion. Therefore, prosthetists should keep this note into their considerations and provide satisfactory alignment.

As it was discussed earlier in this thesis, assessing the gait stability of individuals with lower limb loss using the MoS measure provides improved insight and understanding of adaptation strategies used to promote normal walking when using a prosthesis. In addition, by assessing the MoS, it may possible to identify altered voluntary dynamic control in persons with lower limb prosthesis with respect to misalignments of the prosthesis that potentially could be caused from everyday changes (eg, taking shoes off or changing them). Therefore, it was necessary to conduct a case study to evaluate the effect of prosthetic alignment changes on the dynamic stability and walking patterns of a prosthetic user. Moreover, to investigate if applying slight changes from the optimal would be biomechanically acceptable or would these changes promote more challenges to the prosthetic users.

To the best of the research group' knowledge, the MoS has not been adopted to study the potential impacts on the dynamic stability of prosthetic users when the optimal alignment is amended.

Ultimately, this study may help to design a larger study to look at specific aspects of the prosthetic prescription with respect to balance and stability; and improve the fall prevention programmes in rehabilitation which lead to reduce/prevent falling. This information might be taken into account in the future and may inform clinical decision making in choosing prosthetics components that most suit the individual needs.

5.2 Methods

5.2.1 Inclusion criteria

Inclusion was unilateral (one-sided) below the knee lower limb prosthetic user. Ambulatory without walking aids, no other musculoskeletal problems, for example osteoarthritis, chronic back pain, knee instability affecting the contralateral (other) limb, no current problems with their prosthesis. Have been a prosthetic user walking with a prosthesis for minimum a year; and aged 18 and older. Exclusion was participants suffering from motion sickness. Pregnant or suffer other or weight exceeding 100kg.

5.2.2 Recruitment process

An email invitation was sent to prosthetics users' charities and associations in Greater Glasgow and Clyde. Participant provided written informed consent prior to trials session. The project protocol was approved by the university of Strathclyde ethics committee (UEC).

5.2.3 Participant

One below-knee prosthetic user. The participants' characteristics were: 60 years old,76 kg weight,170 cm height, the affected side was the left, the level of amputation was transtibial, the case of amputation was bone infection and the time since amputation was 31 years. The participant was fitted with a typical patellar tendon bearing socket (PTB) with basic energy storing and return prosthetic foot (ESAR). The method of suspension was generated by means of tight friction between the participant stump and socket as well as by over the socket elastic suspension sleeve (Figure 5.2). The walking speed was found as described in section (2.6) and it was found to be (1 m/s).

5.2.4 Equipment

The CAREN (please see section 2.1 for more details) was used at this test to evaluate the participant during walking with prostheses with different conditions.

5.2.5 Data collection Procedure

The participant was invited to attend four sessions for no longer than (2.5) hour each. This research took place at the National Centre for Prosthetics and Orthotics (NCPO) at the University of Strathclyde. The (Table 5.1) below provides an overview about the sessions.

The first session was a screening session to determine whether the participant could be included in the study. The participant matched all the selection criteria. The participant was then asked to walk over the treadmill (description of the machine in section 2.1) and the test protocol was carried out. The walking protocol was similar to the main study as described in section (2.4). This included one steady state walking as baseline and one self-paced mode.

The participant at that time was asked to walk while wearing his own prosthesis, the reason for this pre-testing trial was to give the participant the experience of what would happen during the study in order to help him to decide whether or not to be part of the study before new prosthesis manufacturing process.

When	Expected total time	Output and Conditions	Procedure
Session 1	2 hours	Duplication of prosthetic socket and measures	• Screening for inclusion / exclusion • Consent for participating • Running the protocol • Duplication for the new prosthesis
Session 2	2.5 hour	Familiarizing	• Checking the socket fitting, comfort and function • Finding the optimal alignment • Familiarizing • Determining the appropriate prosthetic alterations • Treadmill familiarizing
Session 3	2.5 hour	• Optimal alignment • Lengthening • Shortening	**For each condition:** • Acclimatising • Treadmill warm-up + Steady state motion capture • Self-paced session
Session 4	2 hours	• Foot adjusted toe up (dorsiflexion) • Foot adjusted toe down(plantarflexion)	**For each condition:** • Acclimatising • Treadmill warm-up + Steady state motion capture • Self-paced session

Table 5.1 An overview of the sessions.

The participant agreed to take part and a duplication of the participant's prosthetic socket was made by a registered HCPC prosthetist. Measurements of weight and body dimensions were taken during this first visit. The participant's residual limb was free of any complications that may require further intervention (Figure 5.1).

Figure 5.1 The participant’s residual limb. The stump was free of any complications. No swelling. The level of amputation compared to the intact side was medium where more than 35% of the original leg length was reserved.

Following that, the taken negative cast was used to produce a positive mold to fabricate the experimental prosthetic socket similar to the one that the participant uses on a daily basis. The fabrication process of the prosthesis followed the standard patellar tendon bearing (PTB) sockets’ process, nothing different or special was applied (Figure 5.2)

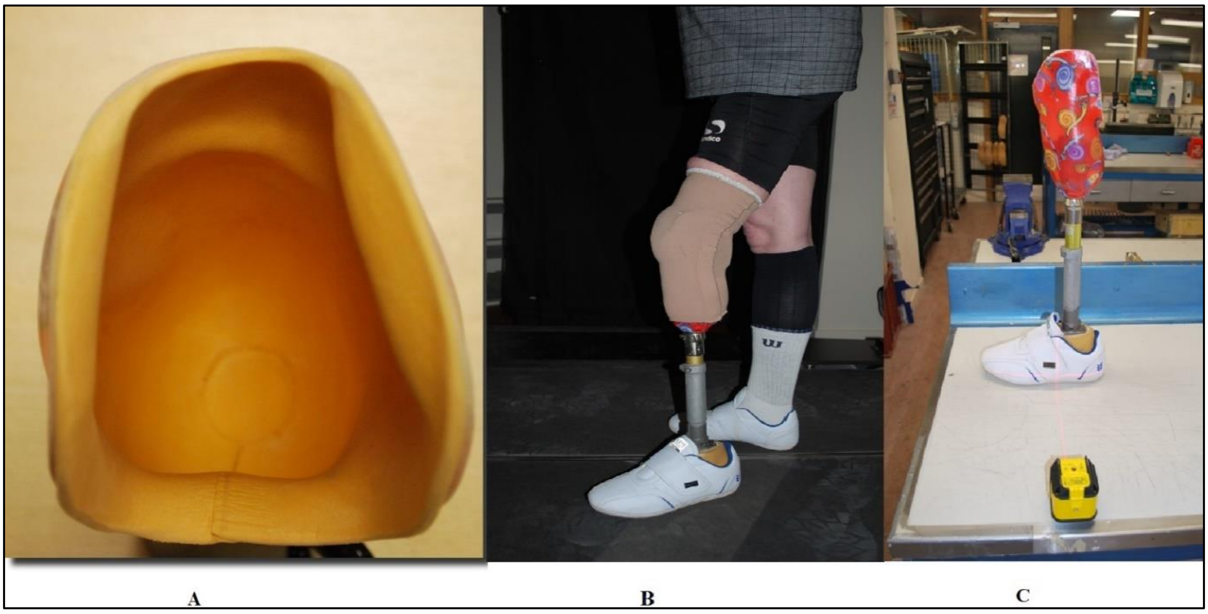

Figure 5.2 The case study's experimental prosthesis. The experimental prosthesis was similar to the participant's own prosthesis that he uses very day. A: patellar tendon bearing. B: the experimental prosthesis fitted to the prosthesis the suspension was obtained from the suspension sleeve and from the friction between residual limb and the socket. C: the experimental prosthesis assembled and benched aligned to the prosthetic manufacturing standards, following the taken measurement of the participant.

In the second session (one week later), the prosthesis was made and ready to be fitted. The participant was invited to visit the NCPO for checking the fitting, comfort and alignment of the prosthesis by the HCPC registered prosthetist. The process of fitting included asking the participant to perform several activities of daily living (ADL) to be familiar with the new prosthesis (Figure 5.3). These included quiet standing, walking and sitting to standing from a chair. During this session, the optimal alignment was found, and the proposed prosthetic variations were determined.

The intended amount of possible changes from the optimal was measured so that these would be ready to be applied during the next motion capture sessions. Whilst measuring these possible changes both component changing limits and the participant management to these changes were taken into consideration.

The optimal alignment was verified by two expert prosthetists, and then measured. Following that, the research team started to change the prosthetic length/foot angle gradually and asked the participant to walk with these changes. The changes were increased based on the participant feedback that he was happy with them as well as the component changing limits then these changes were recorded.

It was found that the possible alignment changes were as follows:

- Prosthesis length lengthened to a maximum of 2% of the leg length (1 cm more)
- Prosthesis length shortened to a maximum of 2% of the leg length (1 cm less)
- Foot adjusted toe up to a maximum of 35% of optimal alignment (a maximum of 5 degrees dorsiflexion)
- Foot adjusted toe up to a maximum of 35% of optimal alignment (a maximum of 5 degrees plantarflexion).

These conditions represent the changes that prosthetic users experience during activity daily living (ADL) according to (Kolarova et al., 2013)

Figure 5.3 Participant was walking with the new prosthesis in the parallel bars to check the fit and comfort as well as the amount of changes that would be applied.

The following sessions were used to record the participant's walking under different conditions.

Prior to treadmill walking under any condition, the participant was asked to practice several ADL to get familiar with the applied condition. Such as sit-to-stand from a chair, quiet standing and over ground walking (between parallel bars).

The participant walked over CAREN (see section 2.1 for full details of the equipment).

Prior to any motion capture session, the participant was asked to change into appropriate tightly fitted clothing in order to attach the reflective markers. Ten reflective markers were applied to the following anatomical landmarks:

- Right and left ASIS
- Right and left PSIS
- Right and left lateral malleolus
- Right and left heels
- Right and left 5th metatarsals.

The places of the markers on the prosthesis were estimated to match the ones on the intact side. The markers' attachment is a low-risk process, to attach the markers allergy free tape (hypoallergenic adhesive tape) was used.

During session 3, two changes were made for the prosthesis; the prosthesis length was increased/ decreased by 1 cm from the optimal length. Then the participant was asked to walk.

During session 4, similarly two changes were made; the prosthetic foot was adjusted by 5 degrees increased dorsiflexion and 5 degree increase of plantarflexion from the optimal foot alignment.

Prior to protocol application for all conditions and sessions, the participant was also asked to walk on the treadmill at a self-selected speed for around 5 to 6 minutes to become acclimatized to the environment/changes. The participant was encouraged to communicate with the research team during the test, in case that the participant found the applied changes stressful to cope, the treadmill would be stopped, and the appropriate adjustments would be made.

Participant walked at a comfortable speed (CWS), similar to the method addressed in section (2.6).

Following that, the treadmill speed was to the CWS and the walking was recorded under these conditions which each last for 3 minutes:

- Baseline fixed speed.
- One self-paced mode trial in which the participant will control the treadmill speed by slowing down or speeding up (section 2.5).

The longest walk trial was less than 5 minutes. The appointment lasted less than 2.5 hours. The participant took rest breaks whenever needed. The participant was given the chance may stop the trials at any point. The participant was not asked to perform any activity which causes distress to him.

Once the participant completed the walking trials, he was assisted from the motion platform immediately after being unclipped from the harness. The reflective markers were then carefully removed.

5.2.6 Data analyses

To study the gait stability the margins of stability (MoS) were calculated along with three spatiotemporal gait parameters (see section 2.10 and 2.11). The CoM trajectories were estimated using four markers on the pelvis (section 2.8). The gait events were identified using

Zeni's velocity-based algorithm (section 2.7). The MoS and gait parameters were calculated at the instant of initial contact (80 steps each limb).

All processing and analyses were performed with custom MATLAB R2019b codes (Mathworks, Inc., Natic, USA).

5.2.7 Statistical analysis

All data was based on the average ± standard deviation. Data management and analysis were performed using SPSS v25 (SPSS Inc., Chicago, IL, USA).

The Shapiro-Wilk test was used to check data normality. The data was normally distributed therefore parametric statistical tests were used to investigate the differences. For all statistical tests, the 95% confidence interval for mean differences was calculated and the *p*-values represented were statistically significant when $p < 0.05$.

In order to investigate the differences in MoS and spatiotemporal parameters between the intact and prosthetic sides, the paired sample t-test was used. Also, the same test was performed to investigate the differences between each alignment change and optimal alignment. The intact and prosthetic sides during the optimal alignment were compared to the corresponding side of each change.

5.3 Results

The participant reported the comfortable walking speed at 1m/s. Whilst for the self-paced mode, the participant has walked slightly faster on average for each alignment change. Optimal SP (1.096±0.065 m/s), short SP (1.030±0.082 m/s), long SP (1.093±0.097 m/s), DF SP

(1.101±0.067 m/s) and PF (1.152±0.073 m/s). the participant completed all conditions without falling.

5.3.1 Baseline walking

Firstly, investigating the between-side symmetry showed that the intact and the prosthetic sides were significantly different from each other in all walking condition parameters when walking at fixed speed of (1m/s) (*p* <0.001). During baseline walking, mean AP MoS for the prosthetic side with increased five degrees of plantarflexion (0.082±0.011m) were significantly larger than the intact side (0.075±0.013 m). Otherwise, mean AP and ML for prosthetic side were significantly less than the intact side. Mean step length for the prosthetics side were significantly greater than intact side for changes. Mean step width for the prosthetic side on the other hand, were smaller than intact side. In terms of step time for the prosthetic side, mean step time were less than the intact side mean for all conditions.

The results of MoS and gait parameter are presented in the (Table 5.2) below. During walking with short prosthesis at fixed speed, it has been noticed that all parameters were significantly different from optimal alignment (*p*<0.018), except mean step time for prosthetic side where no significant differences from the comparable side of the optimal were found (*p*=0.090) . Walking with short prosthesis, could significantly reduce AP MoS ML MoS and step width for the intact side. In contrast, the step length and step time have significantly increased for the intact by (24.9% and 1.9%), respectively. For the prosthetic side compared with the optimal length, it has significantly reduced mean AP MoS, step width and length. Whilst the ML MoS were significantly increased.

Changes	Side	AP MoS (m)	ML MoS (m)	Step length	Step width (m)	Step time (s)
Optimal	Intact	0.105±0.013	0.192±0.010	0.626±0.027	0.152±0.019	0.556±0.012
	Prosthetic	0.085±0.013	0.131±0.009	0.717±0.022	0.131±0.015	0.528±0.015
Short	Intact	0.069±0.012	0.186±0.009	0.661±0.024	0.145±0.020	0.567±0.009
	P	**<0.001**	**<0.001**	**<0.001**	**0.018**	**<0.001**
	Prosthetic	0.074±0.014	0.138±0.007	0.706±0.019	0.121±0.014	0.532±0.015
	P	**<0.001**	**<0.001**	**<0.001**	**<0.001**	0.090
Long	Intact	0.084±0.015	0.203±0.007	0.619±0.022	0.161±0.015	0.565±0.009
	P	**<0.001**	**<0.001**	0.109	**<0.001**	**<0.001**
	Prosthetic	0.068±0.012	0.130±0.009	0.720±0.019	0.139±0.017	0.523±0.010
	P	**<0.001**	0.373	0.366	**0.003**	**0.019**
DF	Intact	0.102±0.011	0.181±0.008	0.617±0.025	0.143±0.015	0.535±0.010
	P	0.211	**<0.001**	**0.034**	**0.003**	**<0.001**
	Prosthetic	0.083±0.012	0.150±0.008	0.684±0.020	0.138±0.012	0.526±0.014
	P	0.408	**<0.001**	**<0.001**	**0.002**	0.371
PF	Intact	0.075±0.013	0.196±0.007	0.648±0.024	0.140±0.013	0.560±0.011
	P	**<0.001**	**0.007**	**<0.001**	**<0.001**	**0.047**
	Prosthetic	0.082±0.011	0.152±0.008	0.697±0.0214	0.133±0.013	0.533±0.015
	P	0.108	**<0.001**	**<0.001**	0.189	**0.020**

Table 5.2 Mean ± standard deviations of AP MoS, ML MoS, step length, step width and step time for the baseline where the participant walked at fixed speed of 1m/s. Parameters for both prosthetic and intact sides of the participant while walking under five conditions. Optimal: no changes were made to the prosthetic alignment. Short: the prosthesis was 1 cm shorter than the optimal. Long: the prosthesis was 1 cm longer than the optimal. DF: the prosthetic foot was in an increased dorsiflexion orientation from the optimal by five degree. PF: the prosthetic foot was in an increased plantarflexion orientation from the optimal by five degree. Significant differences at p <0.05 from the comparable optimal sides.

Walking with longer prosthesis also has significantly changed the parameters' results compared with the optimal length during walking with fixed speed. For instance, mean ML MoS, step width and step speed increased by (5.9%). Meanwhile, mean AP MoS and step length decreased on the intact side by (20%) and (1.1%), respectively. For the prosthetics side, mean AP MoS, ML MoS, and step speed decreased up to (20%). Additionally, the step length and step width were increased by (0.41%) and (6.1%), respectively.

In further, the results of this study showed that walking with prosthesis with five degrees dorsiflexion could significantly reduce all parameters for the intact side apart from mean AP MoS during walking with fixed speed by up 4.7%. For the prosthetics side, mean AP MoS, step length and step time reduced by up to 4.5%; however, the ML MoS and step width were significantly increased by 15.3% and 5.3% respectively.

It has been noted that walking with prosthesis with five degrees plantarflexion could significantly decrease mean AP MoS and step width 28.5% and 7.5%, respectively. In addition, it has increased ML MoS, step length, and step time by 2%, 9.2%, and 0,7% respectively compared to the optimal foot alignment during walking with fixed speed. For the prosthetics side, all parameters were decreased except the ML MoS significantly increased by 16% ($p<0.0$).

5.3.2 Self-paced mode

For between-side symmetry, apart from mean AP MoS during self-paced mode with short prosthesis fitted, the intact and the prosthetic sides were significantly different from each other in all walking conditions parameters ($p < 0.040$). Similar to baseline, mean AP MoS for the prosthetic side with increased five degrees of plantarflexion (0.121±0.013 m) were significantly larger than the intact side (0.116±0.015 m). However, unlike the baseline mean ML MoS of the five degrees of plantarflexion (0.121±0.013 m) was also greater than the intact mean ML MoS (0.116±0.015 m). Mean AP and ML for prosthetic side were significantly less than the intact side for the optimal and other alignment changes. Similar to baseline, mean step length for the prosthetics side were significantly greater than intact side for changes as well as mean step width for the prosthetic side were smaller than intact side. In terms of step time for the prosthetic side, mean step time were also less than the intact side mean for all conditions.

Changes	Side	AP MoS (m)	ML MoS (m)	Step length	Step width (m)	Step time (s)
Optimal	Intact	0.120±0.016	0.188±0.010	0.674±0.030	0.154±0.018	0.553±0.009
	Prosthetic	0.095±0.016	0.131±0.010	0.763±0.033	0.133±0.017	0.524±0.012
Short	Intact	0.075±0.015	0.199±0.011	0.661±0.035	0.147±0.015	0.568±0.012
	P	**<0.001**	**0.010**	**0.014**	**0.009**	**<0.001**
	Prosthetic	0.078±0.015	0.131±0.009	0.723±0.040	0.122±0.019	0.538±0.016
	P	**<0.001**	0.720	**<0.001**	**<0.001**	**<0.001**
Long	Intact	0.100±0.017	0.177±0.010	0.676±0.048	0.156±0.017	0.576±0.017
	P	**<0.001**	**<0.001**	0.186	0.508	**<0.001**
	Prosthetic	0.072±0.016	0.127±0.010	0.772±0.039	0.131±0.015	0.548±0.019
	P	**<0.001**	**0.023**	**0.003**	0.594	**<0.001**
DF	Intact	0.112±0.013	0.177±0.009	0.679±0.027	0.132±0.015	0.548±0.010
	P	**0.001**	**<0.001**	0.199	**<0.001**	**0.003**
	Prosthetic	0.095±0.016	0.140±0.008	0.745±0.024	0.125±0.014	0.533±0.016
	P	0.911	**<0.001**	**<0.001**	**0.008**	**<0.001**
PF	Intact	0.116±0.015	0.183±0.009	0.697±0.029	0.134±0.016	0.549±0.014
	P	**0.046**	0.681	**<0.001**	**<0.001**	0.050
	Prosthetic	0.121±0.013	0.144±0.009	0.753±0.029	0.130±0.016	0.523±0.016
	P	**<0.001**	**<0.001**	0.090	0.263	0.568

Table 5.3 Mean ± standard deviations of AP MoS, ML MoS, step length, step width and step time for self-paced trial where participant controlled the treadmill speed. Parameters for both prosthetic and intact sides of the participant while walking under five conditions. Optimal: no changes were made to the prosthetic alignment. Short: the prosthesis was 1 cm shorter than the optimal. Long: the prosthesis was 1 cm longer than the optimal. DF: the prosthetic foot was in an increased dorsiflexion orientation from the optimal by five degree. PF: the prosthetic foot was in an increased plantarflexion orientation from the optimal by five degree. Significant differences at $p < 0.05$ from the comparable optimal sides.

During SP trial, the participant was responsible to change the speed and walk with the five different conditions. The results of the AP MoS, step length and step width were significantly reduced for both the intact and the prosthetic during walking with shorter prostheses side compared to the optimal walking with the optimal length at self-speed mode (Table 5.3). However, the step time significantly increased for the intact and prosthetics side (4.1%) and (2%) respectively.

Walking with longer prosthesis showed that the AP MoS and ML MoS were significantly reduced for both side by up to 20%. Mean step time, on the other hand, were significantly increased by 4.1% and 4.5% for the intact and the prosthetics side respectively (Table 5.3)

During walking with dorsiflexion, all parameters were increased except mean ML and step time were significantly reduced by 2.6% and 5.4% respectively for the intact side compared to the optimal foot position. In contracts, all variables were decreased except mean ML and step time significantly increased by 6.8% and 1.7% respectively for the prosthetics side during walking with dorsiflexion compared to walking with the optimal foot alignment (Table 5.3) .

During walking with increased plantarflexion alignment, results of the all variables were reduced on the intact side except the step length significantly increased by 3.4%. In contrast, mean AP MoS and ML MoS were significantly increased 21.4% and 9.9% respectively. Other variables were reduced on the prosthetics side compared to walking with optimal alignment.

5.4 Discussion

The aim of this case study was to investigate the potential impacts of misalignments on the dynamic stability. The anticipation was that setting the prosthesis in a misaligned status would put extra challenge on the prosthesis user and may lead to reduced stability. In addition, that the burden on the intact side would be increased as a result of the misaligned status in order to maintain balance. The results of this case study support these hypotheses.

Generally, the results of this case study support the interpretation of the reduced MoS means reduced stability. Since the mean AP and ML MoS for intact side were significantly greater than the prosthetic side for most changes apart from AP MoS for plantarflexion changes in baseline and self-paced mode as well as ML MoS for plantarflexion in self-paced mode.

The greatest AP stability of both intact and prosthetic side were seen during the optimal alignment. These are in line with the common idea suggests that the optimal alignment would be the most biomechanically and clinically stable prosthesis alignment. These results agree with previous work (Schmalz et al., 2002, Kolarova et al., 2013, Boone et al., 2013, Hannah et al., 1984).

The asymmetry results among the parameters between intact and prosthetic sides of this study were in line with several previous reports, (Roerdink et al., 2012). Several reasons were reported for this asymmetry; such as the lack of the distal feedback from the prosthetic side (Viton et al., 2000), the type of the fitted prosthetic foot (Torburn et al., 1990, Agrawal et al., 2013) and metabolic energy expenditure (Mahon et al., 2019).

Mean step length of the prosthetic side were greater than the mean step length of the intact side for all walking condition and changes. These results have been also reported previously in several study (Mattes et al., 2000, Howard et al., 2013, Klodd et al., 2010, Hak et al., 2014).

Studies suggested some factors that could lead to such observations. Hak et al. (2014) suggested that the smaller steps of the intact side consider as a functional way to create a larger AP MoS at intact initial contact. Whilst the longer prosthetic steps were seen as a functional compensation to preserve MoS of the prosthetic side during the phase of double-limb support. Prosthetic socket’s wrong alignment could lead to same results (Jonkergouw et al., 2016). Houdijk et al. (2018a) argued that the type of fitted foot could lead to step length asymmetry.

In their study, it was found that the ESAR feet enhance the step length asymmetry compared to the traditional sold ankle cushioned heel (SACH) feet.

Table 5.4 summarises the effect of each alignment change on the AP and ML MoS compared to the optimal prosthesis in both walking conditions; comfortable walking speed (CWS reported by the participant at 1m/s) and self-paced mode (SP).

Condition	***Side***	***Effect on AP MoS***	***Effect on ML MoS***
Short	Intact	Decreased (CWS 35%*, SP 38%*)	Varied CWS decreased 4%*,SP increased 5%
	Prosthetic	Decreased (CWS 15%*, SP 20% *)	Varied increased CWS 6%*, SP no change
Long	Intact	Decreased (CWS 20% *, SP 16% *)	Varied CWS increased 5%*, SP decreased 6%*
	Prosthetic	Decreased (CWS 20% *, SP 20% *)	No noticeable change
PF	Intact	Decreased (CWS 25% *, SP 3%)	No noticeable change
	Prosthetic	Varied CWS no change, SP increased 27% *)	Increased (CWS 16%*, SP 10%*)
DF	Intact	Varied CWS no change, SP decreased 6% *	Decreased (CWS 6%*, SP %10*)
	Prosthetic	No noticeable change	Increased (CWS 16%*, SP 10%*)

Table 5.4 Summary of each condition effect on MoS compared to optimal prosthesis. (*) represents significant difference ($p<0.05$).

In terms of the applied changes, short, long and plantarflexion were successfully managed to significantly decrease the AP stability on both sides compared to the optimal (Table 5.4). Whilst the dorsiflexion did not greatly affect the AP stability since the it reduced the stability by only (2.5%). The most challenging changes compared to the optimal alignment for the intact AP stability was during short prosthesis by approximately (35%-38%).

The applied changes did increase the demand on the intact side to maintain balance and challenged the participant stability. The opposite results were reported in a study of (Kolarova et al., 2013) who adopted the same applied changes to challenge the below-knee prosthetic users. Kolarova et al., (2013) argued that the applied changes have little overall effects on stability. However, in their study, the authors adopted a different outcome measure to quantify the stability. A measure called 'limits of stability' was used. This measure has not been adopted commonly in the stability studies. Also, in their study, the participants were tested while standing on forceplates only whereas no walking tasks were involved, and basically the study aimed to assess the standing (postural) stability rather than the dynamic stability.

Compared to the optimal alignment (intact= 0.152±0.019 m, prosthetic side =0.131±0.015 m), the participant walked with wider steps on both intact (0.161±0.015 m) and prosthetic sides (0.139±0.017 m) during the long prosthesis session. This is in line with previous reports related to one of the most common gait deviations for prosthetic users (Esquenazi, 2014, Rábago and Wilken, 2016, Varrecchia et al., 2019). A gait deviation called 'abducted gait', in which the participant is voluntarily walking with a wider base of support. In fact, this deviation is considered a diagnostic way for the prosthetist to know that the fitted prosthesis is too long. Walking with a wider base of support has resulted in increased mean ML MoS.

While walking with short prosthesis, the participant took longer, narrower and slower in time steps on intact side, the same was for the prosthetic side except that the steps were shorter. Lateral trunk bending is another common gait deviation seen in the below knee prosthetic users may help in understanding these results. When a prosthetic user is being fitted with a short prosthesis, the trunk tends to flex towards the prosthetic side (Varrecchia et al., 2019). This deviation is known to produce excessive and intolerant pressure on the prosthetic side (Michael and Bowker, 2004). In order to avoid the pain and discomfort, the participant was depending more on the intact side for the forward progression thus the intact step length and time were increased from the optimal. Mean ML MoS for the intact side was reduced whilst the prosthetic

ML MoS was increased although the step width of both sides was not increased. The lateral trunk bending may have also caused the CoM to be directed towards the prosthetic side thus the ML MoS of the prosthetic side was increased.

The effects of the increased dorsiflexion on the intact side were walking with shorter, narrower and faster steps compared to optimal. Whereas the prosthetic side steps were shorter, wider and slightly faster in time compared to the optimal. These adaptions may have resulted in reduced ML MoS for the intact side and increase the ML MoS for the prosthetic side. Kolarova et al. (2013) has showed that excessive dorsiflexion of the prosthetic foot also leads to poorer movement strategy compared to the intact side and able-bodied. Other gait deviations have also been linked with excessive dorsiflexion for the below-knee prosthetic users. These are an excessive knee flexion at initial contact and drop off (where the heel leaves the ground too early resulting in early knee flexion) (Varrecchia et al., 2019).

The effects of the increased plantarflexion on the intact side were walking with longer, narrower, slightly slower steps. Meanwhile, the prosthetic steps were shorter, wider and similar in time. Walking with increased plantarflexion appeared to be reducing the intact-prosthetic side asymmetry and increasing the stability of the prosthetic side. Also, excessive foot plantarflexion has been linked with insufficient or absent knee flexion (Michael and Bowker, 2004).

The research team was aware that the generalisability of the results may be limited due to the fact that this case study included one participant only. This case study aimed to provide preliminary insights regarding the potential alignment changes and responses to these changes. Altogether to help in designing a bigger study with an appropriate sample size.

The applied changes and the intensity of these changes could be strong enough for this study participant to trigger gait responses. But on the other hand, these changes could be within the acceptable range of changes for other prosthetic users. Or they might even be considered extreme changes to other prosthetic users. Therefore, a further study with a bigger sample size would be beneficial to find to what extend these changes may affect the gait pattern of prosthetic users.

One could anticipate that the longer a participant has the prosthesis the better the stability is. Although the time since amputation for this participant was relatively long (31 years), the applied changes triggered gait response. It might be as a result of aging, as it was found that the older participants have reduced stability (Deandrea et al., 2010). Nonetheless, it would be useful to investigate the effect of time since amputation on the outcome measures.

Some studies included the acclimatizing time for the applied changes as a potential limitation in their conclusions. It might be the same case in the present case study. However, in this study the participant was given some time to perform several ADL tasks to be more familiar with the applied change. In addition, since the study aimed to investigate the immediate impacts of the applied changes on gait stability and parameters, acclimatizing time limitation might be neglected.

5.5 Conclusions

MoS can be adapted as an outcome to assess alignment changes for prosthetic users. In addition, quantifying MoS could help the researchers and prosthetists to understand how a given alignment may affect the gait stability of the prosthetic users. The adopted alignments changes can be challenging. Therefore, including them in a bigger research would be beneficial in assessing the potential stability impacts of misaligned prosthesis. The results of this study may assist the prosthetist in identifying unacceptable alignments.

The short prosthesis seemed to be more challenging than other changes for both prosthetic and intact side. However, all changes have imposed extra challenge for the participant. Finally, this case study provides preliminary evidence that inappropriate prostheses alignment including improper length and foot angle may reduce gait stability and impose further challenges for persons with lower limb prosthesis.

5.6 Clinical implication summary

The results of this case study demonstrated that provision of a prosthesis of incorrect length may have an adverse effect on the walking stability of a prosthetic user. The impact of prosthetic length appeared to be much more challenging when compared to other alignment changes such as the prosthetic foot position in relation to the prosthetic socket in the sagittal plane. These preliminary results indicated whilst adjustment of the angle of the prosthetic foot in extra dorsiflexion or plantarflexion may be tolerated, a short prosthesis may result in decreased stability. Further research is necessary with an appropriate sample of participants, however, this work does indicate that particular care must be taken to get the correct prosthetic length during prosthetic fitting and delivery. In addition, further research would be of interest to determine how stability of prosthetic users of higher amputation level may be affected as ‘above knee’ users generally are expected to use a prosthesis of shorter length than the sound side.

6 Lower limb congenital absence & anomaly

6.1 Introduction

Amputation or absence may be due to acquired causes such as trauma or congenital related causes. Whilst several studies and reports have discussed the first group, a limited number of studies have talked about the second group.

Jain (1996) stated that the most common of limb absence or loss in children under 10 years old was congenital abnormalities. These reports promote the need of further investigation for this group. This chapter discusses congenital anomalies.

The aim of this chapter was to examine the dynamic stability of a person with congenital below knee anomaly. Stability results pertaining to the main study of a case study for a prosthetic user with a congenital related limb loss are presented. The assumption was that a prosthetic user with a congenital related limb loss would show improved dynamic stability performance compared to acquired related prosthetic users. This is because a prosthetic user with a congenital related limb would be fitted with a prosthesis for a longer time nearly since birth. The results of this chapter may help in providing reference knowledge to the rehabilitation team specially prosthetists and physiotherapists which may help inform more directed prosthetic component prescription and rehabilitation programmes to fit the requirements of this group.

6.2 Case study

This section discusses a case study of prosthetic user with congenital related limb anomaly. The aim of this study is to examine the dynamic stability on a person with congenital below knee anomaly. Then, to compare the status of the prosthetic user to the groups in chapters three & four. To the best of the researchers' knowledge, this aspect has not been discussed previously in the literature. The outcome of this case study may help prosthetists and physiotherapists to have better understand of stability status of this group. In addition, to have insight of the potential differences compared to able-bodied subjects and prosthetic users with other aetiology.

The methods used in this study followed the same methods for participants chapter three & four. This included the protocol and equipment (please see section 2.1 for more details).

The criteria for participation as well as method of recruitment were similar as in chapter four (section 4.2.2 & 4.2.3). The participant was randomly selected.

Patient characteristics: A 47 years old female participated in the study. 65 kg weight and 157 cm height. The case presented with transverse both upper and lower limb absences (previously as phocomelia) related limb absence case. Both were right-sided. The lower limb deficiency level was below the knee (middle third according to Figure 6.1). Whilst the upper limb deficiency was can be considered similar to level of through the elbow amputation however, with the presence of partial hand.

According to Frantz & O'rahilly classification the limb loss in upper limb can be classified as phocomelia type III, where the partial hand was attached directly to the arm with complete absence of the forearm bones. Whilst the lower limb can be classified as phocomelia type I, with complete absence for the ankle and foot.

The case did not undergo any surgical or revision procedure. There was no similar complaint in any of the family members or close relatives. The case was able to achieve independent mobility, static and dynamic balance. The case was free from any contracture deformity that can affect RoM and performance.

Prosthesis design; the participant was fitted with total surface bearing socket and with a locking pin silicon liner for suspension. The type of the foot was a Vari-Flex foot (Össur, Iceland). At the time of the study, the participant was fitted with the used prosthesis for five years.

Comfortable walking speed was found to be 1.2 m/s (CWS was found as in section 2.6). Whilst the average walking speed at SP was (1.27 m/s).

6.2.1 Results

The results of the stability in AP and ML directions are represented in (Figure 6.1, Figure 6.2 ,Figure 6.3, Figure 6.4 and Figure 6.5). During baseline walking, the participant exhibited approximately 30% increased mean stability in AP direction on intact side than the prosthetic. AP MoS at the initial contact for the intact side was (0.148±0.019 m) whilst at the prosthetic side initial contact was (0.104±0.016 m).

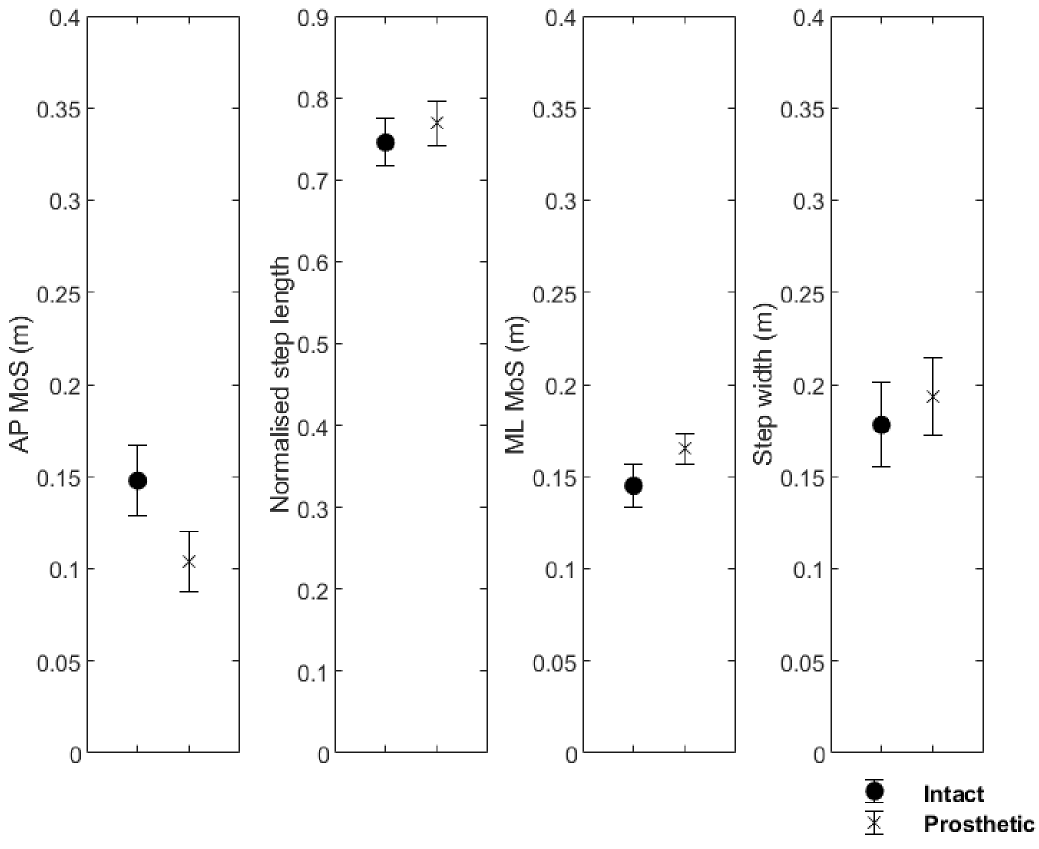

Figure 6.1 MoS in AP and ML directions along with step width and length for congenital prosthetic user during baseline walking. (●) represent mean values for intact side whilst (x) represent mean values for prosthetic side. Error bars represent subject standard deviation.

The participant took longer steps on the prosthetic side (0.770±0.026) than the intact side (0.745±0.029).

In the ML direction, the participant exhibited larger mean ML MoS at the initial contacts on the prosthetic side (0.165±0.007 m) than intact side (0.145±0.012 m). In addition, the participant took wider steps on the prosthetic side (0.193±0.020 m) than the intact side (0.178±0.023 m).

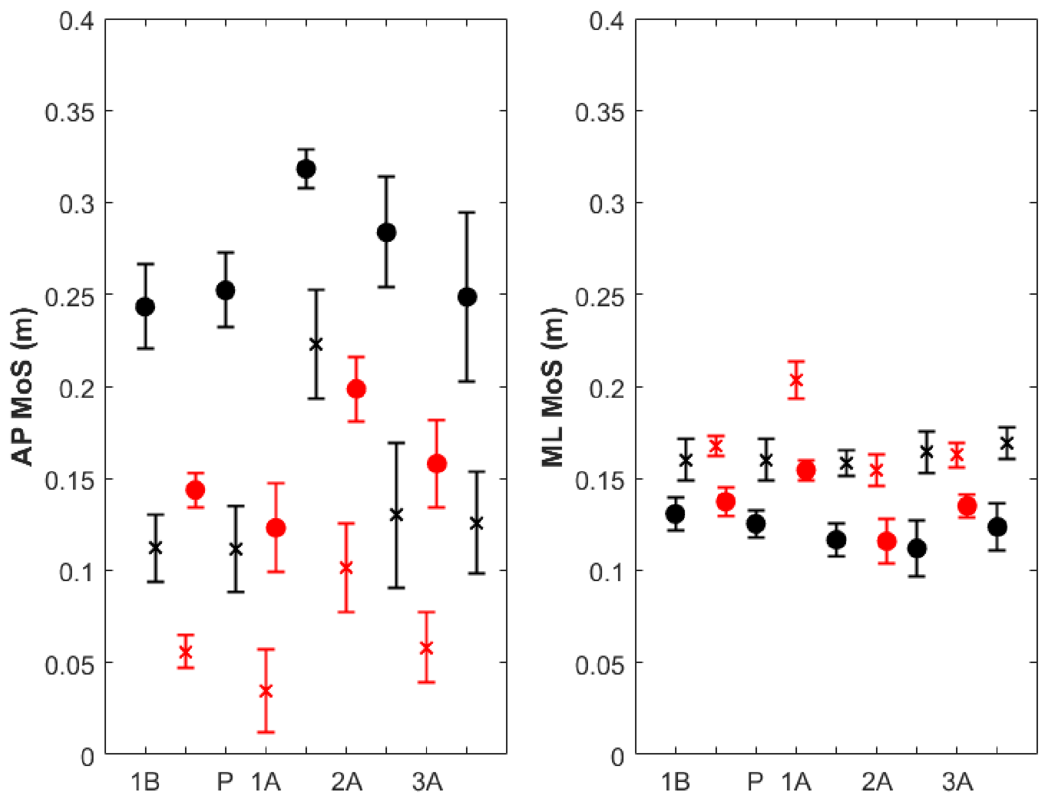

Figure 6.2 Mean and standard deviations of AP and ML MoS at initial contact for congenital prosthetic user when imposing the deceleration slips. (●) represent mean values of intact sides. (x) represent mean values of prosthetic sides. Error bars represent subject standard deviation. The black error bars represent ipsilateral steps whilst the red error bars represent contralateral steps. (B): steps before, (A) steps after.

Overall, the ipsilateral and contralateral steps for both sides followed the same pattern and showed the same reaction to recover from deceleration (Figure 6.1). In addition, the prosthetic mean AP MoS of all steps were less than the intact side. Compared to steps before and baseline, the participant exhibited increased mean AP MoS for the following three steps when the participant was challenged by deceleration slips. However, by the third ipsilateral step after (I3-A) deceleration, the participant managed to return to the status of steps before. As the mean AP MoS for I3-A of both sides (intact=0.248±0.045 m, prosthetic =0.126±0.027 m) were very similar to steps before (intact=0.243±0.022 m, prosthetic =0.112±0.018 m). The greatest deviations from steps before and baseline were seen at the initial contact of the first ipsilateral step after deceleration (intact=0.318±0.010 m, prosthetic =0.223±0.029 m). The intact and prosthetic I1-A step were approximately 70% greater than steps before whilst they were 100% greater than baseline steps. Then the AP MoS gradually started to go to the steps before values during I2-A (intact=0.283±0.030 m, prosthetic =0.130±0.040 m).

For contralateral steps, following deceleration slips, the AP MoS was decreased initially at the first contralateral steps which were during the perturbations. Intact side was decreased by approximately 14% to be 0.123±0.024m compared to 0.143±0.010 m. Whilst the first prosthetic contralateral step was affected more than the intact by 40% compared to steps before (0.056±0.010 m) to be the smallest AP MoS among all steps (0.035±0.022 m). Similar to ipsilateral steps, AP MoS for the following contralateral steps increased at C2-A and then returned to the steps before status at C3-A.

Comparable to the baseline steps, mean ML MoS of all prosthetic side steps were greater than the intact side. Generally, the MoS in the ML direction seemed to be slightly changed compared steps before and after. Mean ML MoS for all ipsilateral intact steps were less than baseline also the three ipsilateral steps were less compared to steps before. Similar to AP MoS, the participant exhibited comparable reaction for the both sides following the deceleration. The variation from the steps before was approximately less than 8% for the steps after whilst the car variation from baseline was approximately 15%. On the other hand, when the prosthetic side was challenged by deceleration, the variation form both baseline and steps before were less than 5% for the steps after. The contralateral steps followed the same pattern as in ipsilateral apart form C1-A where the mean ML mean was considerably increased by 40% compared to steps before.

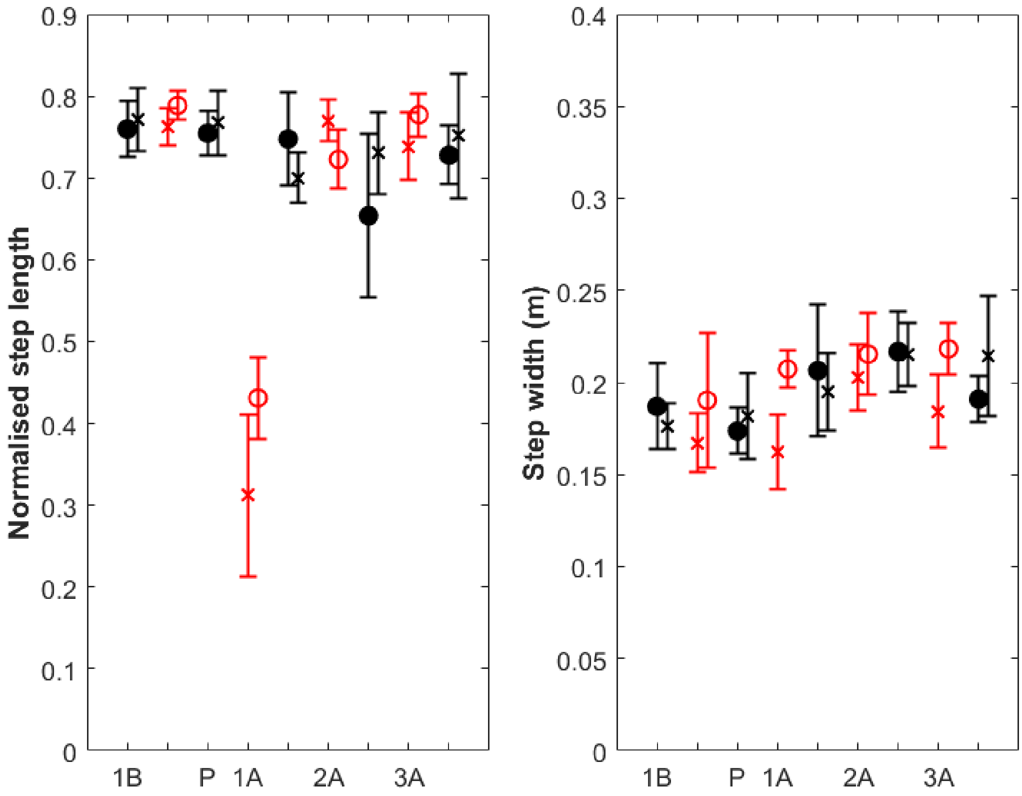

Figure 6.3 Mean and standard deviations of step length and width at initial contact for congenital prosthetic user when imposing the deceleration slips. (●) represent mean values of intact sides. (x) represent mean values of prosthetic sides. Error bars represent subject standard

deviation. The black error bars represent ipsilateral steps whilst the red error bars represent contralateral steps. (B): steps before, (A) steps after.

Participant in general, exhibited slight changes from the baseline and steps before (Figure 6.3). The participant took approximately (<10%) shorter steps on the first and second ipsilateral steps after slips for the both sides. The noticeable changes were seen at the initial contacts of the C1-A of both prosthetic and intact side. C1-A steps were shorter by at least 45% from steps before and baseline. The prosthetic side C1-A steps were shorter than the intact sides, 55% from the baseline and steps before. By the third step after the slips, the steps were comparable to the status before.

In the terms of step width, the participant took slightly wider ipsilateral steps following the deceleration (Figure 6.3). The increase from steps before and baseline was approximately between 15-20% for the three steps after. Even after three steps, mean step width was greater than steps before and baseline. During slips, the intact C1-A step was wider (0.207±0.010 m) than the steps before (0.190±0.036 m) and also wider than the corresponding prosthetic step (0.162±0.020 m). Unlike the baseline, the intact steps were wider than the prosthetic sides especially for the contralateral steps.

Comparable to baseline results and deceleration slips, mean AP MoS of the intact sides of all steps were greater than prosthetic side steps when the participant was challenged by the acceleration slips (Figure 6.4). Both sides followed the same pattern to cope with the imposed perturbation. The participant managed to recover from the acceleration perturbation after one ipsilateral step. The participant showed increased MoS in AP at the initial contacts of the first ipsilateral step (intact=0.285±0.020 m, prosthetic=0.185±0.030 m) compared to steps before (intact=0.250±0.007 m, prosthetic=0.108±0.019 m).Mean AP MoS for the second and third ipsilateral steps were very similar to the steps before. Compared to baseline, the intact steps were greater at least by approximately (80%). The prosthetic ipsilateral steps were slightly greater than baseline. The prosthetic contralateral one step before, two and three steps after were less than the baseline by almost (50%).

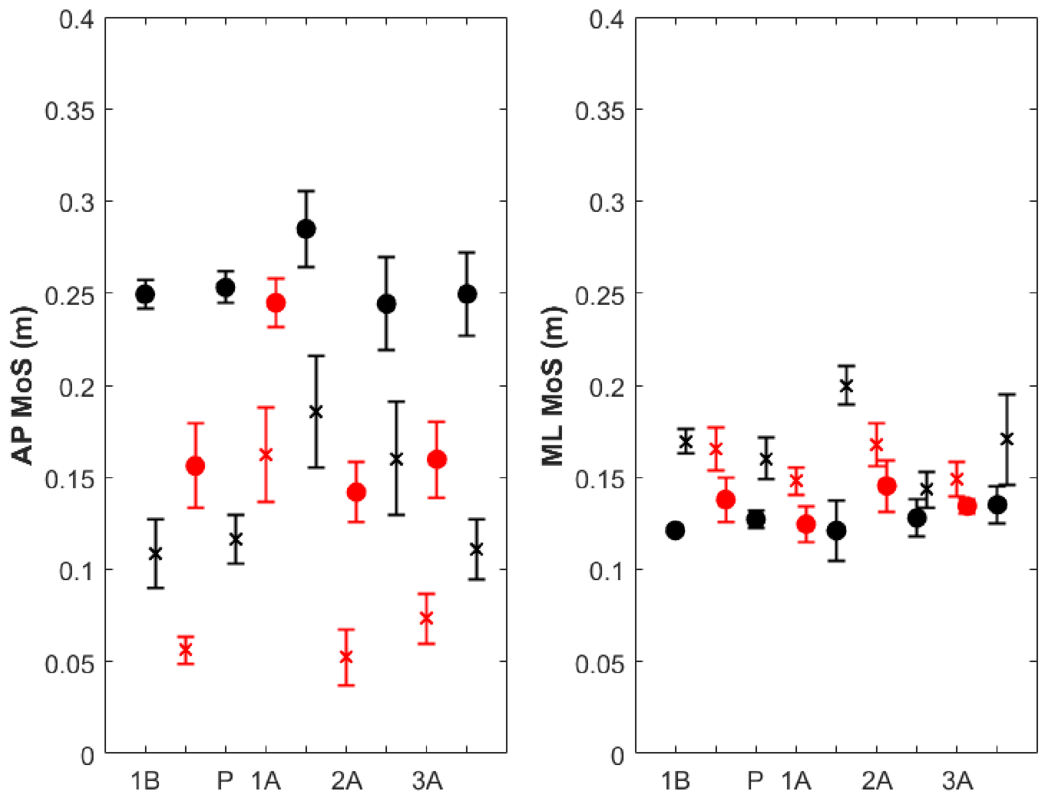

Figure 6.4 Mean and standard deviations of AP and ML MoS at initial contact for congenital prosthetic user when imposing the acceleration slips. (●) represent mean values of intact sides. (x) represent mean values of prosthetic sides. Error bars represent subject standard deviation. The black error bars represent ipsilateral steps whilst the red error bars represent contralateral steps. (B): steps before, (A) steps after.

For the MoS in the ML direction, mean ML MoS of all prosthetic steps were greater than intact steps. ML MoS for ipsilateral steps of both sides were very similar to the steps before apart from the first prosthetic after the acceleration (0.200±0.010 m). Mean ML MoS for the intact ipsilateral steps were about 0.120 m, only the variability was different among these steps for the intact steps (±0.016 m). Compared to baseline steps, the participant showed decreased in mean ML MoS for ipsilateral before and after steps. The contralateral steps notable observations were seen at the initial contacts of both sides for the first step after acceleration (intact=0.124±0.009 m, prosthetic=0.148±0.007 m) which were less than the corresponding steps before (intact=0.138±0.012 m , prosthetic=0.165±0.012 m).

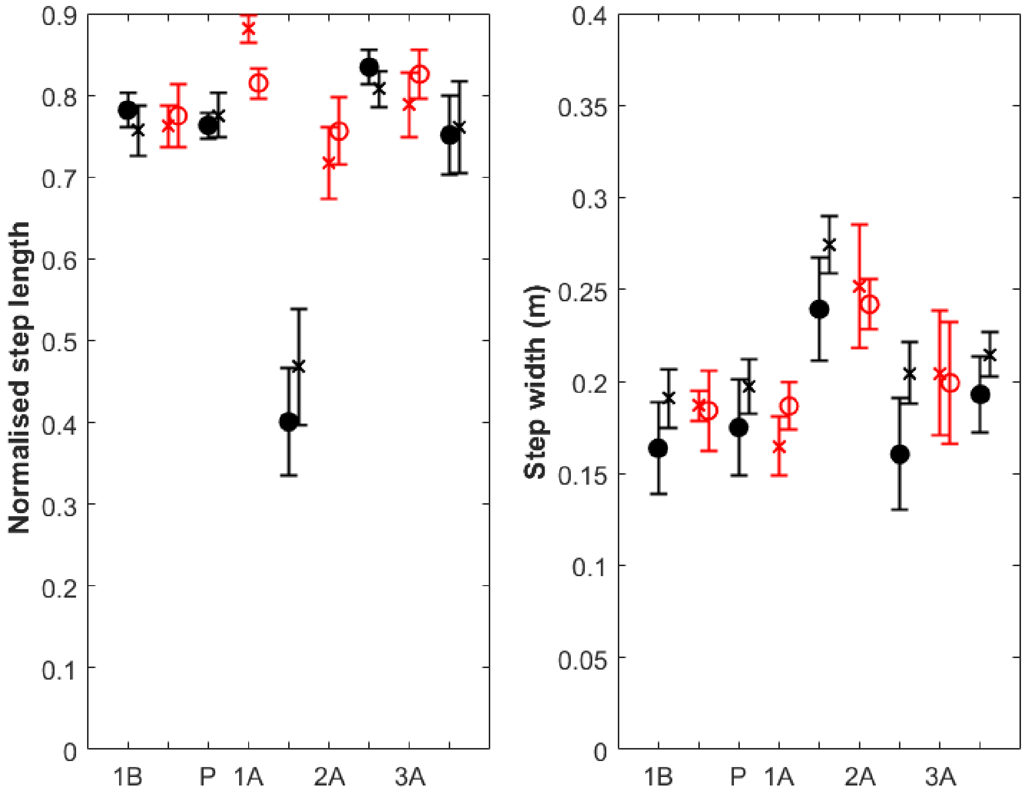

Figure 6.5 Mean and standard deviations of step length and width at initial contact for congenital prosthetic user when imposing the acceleration slips. (●) represent mean values of intact sides. (x) represent mean values of prosthetic sides. Error bars represent subject standard deviation. The black error bars represent ipsilateral steps whilst the red error bars represent contralateral steps. (B): steps before, (A) steps after.

Compared to ipsilateral steps before (intact=0.781±0.020, prosthetic=0.756±0.030), the participant initially took significantly shorter steps on the first ipsilateral intact (0.470±0.071) and prosthetic sides (0.400±0.066) when they were challenged by acceleration (Figure 6.5). Then the participant took longer steps on the second (intact=0.834±0.021, prosthetic=0.807±0.022) before returning back to the steps before at the third ipsilateral step after. Compared to baseline, mean step length for ipsilateral steps before, I2-A and I3-A of the intact side were greater. The prosthetic side ipsilateral steps I2-A and I3-A were all longer than the baseline steps.

In terms of the contralateral steps of both sides, mean step normalised length were greatly increased at first contralateral step (intact= 0.880±0.016 prosthetic= 0.815±0.018) from the steps before. Following this increase mean step length were decreased at C2-A before going to be longer compared to steps before values at C3-A.

The participant users took significantly wider steps on both first ipsilateral sides after (intact= 0.240±0.015 m, prosthetic= 0.274±0.029 m) in order to keep balance. Compared to steps before (intact= 0.164±0.025 m, prosthetic= 0.191±0.016), the step width was significantly increased by approximately (50% intact, 40% prosthetic) at I1-A. The intact and prosthetic I2-A and I3-A were slightly wider than steps before by (5%). Whilst compared to baseline, mean step width of intact and prosthetic I1-A were significantly increased by (30% and 45%) respectively. Mean step width for contralateral steps decreased by approximately (intact less than 5%, prosthetic 10%) from the steps before at C1-A. Mean step width of intact and prosthetic C2-A (intact= 0.242±0.013 m, prosthetic = 0.252±0.034 m) steps were significantly greater than the corresponding steps before . Intact and prosthetic C3-A step width were greater by (20%) than steps before.

6.3 Discussion

The aim of this chapter was to provide information about the congenital prosthetic users in terms of classification systems, causes, rehabilitation and prosthetic intervention. In addition, to provide knowledge about the dynamic stability for a congenital prosthetic user as a case study and to compare the results to groups in chapter three and four. Mostly in the clinical practice prosthetic users with congenital anomalies show increased mobility and better stability compared to other prosthetic groups. The results supported this observation.

In line with the results in chapters four and five, the case study results of mean AP MoS for the intact side was greater than the prosthetic side (Figure 6.1, Figure 6.2 , Figure 6.3 and Figure 6.4). These results agree with the interpretation of the increased MoS reflecting increased stability in that direction.

Compared to able-bodied participants in chapter three (Table 3.2), the participant of this case study showed increased AP MoS on the intact side during baseline walking (0.148 m) compared to (approximately 0.130 m) for the left and right sides. As anticipated the prosthetic side was the smallest AP MoS among all sides (0.104 m). Whilst compared to prosthetic users in chapter four (Table 4.2), the participant also showed increased AP MoS on both sides compared to (intact 0.126 m, prosthetic 0.099 m). These results suggested that the congenital prosthetic user was more stable on the intact side than the prosthetic users in the chapter four to a level that even more stable than the average AP MoS for the able-bodied subjects in chapter three.

Similar to the prosthetic group in chapter four, the participant took longer steps on the prosthetic side (0.770) than the intact side (0.745). As discussed earlier in chapter four and five, this trend is reported to be a functional compensation to maintain the stability in the AP direction (Hak et al., 2014).

The participant unlike the prosthetic group in chapter four (Table 3.2), showed significant increased MoS in the ML on the prosthetic side (0.165±0.007 m) compared to intact (0.145±0.012 m). Mean MoS for the prosthetic side and intact was approximately (0.160 m). In addition, the prosthetic side mean MoS in the ML was greater than able-bodied subjects ML MoS (left=0.122 m, right=0.146 m). These results indicate that the participant needed to have greater stability on the prosthetic side. Especially that the participant took considerably wider steps on the prosthetic side (0.193±0.020 m) compared the intact side (0.178±0.023 m). Which was wider than both left and right (0.118 m, 0.128 m) steps for able-bodied subjects as well as prosthetic and intact sides from chapter four (0.185 m).

In terms of recovering from the perturbation, the participant showed improved reaction closer to the able-bodied subjects (Figure 6.1, Figure 6.2, Figure 6.3, Figure 6.4 and Figure 6.5). The participant recovered from the deceleration and acceleration perturbations in the same way by the both sides similar to the able-bodied. Contrary, the prosthetic users in chapter four showed irregularity in recovering from the perturbations. Following the deceleration slips, the participant returned to the step before after two ipsilateral steps (Figure 6.2 and Figure 6.3). Whilst the prosthetic users did not return to the steps before balanced status even by the third ipsilateral steps (Table 4.3 and Table 4.4). The able-bodied managed to recover by the end of the first ipsilateral step after (Table 3.3 and Table 3.4).

In terms of the stepping mechanisms to retrieve stability after deceleration, the participant of this case study took shorter and wider comparable to the groups in chapter three and four. However, some variations were noticed. Able-bodied participants were affected less than the prosthetic participants. During deceleration, mean step length for C1-A steps (the first contralateral step which was during the perturbations) were the shortest among all steps. The able-bodied right and left C1-A step length (0.55) were the greatest compared to prosthetic side (0.55) and intact side (0.45) of the group in chapter four. Whilst the prosthetic side (0.35) and intact side (0.45) for this case study participant. The prosthetic side of this case study participant seemed to be most affected side among all. The involvement of this side to recover is more than the prosthetic side for the group in chapter four. For example, prosthetic C2-A step of case study participant (0.8) was longer than steps before (0.75) as well as intact side (0.71). This trend is comparable to the right and left sides C2-A step (0.72) of the able-bodied. Unlike the group in chapter four this step was shorter (0.65) than intact side (0.72) and similar to steps before.

The participant of this study mostly took wider steps on the intact side than the prosthetic side expect I2-A steps in which the step with of both sides were equal and I3-A which the prosthetic sides steps were wider than intact (Figure 6.3). This was comparable to able-bodied participants which the right side steps were always wider than left side (Figure 4.13). In addition, this was opposite to the group in chapter four in which step width was comparable between sides.

Similar to the overall results in chapter three and four, the acceleration perturbation was less challenging than deceleration. The participant managed to recover from acceleration perturbation after one ipsilateral step. The participant of this case study took shorter and wider steps comparable to the recovery mechanism for groups in chapter three and four. The participant exhibited step length of (intact=0.470±0.071 which was 40% less than steps before, prosthetic=0.400±0.066 was 50% less than steps before). Also exhibited step with of (intact=0.240±0.015 m which was an increased by 50% from steps before, prosthetic= 0.274±0.029 m 40% which was an increased from steps before).

Able-bodied participants step length after acceleration was (left and right sides =0.608±0.099, which was a decreased by around 16% from steps before) whilst step width was (left=0.145±0.047 m, right=0.158±0.047 m which again was around 17% more than steps before). The prosthetic users in chapter four on the other hand exhibited step length of (intact=0.503±0.115 which was 10% shorter than steps before, prosthetic=0.543±0.150 which was around 15% shorter than steps before. For the step width the participants step width (intact=0.235±0.066 m which was 20% wider than steps before, prosthetic=0.226±0.051 m which was 15% wider than steps before).

During acceleration, mean step length for C1-A were the longest among all steps. The able-bodied right and left C1-A step length (0.803±0.064) were the greatest compared to prosthetic side (0.55) and intact side (0.45) of the group in chapter four. Whilst the prosthetic side (~0.9) and intact side (~0.8) for this case study participant. The prosthetic side of this case study participant seemed to be most affected side among all. Again, the involvement of this side to recover is more than the prosthetic side for the group in chapter four.

Unlike the deceleration, the participant of this study mostly took wider steps on the prosthetic side than the intact side (Figure 6.5). In addition, this was opposite to the group in chapter four in which step width was greater on the intact side.

The results of this case study demonstrated that the prosthetic intervention for individuals with lower congenital limb anomaly or defect is sufficient to achieve satisfactory level of walking stability and to enhance the quality of life. The prosthetic intervention can replace and minimise the need for lower limb amputation.

The results also supported the clinical beliefs that prosthetic users with congenital anomalies have increased mobility and better coping strategies to unbalanced situations.

Nonetheless, it should be noted that this case study only included one participant with relativity fewer overall complications that can be seen in other lower congenital limb anomaly individuals. The partial absence of the upper limb might have affected the balance of this participant. Previous work by Tung et al. (2011) showed that upper limb may play vital role in frontal plane stability. The results of this case study therefore only indicative of this user and cannot be generalizable.

Hence, further investigations for this particular group are needed to help in fully understanding the adopted walking patterns and how these might differ from prosthetic users with other aetiologies. Altogether to help in providing the optimised rehabilitation programmes and prosthetic intervention.

6.4 Conclusion

There is no single prescription and/ or method for congenital lower limb deficiency/anomaly. The prosthesis must be built and customized from child to child who will present with an individual functional deficit which must be managed appropriately in each case. Children also encounter growth changes and therefore children require more frequent follow ups and prosthetic amendments. The support of parents and other family members is required. Some of children with congenital lower limb deficiency, limb amputation may be advised and required.

Paediatric limb deficiency management requires anticipatory efforts. Because the child will go through numerous development stages and various which need to be considered. These include; age, terminal overgrowth, growth spurts, leg length discrepancies, body image and guidance for the family. Prosthesis weight, cosmetic, comfort and prosthesis donning and doffing are important aspects that should be considered to maximum the acceptance of the prosthesis.

In terms of the stability and walking, individual analysed in this case study demonstrated improved MoS results compare to prosthetic users in chapter four. This was obvious in the AP direction. The AP MoS of the intact side for individual analysed were greater compared chapter three and four participants. The recovery mechanisms from deceleration and acceleration perturbations were similar in fashion to able-bodied and prosthetic users participants of chapter four. However, the participant showed improved reaction closer to the able-bodied subjects. The participant returned to the stability status of the steps before perturbations in a smaller number of steps compared to participants in chapter four. In addition, the involvement of the case study prosthetic side was less limited compared to the group in chapter four.

Due to the individual presentation of such prosthetic users it remains uncertain if results would be in any way generalizable even in a study of appropriate size. However, results may indicate that prosthetic users with congenital anomalies have greater stability compared to other prosthetic groups. This corresponds to anecdotal clinical evidence and may be useful in providing insight that may help to provide optimised rehabilitation programmes and prosthetic intervention.

6.5 Clinical implications summary

The prosthetic user with lower limb congenital related limb loss/anomaly of this case study showed enhanced stability and recovery mechanism in response to slips compared to other prosthetic users with different aetiologies (Chapter 4). It is challenging to draw specific conclusions from such a case study since results may not always be generalizable. Each prosthetic user is individual. Prosthetic users of congenital aetiology may also have very individualised (almost unique) presentation. The prosthetic care and rehabilitation for this group should not be always similar with the same concepts and programmes. Nevertheless, results of the case study do support anecdotal opinions about how congenital prosthetic users do seem adaptable, active and very stable. Further research with appropriate inclusion criteria of such a specific group would be of interest. However, due to the rarity of such conditions it may be extremely challenging to recruit sufficient participants.

7 General study conclusions, clinical implication, limitations, and future work

The findings in this study highlight the importance of having a better understanding of the dynamics of prosthetic users. Previously in prosthetics there has been comparatively little literature addressing dynamic stability. Nonetheless, the results for those studies regarding this group were found to be inconsistent. This research has highlighted that better understanding of stability is clearly important for prosthetic rehabilitation and optimising outcomes.

This general discussion chapter addresses how this thesis has made important contributions in exploring the dynamic stability of lower limb prosthetic users. Based on the findings and on limitations in the study, clinical implications of results were provided. In addition, recommendations were provided for the future scientific research and development.

This thesis aimed to discuss the dynamic stability for the prosthetic users. Based on the literature (Chapter one) it was found that further work was required towards better understanding the stability of this group. Several measures were previously adopted to quantify the stability. Margins of stability (MoS) was found to be simple and reliable tool to assess the stability. The MoS results of this study (chapters three, four, five and six) supported this. The literature (chapter one) also showed that slips were more challenging to stability, therefore they were selected to challenge the stability of participants in this study. A previous method of perturbation that was developed in the University of Strathclyde showed promising results for able-bodied individuals and hence was selected. The method included sudden change of the walking speed on each side by the means of acceleration and deceleration. The slips were imposed using the advanced CAREN system.

7.1 Contributions in exploring the use of the mechanical perturbations to assess stability for adult able-bodied

In literature, MoS is typically assessed only in one direction either in AP or ML directions. This may not address the entire picture of the stability status for individuals while walking and facing unbalanced situations such as slips. Additionally, this may not help in making comprehensive suggestions for less able-bodied people. Accordingly, the method followed in this thesis included both AP and ML directions. In addition, the recovery step/s were presented in more detail than in other published studies.

The MoS following slip perturbations was presented for the three steps of both contralateral and ipsilateral sides. This approach provides important insight about the dynamic stability from step to step and allowed tracking of changes to help understand how stability was recovered. Other studies generally present the MoS or the stability by averaging several steps after the perturbation which may mask such changes.

Results from Chapter 3 revealed that able bodied participants required two steps; one ipsilateral step after and one contralateral step after perturbations to recover. The results also agreed with the results of the original paper (Roeles et al., 2017) which demonstrated that deceleration slips were more challenging than acceleration as more steps were needed to recover from the perturbation.

The greatest effects in MoS were seen in the AP direction which is understandable taken that the perturbations were imposed in the AP direction (section 3.3). However, some noticeable changes were also seen in the ML direction (section 3.3). In terms of recovery mechanisms, able-bodied participants took wider and faster steps to maintain the balance regardless of the applied perturbation. The step length was increased after deceleration whilst decreased after acceleration. The clinical implication of these results may include improved awareness that the stability of individuals may benefit from taking wider and faster steps in response to slips regardless of their direction. In order to avoid a fall as a result of backward slips (acceleration perturbations), shorter steps can help in maintaining forward balance (MoS). Whilst for forward slips (deceleration perturbation), longer steps are required to avoid a fall. These insights, could potentially inform and guide clinicians in developing fall prevention programmes.

The idea of which side is more stable for able-bodied is investigated in Chapter 3. Results demonstrated that the stability of the dominant side is greater than the other side (section 3.3, tables 3.2, tables 3.3 and table 3.4). In addition, the contribution of the dominant side in terms of recovering from the slips was greater than non-dominant side. Again, such information is helpful in clinical practice. The results may help in developing fall prevention programmes. These results raised a question : do individuals tend to rely more on the non-dominant limb while performing certain activities? For instance, kicking a ball, right sided individuals tend to kick the ball using the right side, trusting the left side to maintain their stability while doing this, acting like an anchor (Manolopoulos et al., 2006). Arguably, this may indicate that in this case the left (non-dominant) side is more stable in a person with right sided dominance (and vice versa). Further investigations are required to fully understand the concept of stability. Presumably, it may not be an issue for able-bodied as both sides were found to be able to recover from the slips efficiently. However, for less able-bodied such as prosthetic users it is important to have a clear picture of the stability. Since this idea was not one of the aims of this thesis, the previous dominant side for the prosthetic users prior to amputation was not recorded. Further future studies in this direction may help can produce important insight.

7.2 Contributions in exploring the use of the mechanical perturbations to assess stability for below the knee prosthetic users

In chapter four, the dynamic stability of prosthetic users was studied.

Initially, stability was assessed for the whole group. This work contributed to the literature by providing evidence regarding the stability of this group in several aspects. Firstly, the effects of the mechanical perturbations slips method on prosthetic users were not previously assessed. The results showed that both deceleration and acceleration slips were challenging for this group. In addition, it was found that the involvement of the prosthetic side in recovering from the perturbations is limited as anticipated. Since, all participants managed to recover from the perturbations using a combination of stepping mechanisms, without reporting any fall, this can guide the prosthetist and clinicians during prosthesis fitting and training. The stepping mechanisms employed included walking with wider and faster steps with no significant change in step length. These helped the prosthetic users to recover from slips and to avoid falling, therefore individuals following prosthesis fitting can be trained to react using these recovery adaptations to maintain their stability and to avoid a fall (section 4.3.3).

The results of this chapter also contributed to have better understanding of the potential benefits of having a hydraulic ankle prosthetic foot. It was found that individuals fitted with Pro-Flex foot exhibited improved stability. The results were in line with previous results that assessed this novel foot. On the other hand, the results of the MoS when walking with the Echelon VT and Elan feet disagreed with the presumption that. individuals fitted with those feet would exhibit increased MoS (i.e. increased stability). The reasons of these results may be due to lower activity for the individuals fitted with them. This raised the question regarding underuse of the prescribed prosthetic components.

Individuals should be clinically assessed prior to provision of certain prosthetic components in order to make sure that they will use the provided component to its full potential. The assessment should take into account the individual’s medical history, medication, functional mobility and social history.

Based on the assessment, clinically fitting an advanced prosthetic foot such as Pro-Flex for an individual who can utilize it to its full functional capacity can result in improved walking and activity performance especially in terms of stability and recovering from slips .

In addition, ESAR feet demonstrated a satisfactory performance compared to the other feet that have ankle motion in relation to recovering from perturbations by preserving the margin of stability. In prosthetic clinics where the advanced feet are not available, the ESAR can still be the foot of choice.

The thesis findings may suggest that certain foot types may be capable of reacting/adapting better to slips. This knowledge with further investigation should allow better informed prescription of such components.

Finally, in case that a motion capture controllable treadmill, such as the CAREN system, is available in the prosthetic clinic, this study protocol can be use in evaluating the provided intervention. Finding an appropriate method to evaluate the outcome of the provided prosthetic care is one of the areas that still needs further investigation. In addition, it can be used for anti-falling training.

7.3 Contributions in exploring the potential effect of foot alignment changes and prosthesis length on stability.

The effect of changing the foot position in relation to the prosthetic socket in the sagittal plane as well as prosthesis length on dynamic stability has not been evaluated before. Chapter five contributed to the literature about the potential effect on stability of setting the foot in extra dorsiflexion and plantarflexion from the optimal alignment and the effect on stability of providing longer and shorter prosthesis.

The results indicated that different prosthetic alignment and length may affect the MoS significantly. In particular, the short prosthesis was found to be more challenging than other changes in this study. Therefore, special care should be taken to get the correct prosthetic length during prosthetic fitting and delivery. Many prosthetic limbs at higher levels of amputation (transfemoral prosthetic users) are generally set up to be up-to 10mm shorter to facilitate ground clearance. Previously, the effect on the MoS may not be fully understood and this group would benefit from future research using this research protocol.

Further investigation using an appropriate participant sample size is needed to verify these results.

7.4 Contributions in exploring the stability for congenital related limb loss prosthetic users

In chapter six, the dynamic stability of a person with congenital below knee anomaly was discussed. This aspect was not previously assessed, leading to a paucity of evidence for this group. This may be especially important in countries where congenital related limb loss/anomaly is still seen as major cause for limb loss particularly for those aged under 18 years old. Improved understanding and future research may have potential to enhance the level of prosthetic/rehabilitation services which subsequently may improve the quality of life for this group.

The participant of congenital aetiology demonstrated improved MoS results compared to both groups in chapter three and four. These results correspond to the medium to high functional appearance commonly seen during clinical observation of people with congenital limb anomaly. Results also indicated that clinically this group may need/ benefit from advanced rehabilitation programmes, such as physical fitness and rehabilitation for sports and/or competition. In addition, for selection of prosthetic components, there may be more benefit to this group to be fitted with more complex components such as hydraulic feet. Therefore, such results may guide the clinician and prosthetists in aspect of component selection. Contraindications include the additional space requirements for such components. Consequently, designers need to think about design of appropriate ankle joints with limited build-down space for improved reaction to slips within this active group.

Despite these improved results, due to the individuality of congenital presentation (case study), results may not be generalizable but do provide interesting insight which may be useful clinically. The prosthetic care and rehabilitation for this group may not be always similar with the same concepts and programmes. The aim to achieve the 'typical' gait stability and walking pattern is not always possible since every case can be unique.

7.5 Contributions towards understanding the meaning of the MoS value

Based on the literature in chapter 1, it was found that the interpretation of the MoS is yet not clear and consistent as some studies suggested that lower MoS indicates lower stability whilst other suggested the opposite. The results of chapters 3-6 indicated that increased MoS was a sign of increased stability. This agrees with the original interpretation when the MoS measure was first introduced by Hof et al. (2005).

In chapter 3, in the able-bodied control group, the MoS of the right sides for the control was greater than the left side. Since all participants were right sided it is what would be expected to be noticed that the dominant side to be more stable than the non- dominant side.

In chapter 1, it was found that the comparison results between able-bodied individuals and prosthetic users were inconsistent and un-clear. The results of this thesis contribute to the literature regarding this comparison by providing a better understanding. The recovery mechanisms of all participants were very similar; using wider and faster steps regardless of the imposed perturbations. Additionally, both groups showed decreased step length after acceleration slips. However, step length was greatly increased after deceleration for the able-bodied participants whilst it was approximately the same as steps before the perturbation for the prosthetic users.

Whilst there was similarity in stepping mechanisms, several differences indicate that able-bodied participants were more stable than prosthetic users. The MoS of the prosthetic side were found to be the lowest among all limbs including the intact side, left and right sides of the able-bodied which is what would be reasonably anticipated in such comparisons. In addition, prosthetic users showed increased irregularity between steps (step-to-step) compared to the controls. Also, the results showed that the prosthetic users could not manage to return fully to the baseline status as the controls could. More corrective steps were needed to recover from slips. Finally, the acceleration was more challenging to the prosthetic users than the controls.

The results for the participant in chapter 5 also support this interpretation. The mean AP and ML MoS for intact side were significantly greater than the prosthetic side for most changes apart from AP MoS for plantarflexion changes in baseline and self-paced mode as well as ML MoS for plantarflexion in self-paced mode. In addition, the greatest AP stability of both intact and prosthetic side were seen during the optimal alignment.

The results of participant in chapter 6 were as anticipated and in line with the clinical appearance of this group when they attend the care unit for prosthesis services.

7.6 General limitations and suggestions for future work

In addition to the potential limitations discussed specifically in each chapter, some general points were also noted. None of the prosthetic users dynamic MoS studies discussed or addressed the potential effect of the walking speed. The treadmill walking speed of the participants was selected by; self-selected fixed speed by taking the average of two reported comfortable speed points similar to the study protocol of Houdijk et al., (2018), walking at a fixed speed scaled to leg length (Beltran et al., 2014) or using self-paced mode (Hak et al., 2014). Several factors may explain this, it would be difficult to select a fixed speed that suits all participants for example, 1 m/s. It may be considered too fast for some prosthetic users (n=3 of this study who reported CWS at less than 0.86 m/s). Or it may be significantly slower for other prosthetic users (n=6 participants in this study who walked at more than 1.25 m/s). Also, such a walking speed may be significantly slower than CWS for the able-bodied who all managed to walk more than 1.2 m/s in this study. Hence, the slow speed may not trigger a considerable reaction. In addition, walking at slow speed may also affect the performance of the prosthetic feet. Prosthetic feet need to be appropriately loaded in order to perform and provide the sufficient push-off (Müller et al., 2019, Powers et al., 1994). Reduced push-off would increase the load on the intact side (Morgenroth et al., 2011). No significant differences were noticed in the frontal plane parameters during the unperturbed 1 m/s session for the able-bodied participants (Table 3.2). Also, consistent results were seen when they faced the highest intensity perturbations walking at 1 m/s. The changes in MoS were seen in the AP direction, however the participants followed the same pattern compared to walking at CWS, and also the 1 m/s was less challenging on the contralateral side than CWS. The deceleration perturbations were more challenging than acceleration perturbations. The participants managed to recover from the slips after one ipsilateral step. The results of the prosthetic users who did manage to walk at a speed of 1 m/s were also consistent. Nonetheless, further investigation regarding the walking speed effect of the MoS is needed. To achieve this, a larger study sample size is required.

Although the perturbation types and intensities were presented in a randomized pattern in this study, the possibility of the participants to be familiar with the upcoming perturbations is still possible. This could be seen as potential limitation of the protocol that might have an effect over the results. If so, it is not clear to what extent the participants might have been affected. If this method will be used in the future in studying the effect of sudden perturbations of a given group, adding a different type of perturbation can potentially overcome this limitation.

On the other hand, this potential limitation can be used to improve the stability of participants. Knowing that participants, especially those with a history of falling, can proactively enhance their stability when facing repeated perturbations, this can be used as rehabilitation tool. When instrumented treadmills are available in a rehabilitation facility, applying this study protocol can play a vital role in enhancing the stability of the participants. Thus, the knowledge of the basic use of such machines can be a helpful skill for the rehabilitation team including prosthetists and physiotherapists. Further research in this field may reveal important insights.

Initially, the slips were aimed to be applied at exact instance of initial contact. However, there was a delay between the software and the hardware. The processing following the motion capture systems' marker tracking, identification of initial contact event and subsequent commend signal to the treadmill introduces a delay before the initiation of the belt speed change.. The slips were triggered at about 15% of the gait cycle. Nonetheless, the perturbation onset was constant for all participants therefore this limitation can be acceptable. On the other hand, this has raised another question regarding the effect of the onset time on gait stability. Hence, further investigation is needed to examine the effect of different onset time on the walking pattern and stability.

Whilst some stability studies have included above the knee prosthetic users, despite being the second most common level of amputation (Sabzi Sarvestani and Taheri Azam, 2013), most work has focussed on trans-tibial amputees. Adopting the study same perturbation method for above the knee prosthetic users would also be advantageous particularly in relation to length of prosthesis. Providing insights about the performance of this group may help in developing and providing improved prosthetic and rehabilitation services.

It was noted that in the literature, stability studies adopt either MoS alone or the gait kinetics mainly GRF. It would be beneficial to see how individuals cope with sudden slips and trips in terms of kinematics (joint angles). For instance, would an advanced prosthetic foot with enhanced ankle movement enable better reaction to slip in terms of RoM? How would the RoM play a role in the recovery mechanism? The gait kinematics and kinetics combined with the concept of MoS could therefore reveal important information in two fields. Firstly, from prospective of biomechanics it would clarify and strengthen the concept. Secondly, it would show how prosthetic users control their walking in order to maintain stability. Hence, further research is needed in this topic.

Finally, individuals require a familiarization period for treadmill walking (Schellenbach et al., 2010), limiting the time frame over which the data is selected could impact the findings. However, it is unclear to what extent treadmill familiarization affects the selected variable in this study. Further investigation that correlate the stability by means of MoS to the familiarization period for treadmill are therefore needed.

The outcomes of this study are novel and have potential to improve the quality of life for lower limb prosthetic users (acquired and congenital). It is envisaged that greater understanding of different adaptation strategies of the human body may help influence future prosthetic treatment, prescription, alignment, and potentially component design.

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Appendix I

Supplementary material: International Standard as a description method

The term ray may be used to describe the metacarpals or metatarsal with its related phalanges. The level of a bone deficiency mentioned starting from the most proximal to the most distal bone.

In case of the girdle's bones (shoulder and pelvic), the totally or partially absent bones mentioned. For the partial types the missing bone part should be named. The same way used to describe the humerus, radius ulna, femur, tibia and fibula bones, but it for the partially types, the estimated fraction of the missing bone as well as the position (such as proximal or distal) of the deficiency mentioned.

For hand and foot bones, the description includes which carpals, tarsals, metacarpals, metatarsals and phalanges are totally missing and which are partially. Nevertheless, in case of a bone that has not been stated as totally missing or partially, the hypoplasia presence should be stated if possible. The names of the hand digits can be used to describe them such as ring finger (finger number 4).

Appendix II

Supplementary material: PFFD classifications and management

Gillespie & Torode suggested classification of majority of patients into two group which can be distinguished on clinical bases.

Group I known as (Congenital short femur), patients in this group characterized with;

- The effected leg is not short as in true PFFD (Group II) as well as the foot is roughly at the level of mid tibial of the sound limb.
- The leg is set in flexed, abducted, and laterally rotated position.
- There is laxity in anteroposterior of the knee accompanied with valgus deformity.
- The X-ray image shows, varus and retroversion of the femoral head and neck.
- It is possible for the knee and the hip to be functional in some cases at least, equalization in leg is possible as well.

Group II known as (True PFFD), patients in this group characterized with;

- A weak cartilaginous link exists between the head of femoral the proximal shaft of femur.
- Extremely short thigh, overall discrepancy is 35%-50%.
- The leg is set in abduction and external rotation position.
- There is Flexion contracture in both hip and knee.
- The surgical management aimed to help the prosthesis fitting.

Amstutz and Wilson classification for PFFD; it is similar to the Aitken with further subdivision if the Aitken type A; in the classification, the type A is divided into 1 and 2. The type 1 is used to indicate PFFD patient with the present of milder form as well as simple femoral shortening and coxa vara. Meanwhile type 2 is used to indicate PFFD patient with the present of subtrochanteric pseudarthrosis.

PFFD management

In type 1 or 2; sub-trochanteric valgus osteotomy has a good result in improving the coxa vara deformity (Amstutz, 1969, King, 1969, Meyer et al., 1971).

In type 3; the presence of pseudarthrosis showed difficulties in management, many options can be done, in the study of (Panting and Williams, 1978), a procedure of excision the cartilaginous element as well as a fixation of the bony elements in contact by means of a small rod or nail-and-plate, an union occurred. Moreover, in order to provide bone-to-bone contact the cartilaginous element should be excised.

In type 4, surgical exploration according to (Panting and Williams, 1978), has been unrewarding.

In type 5, since the acetabulum is absent, femoral reconstruction can be difficult. Alternatively, many opinions have been offered.

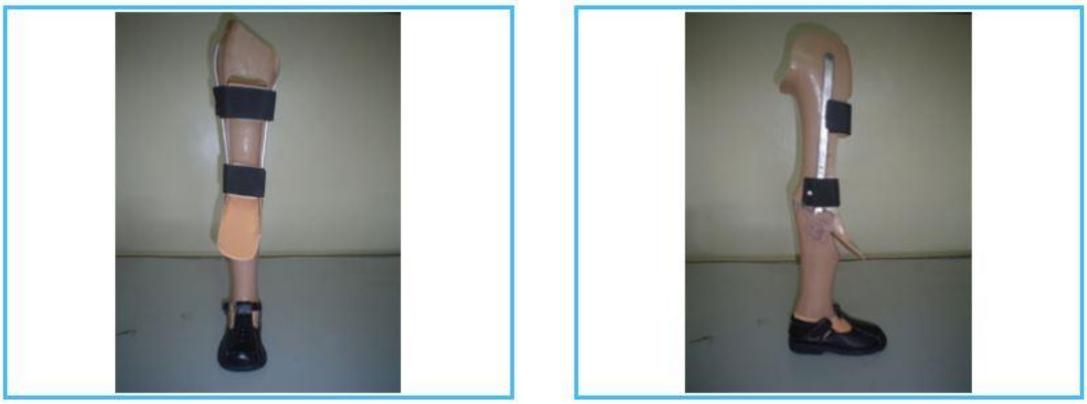
Panting and Williams (1978) options were to neglect the bony defect (do nothing surgically) and depending on the stability that came from the increased muscular development as a result of active weight-bearing. However, Femoro-pelvic arthrodesis may be indicated in gross instability exists and, in the case, that the femoral proximal migration threatens to penetrate the skin.

Mal-rotation and proximal inadequate musculature. These difficulties are related to each other. Most of the PFFD patients show special posture which is flexion deformity (30 to 40 degrees), lateral relation (45 degree or more) and sometimes abduction deformity at the hip.

Appendix III

Supplementary material: Orthoprosthesis

In some cases, individuals with congenital lower limb loss are fitted with a unique design. A device called orthoprosthesis which is hybrid solution to get the benefits of both orthosis and prosthesis. An example of this device is shown in the figure below. This term was first introduced by (Marquardt, 1963). The device aims to modify structural and functional conditions of the partially or totally missing skeleton or neuromuscular part. The orthosis provides the required support for example, instable knee. Whilst the prosthesis compensates the missing part or/and function for example, the foot. This device was introduced as alternative solution to the amputation and to preserve as much as possible of the skeletomuscular system (Wenz et al., 1998, Boonstra et al., 2000). However, orthoprosthesis manufactory and fitting can be challenging. Since almost each case is a unique case. Each orthoprosthesis is made based



on the specific malformation of the individual at the same time taking into consideration that the device must be easy to use (Wenz et al., 1998). In addition, other complications could influence the decision of which rehabilitation plan should be adopted. These included skin problems, osseous overgrowth and bone growth (Boonstra et al., 2000).