BIOMECHANICAL ANALYSIS OF TRANSFEMORAL AMPUTEE'S SPRINT RUNNING AND BLOCK START

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ABSTRACT

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Although a large number of studies about human locomotion have been reported, little is known about how to run with a prosthesis limb. The purpose of the study was to examine the difference between the healthy and prosthesis leg and between different prostheses during the sprint start and maximal running speed. One top level male sprint runner (100 m personal best 13.11 s) with the unilateral transfemoral prosthesis participated in this study. Two specific sprint prostheses (A: mass = 2.3 kg and B: mass = 2.7 kg) were used. Ground reaction forces (GRF) were recorded together with high-speed video recordings that were used for three-dimensional analysis. The first six steps from the sprint start and a step cycle at the maximal speed were measured for both healthy and prosthesis legs.

The highest running speed measured was 7.45 m/s, which was acquired by prosthesis A, although the differences between prostheses in measured parameters were minor. The results showed great asymmetry between the healthy and the prosthesis legs. The knee joint kinematics of the prosthesis leg differed from the sound limb as the prosthesis leg must land with the knee joint fully extended. The upper body was bent backwards and the prosthesis hip extended to assist in landing with the straightened leg causing a shorter step length on the prosthesis leg. The main compensation method for the subject to be able to run was to adjust the swing time to contact times of each step cycle making the step cycle time constant. Horizontal and vertical GRF during impact and push-off phase were smaller for prosthesis than for the sound leg as a consequence of the lack of force producing muscles and limited prosthesis properties. The length of the prosthesis was too long and caused a sideways swing of the prosthesis. As a solution the prosthesis length was shortened and the subject was advised to do strength training to achieve a fore foot contact on the sound side. In addition to these on an upright body posture was suggested to improve the running technique of the subject. A few months later the subject made his new record with a 0.26 s improvement.

Keywords: Transfemoral prosthesis, amputee, sprint running, block start, ground reaction force, 3D-movement analysis

SUOMENKIELINEN ABSTRAKTI

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Ihmisen liikkumiskyky on ollut monen tutkimuksen aiheena, mutta alaraaja-amputoitujen pikajuoksusta on vain vähän tutkittua tietoa. Tämän tutkimuksen tarkoituksena oli selvittää proteesijalan ja terveen jalan sekä kahden eri juoksuproteesin välisiä eroja pikajuoksun lähdössä ja maksiminopeusvaiheessa. Koehenkilönä oli yksi huipputason miespuolinen reisiamputoitu pikajuoksija, jonka 100 m ennätys oli 13,11 s. Tutkimuksessa käytettiin kahta erityistä juoksuproteesia (A massa = 2,3 kg ja B massa = 2,7 kg). Reaktiovoimat ja lihasaktiivisuus tallennettiin yhdessä videodatan kanssa kolmiulotteista liikeanalyysiä varten pikajuoksun telinelähdöstä kuudelta ensimmäiseltä askeleelta ja maksiminopeusvaiheesta.

Suurin juoksunopeus 7,45 m/s mitattiin proteesilla A, joskin erot proteesien välillä olivat pieniä. Tuloksista havaittiin suurta epäsymmetriaa terveen ja proteesijalan välillä. Proteesijalan polven liikeradat erosivat terveen jalan liikeradoista proteesipolven täyden ojentumisen vuoksi askelkontaktin aikana. Ylävartalon heilahdus taaksepäin ja proteesijalan lonkan ojentuminen olivat strategioita, jotka helpottivat astumista proteesijalka ojennettuna, tosin ne aiheuttivat proteesijalan askelpituuden lyhentymisen. Tärkein kompensaatiomenetelmä juoksun mahdollistamiseksi oli kyky rytmittää heilahdusaika kontaktiaikaan, jolloin jokaisen askelsyklin pituus pysyi vakiona. Jarrutus- ja voimantuottovaiheen horisontaali- ja vertikaalikontaktivoimat olivat pienempiä proteesijalalle kuin terveelle jalalle voimantuottolihasten puutteen ja proteesien rajoittuneiden ominaisuuksien vuoksi. Proteesin pituus oli säädetty samaksi kuin oikean jalan pituus seistessä päkiöillä. Koehenkilö ei pystynyt juoksemaan päkiällä m. triceps suraen lihasheikkouden vuoksi. Liian pitkäksi säädetty proteesi aiheutti ongelmia, koska koehenkilö joutui heilauttamaan proteesijalan sivukautta. Ratkaisuna näihin ongelmiin proteesin pituutta säädettiin lyhyemmäksi ja koehenkilöä neuvottiin tekemään voimaharjoittelua, jotta hän pystyisi juoksemaan terveen jalan päkiällä. Lisäksi häntä kehotettiin pitämään vartaloa pystylinjassa. Muutaman kuukauden päästä palautteen antamisesta hän paransi ennätystään 0,26 s.

Avainsanat: Reisiproteesi, amputoitu, pikajuoksu, pikajuoksun lähtö, kontaktivoimat, kolmiulotteinen liikeanalyysi

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1 INTRODUCTION

An ability to walk and return to normal life is the objective of rehabilitation after an amputation. Loss of a limb necessitates new adaptations in motor control and leads to a unique gait pattern. Although running and sprinting are big challenges, many of the lower-leg amputees have the desire to practice and compete in the same way as the able-bodied people. A lower-limb amputee sprinter has run the 100 m in under 11 s and this level of performance is reflective of not only a highly trained athlete but also an advanced prosthesis development.

As sprint running is a very difficult task for a transfemoral amputee there is a limited number of amputee sprint runners and studies concerning them. Only a few studies have been reported about transfemoral amputee's sprint running gait and no reports were found on block start.

The present study aims to clarify the sprinting technique and the level of asymmetry in a performance of a top level Paralympics transfemoral amputee. Comparison between sound and prosthesis legs was done with two different prostheses. As every amputee has a unique gait and even the sound side kinematics is not simply comparable to the able-bodied subjects, there was no reason to compare this subject’s data to any other amputee or an able-bodied sprinter. Therefore, a case study settings and a descriptive study was done.

2 LOWER-LIMB AMPUTEES

2.1 Different amputations

An amputation is defined as removal of either a limb or a body part. The Latin word *amputare* means cutting, shortening, switching or pinching. An amputation is made when an injured limb can not be recovered to function after trauma or disease. (Solonen & Huittinen 1992, 21.) Most of the amputations are performed on elderly people because of the complications brought on by diabetes or other vascular diseases (Alaranta et al. 1998). It was reported that over 80 % of the lower-leg amputations were due to ischemic vascular disorders, mainly diabetes, during the year 1995 in Finland. Amputations were done on 366 people, with an average age of 71.4 yrs, and the amount of transtibial and transfemoral amputations were 101 and 106, respectively. The incidence of amputation was 28 / 100 000 people which is similar in other European countries. (Alaranta et al. 1998.)

The number of amputees is much higher, up to 2.5 million, in North America than in Finland. The number of annual amputations is about 150 000 cases in the United States. The vascular diseases are the main causes for amputations, but almost 50 % of amputations in the United States are caused by trauma, tumors or infections. Commonly, these causes and congenital disorders affected people under 45 years old. (Smith & Fergason 1999.) Alaranta et al. (1998) mentioned trauma to be the cause for lower-leg amputation only in 4.1 % of the cases in the year 1995 in Finland.

Amputations occur more commonly in the lower-limbs than in the upper extremities. The most common type is the transtibial (TT, below-knee) amputation by about 40 000 annual surgeries in the United States followed by transfemoral (TF, above-knee) amputations in the U.S, UK, Australia (Smith & Fergason 1999.) and also in Finland (Alaranta et al. 1998). Amputation is usually a life-saving operation when no other choices are available, and should not be considered as a failed treatment (Vänttinen 1991). Most of the amputees are elderly (Alaranta et al. 1998). Prevention of amputation by reconstructive artery surgery, treatment of diabetes, cessation of smoking and exercising is recommended (Vänttinen 1991). On the other hand, the large number of active amputees in the age group of 21 - 65 creates high demands for recreational and athlete prosthetics (Michael 1990).

2.2 Amputees in Paralympics

Disabled sports emerged after the World War II when many young people got physical and sight injuries. First competitions for those with spinal cord injuries were held in 1948 in Stoke Mandeville, England, and were followed by international games four years later. The first Olympic Games for spinal cord injured were organized in Rome in 1960 after the Olympics. Other groups of disabled athletes including amputees joined the games, now called Paralympics, in 1976 in Toronto. The first Paralympic Winter Games were held also in the same year in Sweden starting a fast growing movement towards current multiple elite events. The number of participants has grown from 1960 Games from 400 to almost 4000 athletes in Beijing 2008. (IPC History 2010.)

In the Paralympics athletes compete in various classes depending on their level of functionality and disability. There are six different categories; amputee, cerebral palsy, spinal cord injuries, intellectual disability and the others who do not belong to any of the categories mentioned (les autres). Classes are sport specific and classification is a part of the rules in each sport. The athletes are classified before entering competition and throughout their careers. Classes T (Track) and F (field) 40 – 46 are for amputee athletes. (IPC Classification 2010.) Class T42 is for unilateral transfemoral amputees and also includes the athletes with knee disarticulations and arthrodesis. Unilatelar (amputation in one limb) transtibial amputees compete in a T44 class and bilateral (amputation in the same limb on both sides) transtibial amputees are classified as T43. (Pailler et al. 2004.)

Since the first participation in 1976 Paralympics in Toronto amputees have competed in separate classes in the 100 m sprint running. The world record for transtibial amputees has improved from 14.3 s by Israeli Danziger in the 1976 to the current world record ran by South African double transtibial amputee Oscar Pistorius 10.91 s in 2007. The world record in transfemoral amputees has advanced even more, the first winner in 1976 Canadian Harrison ran 18.9 s, and American Earle Connor ran the current record 12.14 s in 2002 in Germany. The development of the world record occurred fast after the innovation of the energy restoring carbon fiber prosthesis which was used first by American Denis Oehler in 1988. (Pailler et al. 2004; IPC Athletics records 2010.)

2.3 Lower-limb prosthetics technology

The regular everyday prosthesis may not be adequate for certain sport activities, therefore some specific prostheses components are made for sports. Running, sprinting and jumping requires high force absorbing and durability from the prostheses. Also weight bearing, suspension and light weight are important properties of sport specific prostheses. New materials such as carbon fiber, titanium and graphite together with prostheses development have made running possible for many lower-limb amputees. (Webster et al. 2001.)

Transfemoral prosthesis consists of three main parts: a socket, a knee and a foot (figure 1). There are also different components and adapters. Liner is the part of the prosthesis which is attached to the residual limb. The liners are made from soft materials such as silicone, gel or polyurethane to provide suction suspension, comfort and to stabilize soft tissues. The liner is covered by hard socket and attached to the distal end of the liner. The socket is individually made to fit and support the body weight equally (total weight bearing socket). The prosthetic knee is connected to the socket and the proper selection of the knee joint depends on the patient’s mobility grade and demands. Different knee joints include hydraulic, pneumatic and mechanically controlled knees with a variety of functions such as stance-phase controller and locking abilities. The lowest part of the prosthesis is usually made of carbon fiber which enables energy restoration and releasing, stability and natural gait even in varying terrain. Different adapters are usually connected between the knee joint unit and the foot. Adapters may provide better torsion, rotation and shock absorbing abilities for the prosthesis user. (Össur 2010; Otto Bock 2010.)

FIGURE 1. Transfemoral prosthesis consists of (1) the liner, (2) the socket, (3) the knee joint unit, (4) rotation and shock absorbing pylon and (5) the foot (Össur 2010).

A connection between the residual limb and the socket is very important for the function of the prosthesis. Any movement of the residual limb will subsequently swing the prosthesis limb. During the foot contact with the ground, the residual limb-socket interface transmits the load-bearing ground reaction force back to the amputee. Therefore, the performance of the amputee is dependent on the effectiveness of this interface and the proprioception which it provides. The mechanical benefits of the advanced technology may not be used if the control of the prosthesis is inadequate. (Burkett 2010.)

2.3.1 Prosthesis knee

Two important stance phase movements in normal gait are controlled knee flexion in weight bearing and gradual extension under load. Together with ankle plantar flexion, knee flexion decreases the time between heel contact and foot flat. Knee flexion cushions the gradual transition of the body weight to the single leg during stance and limits the rise of the center of gravity thus allowing more efficient gait. Because the transfemoral amputees lack the physiological function of the rectus femoris muscle as a knee controlling and a hip stabilizing muscle, the knee flexion during stance is incomplete. This leads to an abnormal gait and yields high demands for prosthesis knee joints. (Blumentritt et al. 1997.)

The prosthesis knee joint is stable because of the extension moment produced whenever the ground reaction force (GRF) is anterior to the center of rotation of the joint. In a mechanical single axis knee joint the GRF line must pass anterior of the center of knee axis. Therefore the mechanical single axis knee joints do not allow any control in knee flexion under weight bearing following gradual extension of the knee in the stance phase. The knee joint has to be fully extended during weight bearing and therefore causing inappropriate rise of the center of gravity. The figure 2 illustrates the mechanisms for the single axis and polycentric knee function during stance. The ground reaction force (GRF) vector acts to lock the knee straight when passing anterior to the center of rotation (CR) in the single axis knee. If the GRF vector passes posterior the center of rotation, it would cause uncontrolled knee flexion and consequently falling down. (Blumentritt et al. 1997.)

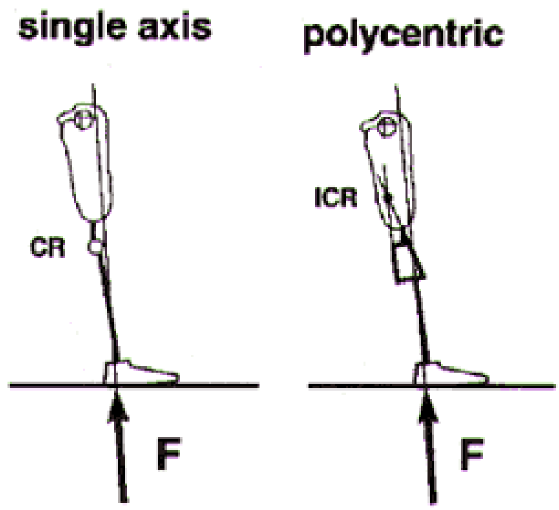

FIGURE 2. The mechanics of the single axis and the polycentric knee during stance. (Blumentritt et al. 1997.)

In the case of polycentric knee joint, the GRF vector must pass in front of the instantaneous center of rotation (ICR) to stabilize the knee extension (figure 3). One of the first polycentric knees developed is the Otto Bock 3R60 prosthetic knee joint. This 3R60 knee has a pivoting posterodistal axis with a spring. The posterodistal linkage moves through an arch during a stance phase. After the heel contacts the ICR moves to more proximal and posterior position providing better stability for the prosthesis knee joint than at the heel contact (figure 3). A prosthetic joint may pivot about 15 degrees about its anterodistal articulation. The GRF may fall posterior to the mechanical axes of the knee, but if it stays anterior to the ICR the knee will be stable. Therefore the stance phase knee flexion occurs when the GRF falls in between the mechanical center and the instantaneous center of rotation (figure 4). In that position the generated torque causes the upper part of the knee to pivot in counterclockwise allowing the knee to flex. Later in the stance phase, when the GRF falls in front of the anterodistal axis resulting in extension moment, the upper-knee frame rotates back to its original position causing the knee extension. (Blumentritt et al. 1997.)

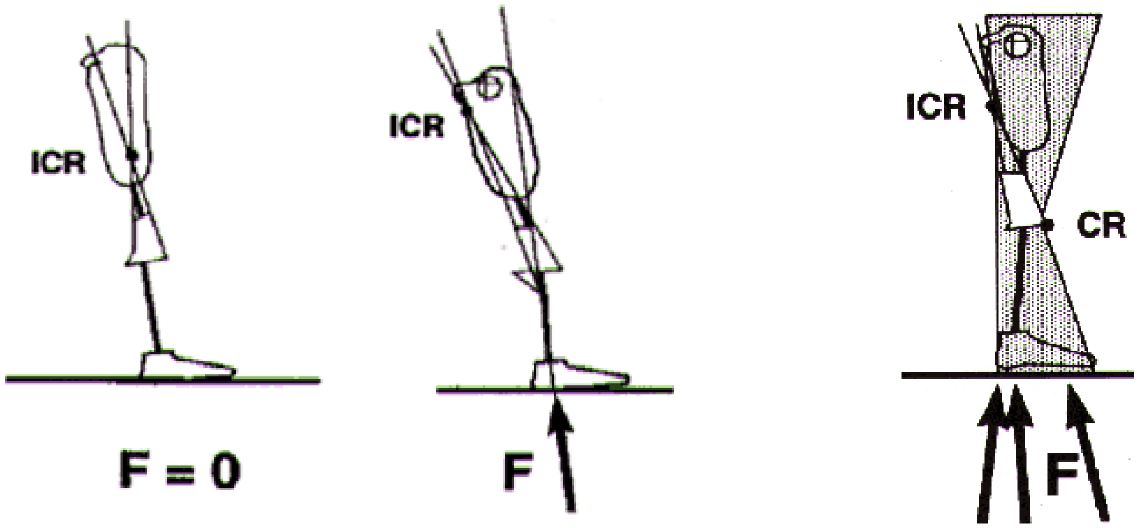

FIGURE 3. The instantaneous center of rotation (ICR) movement to more proximal and posterior position (from the left figure to the middle) during stance phase. On the right, the stance phase knee flexion whenever the ground reaction force (F) acts in between the center of rotation (CR) and the instantaneous center of rotation. (Blumentritt et al. 1997.)

Figure 4, shows changes in the knee angle during one gait cycle of a transfemoral amputee using Otto Bock 3R60 prosthetic knee joint. At the heel contact the prosthetic knee is fully extended and the GRF is located anterior of the knee. As the stance phase progresses and weight is transferred to the prosthesis the GRF moves posterior to the anterodistal axis and the controlled knee flexion begins. At the moment of forefoot touchdown, the GRF moves anterior and extends the knee joint. During the swing phase the knee flexes as the amputee produces enough hip flexion moment to move the GRF posterior to the ICR. (Blumentritt et al. 1997.)

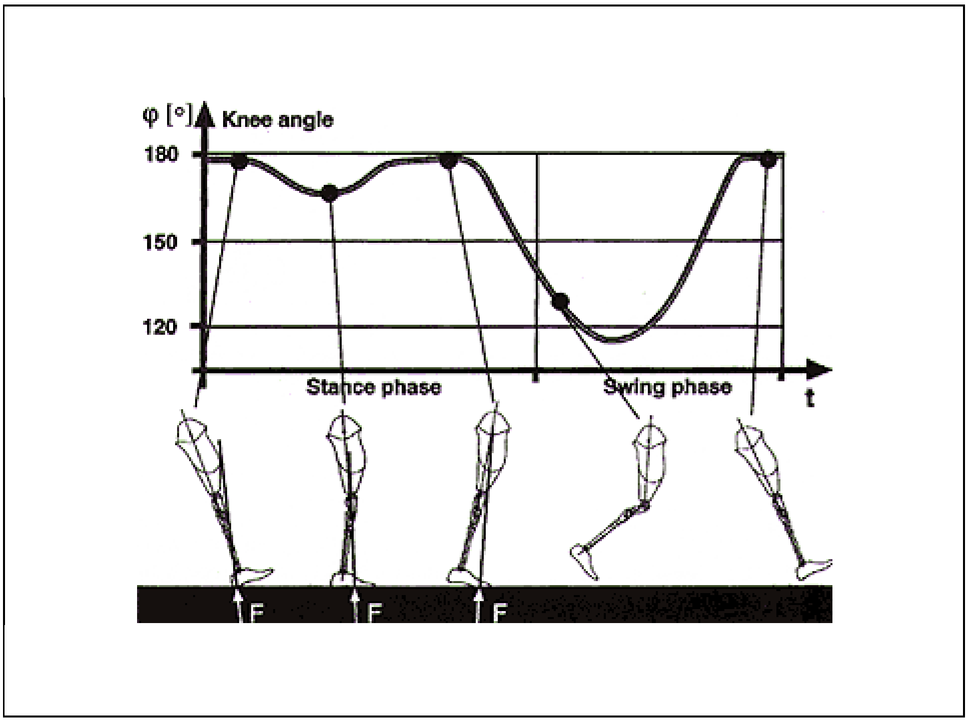

FIGURE 4. Knee angle change during gait cycle of a transfemoral amputee using polycentric knee joint. (Blumentritt et al. 1997.)

2.3.2 Prosthesis foot

The lowest part of the transfemoral prosthesis is the foot. There are a few specially made feet for track and field (figure 5 and 6). Most of the feet are usable for both transtibial and transfemoral amputees. The transtibial prosthesis consists of the socket and the foot with adapters. The sockets and the liners are similar to the ones for trans-femoral amputees. The Össur prosthetics company developed Flex-Sprint I (figure 5B) as the first foot made especially for running. Compared to standard Flex-Foot, Flex-Sprint I had 15° plantar flexed alignment, the heel part was eliminated and the stiffness was altered by a different sequence of carbon layers to suit better for running. The development of running prosthesis continued with Flex-Sprint II and III (in 1998) which is called Cheetah (figure 5A). The Cheetah foot was curved like a hind-foot of the Cheetah. The idea was to provide more forward directed propulsion with a delayed return from deflection to improve running times. Cheetah foot is customized for transtibial amputees. The world record in 100 m for transtibial amputees was set for the first time under 11 s by the Cheetah-foot. (Lehler & Lilja 2008.)

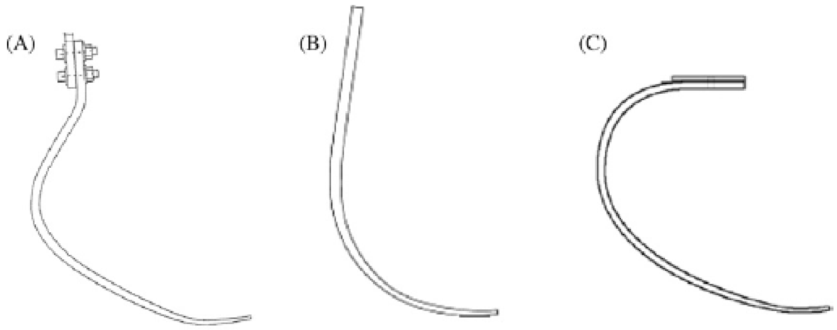

FIGURE 5. Different sprint foots (Össur): (A) Cheetah, (B) Flex-Sprint I and (C) Flex-run (Nolan 2008).

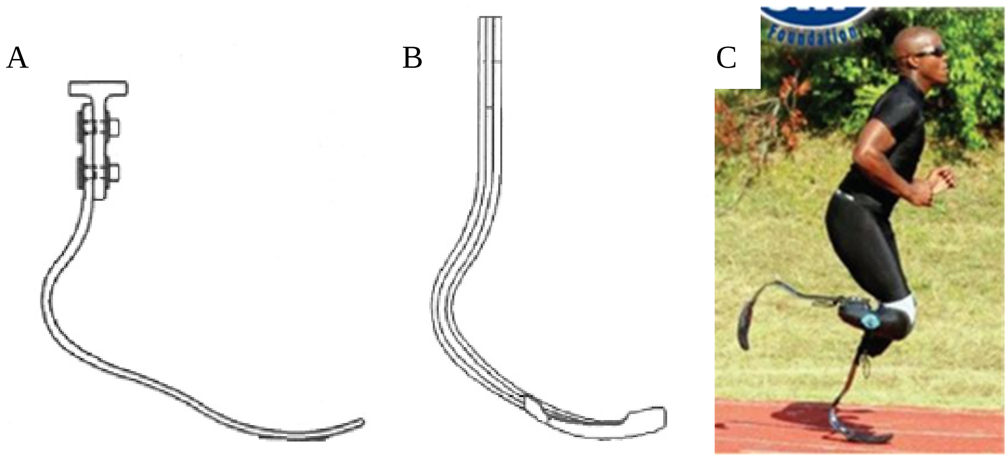

FIGURE 6. Different sprint foots (Otto Bock): (A) Sprinter and (B) C-Sprint (Nolan 2008). A bilateral transtibial sprinter with Cheetah feet (C) (Calabro 2009).

Buckley et al. (2000) compared sprinting kinetics in two transtibial amputee sprinters with both Cheetah (figure 5A) and Flex-Sprint prosthesis (figure 5B). One subject was able to run at a mean speed of 6.81 and 6.95 m/s and another subject at a 6.84 and 7.05 m/s by using the Cheetah and Flex-Sprint respectively. Both prostheses produced similar kinetics but they differed from the sound leg. The prostheses were able to produce ankle extension moment through the stance period in a similar way as the sound limb, absorbing energy during first half and generating energy during later half of the stance. The sound limb was able to generate more power than to absorb in contrast to each of the prostheses with about the same magnitude of absorbing and generating power. (Buckley et al. 2000.)

In the Flex-Run prosthesis, the shape of the blade is a C-shaped (figure 5C), which was made for recreational sports. Because of the C-shaped outline, the athlete is able to progress with the centre of gravity during deflection over the foot and then to use the stored energy for propulsion. (Lehler & Lilja 2008.) Flex-Run is recommended for distance running and it is durable and energy efficient (Össur tuotekuvasto proteesit 2010). The flexibility of the foot is selected individually in regards to the user’s body weight, running speed and technique (Pailler et al. 2004). A bilateral transtibial sprinter is using Cheetah feet in figure 6C (Calabro 2009).

The type of prosthesis foot used affects the energy consumption of the amputee in running. Brown et al. (2009) measured oxygen consumption and heart rate for six amputees and six control group subjects on a running treadmill. The trans-tibial amputees used both conventional and running specific prostheses. Energy cost was 15 % lower and heart rate 10 % lower in submaximal speeds when amputees used a running prosthesis rather than a conventional prosthesis. The amputees exhibited slightly higher VO_2 and heart rate when running with running prosthesis than the matched-pair control group at the same submaximal speeds. (Brown et al. 2009.)

3 GAIT CYCLE IN ABLE-BODIED AND LOWER-LIMB AMPUTEES

3.1 The definition of gait cycle and its temporal parameters

Forward human locomotion consists of walking and running (Enoka 2002, 179; Novacheck 1998). Gait was defined as “the manner of walking” in Vaughan’s (1984) article. Human gait has been referred to as repetitive cycles of one foot contacting the ground and supporting the body weight, and then followed by the other foot. The main difference between walking and running is the relative foot contact time for each cycle. In walking, one of the limbs is always in contact with the ground (*single support phase*) and for a short time both the feet simultaneously touch the ground (*double support phase*). However, in running, there is no double support phase, but there is *nonsupport phase* or *double floating phase* when either leg does not touch the ground. This nonsupport phase alternates with a single support phase. (Enoka 2002, 179; Novacheck 1998.)

The support phase is defined as the *stance phase* and it begins at the moment the foot contacts the ground (*foot touch down*). Stance ends at to *toe-off*, the moment when the foot leaves the ground. The nonsupport phase is the *swing phase* and it begins from the toe-off and ends at the foot touch down. One *gait cycle* is defined to include both stance and swing phases for both limbs, for example, it begins when one foot strikes down and ends when the same foot touches down. Gait cycle can be defined either way, to begin from stance or swing phase. (Enoka 2002, 179; Novacheck 1998.) DeVita (1994) defined gait to begin from swing phase and followed by stance phase because net joint torques and EMG were greater in transition from swing to stance. Enoka (1997, 179) verified one cycle as being a *stride* which consists of two steps. One *step* duration is from toe-off to the opposite limb's toe-off. (Enoka 2002, 179; Novacheck 1998.)

The absolute time of each cycle depends on the walking or running speed and decreases as the speed increases. In walking, the stride lasts about one second and the stance – swing ratio is 60 - 40 at the speed of 1.5 m/s (figure 7) (Vaughan 1984). In a study of two sprinters, five joggers and six elite long distance runners, the gait cycle time was 1 s during walking at the speed of 1.3 m/s, 0.6 s for running at 5.4 m/s and less than 0.6 s for sprinting at the speed of 7.7 m/s (Mann and Hagy 1980). In walking, the double support phase occurred twice and was about 10 % in each cycle. The double support phase disappeared in race walking as the swing and stance were equal in duration, but regarding the rules there should be a short double support phase. (Vaughan 1994.) The changes in the actual swing phase time as the speed increases are small, thus the main decrease in stride time occurs in the reduction of the length of the stance phase (Enoka 2002, 180). Mann and Hagy (1980) discovered a decrease in the length of stance phase as the speed increased. The stance phase was 62 % of the gait cycle time for walking at the speed of 1.3 m/s and it decreased to 31 % at 5.4 m/s to 22 % of the cycle time for sprinting at the speed 7.7 m/s. The body's center of gravity lowers as the speed increases due to increased flexion in the hip and knee joints and dorsiflexion in the ankle joint. The body's center of mass moves horizontally a longer distance during stance in walking (0.9 m) than in running (0.6 m). (Vaughan 1994.)

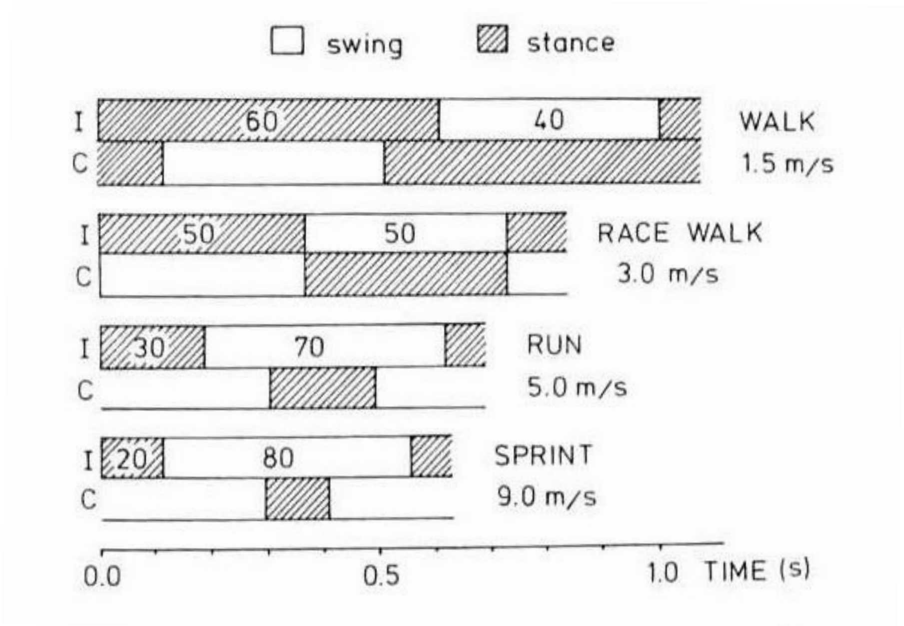

FIGURE 7. Swing and stance phase duration in for both legs in walking, race walking, running and sprinting. On the left column, I refers to ipsilateral and C to contralateral leg. (Vaughan 1984.)

The velocity of walking and running is related to step frequency and step length in a simple equation: $v = l_{step} * f_{step}$,

in which v means velocity, l_{step} means length of one step *(step length)* and f_{step} means *step rate* or *frequency*. Step frequency is simply the inverse of the step time (1/step time) (1/s). Velocity can be increased by increasing either step length or frequency or both. When velocity is increased in walking, walker decreases the cycle time by decreasing both the stance and swing times. In running, the cycle time is decreased by shortening of the contact time and even lengthening of the swing time. (Vaughan 1984.)

3.2 Lower-limb amputee walking

A lower-limb amputation causes slower walking speed and adaptations to normal gait patterns. Jaegers et al. (1995) studied transfemoral amputees walking gait and observed the comfortable walking speed (v_{comf} 1.01 m/s) to be 29 % slower than in the able-bodied subjects (1.42 m/s). The increase from the comfortable speed to a rapid speed (v_{rapid} 1.25 m/s) was achieved mainly by increasing the step length from 1.33 m at the v_{comf} to 1.50 m at v_{rapid}, particularly in the amputees with a short stump. The step frequency at the v_{comf} was 1.49 steps/s and at the v_{rapid} 1.65 steps/s. Seven of the eleven amputees walked with even longer step length but slower frequency than the normal-bodied subjects (1.28 m and 1.50 steps/s at the 0.95 m/s). (Jaegers et al. 1995.)

One study compared four different prosthetic feet in two transfemoral amputees and reported differences between the feet in step length varying from 1.43 m to 1.59 m in the same subject for the normal speed (1.07 - 1.26 m/s). The step length was longer for the sound side, except in one prosthesis foot trial, and the difference between legs increased as the subject walked at the faster speed (1.48 - 1.66 m/s) (Van der Linden et al. 1999). Jaegers et al. (1995) reported that the step width (range 18 - 30 cm) was greater at the comfortable speed for the amputees than in the controls (16 cm) and correlated with the step frequency and walking speed but not to the length of the stump. Contrary to other studies, the step width was less during the high comfortable walking speed. Some amputees walked with a strong lateral flexion towards the amputated limb during the stance phase; this and the large step width are possible causes of the atrophied hip abductors. (Jaegers et al. 1995.)

In the study by Jaegers et al. (1995) the able-bodied subjects walked with a symmetrical gait as the mean stance phase being 58 % and the mean swing phase 42 % for both legs at the comfortable speed (Figure 8) and 54 % and 46 %, respectively, for the rapid speed. In contrast, the temporal parameters differed for transfemoral amputees between the legs: the proportional duration of the stance phase was 63 % for the intact limb and 58 % for the prosthetic limb in comfortable speed. Likewise, the double support phase displayed similar asymmetry in some subjects as the sound leg phase was 10 – 30 % longer than the prosthetic side phase. The stance phase increased and the swing phase decreased at both sides as the length of the stump shortened. The stance, swing and double phases were more symmetrical at the faster comfortable speed. (Jaegers et al. 1995.)

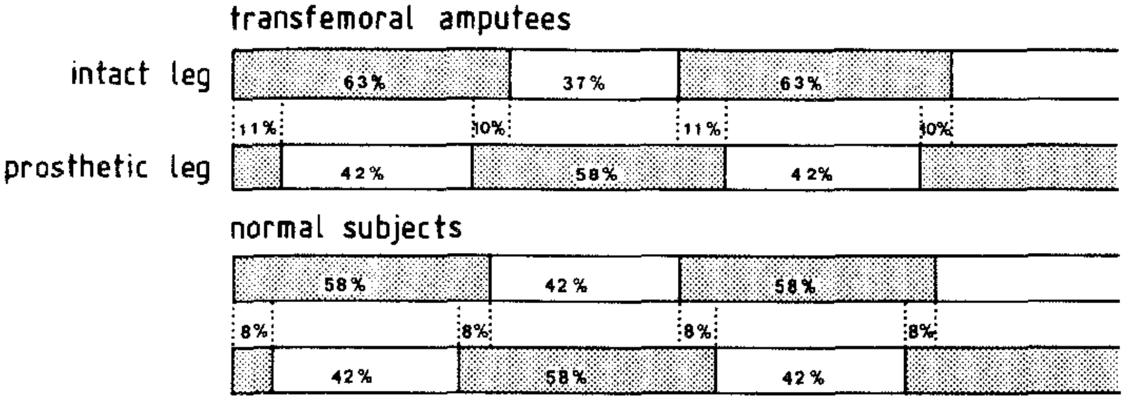

FIGURE 8. Stance (grey) and swing (white) phase percentage duration during gait of transfemoral amputee and normal subjects. (Jaegers et al. 1995.)

Asymmetry in walking gait was even greater in Van der Linden et al. (1999) research when the prosthetic stance duration ranged from 55 - 59 % and the sound side stance 67 - 68 % of the step in normal speed and 52 - 55 % and 67 - 68 %, respectively, in fast speed for different prosthesis feet used. Symmetry index (prosthesis / sound stance duration) varied therefore from 81 to 87 for normal speed and 78 - 83 for fast speed (Van der Linden et al. 1999).

3.3 Lower-limb amputee running

Little is known about the amputee running gait especially in sprint running. Some studies in the transtibial amputees in moderate running speed have shown only slightly asymmetrical gait. Sanderson and Martin (1996) showed that the transtibial amputee gait is similar to the normal-bodied gait with only a little asymmetry and differences in temporal and kinematic parameters. At two different running speeds, six amputee subjects had shorter step lengths and larger step frequency than the normal bodied controls. At the slower speed (2.7 m/s) the stance phase was longer for the prosthesis leg but as the speed (3.5 m/s) and ground reaction forces increased, it became shorter than the duration of the stance in the intact limb. This may have been a method to protect the stump and joints from excessive loading. (Sanderson & Martin 1996.)

In contrast to the relatively symmetrical transtibial amputee's gait, the transfemoral amputation severely affects the ability to run. Burkett et al. (2003) investigated inter-limb asymmetry and differences in walking and running in four transfemoral Paralympic athletes. The results showed greater asymmetry between the legs in running than walking. Similar to the data of Sanderson and Martin (1996) two fastest sprinters (mean speed 4.33 and 4.05 m/s) had shorter stance phase for the prosthesis leg in running. However, the stance phase was longer in two slower subjects (mean speed 2.47 and 3.22 m/s) and in walking for all subjects. The running gait of the two slower runners included a double-support phase and it was therefore different than the gait for the two fastest sprinters. The running gait for the two slowest runners might have been more like a hop skipping with the prosthesis severely limiting the running pattern. The swing time for the prosthesis limb was longer than for the intact limb in all subjects when sprinting. Interestingly, the step and flight time showed great asymmetry in all subjects. The step and flight time were shorter for prosthesis than the intact limb in the fastest sprinter but longer for the second fastest runner. (Burkett et al. 2003.)

In their previous study Burkett et al. (2001) reported that the limitations in the prosthesis properties may be the cause for the low number of running trans-femoral amputees. They also mentioned the strongly asymmetrical gait, especially the delay in the swing, causing a typical hop-skip running. The effect of the prosthesis knee vertical alignment with four different knee positions to the running velocity and gait symmetry was analyzed. Four athletes ran with normal prosthesis settings and with three lower adjustments for the knee. A symmetry index was calculated to express left-to-right leg symmetry. The symmetry index revealed clear improvements in gait parameters for the optimal alignment. The optimal knee alignment was found by lowering the knee axis 0.13 - 0.24 m from the normal knee location and this caused a significant increase in running velocity. The fastest athlete increased his average running speed from 4.33 m/s to 5.47 m/s and the range of increase in running speed was 19 - 40 % (26 % average) with the most optimal prosthesis alignment. (Burkett et al. 2001.)

4 SPRINT RUNNING PHASES

Sprint running performance can be divided into four phases which are start, acceleration, constant speed and deceleration phases. Each phase has its special biomechanical characteristics and different durations regarding to sprinting distance and the abilities of the athlete. Start phase includes reaction time 0.11 - 0.18 s and the first movements from the starting blocks. Acceleration phase is the longest, 50 - 70 m, in the 100 m sprint running and is followed by 20 - 25 m long constant speed phase when the maximal velocity is achieved. Speed decelerates during the last 20 - 35 m. In a 100 m run for the top male sprinters the average velocity is 10.16 m/s. The average step length is 2.10 - 2.27 m, average step frequency 4.40 - 4.58 and the amount of steps is 43 - 46. (Jouste 1997, 388.)

4.1 Start and acceleration phases

Block phase can be defined as the phase when the sprinter is in contact with the starting blocks. In the efficient set position the sprinter's center of gravity is high (0.60 - 0.66 m) and close to the starting line (0.16 - 0.19 m horizontally from the starting line). The center of gravity moves forward and upward when the force production begins. The stronger the arms, the more forward the sprinter can lean and the shorter the horizontal distance from the starting line. (Mero et al. 1983; Mero et al. 1992.) A mean hip angle of the front leg varies between 41 – 52 ° and 80 – 89 ° for the rear foot. After the start signal the knee and hip angles increase when producing force while the ankle angles decrease initially allowing the calf muscles to pre-stretch. This may lead to performance enhancement by using the stretch-shortening cycle in force production. The fastest sprinters have the smallest hip angles and therefore they can use their hip extensors more and produce force for longer time than slower sprinters. (Mero et al. 1983.)

Force production in the blocks lasts 0.34 - 0.37 s in male sprinters. The rear leg (the first leg out of the blocks) produces force 45 % of the block phase. The elite sprinters produce greater forces and have greater initial block velocity than the less trained sprinters. (Mero et al. 1992.) Horizontal force impulse was 234 Ns for the best of the three groups. The duration of force production in blocks was 0.361 s for the best group but there were no significant differences between the groups. The mean force values differed significantly between the groups (650 - 531 N) and therefore the results indicated that the level of horizontal force produced in the blocks is more important variable than the time to produce the force. Net vertical force values were lower than the horizontal force values. (Mero et al. 1983.) The block phase velocity has varied from 2.90 m/s to 3.94 m/s for males in different studies. (Mero et al. 1992.)

Acceleration phase is followed immediately after the take-off from the blocks and it is 30 - 50 m long in elite sprinters (Jouste 1997, 388). At the beginning of the two first stance cycles, the body's center of the gravity is ahead of the ground contact point but since from the third contact, the center of gravity is already behind the contact point. Since the first ground contacts, the center of gravity falls during the early stance (braking phase) and rises at during the later part of the stance (propulsion phase). (Mero et al. 1983; Mero et al. 1992.) During acceleration phase step length, step frequency and flight time increase and contact time decrease and approach the maximal velocity phase values (Mero et al. 1992).

Ito et al. (2006) studied running patterns of 18 male sprinters in Athletics World Championships in 2005. Running velocity and step length increased from the start to maximal velocity phase. High performance group (race time 10.12 - 10.32 s) had longer step length than the low performance group (10.40 - 10.90 s) in acceleration phase. The step frequency was almost constant (4.56 steps/s) through the acceleration and full stride phases. The high performance group had reached a higher velocity in full stride phase at 60 m than the low performance group. The step width decreased from the first step (0.39 m) to 0.17 m in maximal running speed. Wide step width may be advantageous for developing driving force during acceleration phase. Narrow step width is optimal to produce driving forces during the short foot contact periods in maximal running speed. (Ito et al. 2006.)

4.2 Maximal velocity and deceleration phases

The maximal velocity phase occurs after the acceleration phase about 50 - 70 m after the start and it is about 20 - 30 m long in 100 m sprint. The maximal speed achieved by the elite male sprinters can be 12 m/s in a 100 m run. The step length in the maximal velocity phase have been 2.20 - 2.65 m, step frequency 4.5 - 5.5 Hz, the contact times 0.08 - 0.09 s and the flight time 0.12 - 0.14 s. (Jouste 1997, 388.) The increments in stride length and step frequency are linear up to the speed of 7 m/s, but at higher speeds the velocity increases mainly by increasing the step frequency. The body’s center of gravity decreases at initial foot contact during braking phase and increases at propulsion phase. Running velocity decreases initially at braking phase and increases at propulsion phase. Vertical peak-to-peak displacement of the center of gravity decreases as the running speed increases. (Mero et al. 1992.)

Maximal velocity phase is followed by deceleration phase at the end on the 100 m race. Velocity loss from the peak velocity ranges from 0.9 to 7.0 % at major championships. Variability between sprinters depends from the performance capacity and can vary from 0 to 9.3 %. Step frequency decreases but stride length slightly increases. Contact and flight times increase at the end of the race. Also, increases are seen in braking distance, vertical descent of body's center of gravity and velocity loss during braking in deceleration phase. (Mero et al. 1992.)

5 SPRINT RUNNING KINEMATICS OF THE LOWER-LIMB AMPUTEES

Buckley (1999) analyzed the kinematics of five amputee subjects, four transtibial and one transfemoral amputee, and five able-bodied subjects, who all were experienced sprinters. The 2D-video data (50 Hz) and kinetic data were collected from three maximal velocity sprints for both prosthesis and sound leg landing on the force plate. The personal best for 100 m in the transfemoral subject was 15.1 s. He used his daily prosthesis (Endolite Hi activity prosthesis, a CaTech hydraulic swing and stance control unit, a Flex-Foot Modular III and an ischial total contact socket) in the trials. Thigh-knee angle-angle diagram is presented in figure 9d. It shows that the knee was overextended for a prolonged period of time (increase in vertical direction) and that there was a reduced motion of the thigh (decreased width in the plot) in the prosthetic side. The sound leg showed no noticeable differences compared to able-bodied, expect a reduced knee flexion angle, and therefore there was no evidence of compensation mechanisms. The reduced flexion knee angle during swing was maximum 79° for the prosthesis leg and bigger for prosthesis leg (58°) and normal-bodied legs (38°) (figure 10). The prosthetic knee was extended straight early in the swing and stayed extended about 80 % of the cycle (figure 9b). (Buckley 1999.)

Ankle angular displacement (figure 9a) displayed a phase shift in the flexion-extension model on the prosthetic foot. There was a small plantarflexion seen in the foot touch down, which was not seen in any other subject. Thigh angles differed for the prosthetic side compared to sound and able-bodied legs (figures 9c and 10). In foot touch down, the prosthetic leg was 16° less flexed and began to flex before toe-off during stance phase extension period, which was different than in any other subject. (Buckley 1999.)

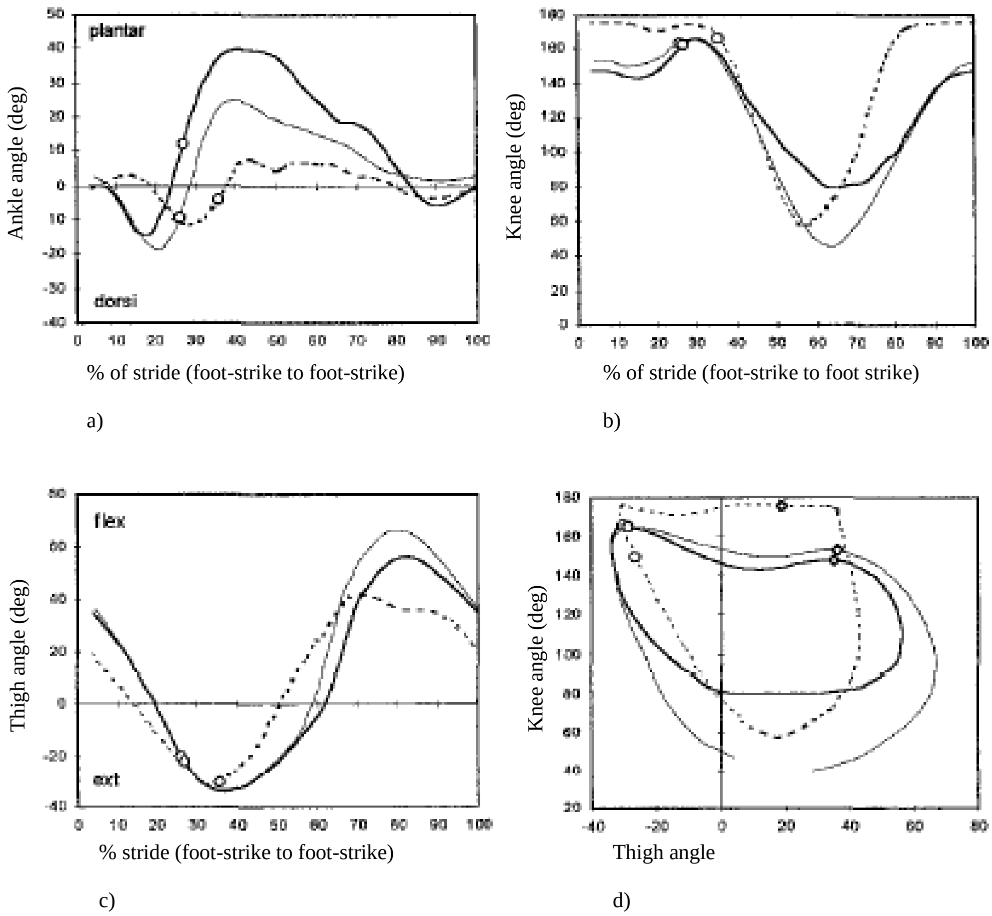

FIGURE 9. Ankle (a), knee (b) and thigh (c) angular displacement graphs and thigh-knee angle-angle diagram (d) for transfemoral amputee sound leg (solid bold line) and prosthetic leg (dotted line) and for able-bodied subjects mean (solid thin line). ○ = toe-off, ◊ = foot-strike. (Buckley 1999.)

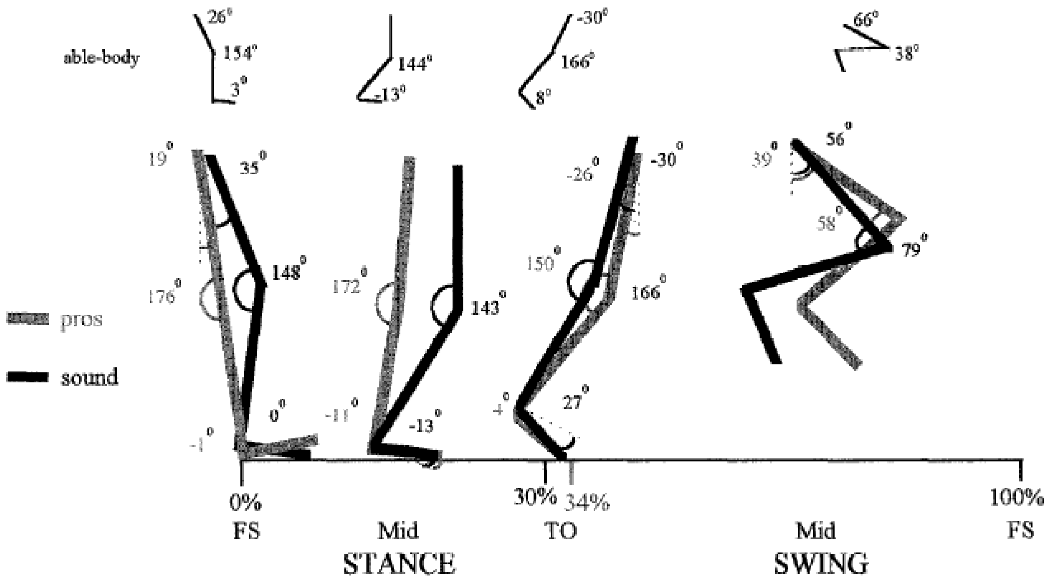

FIGURE 10. Angular displacement data at the key moments during sprint cycle of transfemoral amputee sprinter for both prosthetic (grey) and sound (black) legs. Above there are angular displacements of the able-body sprinter. FS = foot strike, TO = toe-off. (Buckley 1999.)

Figure 11 presents maximum angular velocities of the joints. Asymmetry between legs was seen in every joint and it was pronounced in knee joint angular velocities.

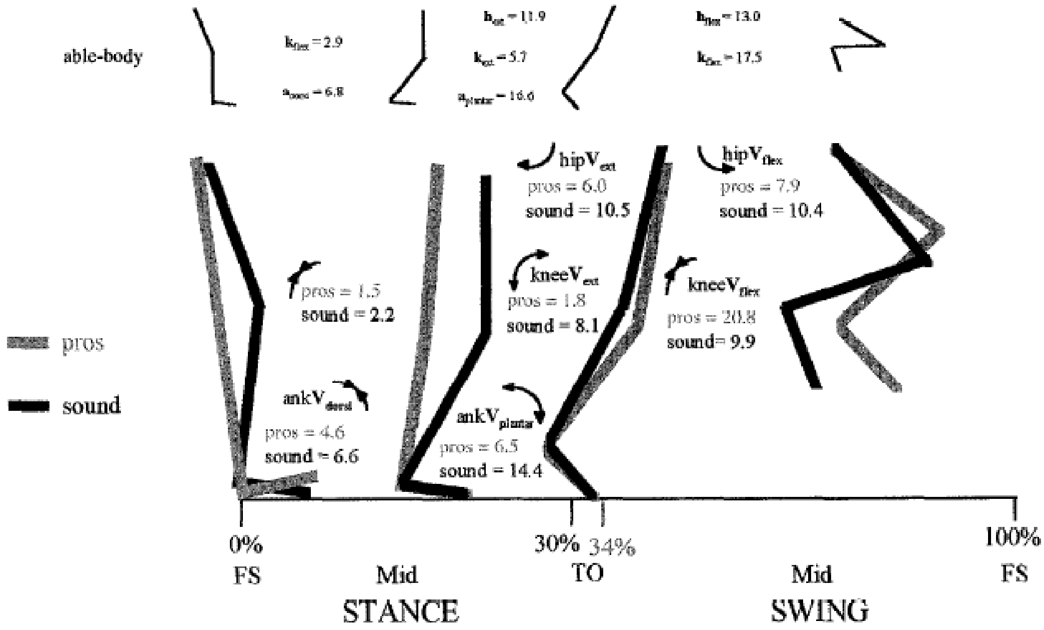

FIGURE 11. Maximum angular velocities (rad/sec) at the key moments during sprint cycle of transfemoral amputee sprinter for both prosthetic (grey) and sound (black) legs. FS = foot strike, TO = toe-off, v = velocity, ank = ankle (Buckley 1999.)

The knee extension from the early swing until late stance phase reveals the prosthetic limb to be a rigid supporter. The extended leg at touch down caused the keel contact to the ground at first and because of the function of the prosthesis, the keel flexion caused plantarflexion of the ankle to happen at the touch down. (Buckley 1999.) This kind of movement is beneficial to stabilize the knee as the instantaneous ground reaction force vector passes anterior of the center of rotation of the knee (Blumentritt 1997). The prosthesis knee clearly limits the subject’s ability to run and the future challenge would be to develop a knee joint unit which enables knee flexion at foot strike and at early stance and optimal swing phase control settings. The prosthesis leg should operate optimally in a range of speeds up to maximal sprinting velocity. (Buckley 1999.)

6 SPRINT RUNNING KINETICS

The ground reaction force (GRF) is the response of the ground or supporting surface to the actions of body segments. GRF is the resultant force of these actions and reflects the acceleration of the body’s center of mass. GRF can be divided to vertical force and two horizontal forces: anterior-posterior and medio-lateral force. (Enoka 2002, 185.) During foot contact, force production lasts a certain period of time and therefore it is useful to calculate the impulse which is the product of average force and contact time. Graphically it is defined as to area under a force-time curve. The Newton’s law of acceleration can be interpreted as the application of an impulse will result in a change in the momentum of the system. (Enoka 2002, 91 - 92.) GRF impulses can be calculated in vertical and in all horizontal directions. Vertical and horizontal braking and propulsion impulses can be calculated relative to body mass, therefore reflecting the change in velocity of the center of mass during the respective periods and in the respective directions. (Hunter et al. 2005.)

Contact times are often defined form GRF data in running and walking studies. In Munro and Miller’s study (1987) of twenty adult runners the GRF data was collected at the running speeds of 3.0 - 5.5 m/s. The contact time decreased from 270 to 198 ms as the running speed increased. The contact time was defined to a period of time when vertical GRF exceeded 16 N. (Munro & Miller 1987.) The definition of the contact time varies as in some studies it is defined to a time during which GRF exceeds 10 N (Hunter et al. 2005) or 50 N (Cavanagh & Lafortune 1980). This affects the contact time which was longer (270 - 198 ms) in Munro and Miller (1987) study compared to (mean time for rearfoot strikers 188 ms and 176 ms for midfoot strikers) the results of Cavanagh & Lafortune (1980) study with different definition.

6.1 Vertical ground reaction force

In running, there is only one major peak in vertical GRF (the second peak) compared to walking when there are two peaks. The difference in vertical GRF in these two actions becomes from the different leg kinematics. In running the knee flexion occurs in the first part of the stance phase to counter the downward movement of the center of mass. The lowest point of the center of mass is about middle of the stance phase. The knee starts to extend towards the end of the stance phase to project the body upwards. The activity of the knee extensor and plantarflexor muscles accelerates the center of mass in upward direction. The net muscle activity during stance phase is in the direction of extension; therefore the ground reaction force is a single-peaked curve in running. In walking, the knee joint is extended with only a small flexion during stance. The leg is assumed to be a rigid structure and this is called an inverted-pendulum model of walking. The center of mass follows an arch of a circle and the leg rotates about the ankle joint. During the middle of the stance phase when the center of mass is over the foot and in its maximum height, the forward horizontal velocity is in its minimum. This gives the minimum value of the vertical GRF during middle of the stance in walking. During walking the center of mass is located at its maximal height about middle of the stance and, in contrast, about the same point, in running at its minimal vertical position. (Enoka 2002, 185.)

Cavanagh & Lafortune (1980) analyzed the ground reaction forces, center of pressure distribution and running patterns of 17 runners whom they divided to mid-foot and rear-foot or heel-strikers. A steady running speed was about 4.5 m/s (range 4.12 - 4.87 m/s). The rear-foot strikers achieved a double peaked curve of vertical GRF. The first peak was about 2.2 times body weight (BW) and it occurred 23 ms after and the second peak of 2.8 BW occurred after 83 ms the touch down. The curves for the mid-foot strikers lacked the first impact peaks and the main peak value was 2.7 BW that occurred 75 ms after touch down. (Cavanagh & Lafortune 1980.) In Munro and Miller's study (1987) vertical GRF curves were double peaked which is characteristic to heel-strikers. In the beginning of the vertical GRF curve there is seen the first impact peak and then decrease to a relative minimum followed by the second peak. (Munro & Miller 1987.)

Average vertical GRF reflects the vertical GRF throughout the whole stance phase and is thereby subject to less intraindividual variance than the maximal vertical GRF. Munro and Miller (1987) found out that the average vertical GRF increased significantly ($p < 0.001$) from 1.4 BW at the 3.0 m/s to 1.70 BW at 5.0 m/s and the maximum values increased from 2.51 to 2.83 BW, respectively. Similar increase in maximal GRF with increasing speed was found by Kyröläinen et al. (2005) with 17 top male sprinters running at different speeds when maximal GRF increased from 2012 ± 142 N at 4.0 m/s to 2366 ± 364 N in maximal velocity (average 8.5 m/s) in braking phase (figure 12).

The vertical GRF data for six transtibial amputees was in a the same range as for the able-bodied runners, the maximal values for a sound leg were 2.39 BW and for a prosthesis leg 2.15 BW in 2.7 m/s speed and 2.58 BW and 2.30 BW respectively in 3.5 m/s (Sanderson et al. 1996). One transfemoral amputee presented about 2.5 BW maximal loading force for the prosthesis leg and about 2.2 BW for the sound leg when running at 2.47 m/s (Burkett et al. 2003).

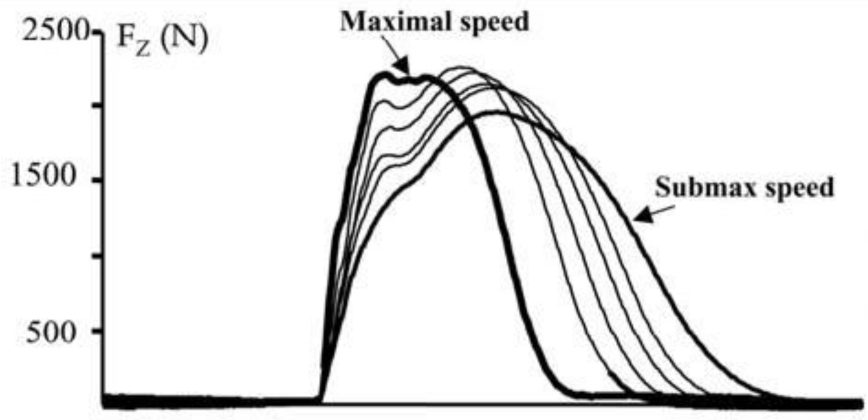

FIGURE 12. The vertical GRF graphs at different running speeds from 4.0 m/s to maximal velocity (average 8.5 m/s). The maximal velocity graph is bold. (Kyröläinen et al. 2005.)

Vertical GRF impulse, when normalized to body weight, reflects the change in velocity of the runner, if the wind resistance is ignored. Vertical GRF impulse can be calculated as the product of average vertical GRF and contact time subtracted by body weight impulse. The vertical GRF impulse normalized to body mass was 0.99 m/s in a velocity range of 7.44 - 8.80 m/s of 28 male sprinters. (Hunter et al. 2005.) Change in vertical velocity tells about a runner’s ability to reverse the downward velocity of the center of gravity at touch down to an upward velocity at take-off. Change in vertical velocity can be calculated by subtracting the body weight impulse from the vertical GRF and dividing by mass of the subject. Change in vertical velocity increased from 1.0 at 3.0 m/s to 1.5 m/s at the 5.0 m/s reflecting the corresponding change in average vertical GRF. (Munro & Miller 1987.)

6.2 Anterior-posterior ground reaction force

The horizontal anterior-posterior ground reaction force has been characterized to as a biphasic during running. The braking phase starts at the foot touch down and continues to about the middle of the stance phase. The direction of the GRF is opposing forward movement and when the GRF turns to forward movement, the propulsion phase starts until the toe-off. Munro and Miller (1987) noticed variety of braking patterns in their study. Five of the 20 subjects had one peak in the braking phase occurring about 25 % of the total stance time. Ten subjects had double peak in their braking phase occurring at about 7 and 24 % of the stance time and five subjects had multiple braking peaks. (Munro & Miller 1987.)

Six amputee subjects showed slightly greater braking than propulsion forces for both prosthesis and intact legs at the 2.7 m/s speed but lower at the 3.5 m/s speed. The peak braking forces were 0.315 BW for the sound leg and 0.167 BW for the prosthesis leg and the propulsion forces 0.327 BW and 0.209 BW, respectively, at the 3.5 m/s. (Sanderson et al. 1996.) The study by Kyröläinen et al. (2005) study the horizontal maximal breaking GRF increased from 339 ± 33 N to 830 ± 210 N as the running speed increased from 4.0 m/s to the maximal velocity of 8.5 m/s (figure 13).

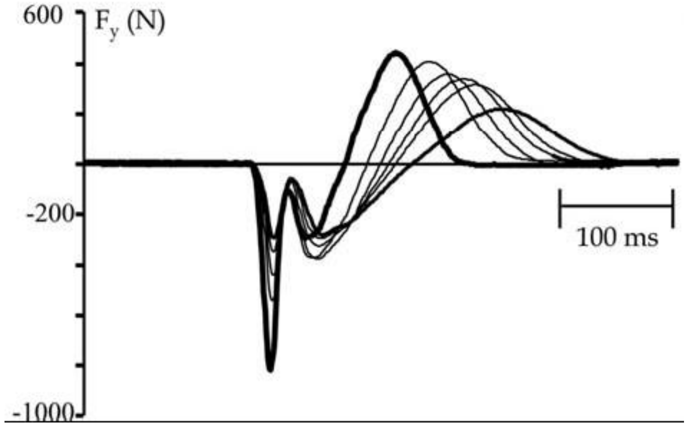

FIGURE 13. The horizontal GRF graphs at running speeds from 4.0 m/s to maximal velocity (average 8.5 m/s) (bold). (Kyröläinen et al. 2005.)

The braking and propulsive impulses are separated when the horizontal GRF changes direction from negative to positive and usually normalized to body weight. The braking impulse increased from 0.15 BW to 0.25 BW and the propulsive phase impulse increased from 0.14 to 0.25 BW as the running speed increased from 3.0 to 5.0 m/s. The time when the braking phase turned to propulsion (zero fore-aft shear) was about 48 % of the contact time. (Munro & Miller 1987.) In 15 transtibial amputees braking impulses were lower for the prosthesis leg (1.97 - 2.34 %BW*s) than the sound leg (2.29 - 2.63 %BW*s), (the prosthesis to sound leg ratios being 0.76 - 0.98), in walking at 1 m/s with four different prosthesis. The difference between legs was greater in propulsion impulses: prosthesis leg 1.71 - 2.05 %BW*s and sound leg 2.71 - 2.79 %BW*s, ratios ranging from 0.61 to 0.79. (Zmitrewicz 2006.)

6.3 Medio-lateral ground reaction force

The horizontal medio-lateral ground reaction forces are relatively small when compared to anterior-posterior or vertical GRFs. In the study of Cavanagh and Lafortune's (1980) the mean peak to peak amplitude was only 9 % of the peak vertical component and 26 % of the anterior-posterior component. The medio-lateral GRF curve displayed two positive and two negative peaks for both mid- and rear-foot strikers. The mean peak amplitude was three times bigger for mid-foot strikers (0.35 BW) than for the rear-foot strikers (0.12 BW). Also the graphs were homogenous for mid-foot strikers but versatile for the rear-foot striker group. (Cavanagh and Lafortune 1980.) The peak to peak amplitudes in Munro and Miller's (1987) investigation varied between 0.20 - 0.50 BW for each subject, the mean value of all trials being 0.29 BW. The average GRF ranged from 0.04 to 0.25 BW for the medial and from 0.06 to 0.31 BW for the lateral GRF for each subject. There was great variability between subjects in the medio-lateral GRF graphs. Also some subjects elicited different patterns for left and right foot. (Munro & Miller 1987.) Hamill et al. (1983) found out a decreasing trend in mean data for the medio-lateral forces as running speed increased from 4.0 m/s to 7.0 m/s.

7 JOINT MOMENTS IN SPRINT RUNNING

Human movement consists of the rotation of body segments about their joint axes which are produced by external forces and muscle activity. The capability of a force to produce rotation is called torque or moment of force. Torque is equal to the magnitude of force times the perpendicular distance between the line of action of the force and the axis of rotation. (Enoka 2002, 62.) Joint reaction forces and moments can be calculated from complete anthropometric data, external forces and kinematic measures by inverse dynamics. The calculation is simplest to begin from the free-moving distal segment when the only acting forces are the proximal joint force and moment and the segment weight. Then it is possible to solve more proximal joint moments. (Vaughan 1984.) This inverse solution of a link-segment model gives information about the net summation of all muscle activity at each joint. Several forces are acting on the link-segment model. Gravitational forces act downward through the center of mass of each segment and are equal to 9.8 m/s^2. The ground reaction or external forces are distributed over the area of the body but since they are presented in scalar vectors they are considered to act on a certain point of body. Muscle and ligament forces act on a certain joint. The net muscle forces at a joint can be calculated in terms of net muscle moments. (Winter 2005, 86 - 88.)

The link-segment model constitutes of all the forces acting on the body. It can be divided to its segmental parts ending to the joint. This represents a free body diagram with the forces acting across each joint (joint reaction forces and muscle forces). According to Newton's third law there is an equal and opposite force acting at each hinge joint in the free body diagram. (Winter 2005, 88 - 89.) The joint reaction force acts into the joint representing the reaction of the adjacent body segment to the compressive force in the joint. It is a three-dimensional force in which one component is normal to the joint and directed into the joint surface and two components are tangential to the surface composing the shear force that acts along the joint surface. The magnitudes of the joint reaction forces can be large. Muscle force is often the most significant part of a free body diagram in segmental analysis. The muscle force acts back across the joint representing the net pulling action of the muscles that cross the joint. Both the agonist and antagonist muscles control the movement around the joint. (Enoka 2002, 119 - 122.)

Simonsen, et al. (1997) studied flexor moment of the knee in the stance phase in walking at 1.25 m/s. They calculated external moment arms related to the ankle, knee and hip joint with respect to the resulting ground reaction force, figure 14.

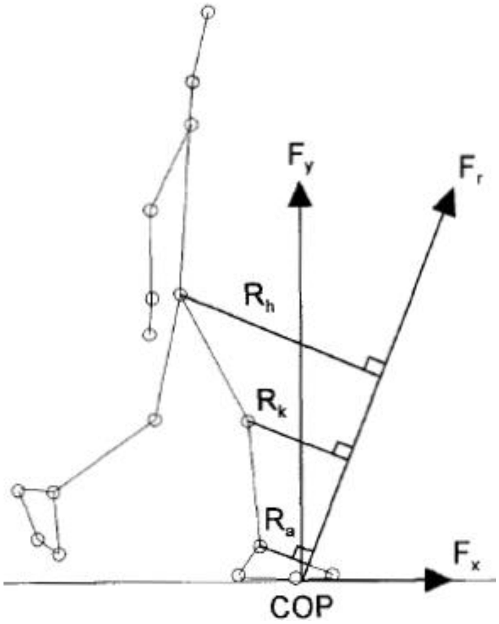

FIGURE 14. A stick-figure of one subject during stance phase of walking. The ground reaction forces the vertical F_y, the sagittal F_x and the resulting F_r) are marked through the point of centre of pressure on the force plate. External moment arms with respect to the hip (R_h), knee R_k and ankle joint (R_a) are drawn. (Simonsen et al. 1997.)

In Simonsen et al. (1997) study, the knee joint flexor moment just after the heel strike functions to direct the ground reaction force towards a desirable direction and minimize the loss of forward speed. The extensor moment in the hip joint takes place at the same time. In dynamic movement the positive support moment formed by the ankle and the hip joint works to keep the posture upright despite of the knee flexor moment. The results showed that a flexor moment about the knee joint can be caused by a large plantar moment of the ankle joint and the knee flexor moment can contribute to an extensor moment of the hip joint. However, there are still great inter-individual differences in the kinematics during foot touchdown in the moments generated by the three major leg joints. (Simonsen et al. 1997.)

Sanderson & Martin (1996) found out that transtibial amputees achieved similar running gait as non-amputees. When running speed increased from 2.7 to 3.5 m/s the amputees modulated the magnitude of the joint moments without markedly altering the temporal sequencing, ground reaction data or joint kinematics. However, the running speed was rather slow compared to sprint running. (Sanderson & Martin 1996.) In Buckley's study (2000) two transtibial amputee sprinters were able to run with a mean speed of 6.81 m/s and 6.95 m/s with two types of different running prosthesis. An asymmetrical transtibial amputee gait showed differences in joint moments between the prosthesis and the sound leg in running. Each prosthesis for both subjects produced ankle extension joint moment during contact phase, figure 15. The prosthesis leg absorbed about the same amount of energy during the first half of the stance then released in the second half. The peak energy values were considerably less for the prosthesis leg than the sound leg and the sound leg generated energy more than absorbed. The knee of the prosthesis leg produced an extension moment throughout stance in subject 1 with Sprint Flex prosthesis and in Subject 2 with either prosthesis used. The hip of the prosthesis leg produced an extension moment which was maintained until late stance. This prolonged extension moment was created from concentric action resulting positive power. This finding was in contrary to the sound side in which only a short duration extensor moment occurred in early stance and then followed a flexion moment which was associated with negative power. (Buckley 2000.)

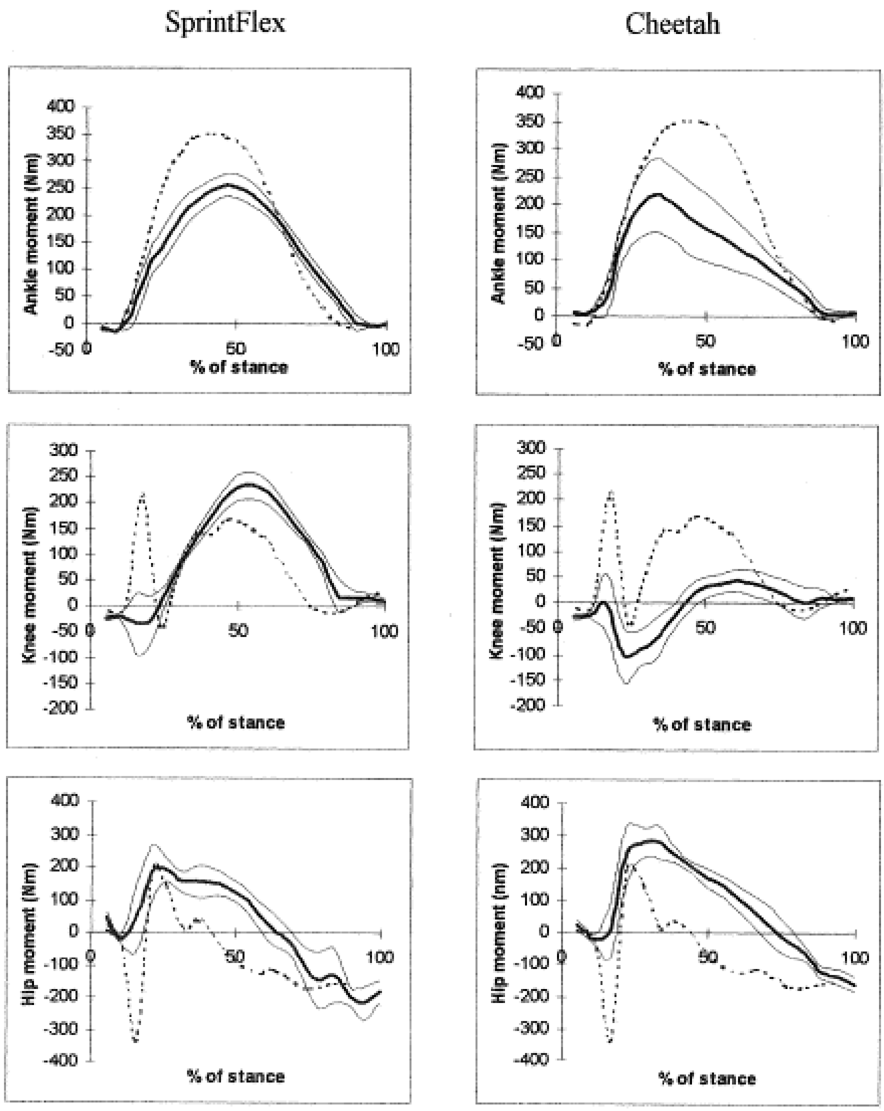

FIGURE 15. Joint moments of the ankle, knee and hip of the sound (dotted line) and prosthetic limbs (solid line) in a transtibial amputee sprinter when using either SprintFlex or Cheetah prosthesis legs. The mean is shown along with ±1S.D. for the prosthesis leg, but only the mean is shown for the sound leg. Moments that tended to extend a joint are drawn as positive, and those that tended to flex a joint are drawn as negative. (Buckley 2000).

8 PURPOSE OF THE STUDY

The purpose of this study was to examine the difference between the sound and prosthesis leg and between two prostheses (A & B) during the sprint start and maximal running speed in a top level transfemoral amputee sprinter. Another purpose of the study was to develop the running technique of the subject. The analyzed data consisted of temporal parameters, kinetic and kinematic data. Results of the study are compared to previous results of transfemoral amputee and able-bodied sprint running studies. The results of this study are descriptive in their nature and can not be generalized for all transfemoral amputees running.

Research questions

1. Is the running gait symmetrical between the sound and the prosthesis legs?

Hypothesis 1: There is an asymmetry between the sound and the prosthesis legs in all measured variables.

2. Are there differences in temporal, kinetic and kinematic data between the two different prostheses?

Hypothesis 2: There are differences between the prostheses.

9 METHODS

9.1 Subjects

One male sprinter with unilateral transfemoral amputation voluntarily participated in the study. His age was 24 years, body mass (without prosthesis) 56.0 kg and height 166.7 cm. Traumatic amputation for subject's left leg had been done seven years earlier and he had experience in sprint running for five years. His personal best in 100 m sprint running was 13.11 s. He used a dedicated sprint prosthesis which had an ischial total weight bearing socket with suction suspension (Matsumoto, Japan) and Otto Bock 3R55 knee joint unit (Germany). In the measurements the subject used two different sprint legs: prosthesis A (Imasen, Japan, m = 2.3 kg) and prosthesis B (Otto Bock, Germany, m = 2.7 kg). He had previously used each of them in daily training. Length of the sprint prosthesis was adjusted so that it was the same as the sound leg when standing on tiptoe. Knee joint of the prosthesis had been lowered to assist in swing phase. The subject was informed of the purpose of the study and signed a written consent to participate. He had right to withdraw from the study at any time. The study was conducted according to the declaration of Helsinki approved by the Ethics Committee of the Osaka University of the Sport and Health Sciences.

9.2 Protocol

The measurements were taken at an outdoor all-weather track and field court in a single session. The preparations started early in the morning four hours before the actual measurements started. Two force plates (60 cm x 90 cm, Kistler Instruments AG Switzerland, model type 9287 and 9287B) were set in series under the track to form a force platform system. Four cameras were placed on correct positions. Two of the four cameras were digital high-speed cameras (200 Hz) (NAC Memorecam, Tokyo, Japan) and two other were digital video cameras (60 Hz) (Sony HVR-AJ1, Tokyo, Japan). Three of the cameras were set up around the performance area about 90° angle of each one, two high-speed digital video cameras (cameras 2 & 3) were behind the sprinter and one camera (camera 1) in the front side (figure 16). One digital camera (60 Hz) was used only to measure 10 m time from the start and it was positioned perpendicular to the running direction. A calibration was done with a calibration pole for later three-dimensional analysis in 17 selected points at the performance area (appendix 1). The computers for data collection were set on the table close by the track.

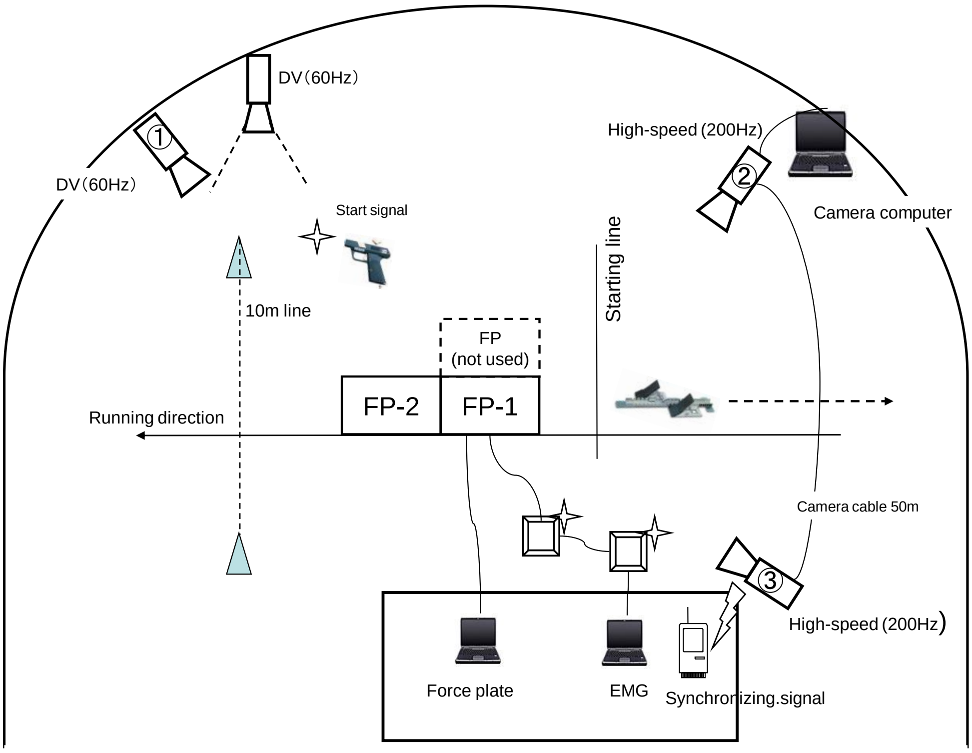

FIGURE 16. Measurements settings. FP = forceplate.

The subject was prepared for the measurements by placing four reflective spherical markers on the subject's skin over the crest of os. ilium to the posterior and anterior superior iliac spine to both sides. White tapes were added to subject's wrists to help in manual digitization analysis. The subject was instructed by the measurement protocol.

The subject performed a few practice starts from start blocks before testing trials. Both the block start and maximal velocity sprint running were experienced with two different sprint leg prosthesis used. The sprint start was studied for the first six steps including both swing and contact phases. In the first step trials the subject performed sprint start from starting blocks so that the first step landed on the force plate (1st step). Starting blocks were set in a close distance to the force platform. In the second step trials (2nd step) the second step landed on the force plate and so on until the sixth step. The subject always changed the sprint leg (prosthesis A and B) after each start on the way that the both 1st step trials were run before 2nd step trials and so on. Ito et al. (1997) and Baba et al. (2000) had done their measurements by recording 1st, 3rd, 5th and 9th step from the start of normal bodied sprinters. In these measurements also the 2nd, 4th and 6th step were recorded as the transfemoral amputee's prosthesis and sound leg kinematics and measured parameters might be different (Burkett et al. 2001; Burkett et al. 2003; Buckley 1999).

The starting blocks were set up in front of the force platform to a certain distance so that the intended step would land in the force platform system without aiming. Accepted trials were those in which maximal effort was put up and those in which the whole step landed on the force plate without visible targeting (Burkett et al. 2003). For each start one accepted trial was recorded with both sprint legs used. In the sixth step trials the first 10 m running time was also measured. Maximal sprinting velocity trials were repeated twice to acquire both sound and prosthesis leg landing on the force plate. Totally four maximal velocity and 12 start starts were recorded.

9.3 Data collection and processing

Ground reaction force components were measured in three directions, vertical, anterior-posterior and medio-lateral direction and recorded with sampling frequency of

1000 Hz. From the acquired video data of two high-speed digital video cameras, 25 body landmarks were manually digitized at 100 Hz frequency by DKH Frame Dias II software (Tokyo, Japan). Three-dimensional link segment model was created. Lower leg segment lengths were calculated and checked from coordinative data. After checking segment length, smoothing was done with Butterworth type 4th-order low-pass filter with 8 Hz cut-off frequency. Temporal data was calculated from X- and Y-coordinate data of toes. Three-dimensional kinematic data was analyzed by Sprint3d-program (3-jikendousabunseki-softo V6.34, Fuchimoto, T. and Fujihara, T., Osaka University of Health and Sport sciences).

9.4 Data analysis

9.4.1 Temporal data

One successful trial of each step landing on the force plate was performed with both prosthesis A and B. Therefore there were in total 12 starts and 4 maximal velocity trials analyzed. The definitions for terms used in this study and how they were analyzed are explained here:

Maximal running velocity (MaxV) = Average velocity of four consecutive steps in maximal running velocity trials.

Gait cycle = consists of contact phase and swing phase and the cycle ends when the same foot, which was in ground contact in the beginning of gait cycle, contacts the ground again.

Cycle time, *cycle speed* and *cycle length* were used to characterize the gait cycle.

Contact time = analyzed from the force plate data, the duration of the stance as the period of time when GRF differs from zero or noise and it was manually analyzed for each step.

Swing time = The duration of the swing phase, from toe-off to the next touch down

Step time = From toe-off to opposite leg toe-off

Step speed = Step length / step time

Step frequency = 1 / step time

Step length and *width* = calculated from mid-contact to mid-contact from toes. The step width and length were calculated as a following method during start and maximal running velocity trials:

0-1 Step = starts from the start block position and includes the swing of the first step and the first step contact (prosthesis leg)

0-2 Step = starts from the start block position and includes the sound leg swing and the second step (sound leg) contact

1-2 Step = from the first step contact to the second step contact (sound leg),

2-*3 Step* = from second step contact to third step contact (prosthesis leg), etc.

MaxV SP = maximal velocity from sound to prosthesis leg

MaxV SP = maximal velocity from prosthesis to sound leg

9.4.2 Kinematic data

Joint angles, angular velocities and accelerations were analyzed in Microsoft Excel program (version 12, 2007, WA, USA). Four consecutive steps in each trial were selected in the movement analysis and the kinematic data was calculated from those steps. The analyzed joint angles and joint angular velocities are presented in figure 17 for the swing leg and in figure 18 for the supporting leg. The results presented represent the maximal and minimal values of the analyzed parameters and the average values of hip, knee and ankle angles at the touch down, middle of the stance and the take off.

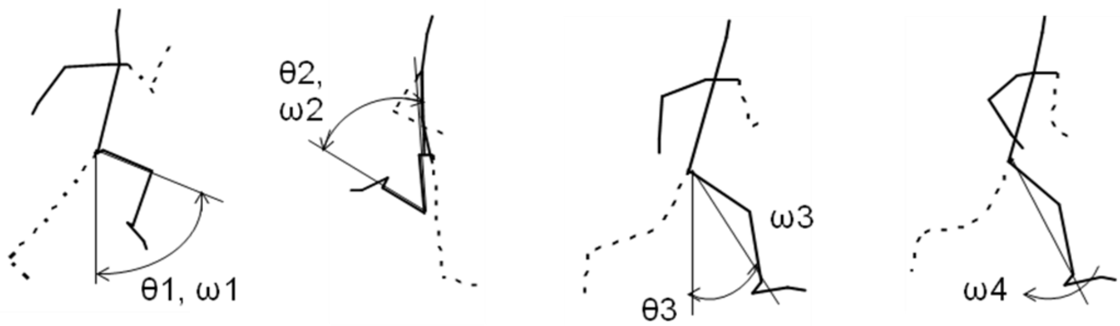

FIGURE 17. Analyzed kinematic parameters of the swing leg: maximal thigh lift angle (θ1) and its' maximal velocity (ω1), minimal knee angle (θ2), maximal knee flexion velocity (ω2), maximal leg angle (θ3), maximal knee extension velocity (ω3) and maximal leg touch down velocity (ω4).

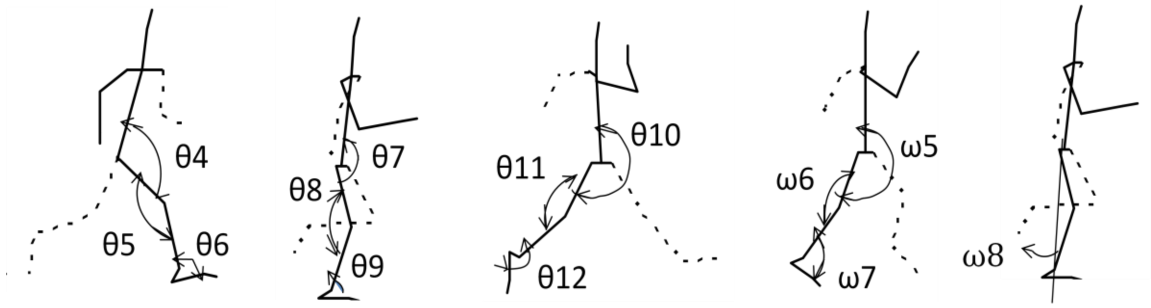

FIGURE 18. Analyzed kinematic items of the support leg: Hip (θ4), knee (θ5) and ankle (θ6) joint angles at the moment of foot touch down, minimal angle of the knee (θ8) and the ankle (θ9), and hip (θ7) joints at the middle of the stance phase, hip (θ10), knee (θ11) and ankle (θ12) joint angles at the moment of foot take off. Maximal extension velocity of the hip (ω5), knee (ω6) and the ankle (ω7) joints during the stance phase and maximal leg back swinging velocity (ω8) of the support leg.

9.4.3 Kinetic data

The GRF data was analyzed in Microsoft Excel, (version 12, 2007, WA, USA). The touchdown on the force plate was searched from the data. The stance phase was defined to the period of time when GRF differs from zero or noise and it was manually analyzed for each step.

Peak loading force, vertical average force and vertical impulse and vertical braking and propulsion impulses were analyzed from the vertical GRF data. The vertical impulse was calculated as the product of average force and contact time subtracted by body weight impulse. Vertical impulse: $I_{vert} = F_{aver}\, t - gmt,\ g = 9.81\ m/s^2$

Velocity change was calculated from the vertical impulse, by dividing it by body mass. Vertical braking and propulsion impulses were defined from horizontal braking and propulsion phase durations.

Peak forces, average forces and impulses in braking and propulsion phases were calculated from the horizontal anterior-posterior force data. Propulsion impulse was defined to include all negative force data and braking impulse to all positive data during stance phase. The medio-lateral peak force, average force and impulse were defined from horizontal medio-lateral GRF values. All values were normalized to the body weight (product of body mass and gravity of Earth (9.81 m/s^2). The body weight included the weight of the prosthesis used.

10 RESULTS

10.1 Temporal parameters

Temporal parameters displayed asymmetrical running gait. Inter-limb asymmetry was great and it was greater in prosthesis A data than prosthesis B in most of the parameters. Differences between prostheses were small in all parameters. Symmetry was found in cycle time (swing and stance) being constant 0.49 s for both legs and prostheses.

In maximal running velocity, the transition from sound to prosthesis leg stance (SP), the running step velocity was slower (average of all measured steps A 6.71 m/s and B 7.02

m/s) than from prosthesis to sound leg (PS) (A 8.07 m/s, B 7.56 m/s). The total velocity was faster in prosthesis A, 7.45 m/s, than B, 7.21 m/s, trials. The 10 m time in start for prosthesis A trial was faster 2.61 s than for prosthesis B 2.71 s. Running velocity increased linearly during the first seven steps in start from the first step 3.2 m/s to 5.6 m/s in A (figure 19). The following data is presented only for prosthesis A, because the prosthesis A was better as the running speed was faster and the differences between prostheses were small.

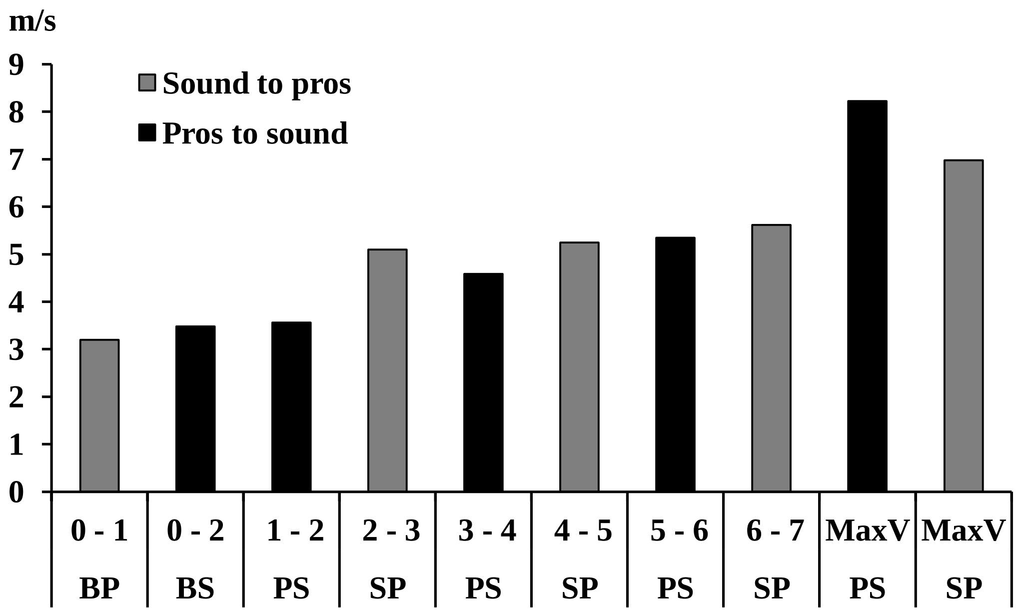

FIGURE 19. Running velocity increased linearly during the first 7 steps in start. The maximal running velocity differed for both legs in prosthesis A trials. B = from the starting blocks to the P = prosthesis leg or S = sound leg, SP = sound to prosthesis leg, PS = prosthesis to sound leg.

Contact times were longer for prosthesis leg (A 0.130 s, B 0.126 s) than sound leg (A 0.098 s, B 0.107 s) in maximal velocity trials (Figure 20). The difference between legs was greater in trials with prosthesis A than B. The contact times decreased during start from the 1st Step 0.249 s to the 6th Step 0.142 s in prosthesis A although the decrease was not linear. The 3rd and 5th step contact times for the prosthesis leg were longer than the previous 2nd and 4th step contact times in prosthesis A. The decrease in contact times for prosthesis B was linear during start.

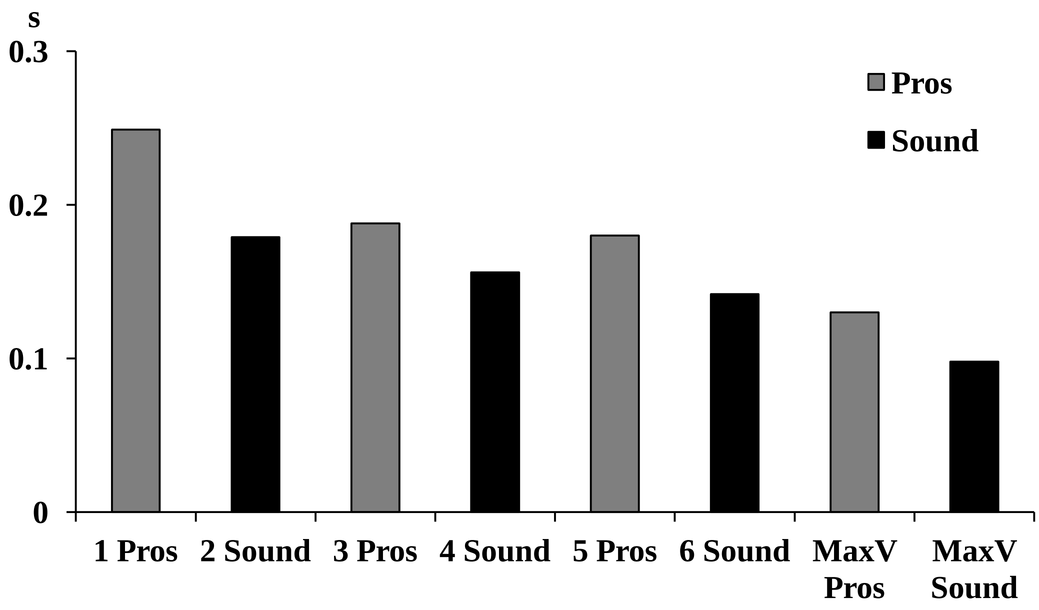

FIGURE 20. The contact times in the start from the 1st step to the 6th step and in maximal sprinting velocity for prosthesis A. Pros (grey) = Prosthesis leg, Sound (black) = sound leg.

Asymmetrical running pattern was also seen in step length and step frequency as SP step length was longer than PS but step frequency was bigger in SP than in PS. Step frequency showed greater difference for prosthesis B (SP 3.6 and PS 4.4) trials than for prosthesis A (SP 4.4 and PS 3.9). The step length varied as SP of prosthesis A was 1.87 m and prosthesis B 1.96 m and PS prosthesis A 1.70 m and prosthesis B 1.72 m. During start the differences between legs were accentuated as the prosthesis to sound leg steps were short and the sound leg to prosthesis leg steps were long (figure 21).

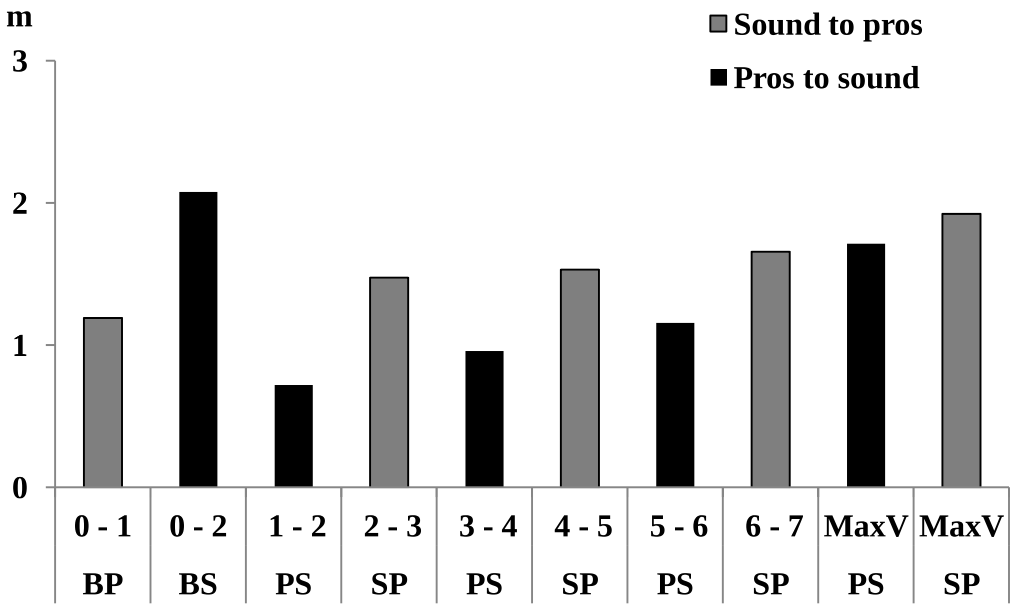

FIGURE 21. Step lengths during start and maximal running velocity. B = from the starting blocks to the P = prosthesis leg or S = sound leg, SP = sound to prosthesis leg, PS = prosthesis to sound leg.

Average step width was 0.30 m (0.22 m - 0.40 m) in maximal velocity for all trials. The average step widths varied more for prosthesis A than for B: from sound to prosthetic leg 0.27 m and from prosthetic to sound leg 0.34 m for prosthesis A (figure 20) and 0.30 and 0.31, for prosthesis B, respectively. During start the step width decreased from initial 0.60 m to 0.41 m in the 5 - 6th step (figure 22). The largest step width was 0.65 m for 1 - 2nd step.

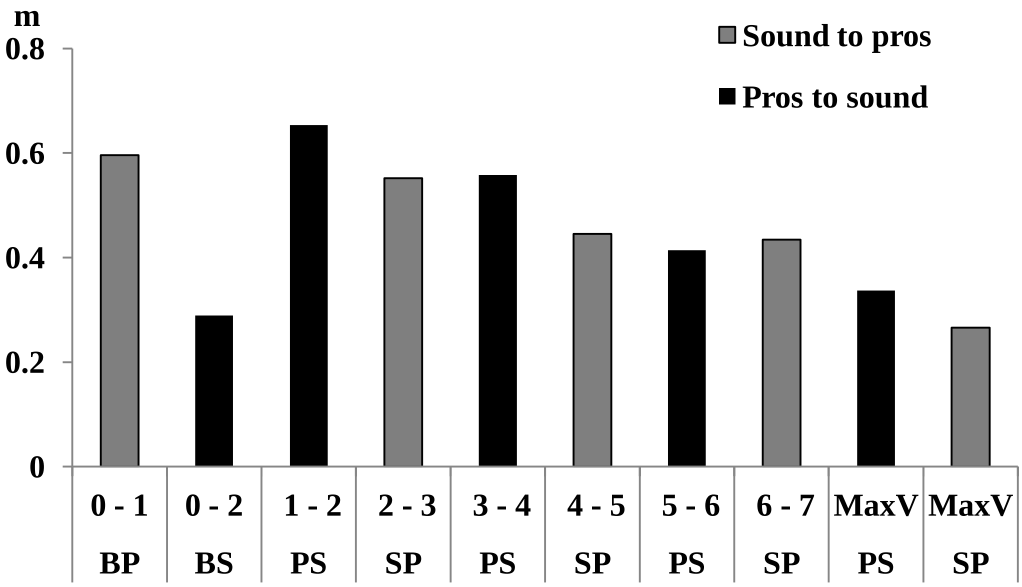

FIGURE 22. The average step widths during start and maximal velocity trials for prosthesis A. B = from the starting blocks to the P = prosthesis leg or S = sound leg, SP = sound to prosthesis leg, PS = prosthesis to sound leg.

10.2 Kinematic parameters

Kinematic data differed markedly between sound and prosthesis legs and there were small differences in some parameters between prostheses A and B. The results are shown only for the prosthesis A trials of the sound limb as there were no differences in sound limb data between prostheses A and B trials, and for both prostheses legs. The graphs were chosen to represent typical kinematics of the gait cycles. The maximal or minimal values found in the whole data are not necessary included in the presented graphs.

Stick figures from the main points of the gait cycle of the prosthesis A trials are presented in the figure 23. These figures clearly show the asymmetrical gait between the legs and also upper body compensation mechanisms.

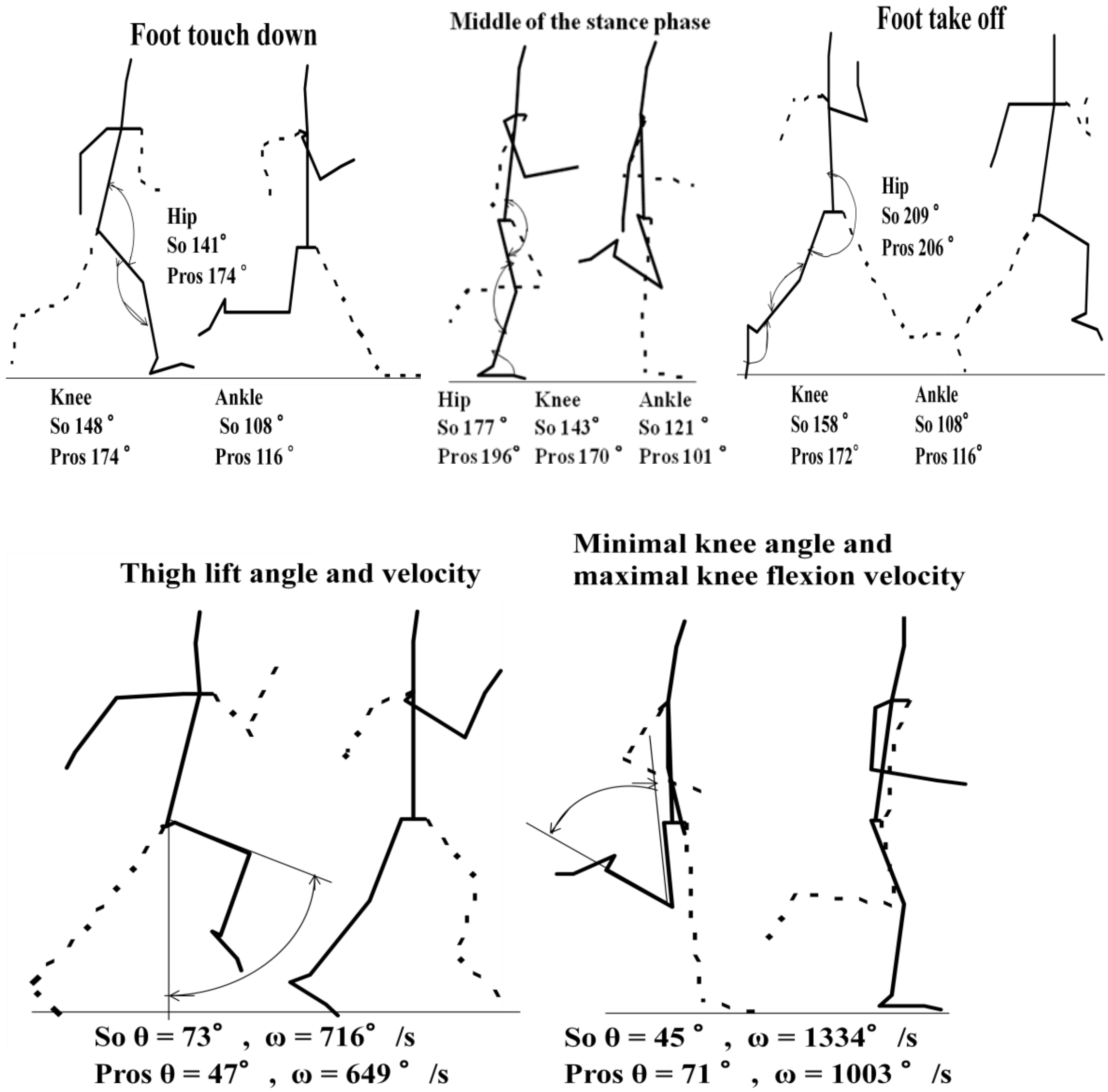

FIGURE 23. Upper row: Hip, knee and ankle joint angles during stance phase at foot touch down, middle of the stance and at foot take off. Lower row: Thigh lift angle and its' velocity and maximal knee angle and maximal knee flexion velocity during leg swing phase. The left figure is the sound leg stick picture and the right figure the prosthesis leg stick picture. Dotted line = prosthesis leg, solid line = sound leg.

Hip angular displacement data during a step cycle was similar for both prostheses (figure 24). Hip angles during later half of the swing phase were markedly asymmetrical, minimal flexion angle was 140° for the prosthetic leg and 100° for the sound leg. Greater hip flexion at the end of the swing phase resulted more flexed hip at the foot touch down for the sound leg (141°) than for the prosthesis leg (173°). Hip angles were similar at the foot take off; therefore the hip extension velocity of the sound leg was faster than that of the prosthesis leg during the stance phase. The upper body extended slightly backwards during the prosthetic leg swing to facilitate stepping on to a straightened leg.

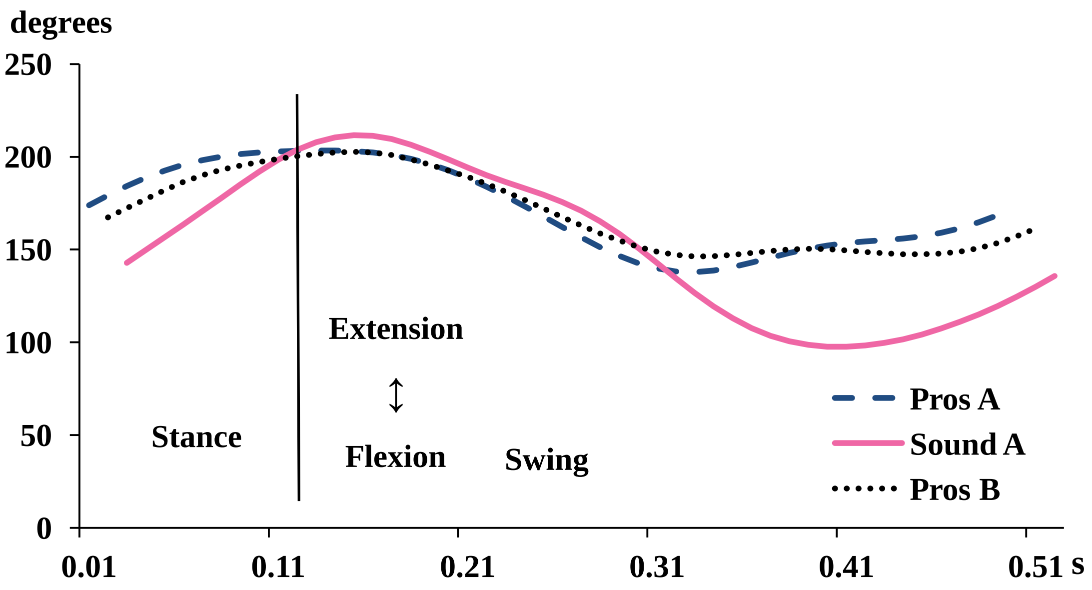

FIGURE 24. Hip angular displacement data of one step cycle for sound leg for prosthesis A trial (Sound A) and for both prostheses in prosthetic leg (Pros A & B). Vertical line is set to foot take off. Angle 180° means full extension of the hip joint, > 180° overextension and 0° flexion.

Knee joint angles were asymmetrical because of the full extension of the prosthesis knee during the contact (figure 25). The prosthesis leg stayed extended through the stance phase and extended in the swing phase. The small variability between 180 - 170° in the extended knee angle was due to prosthesis leg knee joint unit being located more anterior than in the sound leg and therefore giving slightly misleading results in the digitizing process. Prosthesis leg knee also stayed more extended through the swing phase than the sound leg. Minimum knee flexion angle showed asymmetry: prosthesis leg 71° and sound leg 37°. Knee angles were slightly more flexed for prosthesis B than A throughout the cycle, the main difference being the minimum knee flexion angle of the swinging leg which was smaller for prosthesis B 39° than A 71°. As the subject had no direct volitional control of his knee due to amputation, the minimal knee angle depended on the prosthesis mechanical swing stopper. The stopper was located on the back side of the prosthesis thigh and because of different shapes of the prosthetic legs; the minimal knee flexion angles were greatly different for both prostheses.

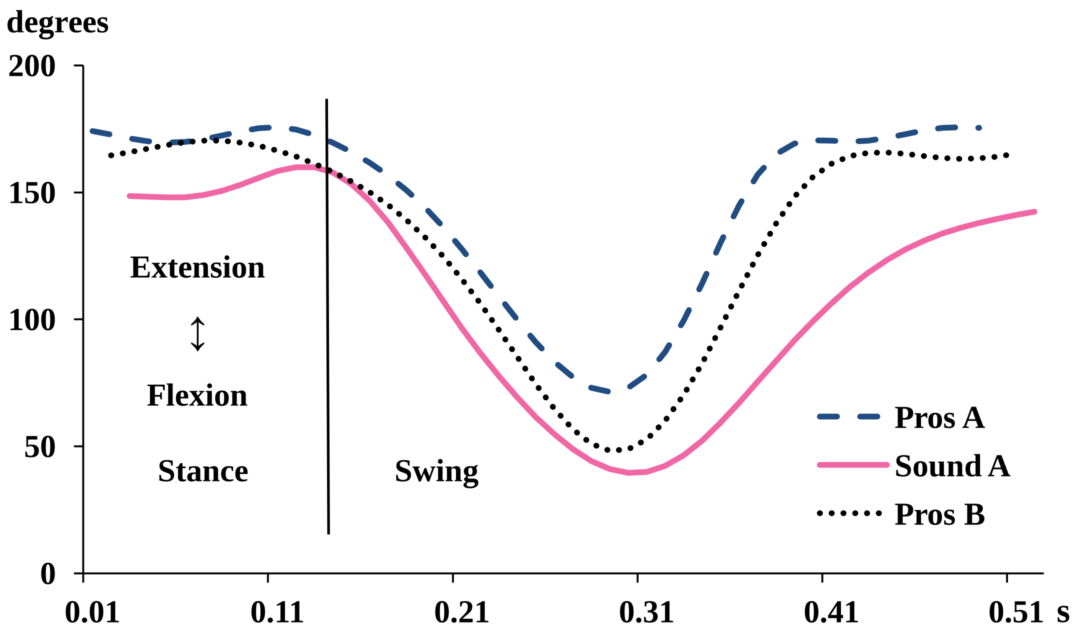

FIGURE 25. Knee angular displacement data of one step cycle for prosthesis A trial (Sound A) and for both prostheses in prosthetic leg (Pros A & B). Vertical line is set to foot take-off. Angle 180° means full extension of the knee joint and 0° flexion.

The ankle angular displacemant also showed asymmetry between the legs, especially in the swing phase (figure 26). The ankle angles were similar at the moment of the touch down and in the middle of the stance phase, but the ankle of the sound leg extended more in the end of the stance phase and stayed extended to a 145 - 150° until late swing phase. The ankle angles of the prosthesis leg extended slowly through the swing phase, but the maximal entension angle was less than for the sound leg and occurred later in the swing. There were small differences in the angular displacemant data of the ankle between the two prostheses. The ankle was about 10° more flexed throughout the whole step cycle with prosthesis B than A. The prosthesis leg is very different in the shape and also in the function than the normal ankle. The prosthesis leg flexes during early stance similarly than the ankle of the sound leg but in the push-off phase a normal ankle extends fast and reaches maximal extension about 150° in the early swing phase. The prosthesis leg behaves like a spring model and therefore it extends during propulsion and swing phase, but extension is slower and limited compared to the sound ankle.

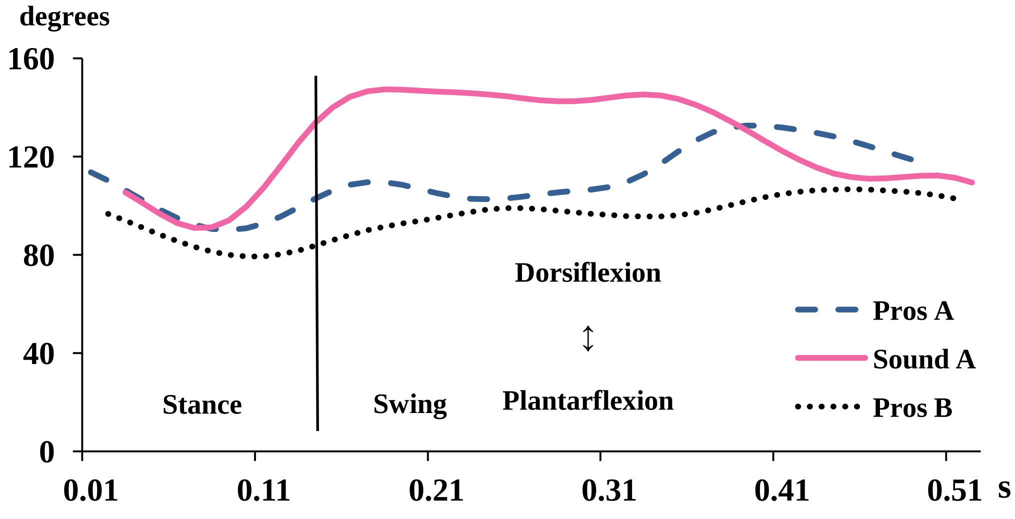

FIGURE 26. Ankle angular displacement data of one step cycle for prosthesis A trial (Sound A) and for both prostheses in prosthetic leg (Pros A & B). Vertical line is set to foot take-off. Angle 180° means full extension of the ankle joint and 0° flexion.

Thigh and leg angular displacement graphs are shown in the figures 27 and 28. Maximal thigh lift angle was much greater for the sound leg (73°) than for the prosthesis leg (prosthesis A = 47°, B = 42°) during swing phase. Thigh lift velocities varied from 610 deg/s to 750 deg/s of all trials. The highest velocity achieved (750 deg/s) was in the prosthesis leg in prosthesis B trial. The average of four analyzed step cycles was 680 deg/s for both the prosthesis and the sound leg. The maximal angular velocities of the sound leg and prosthetic legs of prosthesis A trials were 720 and 650 deg/s, respectively.

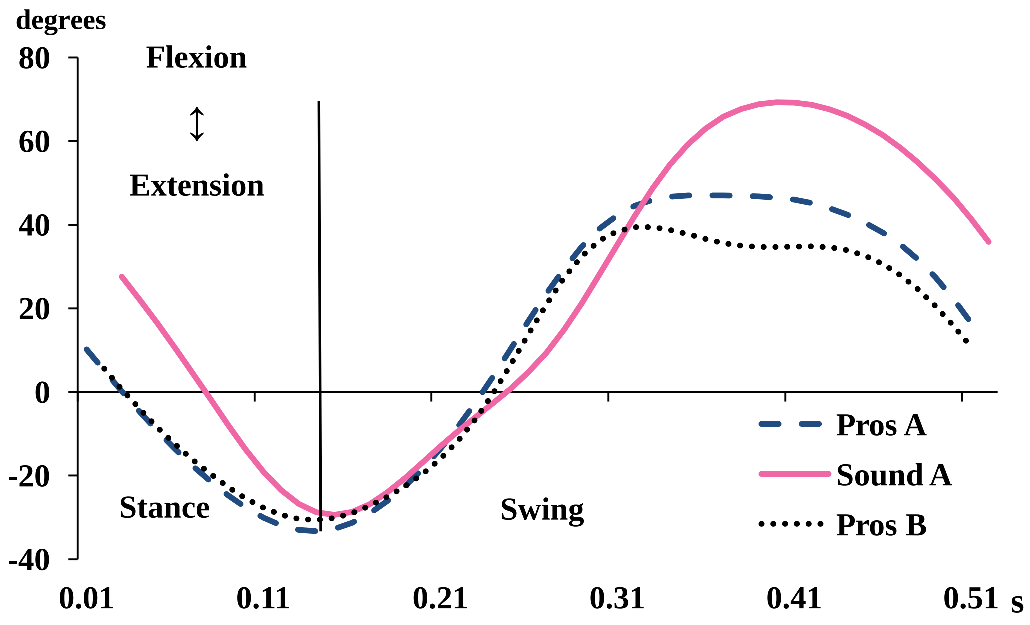

FIGURE 27. Thigh angular displacement data of one step cycle for prosthesis A trial and for both prostheses in prosthetic leg. Vertical line is set to foot take-off.

Leg angular displacement data was very similar between sound and prosthesis legs and between prosthesis A and B. The differences were found at the end of the swing phase when the maximal leg angle was bigger for prosthesis leg 47° than for sound leg 37° (prosthesis A) and contrary to this, smaller for the prosthesis (34°) than the sound leg

(37°) in the trials with prosthesis B. Maximal leg touch down velocities were slightly bigger for prosthesis A (480 - 490 deg/s) than B (410 - 460 deg/s), but there were no marked differences between sound and prosthesis legs.

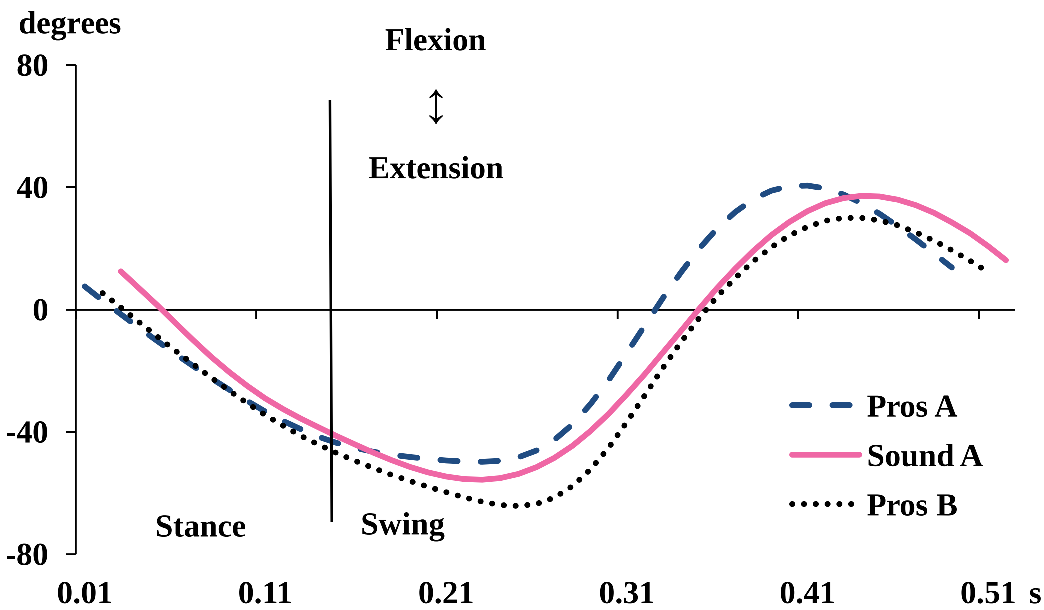

FIGURE 28. Leg angular displacement data of one step cycle for both prosthetic and sound leg for prostheses A and B trials. Vertical line is set to foot take-off.

As the distribution of mass of an amputee is not symmetrical, the center of mass is not located in the center line of the body and calculating the center of mass is a very difficult task. Therefore, the middle point of the hip is used instead of center of mass to describe the vertical movement of the center of mass. There was some variation in the graphs drawn from different trials, but no systemic differences between prostheses A and B was found. One graph from prosthesis A trial is presented on figure 29 to describe the typical movement of the middle point of the hip. The lowest point of the middle point of the hip during gait cycle was during the stance phases and the highest point in the middle of the swing phase when both legs were in the air after prosthesis leg contact.

During the stance phase the middle point of the hip was lower for the sound leg than for the prosthesis leg, although the difference was minor. The measured values varied between 74 - 84 cm for prosthesis A and 75 - 86 cm for prosthesis B.

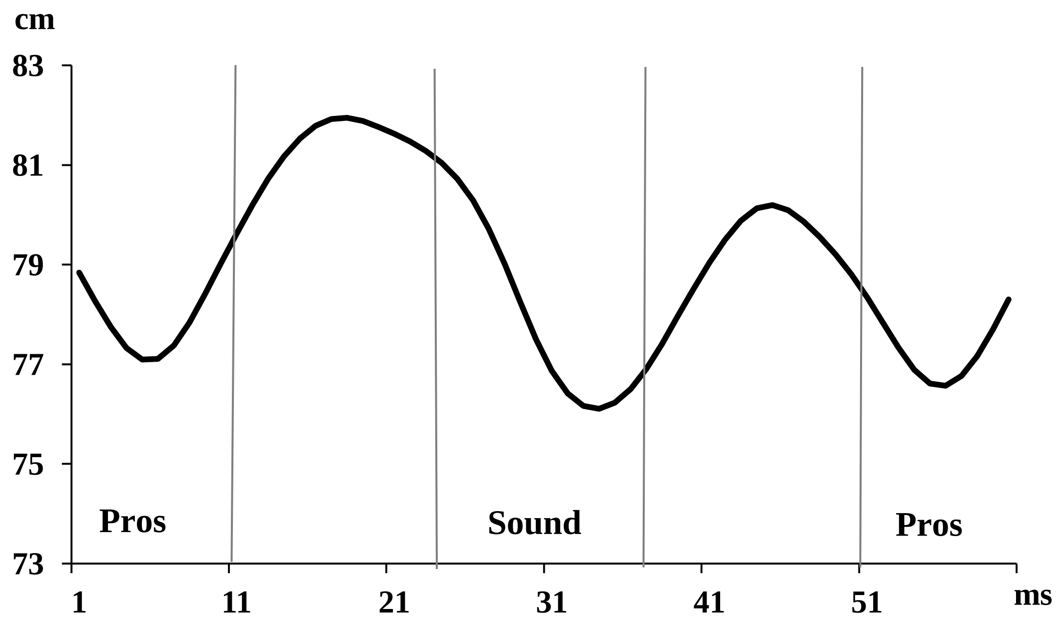

FIGURE 29. Vertical displacement of the middle point of the hip in prosthesis A trial. Pros = prosthetic leg stance, Sound = sound leg stance.

The horizontal medio-lateral displacement of the head, hips, knees and toes is drawn in figure 30. Displacement data of the head is presented in both curves (black line). The relation of the head to the hips is presented in the left graph and the relation of the knees and toes to the head on the right side. The hip displacement of the sound side was nearly equal to the head during late swing and sound side stance. In contrast, the prosthesis limb displacement data was 10 - 15 cm further from the head displacement; therefore the subject was leaning strongly on the sound side during most of the cycle and even during prosthetic leg stance. The left graph indicates displacements of the toes and the knees. The prosthesis knee was located about 15 cm laterally from the head, sligthly further than the hip, and the toe even further highlighting the fact that the subject was leaning on the sound side during sprinting. Interestingly, quickly after the stance phase in both legs, the toes and the whole foot was turned medially inwards. The knee of the prosthesis leg remained on its straight position, but the sound leg knee joint turned outwards to compensate for the foot movement. Hip and thigh abducted after the stance phase and adducted in the late swing phase of the sound side.

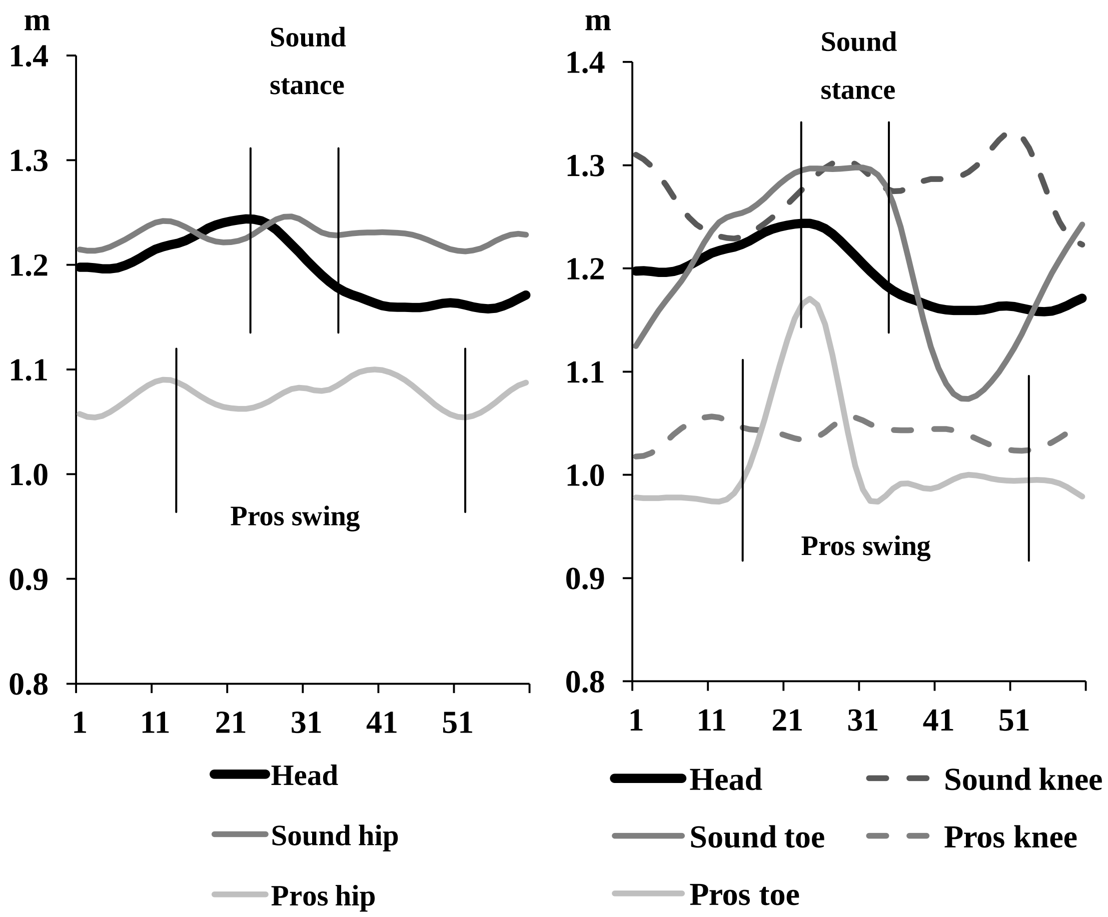

FIGURE 30. Horizontal medio-lateral displacement of head, hips, knees and toes. On the left graph, displacement of the head and both hips are presented and on the right, the displacement graphs of the knees and the toes related to the head are presented. Stance and swing phases are separated by vertical lines, Sound = sound leg, Pros = prosthesis leg.

10.3 Kinetic parameters

10.3.1 Vertical ground reaction force

The pattern and magnitude of the recorded kinetic data showed similar inter-limb asymmetry as the temporal data. A peak vertical GRF reached 2760 N for the sound leg but only 1500 N for prosthesis leg in maximal velocity for prosthesis A (figure 31). Normalized to body weight the values were 4.76 and 2.59 BW, for the sound and the prosthesis legs, respectively. The GRF for prosthesis B was bigger, 3000 N (5.14 BW) for the sound leg and 1940 N (3.31 BW) for the prosthesis leg, at maximal velocity than for prosthesis A. The average forces for the sound leg were 1490 N (2.71 BW) for prosthesis A, 1480 N (2.69 BW) for prosthesis B and smaller in magnitude for the prosthesis leg 930 N (1.69 BW) for prosthesis A and 1190 N (2.16 BW) for prosthesis B. The force curves for both prostheses were similar, therefore only the prosthesis A trials are shown in figures.

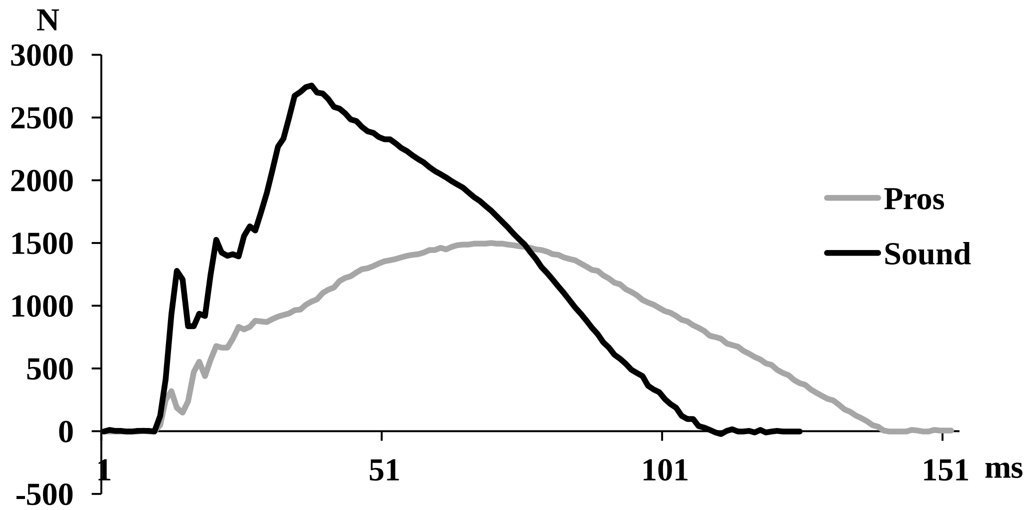

FIGURE 31. Vertical ground reaction force curves for both sound and prosthesis (Pros) legs when using prosthesis A.

The GRF curve for the sound leg in maximal velocity and also in start trials had typically the first impact peak occurring quickly (0.003 s) after the touch down and followed by the second smaller impact peak at 0.010 s and reaching the maximum value at 0.027 s in prosthesis A maximal velocity trial (figure 31). The prosthesis B maximal velocity trial of the sound leg had almost the same curve shape and timing for the impact peaks as the first impact peaks occurred at 0.005 s, the second at 0.012 s and the maximum peak at 0.026 s. In contrast to the sound leg GRF curves, the curves differ in shape for the prosthesis trials in maximal velocity in the prosthesis leg prosthesis A trial (figure 31) there was a small impact peak at 0.002 following some small peaks before reaching the highest peak value 1500 N at s 0.059 s.

Vertical impulse values were greater for the sound leg than the prosthesis leg during stance phase and also in braking and propulsion phases. The vertical impulse was negative in the first and third steps in the start and turned to positive in the fifth step and in maximal velocity for the prosthesis leg contacts. In the sound leg steps during start and maximal velocity, the vertical impulse was positive and significantly greater than the previous prosthesis leg impulses. The vertical impulse in braking phase relative to body mass increased as the running velocity increased (figure 32a). The vertical impulse in propulsion phase relative to body mass decreased as the running velocity increased (figure 32b).

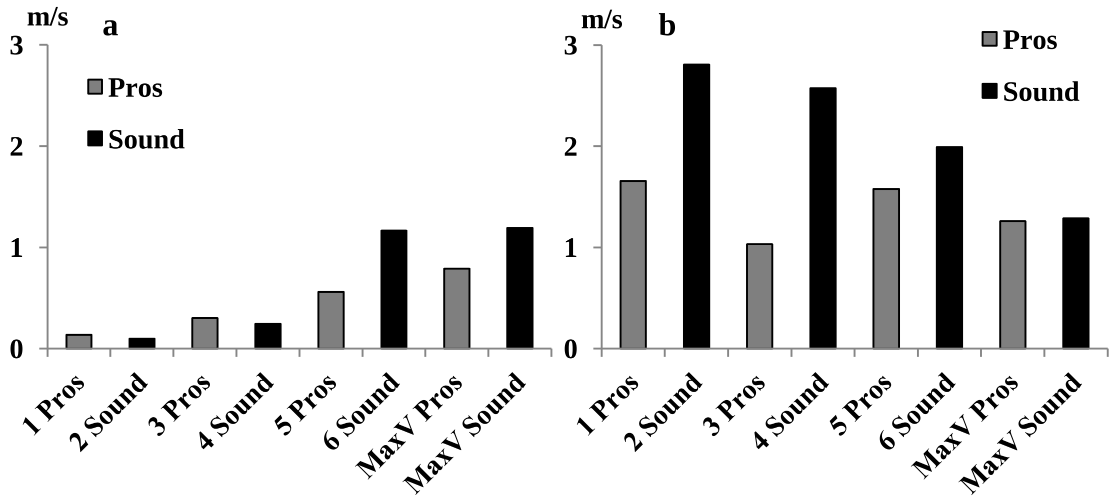

FIGURE 32. Vertical braking (a) and propulsion phase (b) impulses relative to body mass for prosthesis A trials in start and in maximal running velocity for each step. Pros (grey) = Prosthesis leg, Sound (black) = sound leg.

10.3.2 Anterior-posterior ground reaction force

The curve for horizontal anterior-posterior ground reaction force at maximal running velocity for prosthesis A trials for both legs is presented in figure 33. The forces in the braking and propulsion phases are much lower for the prosthesis leg than the sound leg as the peak braking forces are 970 N (1.77 BW) for the sound leg and 330 N (0.59 BW) for the prosthesis leg and peak propulsion forces are 460 N (0.83 BW) and 260 N (0.47 BW), respectively, in prosthesis A.

There was a short propulsion phase before actual braking phase in prosthesis leg; this was due to full knee extension. This can be seen in figure 33 as a short negative phase in the beginning of stance phase in the prosthesis leg prosthesis A. The peak value in prosthesis A trial was 90 N (0.16 BW) in maximal velocity. This pre-propulsion phase duration decreased from the start 0.015 s to 0.03 s in maximal velocity. Prosthesis leg pre-propulsion impulse decreased from the start to the maximal velocity.

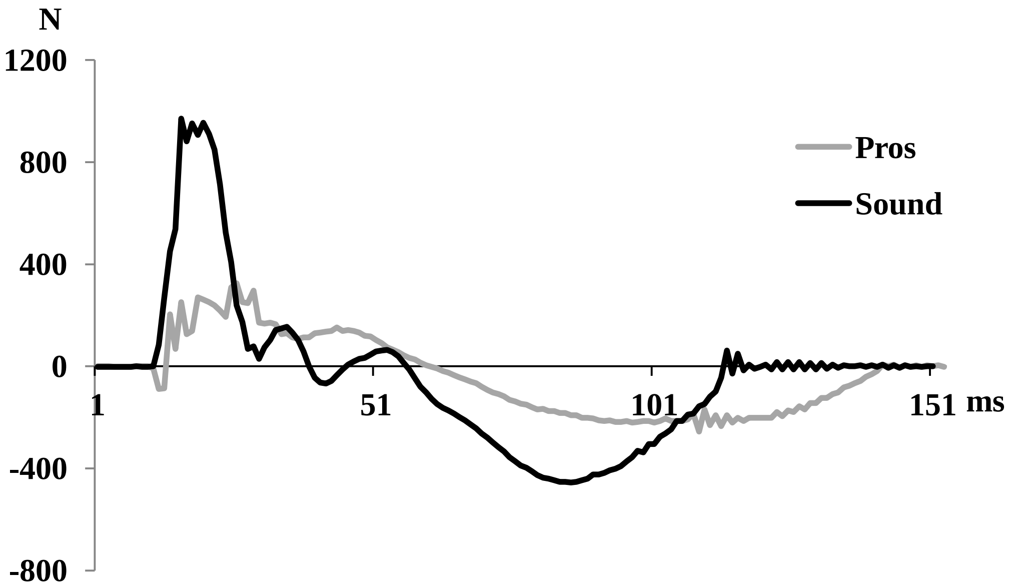

FIGURE 33. Horizontal anterior-posterior GRF at maximal running velocity for prosthesis (Pros) and sound leg for prosthesis A trials.

Horizontal braking phase impulse increased as the running velocity increased (figure 34a). On the other hand, the propulsion phase impulse decreased from the start to the maximal velocity (figure 34b). Average forces of the prosthesis leg were low compared to the sound leg in the braking and propulsion phases.

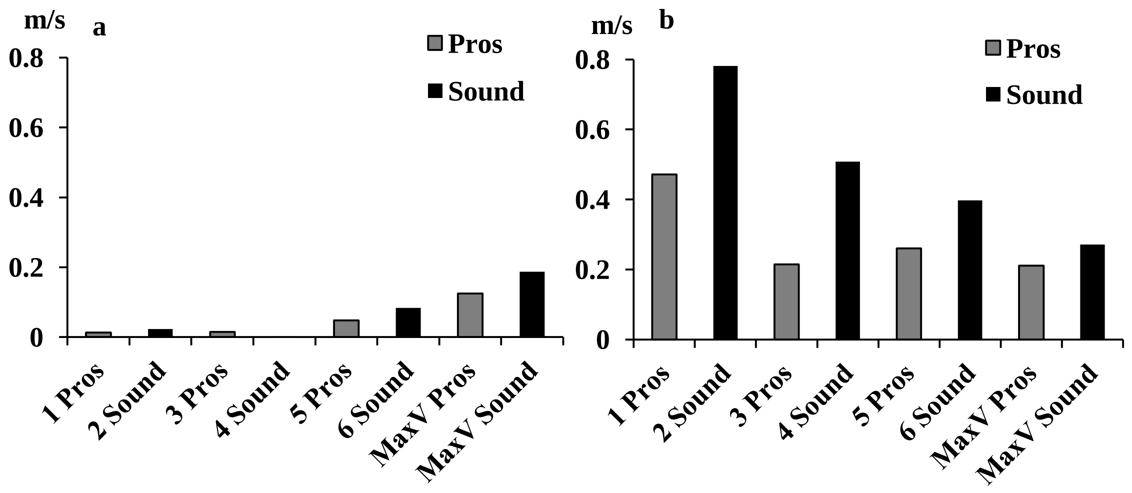

FIGURE 34. Braking phase impulse (a) and propulsion phase impulse relative to body mass for prosthesis A trials in start and in maximal running velocity for each step. Pros (grey) = Prosthesis leg, Sound (black) = sound leg.

10.3.3 Medio-lateral ground reaction force

The medio-lateral ground reaction force values were mainly on the lateral side on both legs in all trials; only just before toe-off the force values turned on the medial side. As the large step width predicted the lateral force was relatively big. The highest peak value was almost 500 N (0.86 BW) for the sound leg in prosthesis A in maximal velocity trial (figure 35).

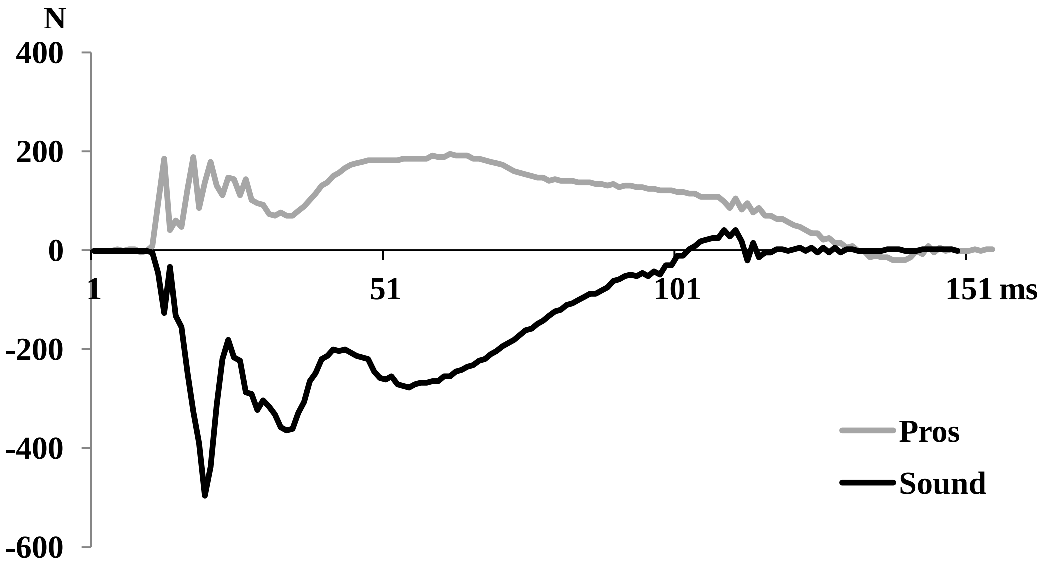

FIGURE 35. Horizontal medio-lateral GRF at maximal running velocity for prosthesis (Pros) and sound leg for prosthesis A trials.

The lateral average forces were rather constant during all trials, varying from 90 N (0.16 BW) to 210 N (0.38 BW) for both prosthesis A and B (figure 36).

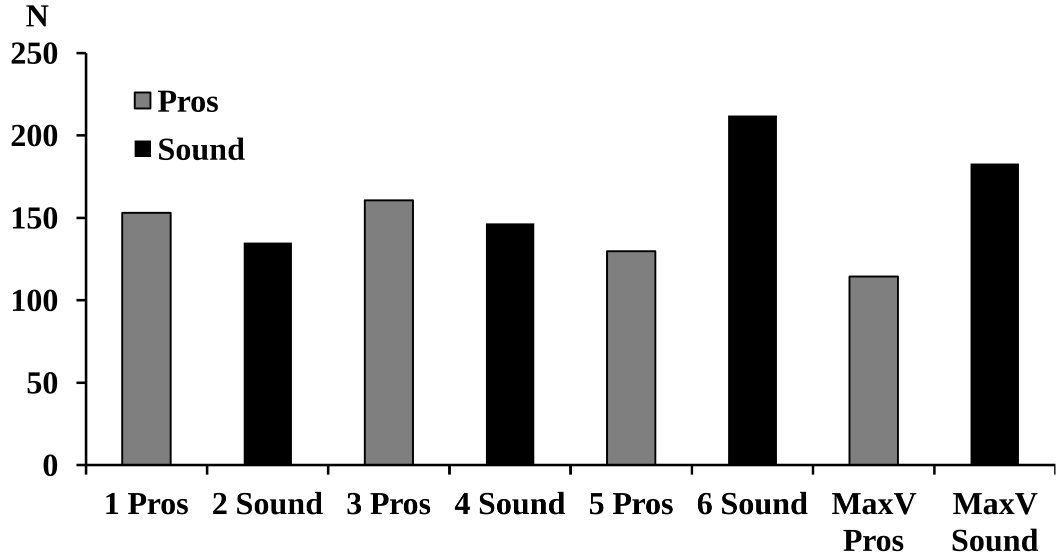

FIGURE 36. Lateral average forces for prosthesis A trials in start and maximal running velocity. Pros (grey) = Prosthesis leg, Sound (black) = sound leg.

10.3.4 Joint moments

Figure 37 displays ground reaction forces for prosthesis leg in maximal velocity, prosthesis A trial. Figure 38 presents free body diagrams from prosthesis leg at touch down in the braking phase and at the toe off in the propulsion phase. During the braking phase, a vertical force component and a vertical knee joint moment are great and both the horizontal force and the moment are small. The ground reaction force vector causes perpendicularly the net external knee moment to extend the knee. Therefore at touch down and beginning of the stance phase the the knee joint produced extension moment. In the propulsion phase the moments are produced on the opposite way, as the vertical forces cause the knee flexion moment and horizontal forces the knee extension moment. At the end of the stance phase the horizontal force component and moment are great and the vertical force component and moment are small, leading to knee flexion at toe-off.

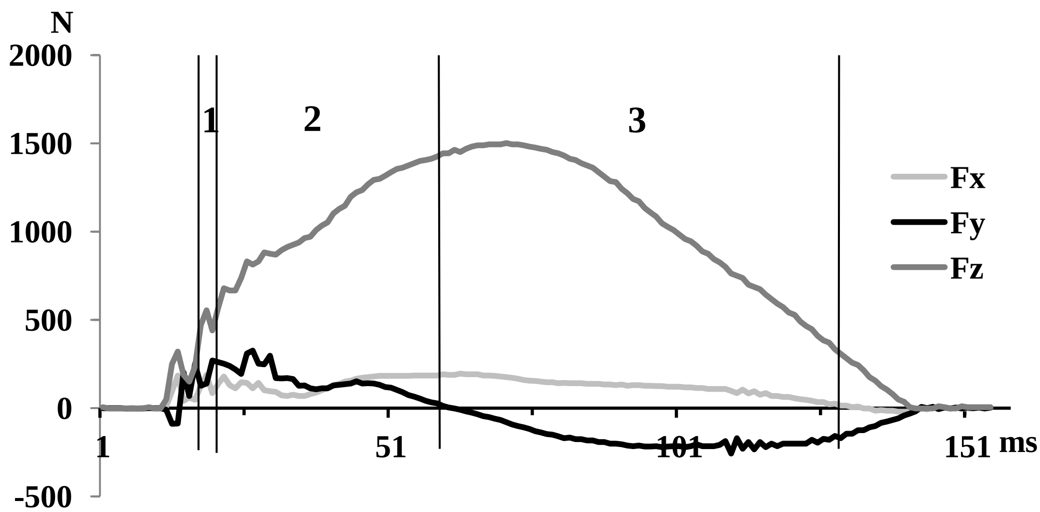

FIGURE 37. Ground reaction forces of prosthesis leg during the maximal velocity trial with prosthesis A. Fx = medio-lateral force, Fy = horizontal anterior-posterior force, Fz = vertical force. 1 = short acceleration phase, 2 = braking phase, 3 = propulsion phase, with regard to Fy.

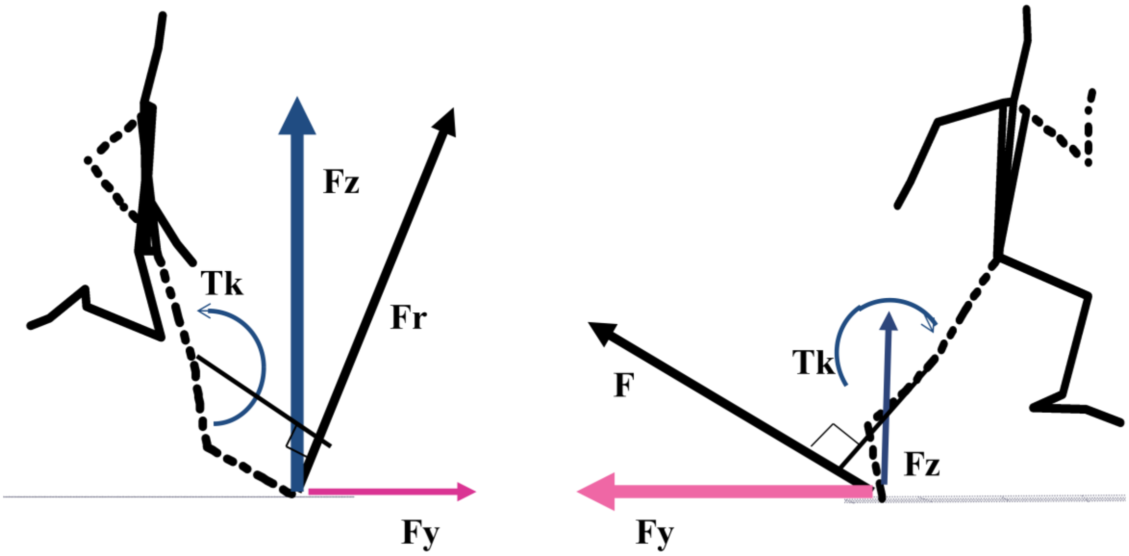

FIGURE 38. Free body diagrams presenting prosthesis foot touch down (left) and toe-off (right) and the forces and knee moments affecting to the body. Fr = gravitational force, F = Force vector affecting to prosthesis foot, Fy = horizontal component and Fz = vertical component, Tk = torque of the knee.

11 DISCUSSION

The results showed a great asymmetry between the legs in all measured outcomes, but differences between the two prostheses were small. The highest total running velocity was acquired with the prosthesis A. The maximal running velocity (7.45 m/s) was the highest ever reported in transfemoral amputee studies.

There was no difference in the contact times in maximal velocity trials between the two prostheses. The contact times were longer for the prosthesis leg than the sound leg. There have been controversial results in other amputee research showing shorter contact time for prosthesis leg (Sanderson & Martin 1996; Burkett et al. 2003). The gait cycle time remained similar with either prosthesis. Timing of the gait cycle becomes from adjustments in the swing phase time.

Previous studies have shown that a delay in prosthesis leg swing time is one of the biggest problems and causes asymmetrical running gait. In earlier studies the researchers have found that by lowering a knee joint (Burkett et al. 2001) and developing hydraulic knee joint units (not allowed in athletics) can assist in prosthesis swing. This study showed shorter swing time and longer stance time for the prosthesis leg. These together made the gait cycle time the same than in the sound limb. Transfemoral amputee's ability to adjust the swing time to contact times makes rhythmic sprint running possible. It seems that the major compensation mechanism enabling running is to adjust the gait cycle time to a constant even the contact times vary between sound and prosthesis legs.

The main result in sprinting kinematics to explain the asymmetrical gait was found in the prosthesis knee joint full extension during stance phase. The function of the prosthesis knee joint does not permit flexion during foot contact and therefore the knee stays extended during stance. This is the case for all transfemoral amputees and it is well reported (eg. Buckley 1999; Jaegers et al. 1995; Burkett et al. 2003). The knee must land as extended and to achieve this, the leg is extended early in the front swing. To assist in knee extension, upper body is bent slightly backwards in the swing of the prosthesis limb and then pushed down and quickly rolled over the supporting leg. The hip flexion of the prosthetic side is reduced at the end of the swing and during stance phase. Reduced hip flexion angle leads to a shorter step length and thus to lower running (Burkett et al. 2003). The center of mass stayed higher during the prosthesis leg stance phase of the sound leg as a result of full knee extension. However, the difference was small because of the heel striking of the sound leg and leaning on the sound side even during prosthesis leg stance.

There was a great inter-limb asymmetry in the kinetic data. Vertical and horizontal braking and propulsion GRFs were smaller in the prosthetic than in the sound leg. Other amputee running studies have presented similar findings for transtibial amputees (Sanderson & Martin 1996) and for transfemoral amputees (Burkett et al. 2001). Those research groups thought that the reduced GRF for prosthesis leg was because of a limitation in the prosthesis foot properties (Burkett et al. 2001; Sanderson 1996) since changing the knee alignment had no improvement in prosthesis leg GRF (Burkett et al. 2001). Burkett et al. (2003) discovered greater vertical force peaks for prosthesis foot which they speculated to be the cause of poor shock absorption properties of the prosthesis. The contact time was also shorter in both walking and running in the study by Burkett et al. (2003) supporting the idea of discomfort and a higher risk for injury because of the limited shock absorbing of the prosthetic foot. Sanderson & Martin (1996) found that as the running speed increased from 2.7 m/s to 3.5 m/s the contact time shortened to be shorter for prosthesis than sound limb in faster speed. This might have been a strategy to reduce joint and stump loads (Sanderson & Martin 1996). The present study was done in 2006 and prosthetic technology has improved since these earlier studies. The running speed achieved in this study was much faster than in any previous study of the transfemoral amputee. Our subject was highly trained sprinter and due to daily training he seemed not to suffer excessive stump loading. As regards to the GRF data the prostheses used in this study had limitations in the properties to store and release elastic energy but not so much in shock absorption. The sprinting prosthesis should store and release elastic energy similarly as an intact limb during contact. During the prosthesis foot contact the body weight rolls over the leg and therefore no active force peaks were seen. The smaller the horizontal forces in prosthesis leg resulted from the lack of force producing muscles and limited properties of prosthesis. In this case a stiffer prosthesis might lead to faster running speed.

In the sprint start, the subject took very wide steps in the first 10 m which became narrower for the maximal sprinting velocity. Wide step width lead to a lower center of mass, greater lateral ground reaction forces, but the propulsion forces of the first and the second steps were great enough enabling the forward movement. Ito et al. (2006) have found that the large step width might be useful in the starting phase to produce high propulsion forces. The elite sprinters exhibited step widths about 30 - 40 cm in the first six steps (Ito et al. 2006), which were narrower than the subject of this study performed. The amputee sprinter was able to start from the blocks, which is not always the case for the transfemoral amputees. The length of the prosthesis was adjusted so that its length was the same as the sound leg when standing on tip-toes. The prosthetic limb was therefore quite long, making the start difficult. The subject had to sway the prosthetic leg lateral sideways to be able to bring it in front fully straightened to make the step. Lateral swaying caused extremely wide starting steps and also large lateral forces were produced. Although smaller lateral force production might lead to greater propulsion force production and faster forward movement due to narrower step width, this may not be the case. The wider step width during a start might be a necessary adaption to be able to perform a block start with a prosthesis leg.

The step width (mean 30 cm) during maximal sprinting velocity was also larger than in the normal-bodied sprinters (17 cm) (Ito et al. 2006). The lateral forces, which are usually minor and have no effect on sprinting velocity, were also great compared to the normal-bodied subjects. The force production laterally may decrease the forward force production and hence affect the running velocity. Another problem in the running technique was at the foot touch down, as the heel landed first in the sound leg. This reduced the hip and the center of mass. Together with the prosthesis adjusted to a long length adjusted prosthesis and the fear to fall down if the prosthesis foot touched the ground during swing phase, the subject swung his prosthesis leg on the lateral side causing the body lean strongly on the sound side. The swinging of the prosthesis laterally and bearing the upper body on the sound side lead to an unbalanced body posture and running technique. As a result of this analysis, the solution could be to shorten the prosthesis length to facilitate the prosthesis leg swing and decrease the lateral swinging and lateral forces produced. The strength training of the m. triceps surae of the sound leg could make the running with a ball contact possible. This could help to keep the hip area and the center of mass higher allowing the prosthesis leg swing straighter and to keep the body posture straight.

The performance of the sound leg and running velocity was higher for the prosthesis A than B and A was therefore regarded as better. There were not many differences in the analyzed data, although the symmetry was better for prosthesis B than A in some parameters. This raises the question weather symmetry between the legs is needed to achieve a higher running speed. It seems that the subject is able to run with his own unique technique with either prosthesis and do small adjustments to running depending on the prosthesis shape and characteristics. The weight of the prosthesis A was 400 g lighter than B. This might be the main reason for higher speed acquired with prosthesis A than B.

The strength of this study was that the subject was a top level athlete competing internationally. The running velocity was much higher than in any previous studies and he was skillful enough to start from the blocks. The subject was experienced sprinter and he was able to succeed from the trials with a reasonable amount of attempts. This study included 3D-motion analysis and ground reaction force data. Lateral forces and movement were analyzed carefully giving important results. The weakness of this study was in manual digitalization. The markers were not used in all of the digitizing points and those points were manually tracked. This may have lead to some errors, although the analysis was done carefully and the lengths of the body segments were checked, and the movement data smoothed. In the future studies, the use of markers could be useful, especially in the prosthesis leg, and also a digitizing program with an automatic analysis.

The main practical reason to do this study was to improve the running technique of the subject. As a consequence of the analysis detailed feedback was given to the subject. Length adjustments were made to the prosthesis and the subject was advised to do strength training to achieve forefoot contact with the sound leg. The subject was able to correct his running posture more upright and after a few months after the study, he ran his new personal record in 100 m, 12.85 s. There was an improvement of 0.26 s on his old record. One year later he performed well in the Beijing Paralympics and positioned fifth in 100m and won a silver medal in long jump.

Findings of the present study produced new information about the sprinting technique of transfemoral amputee especially regarding to block start technique and compensation mechanisms of asymmetrical gait.

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APPENDIX 1

Calibration pole positions