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GAIT ANALYSIS IN BELOW THE KNEE AMPUTEES FOR DISTAL ARTERIAL DISEASE

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ABSTRACT. Atherosclerosis can lead to acute lower limb ischemia that may require amputation with important functional and social burden, as well as increased morbidity and mortality. A group of 8 able prosthetic users were examined using spatial motion detection. Consistent patterns revealed increased compensatory motion at the hip of the amputated limb. We found limited peak dorsiflexion during swing which obligates to increased hip flexion. We also determined that frontal and transversal plane motion is symmetrical for the thighs and knees. Gait parameters summary shows decreased step and velocity compared to able bodied subjects as well as shorter swing phase for the prosthetic limb due to limmited mobility at the knee and angle. In conclusion gait alterations are clearly identifiable in below the knee amputees using conventional spatial motion detection. These measurements show marked abnormalities of the amputated limb compared to contralateral as well as able bodied subjects respectively. These variations are influenced both by patient skills and prosthetic design.

Keywords: gait, amputee, peripheral arterial disease, transtibial

INTRODUCTION

Atherosclerosis is a major health problem. In addition to cardiac and cerebral pathology, many patients develop acute lower limb ischemia that requires amputation. This leads to important functional and social burden, as well as increased morbidity and mortality.

A recent cross-sectional study performed by LAZZARINI et al aimed to identify the prevalence of amputations over one year in a major hospital setting. Overall there was a slight predominance for males, approximately half were first amputations and half major. The mean age was just over 60years. Half associated type 2 diabetes, followed by peripheral arterial disease, trauma, type 1 diabetes and malignant tumours (5%). Trauma patients were young adults and diabetics were old (LAZZARINI et al., 2012).

Waters et al were one of first to describe that self-selected walking velocity is directly related to amputation level. Furthermore, they measured the oxygen consumption and found that for both traumatic and vascular amputees, the cardiovascular burden of gait is also higher the more proximal the level and abnormal the amputation is (WATERS et al., 1976). Skinner et al have also shown marked differences from normal gait in both above and below the knee amputees (SKINNER et al., 1985). Forward walking speed was found lower in the amputee and reversely proportional to the amputation level (SKINNER et al., 1985). Traumatic above the knee amputees ambulate without the symmetry seen in normal subjects which increases the energy cost of ambulation and thus places the transfemoral amputees at their cardiovascular limits and strains other amputees severely (SKINNER et al., 1985).

This was later confirmed by additional studies by PINZUR et al (PINZUR et al., 1993). These authors also identified that subjects with trans-femoral amputations walk at relatively slower speeds than those with distal (trans-tibial) levels (PINZUR et al., 1993). This decrease in both self-selected walking velocity, as well as maximum walking speed, was directly related to the number of retained functioning anatomic joints (PINZUR et al., 1993). By walking at slower speeds, and requiring more oxygen, the metabolic cost of walking is proportionally greater with shorter residual extremities (PINZUR et al., 1993). In addition, as the residual extremity has less functional anatomic joints, the subjects lessen their potential to increase their walking speed (PINZUR et al., 1993). Furthermore, they found the metabolic cost of walking for peripheral vascular insufficiency through-knee amputees to be midway between above and below the knee (PINZUR et al., 1993). This research came with a keen interest in the potential benefits of knee disarticulation, considering this might be a favourable compromise for limited activity patients that might lead to better balance and improved stump weight bearing (PINZUR et al., 1993).

At the same time, studies showed that below the knee amputees increase the electrical activity of their remaining hamstring and quadriceps muscles to compensate for their absent muscle groups (PINZUR et al., 1991). Also, sedentary patients do not adequately use the propulsive capacity of their thigh musculature and may not benefit from their lever arm (PINZUR et al., 1991). Pinzur et al have also found that bilateral

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lower extremity peripheral vascular disease amputees are unlikely to become independent community walkers (PINZUR et al., 1992).

Under these circumstances, we planned to determine the gait alterations in below the knee peripheral vascular disease amputees with standard foot-ankle prosthesis.

MATERIALS AND METHODS

A group of 8 able prosthetic users with an average of 26 months since index surgery (SD=7) were examined using spatial motion detection (Zebris). After mockup tests, the patients were asked to perform 5 crosses through the sensors, with three point markers on their thighs and feet in a predetermined spatial configuration based on their individual anthropometric parameters (fig.1). They all had healed stumps, patellar tendon supporting cups and simple, articulated ankles. The standard surgical technique was used for all cases.

All patients initially presented with acute limb ischemia which required emergency amputation. The mean age was 67years (SD=9.3), 7 patients were male and 6 have had type II diabetes for 8 years (SD=5.5).

A relevant case is presented in fig.1: seen from behind the patient has 2 active markers on each leg, one on the thigh and one on the foot, that are connected to a central unit on the waist. On the left and right are two detectors. The patient anthropometric configuration is calibrated before the markers are set up, using a probe that manually defines the pelvis, the center of rotation of the hip, knee and ankle and the feet. The gait parameters are converted by a central processing unit and stored on a computer, used for data analysis.

Fig. 1 The typical setup configuration of the gait analysis system.

RESULTS

There are consistent patterns with compensatory increased motion at the hip of the amputated limb (fig. 2 - top row, green line). This modification is caused by decreased knee and ankle flexion (middle and bottom rows, green line). In addition, the rocker mechanism of the foot – ankle complex is required for a fluent stance phase. With this standard prosthetic design we can see a limited peak dorsiflexion followed by passive spring plantar flexion during swing which obligates to increased hip flexion.

The gait cycle of the prosthetic limb (right, green) is presented in fig.3. Using the spatial sensors we determined that frontal and transversal plane motion are symmetrical for the thighs and knees (fig.4). In fig. 5 we can see decreased step and velocity compared to able bodied subjects as well as shorter swing phase (green) for the prosthetic limb due to limited mobility at the knee and angle.

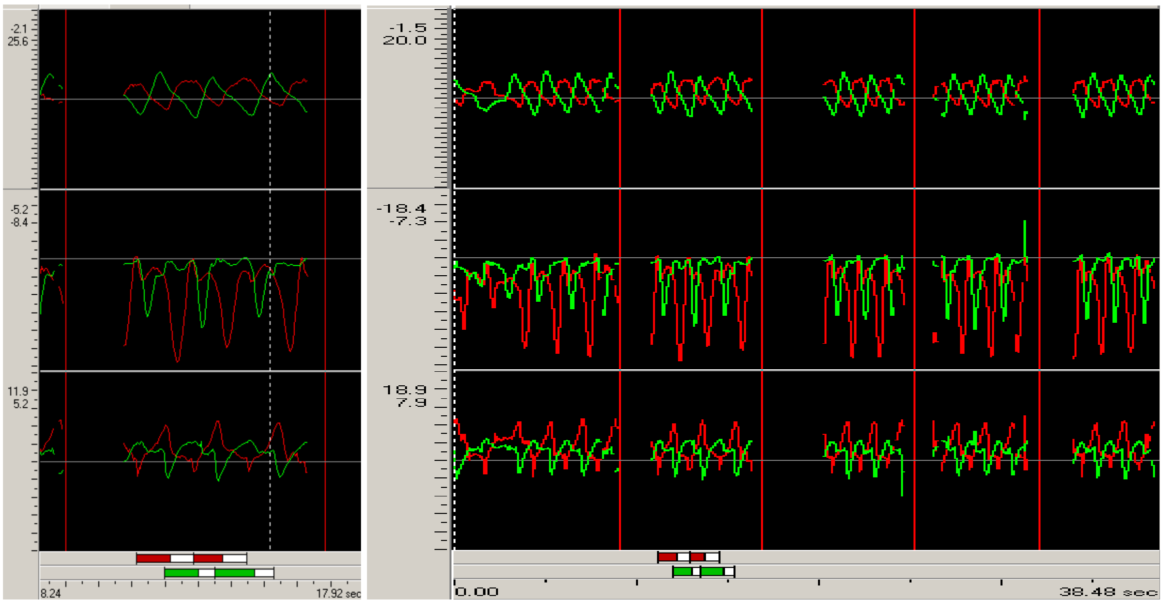

Fig. 2 Single measurement and repeat (5) sagital plane angles.

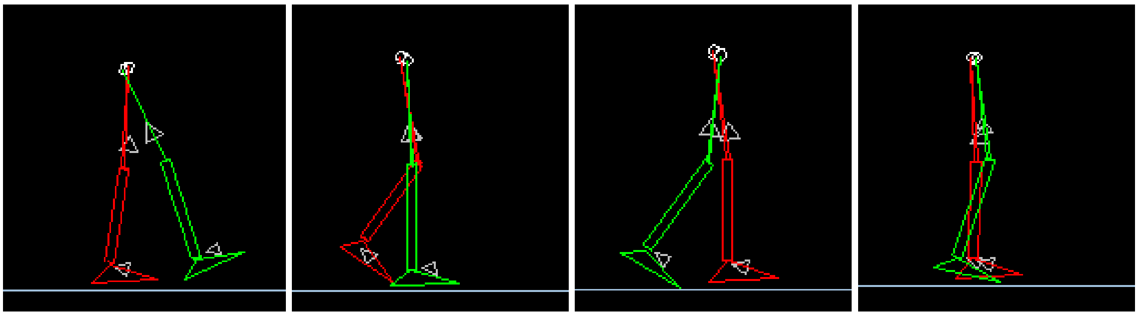

Fig. 3 Heel strike, Stance, Preswing and Swing

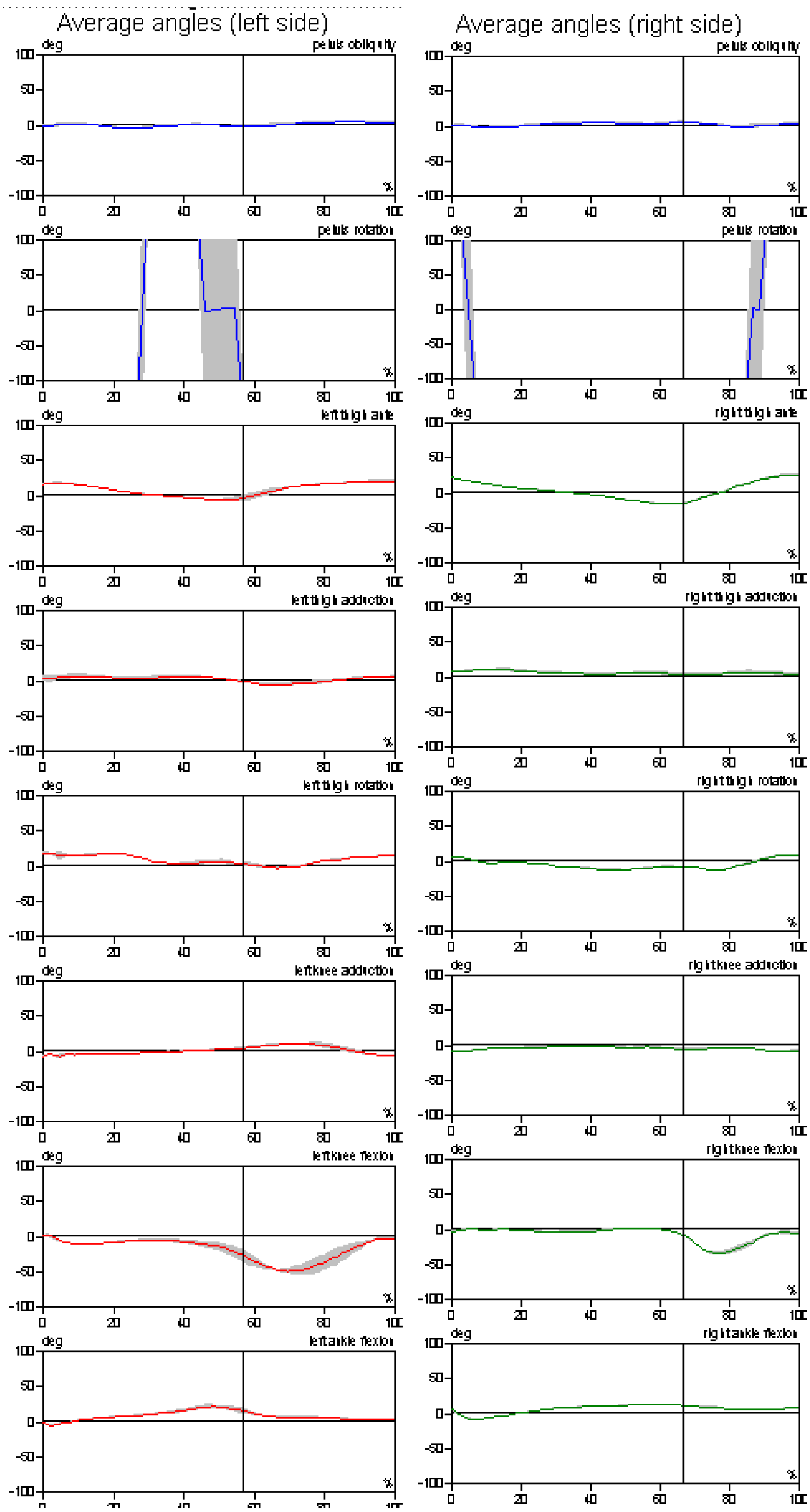

Fig. 4 Comparison of average (5 measurements) angles and standard deviation (SD) for left (red) and right – prosthetic (green) limbs.

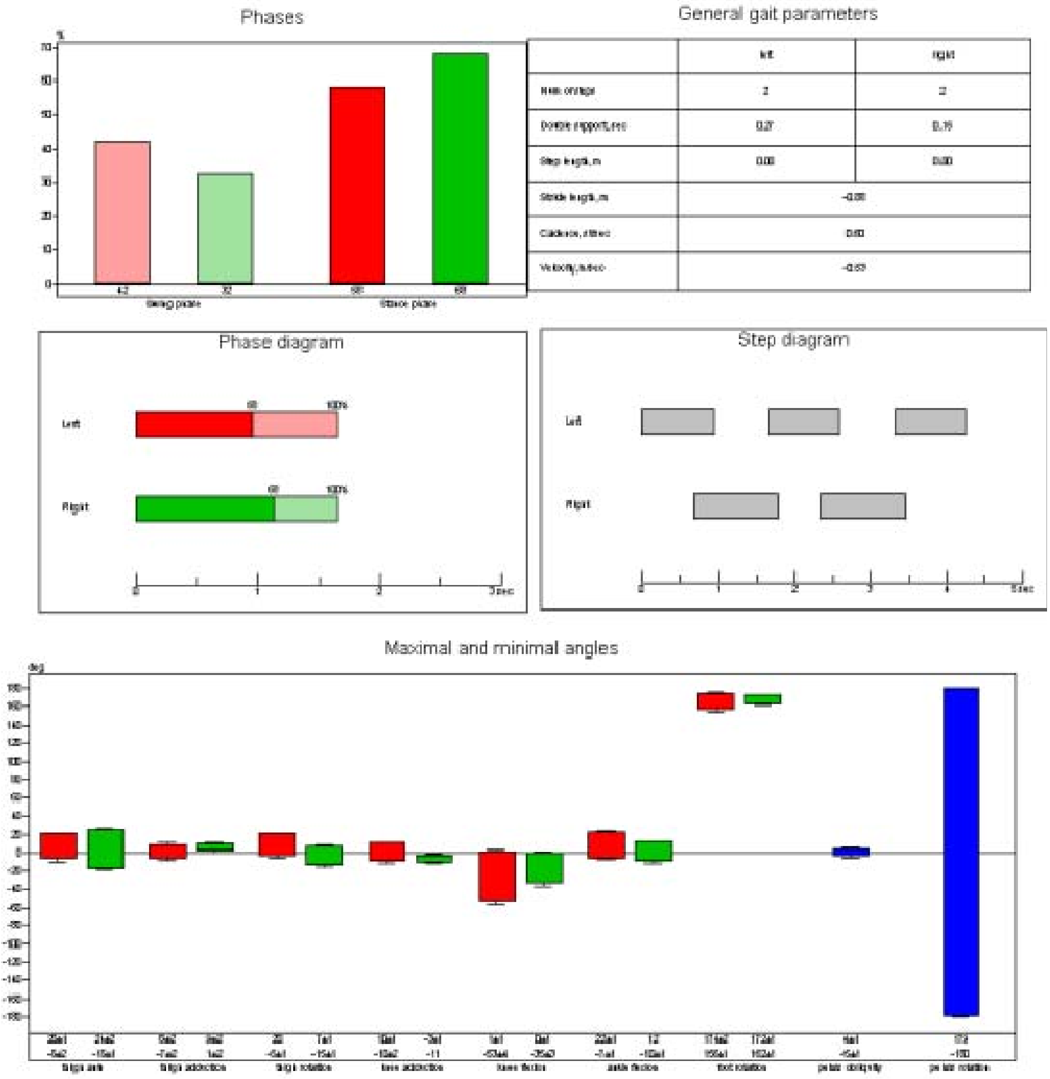

Fig. 5 Gait parameters summary.

DISCUSSION AND CONCLUSIONS

The higher the amputation, the more pelvic tilt excursion was proved for amputees. The same study showed that limb length did not correlate with any other temporal-spatial, kinematic, or kinetic parameter. Therefore, if the femur is at least half the length of the contralateral, above the knee amputation level does not dramatically alter gait (BAUM et al., 2008).

The importance of the ankle – foot prosthetic designs is underlined by the role of this segment for an efficient gait. In the able body patients this functions are active propulsion with the supporting limb during gait initiation, shock dampening at heel strike and forward transfer of motion during stance (single limb support). In addition, the whole limb participates in energy recovery and transfer. The ankle – foot prosthetic designs are generally classified into simple passive motion, passive energy accumulation and active bionics. A study comparing the revolutionary Flex foot design to standard found kinetic differences were limited to ankle joint variables in the sagittal plane with greater peak moments and power during propulsion (UNDERWOOD et al., 2004). Effects were also found at proximal joints and the non-amputated limb (UNDERWOOD et al., 2004). They conclude the dynamic foot able subjects to rely more heavily on the prosthetic foot for propulsion and stability during walking (UNDERWOOD et al., 2004).

Another study found the flexion unit increased ankle sagittal plane motion and positive ankle power. The torsion unit increased transverse plane ankle range of motion (SU et al., 2010). In addition, subjects perceived that the increased prosthetic ankle motion was beneficial for stability on uneven terrain (SU et al., 2010). The authors therefore recommend mobile components especially for persons with bilateral below the knee amputations. Nevertheless, clinicians should assess if patients can benefit from improved mobility while sacrificing some degree of stability (SU et al., 2010).

Pressures at the stump were found to be dependable on knee moments and walking condition (WOLF et al., 2009). Adapting the prosthetic ankle angle modifies joint kinetics and thus pressure distribution at the stump (WOLF et al., 2009). However, large inter-individual differences in local pressures underline the importance of individual socket fitting (WOLF et al., 2009).

Rehabilitation regimens lead to improved gait kinematics. Results in the literature showed these lead to a more normal velocity and increased symmetry in step length, but reduced symmetry in pelvic motion (SJÖDAHL et al., 2003). In the frontal plane, pelvic obliquity increased to similar amplitude to able bodied subjects, but with different timing (SJÖDAHL et al., 2003). Nonetheless, gait appeared more symmetrical, probably due to more efficient pelvic motion, more symmetrical upper-body movements and compensatory increased work with the non-amputated (SJÖDAHL et al., 2003).

In comparison with normal subjects, amputees showed decreased internal rotation moments at the prosthetic limb hip and knee during turning, maybe as a compensatory mechanism to minimize stress on the residual limb (SEGAL et al., 2011). There is also an increase in amputee sound limb hip external rotation moment in early stance, which may be a compensation for the decrease in prosthetic limb internal rotation moment during late stance of the prior step (SEGAL et al., 2011).

VRIELING et al found uphill walking leads to increased hip and knee flexion at initial contact and hip flexion in swing in the transtibial prosthetic limb (VRIELING et al., 2008). In contrast, during downhill, amputees showed more knee flexion on the prosthetic side in late stance and swing (VRIELING et al., 2008). An important adjustment in amputees is decreased hip extension in late stance during inclined amputation, probably related with a shorter step length (VRIELING et al., 2008). In addition, amputees increased knee flexion in early stance in the non-affected limb in uphill walking to compensate for the shorter prosthetic limb length (VRIELING et al., 2008). For downhill walking fewer adjustments are necessary, since the shorter prosthetic limb already lowers the body (VRIELING et al., 2008). During obstacle crossing amputees increase success rate, gait velocity and swing knee flexion of the prosthetic limb (VRIELING et al., 2009). Knee flexion in transfemoral and knee disarticulation amputees is not sufficient for safe obstacle crossing, which leads to circumduction (VRIELING et al., 2009). During initiation and termination amputees increase the anterior-posterior ground reaction force and the centre of pressure shift in the mediolateral direction (VRIELING et al., 2009). The centre of pressure shifts anteriorly before single-limb stance on the prosthetic limb during initiation, and remains posteriorly when leading with the prosthetic limb respectively (VRIELING et al., 2009).

In conclusion we found that gait alterations are clearly identifiable in below the knee amputees using conventional spatial motion detection. These measurements show marked abnormalities of the amputated limb compared to contralateral as well as able bodied subjects respectively. These variations are influenced both by patient skills and prosthetic design.

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