



METHODOLOGICAL PROPOSAL FOR INSOLE PRESCRIPTION TO TRANSFEMURAL AMPUTEES BASED ON BIOMECHANICAL PARAMETERS: A PRINCIPAL COMPONENT ANALYSIS APPROACH

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Resumo

Durante o processo de alinhamento da prótese e treinamento da marcha dos indivíduos com amputação, procura-se um padrão de marcha o mais próximo possível do encontrado nos sujeitos sem amputação. O conhecimento do padrão da marcha em indivíduos não amputados, bem como a influência de ortóteses, como cunhas, parece ser uma importante ferramenta de otimização da marcha para amputados transfemorais (TF). Uma desvantagem na utilização de dados quantitativos para análise de marcha é a enorme quantidade de informações geradas, proveniente da análise de pontos discretos das curvas de dados. A Análise de Componentes Principais (PCA) é uma poderosa ferramenta utilizada para reduzir a informação redundante permitindo a comparação das curvas como um todo. Os objetivos deste estudo foram: a) avaliar a influência das cunhas na força de reação do solo (FRS) e no centro de pressão (COP) durante a marcha dos indivíduos não amputados, b) avaliar a influência das cunhas no momento muscular resultante e na amplitude de movimento durante a marcha dos indivíduos não amputados, c) comparar os parâmetros FRS e COP entre TF e não amputados durante a caminhada/marcha d) desenvolver um modelo biomecânico para otimização da marcha de amputados TF através da prescrição e teste de cunhas. A FRS (plataforma de forças – 1000Hz), deslocamento do COP (plataforma de pressão – 300Hz) e cinemática 3D (videogrametria – 50Hz) foram recolhidas para uma velocidade de marcha auto-selecionada em 20 sujeitos não amputados e 15 TF. A influência de cada uma de 6 cunhas foi analisada nos não amputados e, com base nesses resultados, foi desenvolvido em MATLAB® um programa para verificar se as cunhas influenciariam positivamente a marcha de TF. O modelo biomecânico desenvolvido foi testado em 3 TF. Obtiveram-se os seguintes resultados: a) As cunhas influenciam a FRS e o COP nos não amputados; b) As cunhas influenciam os momentos musculares e amplitude de movimento dos não amputados; c) a PCA mostrou-se válida para discriminar diferenças na marcha entre TF e não amputados; d) O modelo biomecânico mostrou-se válido para prescrever cunhas adequadas a TF. Uma nova proposta metodológica baseada na PCA das curvas de FRS e COP mostrou-se válida para aproximar os valores do membro remanescente de amputados TF dos valores do grupo de não amputados. A aplicação desta proposta poderá auxiliar técnicos e fisioterapeutas a decidir a melhor intervenção a seguir para cada paciente, e assim melhorar a qualidade de vida dos pacientes TF.

Abstract

During the alignment process of prosthesis and amputees' gait training, a gait pattern as close as possible to that found on their able-bodied peers is wanted. The knowledge of the able-bodied gait pattern as well as the influence of gait aid devices such as wedges on this pattern seems to be valuable for transfemoral amputees' (TF) gait optimization. A drawback for using data from quantitative gait analysis is the enormous amount of information provided for extracting parameters from data curves. Principal Component Analysis (PCA) is a powerful method used to reduce redundant information allowing the comparison of the complete waveform. The purposes of this study were a) to assess the influence of the wedges on GRF and center of pressure parameters during able-bodied gait; b) to assess the influence of the wedges on net joint moment and range of motion of able-bodied during gait; c) to compare GRF and plantar pressure parameters between TF amputees and able-bodied subjects during self-selected level-walking d) to develop a biomechanical model for optimization of TF amputees' gait by prescribing wedges and test it experimentally. The ground reaction forces - GRF (force plate – 1000 Hz), COP displacement (pressure plate – 300 Hz) and 3D kinematic (videogrammetry - 50 Hz) were recorded during self-selected speed level-walking of 20 able-bodied and 15 TF amputees' participants. The influence of six wedges was verified on the able-bodied gait pattern and, based on this results, a program was developed in MATLAB® to verify whether or not any wedge would influence positively (shift the TF amputees' parameters towards those of the able-bodied) the TF amputees' gait. After, the biomechanical model developed, it was applied on three of the TF amputees. The results were: a) The wedges influenced the GRF and center of pressure parameters of the able-bodied participants during gait; b) The wedges influenced the net joint moment and range of motion of the able-bodied participants during gait; c) the PCA approach was able to discriminate differences on gait pattern between TF amputees and able-bodies subjects during gait and; d) The developed biomechanical model was able to prescribe successful wedges for TF amputees. A new methodological approach based on PCA analysis of GRF and center of pressure parameters was showed to be successful, by wedges prescription, for shifting the sound limb gait parameters of TF amputees to those found on able-bodied subjects. The application of PCA may help clinicians to decide the best aid device for each patient, and consequently to improve the quality of life of TF amputee patients.

List of Abbreviations

1L	Lateral wedge, 1 cm (see appendix II)
1M	Medial wedge, 1.1 cm (see appendix II)
1P	Posterior wedge, 0.9 cm (see appendix II)
2L	Lateral wedge, 2 cm (see appendix II)
2M	Medial wedge, 2.2 cm (see appendix II)
2P	Posterior wedge, 1.8 cm (see appendix II)
Ankle _{ROM}	Ankle range of motion
Ankle _{SAG}	Joint moment in sagittal plane in the ankle
BW	Body weight
CG	Control group
Cg _{pyl}	Centre of mass for the rigid body representing the pylon
Cg _{res}	Centre of mass for the rigid body representing the socket attached to the residual limb
CI	Confidence interval
CON	Control condition
COP _x	Center of pressure medio lateral displacement
COP _y	Center of pressure antero posterior displacement
Deg	Degrees
D _{cut}	Cutting plane in the tibia
D _{ext}	External diameter of the pylon
D _{int}	Internal diameter of the pylon
DoF	Degrees of freedom
EMG	Electromiography
GRF	Ground Reaction Force
GRF _{ap}	Antero-posterior component of the GRF
GRF _{ml}	Medio-lateral component of the GRF
GRF _{vt}	Vertical component of the GRF

Hip_{ROM}	Range of motion of the hip joint
$Knee_{FRT}$	Joint moment in the frontal plane in the knee
$Knee_{ROM}$	Range of motion of the knee joint
$Knee_{SAG}$	Joint moment in the sagittal plane in the knee
l_{cut}	Distance from the knee to D_{cut}
l_{Tib}	Length of the residual limb
m_{pyl}	Mass of the prosthetic pylon
m_{res}	Mass of the residual limb
m_{Tib}	Mass of the tibia
NJM	Net joint moment
Nm	Newton meter
PC1, PC2, PCn	First, second, n_{th} principal component
PCA	Principal component analysis
PeakAnkle _{SAG}	Peak Joint moment in sagittal plane in the ankle
PeakKnee _{FRT}	Peak Joint moment in frontal plane in the knee
PeakKnee _{SAG}	Peak Joint moment in sagittal plane in the knee
QOL SF36	Quality of life SF-36 questionnaire assessment
ROM	Range of Motion
R_{res}	Radius of the initial shank
SL	Sound limb
SP	Stance phase
T^2	Mahalanobis distance
TF	Transfemural amputees

1 GENERAL INTRODUCTION

The determination of the forces and joint moments in the lower limbs have been the subject of many studies. Since the beginning of the last decade, the inverse dynamics has been one of the most widely used techniques for this purpose, considering that it is noninvasive and allows the analysis of different movements such as walking, running and sports movements. In principle this technique allows the determination of loads on the muscles, tendons, bones and ligaments. Most studies present models for determining the forces and moments only in the sagittal plane (Beynnon et al., 1996; Herzog & Read, 1993; Kim & Pandy, 1993; Lu & O'Connor, 1996; Tumer & Engin, 1993), without being properly assessed or estimated the errors associated with this simplification (Glitsch & Baumann, 1997). More recently, some studies analyzed three-dimensional movements (Engin & Tumer, 1993; Glitsch & Baumann, 1997; Hefzy & Yang, 1993; Loch et al., 1992; Riemer & Hsiao-Weckslar, 2008; Schache et al., 2007).

Insoles and wedges are devices commonly prescribed for compensation of gait deviations. Some studies showed that the use of wedges could alter the gait pattern and decrease the pain caused by muscular and bone diseases (Chiu & Shiang, 2007; Erhart et al., 2008; Schmalz et al., 2006). According to Kerrigan et al. (2002), the use of insoles in the individuals shoes could influence decisively the quality of their gait.

Orthopedic Prosthetic devices are appliances and/or equipment that will replace human body parts, for example, mechanical legs, mechanical arms, etc... The prosthetic devices for limbs that are amputated above the knee are divided into two major categories: conventional prostheses and modular prostheses. Conventional prostheses do not have great mobility in the joints, while the modular prostheses have several mechanical devices to simulate the joint. The prescription of the most adequate device may vary according to the level of activity and mobility of the individual.

Transfemoral (TF) amputees gait is for definition asymmetrical (Nolan et al., 2003; Rabuffetti et al., 2005). The use of a prosthetic device leads to a different pattern of gait, at least for the fact that both limbs are different. This asymmetry causes an overload of the intact limb (Zmitrewicz et al., 2006), where high indexes of injuries like knee and hip osteoarthritis (Melzer et al., 2001; Pieter et al., 2009), scoliosis (Burke et al., 1978) and lumbar pain (Kulkarni et al., 1998; Skinner & Effeney, 1985) were reported. Also, a low bone density in the hip of the amputated leg is reported (Kulkarni et al., 1998; Sherk et al.,

2008). The role of therapists is to help TF amputees to minimize this asymmetrical condition and to turn the become gait pattern as close as possible to that of able bodied subjects. Despite the initial depression that affects virtually all new amputees, this state of depression is quickly replaced by a desire to return to an active life and, if possible, to their previous activities (Christensen et al., 1995).

From the studies that analyses gait patterns, in TF amputees gait, or in the influence of wedges in gait, the results are commonly presented as the analysis of parameters extracted from the kinematic and kinetic curves, generating a huge amount of data (Chui & Lusardi, 2010), that sometimes is difficult to interpret. This approach relies on the definition of discrete parameters that is subjective, and it becomes difficult to extract the same values of all temporal waves, especially in the presence of pathologies (Landry et al., 2007). A significant barrier to the clinical use of gait information is the successful reduction and analysis of data (Chau, 2001).

Deluzio et al. (1997) introduced a novel application of Principal Component Analysis (PCA) to the analysis of kinematic and kinetic data, since then, PCA has become a common method of reducing dimensionality and analyzing waveforms in gait analysis (Muniz & Nadal, 2009). The study of the parameters obtained by inverse dynamics allows the evaluation of the mechanism that is associated with a particular movement and the muscular effort involved. From the analysis of the influence of the different parameters that constitute an insole, such as material and wedges, it is possible to apply these results to build more adequate insoles for individuals with different gait dysfunctions (Kerrigan et al., 2002). Thus, the use of inverse dynamics can help in the construction of insoles that provide an increasingly better quality of life in this population.

Therefore, the general purposes of this thesis are:

- To develop a 3D model of indirect determination of joint moments of the ankle and knee, using the technique of inverse dynamics from the measured reaction force with the ground and measuring the accelerations of the segments involved, to apply in elderly subjects and TF amputees.

- To analyze the patterns of joint moments of the ankle and knee, ground reaction forces and center of pressure displacement in elderly subjects, in the light of a classical approach and of PCA;
- To analyze the influence of six kinds of wedges in joint moments, GRF components and COP displacement in elderly gait;
- To analyze the patterns of joint moments of the ankle and knee, ground reaction forces and center of pressure displacement in TF amputees;
- To present a model for the prescription of wedges for TF amputees that takes into account the patterns of force and pressure in the elderly group, using PCA.

For the accomplishment of these purposes, the results of this thesis were presented as a main part (chapters 2 to 6) and Appendixes (from I to V). A literature review about 2D and 3D inverse dynamics models, biomechanical evaluation of TF amputees, the influence of wedges on gait, and PCA applied to gait waveforms was carried out (Chapter 2). The developed inverse dynamics 3D model for the analysis of net joint moments is presented in Appendix I. Then, the application of the 3D model into gait data of able bodied subjects, the evaluation of the influence of wedges on the net joint moments and angles, and the comparison of the PCA approach to the classical one were performed (Chapter 3).

As alternative of the complex dynamic inverse approach, we decided also to verify whether or not some easier-to-acquire parameters such as GRF and COP would be able to discriminate gait alterations either between groups or as influence of wedges (Appendix II & Chapter 4). As these parameters (GRF and COP) were able to discriminate gait between patterns and one of our aims was to present a tool as simple as possible for clinicians and physical therapists to prescribe wedges for TF amputees, we have adopted these as the main parameters of this study.

The detailed rationale of the developed biomechanical model for wedge prescription for the sound limb of TF amputees, as well as the experimental test of the model were presented at Chapter 5 and Appendix III. Appendix V shows some extra results obtained in the wedges prescription for TF that are not included in Chapter 5. As an

example of other application of the 3D model developed, Chapter 6 shows the results obtained by an ANYBODY™ simulation applied to transtibial subjects, walking with the same protocol and specifications designed in the model presented here. Appendix IV shows the proceedings of the parts of this work presented in conferences.

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2 LITERATURE REVIEW

The literature review is divided into four parts: the first addresses the works presented in the literature on inverse dynamics, its evolution and different models used; the second part presents works related to the evaluation of amputees who use prostheses, and the use of inverse dynamics in these assessments; the third part analyses the influence of the wedges in gait; and the last deals with Principal Component Analysis applied to gait waveforms.

2.1 *Inverse dynamics: 2D and 3D models*

The inverse dynamics is a technique that allows, from Newton's second law, to compute the resulting forces and moments in joints using kinematic and kinetic data (Loss et al., 2002). Through the measurement of accelerations, masses and the external forces, we compute the values of internal forces and moments at the joints.

Equation 2. 1
$$\vec{F} = m \vec{a}$$

Equation 2. 2
$$\vec{M} = I \vec{\alpha}$$

Most of the early studies present models for determining the forces and moments only in the sagittal plane (Beynnon et al., 1996; Herzog & Read, 1993; Kim & Pandy, 1993; Lu & O'Connor, 1996), without being properly assessed or estimated errors associated with this simplification (Glitsch & Baumann, 1997). More recent studies use three-dimensional analysis (Dumas et al., 2009; Engin & Tumer, 1993; Glitsch & Baumann, 1997; Hefzy & Yang, 1993; Loch et al., 1992; Riemer & Hsiao-Wecksler, 2008; Schache et al., 2007) allowing the evaluation of more complex movements (Robert et al., 2007).

Alkajer et al. (2001) made a comparison between joint moments obtained with a 2D and a 3D model and also what influence the position of the axis of rotation would have on the determination of the joint moment. The general shapes of the 2D and 3D joint moment patterns about the ankle, knee and the hip were very similar, but the statistical analysis of differences in the joint moments between the 2D and 3D model showed significant differences with respect to the magnitude of the moments. A dorsiflexor moment was seen in the 3D analysis whereas the 2D calculation showed almost complete plantar flexor

dominance about the ankle joint. The knee joint flexor moment in the middle of the stance phase was larger in 2D than in 3D. For the hip joint moment, the flexor moment in the second half of the stance phase was larger in 3D

Different approaches have been presented, taking into account several parameters. Beynnon et al. (1996) for example, suggested a two-dimensional model with three degrees of freedom. The tibiofemoral joint is represented by two rigid bodies (femur and tibia), assuming no deformation of the articular cartilage. The cruciate ligaments are also considered, being the four elements made of non-linear elastic ligament: anteromedial part of the anterior cruciate ligament, posterolateral part of the anterior cruciate ligament, anterior component of the posterior cruciate ligament, and a component back of the posterior cruciate ligament. Herzog et al. (1993), suggest a model that considers the tibia, femur, cruciate ligaments, collateral ligaments, quadriceps and hamstrings. Lu et al., (1996), complements the model of Herzog et al. (1993) by adding the patella, the patellar ligament, the semimembranosus and semitendinosus hamstrings, as well as the gastrocnemius. Tümer et al. (1993) present a two-dimensional model also, but composed of three rigid bodies (femur, tibia and patella), including the patellofemoral and tibiofemoral joint. In addition to the anterior and posterior cruciate ligaments, are also considered the patellar ligament, the medial and lateral collateral ligaments. It also takes into account the action of three muscle groups: quadriceps, hamstrings and triceps surae. Also in a two-dimensional analysis, Kim et al. (1993) suggested a broader approach, shaping not only the knee, but the lower limb as a whole, considering four segments, foot, leg, thigh and sacrum, which are operated by eight muscles: tibialis anterior, soleus, gastrocnemius, vastus lateralis, rectus femoris, hamstrings and gluteus maximus, and four ligaments: the anterior and posterior cruciate ligaments, and medial and lateral collateral ligaments. Loss (2001) proposed a model composed of three rigid segments representing the thigh, leg and foot. For the determination of joint forces and moments it makes use of two equations of translation and a rotation. These 2D models, restrict themselves to the analysis of movements that occur predominantly in a single plane, eliminating the possibility of analysis with rotations or movements that occur in more than one plane.

Considering 3D models, Hefzy et al. (1993) developed a three-dimensional model for the patellofemoral joint that determines how the motion and joint contact forces vary with knee flexion. The model uses six equilibrium equations and 11 constraints, a total analysis of 17 nonlinear equations in 17 variables. The patella is idealized as a rigid body where three forces operate, originating at: patellar tendon, tendon and suprapatellar contact force with the femur. Loch et al. (1992) model the knee in a three-dimensional approach, considering only the femur and tibia as rigid structures, interconnected by deformable structures, including the anterior cruciate ligament, a cartilage surface, and a "connecting factor" which includes the effects of the meniscus, joint capsule, soft tissue and all ligaments except the anterior cruciate ligament, as previously considered. The model is designed for small displacements of the joint, assuming a linear behavior of all involved structures, and aims principally to predict situations of reconstruction of the anterior cruciate ligament. Glitsch and Baumann (1997) propose an extremely sophisticated model, which includes a three-dimensional anthropometric four rigid segments, pelvis, thigh, leg and foot, connected by the hip, knee and ankle joints, 47 muscles which act as defined from the origin and insertion points and its cross-sectional area.

More recent studies found in the literature are concerned about the optimization and error reduction of the models proposed, comparing with direct measurements (Dumas et al., 2009), different models to determine the planes of movement (Schache et al., 2007) and reducing the error associated to skin markers movement (Riemer & Hsiao-Wecksler, 2008). The models presented in the literature have limitations and considerations, but the mainly purpose is to simulate the human skills the best possible, to understand the mechanisms associated with the motion and, after that, to infer the differences in pathological patterns or different athletes techniques. Specific situations can be studied, as suggested by Engin and Tumer (1993), and Hoshino and Wallace (1987), specifically regarding the absorption of impact by the joint, or in cases of differential stress, which occur during the years of rising (Schuldt K, 1983).

2.2 Biomechanical evaluation performed in individuals amputees using prosthetic leg

In the literature are found many papers which aim to assess the prosthetic pattern of movement. Some works are related to the comparison between the sound and prosthetic limbs (Nolan et al., 2003), others compare amputees with able bodied (Nolan et al., 2003; D. A. Winter & S. E. Sienko, 1988) while others compare different types of prostheses used by the same individuals (Schmalz et al., 2002; David A. Winter & Susan E. Sienko, 1988). The parameters of interest vary between the ground reaction force, kinematic variables of the lower limb joints, plantar pressure as well as muscle forces and moments. Specifically with respect to the manufacture of prostheses, Schmalz et al. (2002) developed a study to define more clearly the influence of prosthetic alignment and different prosthetic components in energy consumption and in moments of force in the sagittal plane of transtibial amputees (TT) and transfemoral (TF) amputees during gait. Five different models of the prosthetic feet have been used, namely one with 5 different alignments of the foot, and two different knees, one conventional and one hydraulic. The results showed that moving the foot anteriorly increases the tendency to have a knee bending moment, while moving the foot posteriorly tends to increase the knee extensor moment. When varying the angular position of the foot, the knee is affected (the foot in plantar flexion increases the peak extensor moment). With regard to oxygen (O_2) consumption, this was not affected by the antero-posterior position of the foot, but the foot angle significantly increases O_2 consumption, regardless of whether plantar flexion or dorsiflexion. This is accentuated with increasing speed. Heart rate was not affected. The foot model 1S71 (Otto BockTM), presented the lowest plantar flexion moment. IC40 (Otto BockTM) and FLEX WALK II (Flex footTM) produced some significantly greater dorsiflexion moments during the toe off phase due to their elastic characteristics. There were no significant differences in oxygen consumption with different feet. With respect to the position of the knee, there was an increase in the duration of the hip extensor moment in the first half of the support phase with the knee 2cm above the baseline. This implies an increase in hip extensor activity to resist the bending moment in the knee caused by this type of alignment, preventing the prosthesis to break. At higher speeds, the alignment of the knee 2cm back of the vertical baseline shows a significant increase in O_2 consumption. By aligning 2cm above baseline, the O_2 consumption is higher at all

speeds. With respect to the consumption of O₂, the model C-LEG (Otto Bock™) decreases it at low and medium speeds.

From the studies evaluating joint forces and moments in the lower limbs of transtibial and transfemoral individuals, we highlight the work of Winter & Sienko (1988) who designed a study to show how a group of transtibial amputees alter the motor patterns of the amputated limb, resulting in a considerable degree of motor asymmetry. Five individuals were analyzed with data collected over three years in the laboratory. The parameters analyzed were the moments and muscle power of ankle, knee and hip, the timing and strength of support (the sum of the three joints) and EMG of the gluteus maximus, biceps femoris, semitendinosus, rectus femoris and vastus lateralis. The results showed that, regardless of the type of prostheses, amputees show hyperactivity of the hip extensors during the early and mid-stance phase, resulting in a power generation above normal during the concentric phase. This compensation is due, in part, to the lack of energy generated by the plantar flexors at the time of loss of contact with the ground.

In another work, Nolan (2003) conducted a study to investigate the effects of increased walking speed in unilateral amputees, particularly in peak vertical force, moments of force, support time and swing time. Four transtibial amputees, 4 transfemoral and 6 normal subjects were asked to walk at speeds of 0.5, 0.9, 1.2 m/s and at maximum speed. A summary of the results obtained in the study is presented in Table 2.1

Table 2.1: summary of results obtained in the study by Nolan (2003). N: Control, P: prosthetic limb; S: sound limb; TT: transtibial; TF: transfemoral; ↑: increase; ↓: decrease

Variable	Among groups	Speed increase	Asymetry
Vertical Force	N > P		
Peak	N < TT/TF	↑ with ↑	↑ with ↑ for TT and TF
Impulse	PTF < PTT < N N < TT < TF	↓ with ↑	
Stance time	N = STT < STF N = PTT < PTF N = TT < TF	↓ with ↑	↓ with ↑ in TF
Swing time	N = PTT < PTF N < PTT < PTF	↓ with ↑	↑ with ↑ for TT and TF

One possible explanation for the lower ground reaction force in PTT and PTF is that amputees may be trying to protect the residual limb, loading it less, and thus increasing the load on the intact leg. Another explanation is that the center of gravity

is closer to the amputated limb, thus more weight is placed on the intact leg during walking, leading to an overload of this limb.

Analyzing the works presented in the literature, it is observed that the existing prostheses still induce movement patterns far from a normal pattern, with high asymmetry between members – able bodied subjects have less than 10% of asymmetry between limbs during walking, while amputees have more than 23% (Melzer et al., 2001), an overload of the sound limb (71% of the amputees reported sound limb and lower back pain (Melzer et al., 2001)) and a higher metabolic cost (oxygen consumption increases 25% in transtibial and 55.65% in transfemoral compared to able bodied (Schmalz et al., 2002). This shows the importance of studies in this area for the manufacture and creation of new prosthetic models inducing a more symmetrical pattern of movement.

2.3 The use of wedges and their influence in gait

The use of insoles as an auxiliary in small gait deviations is a common practice. From the analysis of the influence of the different parameters that constitute an insole, it is possible to apply these results to build insoles adequate to individuals with different diseases (Kerrigan et al., 2002). The studies commonly analyze the influence of the devices in a healthy population, and apply the results to a group with special needs. According to MacLean et al. (2006), this kind of intervention in a normal population of healthy individuals is not enough to show differences, because the human capacity of adaptation is high. Nevertheless, devices like wedges are used in patients with orthopedic diseases and some studies showed that these devices may relief pain in the joints and improve gait patterns (Kakihana et al., 2005; Russell & Hamill, 2011; Schmalz et al., 2006).

Most of the studies are concerned about the influence of lateral wedges in knee varus moment, mostly because of the application of these wedges in knee osteoarthritis patients. McLean et al. (2006) tested 15 female runners wearing custom made foot orthoses, with a 5° lateral wedge. No differences were found in knee varus moment when wearing the orthosis. Differences were found in ankle inversion and knee extensor moments, which decreased with the use of orthoses. Kakihana et al. (2005) tested a

control and an osteoarthritis group wearing a 6° lateral wedge and also found no differences between wedges conditions in knee varus/valgus peak moment. In contrast, Erhardt et al. (2008) analyzed 15 healthy subjects walking with laterally wedged shoes in three different speeds and found differences in peak valgus moment in all the conditions. Kerrigan et al. (2002) analyze osteoarthritis patients walking with 5° and 10° laterally wedged shoes and found that compared with the no insole condition, the 5° wedge reduced the peak knee varus moment values by about 6% and the 10° wedge reduced the peaks by about 8%. In another study evaluating knee moments, Schmalz et al. (2006) analyzed lateral and medial 10° wedges combined with Ankle-Foot -Orthosis in 10 healthy subjects and found that the lateral wedges have no influence in the knee frontal moment, while the medial wedges increase the knee varus moment, both with and without combination with Ankle-Foot-Orthosis.

The posterior wedges are not as widely explored as the lateral conditions, but some studies evaluate the cushioning properties of these devices. Chiu & Shiang (2007) evaluated the cushioning properties of a 2mm posterior insole in healthy subjects and conclude that the use of this insole promotes a higher shock absorption compared to shoes with limited cushioning properties.

Another variable of interest when analyzing the influence of the wedges in gait is the center of pressure (COP) displacement. Balmaseda et al. (1988) found that the COPx trajectory is laterally deviated using an Ankle-Foot-Orthosis. Also, Guldemand et al. (2006) found a laterally deviated COP displacement using custom-made foot orthosis. These findings are opposed to the work of Chevalier et al. (2010) that found no differences in COPx displacement comparing shod and barefoot walking, probably because shod is not a condition that implies different foot angles.

The influence of the use of wedges in gait patterns is not clear. The results obtained are sometimes controversial, and the authors themselves reconized that the healthy subjects have a high capacity of adaptation and can adapt their gait to the new situation (MacLean et al., 2006). In this perspective, healthy individual data should be used carefully as a control behavior for experimental groups. However, some studies found differences in gait patterns even in healthy subjects testing different situations, supporting

the idea of using a healthy group to evaluate the influence of devices or shoes and extrapolate those results to the experimental group.

2.4 Principal Component Analysis in gait

The Principal Component Analysis approach is a technique of multivariate exploratory analysis that transforms a group of correlated variables into a smaller amount of independent variables, which are linear combinations of the original variables, named principal components (Moroco, 2003). In this way, PCA is generally viewed as a method of data reduction, and one of the principal advantages of PCA is to allow synthetizing the information of a number of correlated and sometimes redundant variables into one or more independent linear combinations, that represent the majority of the information contained in the original variables (Deluzio et al., 1997; Sadeghi et al., 2002).

The study of the gait pattern is usually based on the extraction of discrete parameters obtained from the waveforms, like maximums, minimums, ranges and impulses. The main problem with this technique is the generation of a huge amount of discrete parameters, that are subjective in the selection and sometimes difficult to interpret (Sadeghi et al., 2002). In the last two decades, the interpretation of gait data was improved by different methods of multivariate analysis (Deluzio et al., 1997; Muniz & Nadal, 2009; Sadeghi et al., 2002).

Knapp and Comrey (1973) have applied PCA to analyze waveforms in general, as a way to reduce the amount of information. The application of PCA to gait waveforms is a quite recent approach. The main applications of this technique are the classification of the gait pattern of a patient in comparison to normal gait (Deluzio & Astephen, 2007; Deluzio et al., 1999; Muniz & Nadal, 2009; Muniz et al., 2010), and the determination of the portions of the waveform that discriminate a certain group (Lee et al., 2009; Sadeghi et al., 2002). For their general purpose method, Knapp and Comrey (1973) have stated that the ± 0.71 value for the factor loading criteria is considered as a minimum to determine whether the portion of the waveform is relevant.

Deluzio et al. (1997, 1999, 2007) developed a series of studies demonstrating how PCA could be used to analyze gait waveforms (1997) and applied it to two groups to

give examples of its applicability (1999, 2007). The first study (1997) presented detailed information about the methodology used, and showed how PCA could be used to analyze the gait pattern of a patient and compare it to a group of healthy subjects. Deluzio et al. (1999) analyzed a group of 13 patients with knee arthroplasty and used PCA to classify their gait pattern before and after the surgery.

Muniz et al. (2009, 2010), also used PCA to discriminate normal and abnormal gait patterns in a group with lower limb fractures, comparing them before and after a rehabilitation program. They have used the elliptical area that determines the distribution of the normal subjects to classify the patients gait before and after rehabilitation. The results showed that PCA is a powerful method to identify differences in gait pattern and to characterize the patients correctly.

Jones et al. (2008) evaluated the GRF components and lower limb kinematics of the gait patterns from osteoarthritic patients using three methods to classify the subjects: PCA, Dempster-Shafer (DS) based method and an artificial neural network (ANN). The results showed that the results obtained from the three methods are complementary and therefore the best approach is to perform an hybrid analysis of the data.

From the studies evaluating normal gait, the work of Sadeghi et al. (2002) showed how PCA can be used to detect the main functional structure of actions taken by knee muscles in the sagittal plane during gait. The results showed that by using PCA it was possible to show the contribution and importance of a muscle group in independent tasks, namely balance control, foot clearance and shock absorption.

Multivariate techniques of data reduction like PCA are of great importance in the classification and separation of normal and pathological patterns (Chester et al., 2007). According to Hernández-Caraballo *et al.* (2005), PCA is a method that should be used in the analysis of data with the purpose of the reduction of the dimensionality, to remove the redundant information. Nevertheless, the interpretation of the data is also necessary, based on the knowledge of experienced technicians, to be useful in rehabilitation processes and correct application of the results obtained.

2.5 References

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3 INFLUENCE OF WEDGES ON LOWER LIMBS' NET JOINT MOMENT AND RANGE OF MOTION DURING HEALTHY ELDERLY GAIT USING PRINCIPAL COMPONENT ANALYSIS

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ABSTRACT:

The elderly are susceptible to many disorders that alter the gait pattern and could lead to falls and reduction of mobility. One commonly applied therapeutical approach to alter the gait patterns is the insertion of insoles. Principal Component Analysis (PCA) is a powerful method used to reduce redundant information while allowing the comparison of the complete waveform. The purpose of this study was to verify the influence of wedges on lower limbs' net joint moment (NJM) and range of motion (ROM) during the gait of healthy elderly participants using PCA. In addition, discrete values of lower limbs' peak NJM and ROM were also evaluated, to represent the classical evaluation procedure. Twenty subjects walked with no wedges (control condition) and wearing six different wedges. The variables analyzed were PCs from NJM and ROM from the sagittal plane in the ankle and knee and NJM from the frontal plane in the knee. The discrete variables were peak NJM and ROM from the sagittal plane in the knee and ankle. The results showed the influence of the wedges to be clearer by analyzing through PCA methods than to use discrete parameters of gait curves, where the differences between conditions could be hidden.

3.1 *Introduction*

The elderly population is rising in number due to a longer life expectation (Johnson, 2011). The elderly are more susceptible to many disorders (Urs Granacher, 2012) that alter the gait pattern and could lead to falls (R. N. Kirkwood et al., 2011; Marcus, 2012) and reduction of mobility (Guccione et al., 1990), which affect their health and independence. In this perspective, the gait analysis of the elderly healthy population seems to be important in order to constitute reference values for understanding the abnormal gait pattern and to assess the influence caused by different therapeutical approaches on it.

One therapeutical approach applied to alter the gait patterns is the insertion of wedged insoles inside the shoe. Their use has been described as a powerful tool for the compensation of small gait deviations. According to Kerrigan et al. (2002), the use of wedged insoles inside the shoes could influence decisively the quality of the subjects' gait. Analyzing the influence of the different parameters that constitute an wedged insole such as position, height, material and density of the wedge, it is possible to build customized wedged insoles adequate to individuals with different gait deviations (Kerrigan et al., 2002). Nevertheless, there is no consensus about the influence of different wedge features on the normal gait pattern (Van Gheluwe & Dananberg, 2004): some authors found differences in gait waveforms (Chiu & Shiang, 2007; Erhart et al., 2008; Schmalz et al., 2006) while others concluded that there is no influence in the gait of healthy subjects (Kakihana et al., 2005; MacLean et al., 2006).

Most of the recent studies evaluating the elderly gait found in the literature usually present the results of parameters for kinematics (Chui & Lusardi, 2010; Renata N. Kirkwood et al., 2011), for pressure distribution (R. N. Kirkwood et al., 2011) and some combine kinematic and kinetic parameters (Russell & Hamill, 2011; Trombini-Souza et al., 2011). These results are commonly presented as parameters extracted from discrete points in the kinematic and kinetic curves, generating a huge amount of data, that sometimes is difficult to interpret (Sadeghi et al., 2002). This classical approach relies on

the definition of discrete parameters that are subjective, and it becomes difficult to extract analogous information of all temporal waves, especially in the presence of pathologies (Landry et al., 2007). A significant barrier to the clinical use of gait information is the absence of a successful reduction of the data (Chau, 2001). In the last two decades the analysis and interpretation of gait data was improved by the use of different methods of multivariate analysis (Deluzio & Astephen, 2007; Jones et al., 2008; Muniz & Nadal, 2009; Olney et al., 1998; Sadeghi et al., 2002). Principal Component Analysis (PCA) is one of the methods used to reduce redundant information allowing the comparison of the complete waveform (Sadeghi et al., 2002).

Sadeghi et al. (2002) compared the results from a classical approach and from PCA analysis concerning the range of motion of the knee and ankle joints, and showed that the second approach is more sensitive to differences between loaded and unloaded walking. However, to the best of our knowledge, none of the studies available in the literature analyzed a group of healthy subjects wearing different insole conditions in the light of the PCA approach.

The purpose of this study was to verify the influence of wedges on lower limbs' net joint moment and range of motion (ROM) during the gait of healthy elderly participants using the PCA method and the classical approach. In addition, both methods (PCA and classical) were analyzed in order to verify their sensitivity to identify differences between control and experimental conditions. We hypothesized that wedges can measurably influence the gait of healthy elderly people and that the PCA method is more sensitive to identify differences between conditions.

3.2 *Methods*

3.2.1 *Participants*

As inclusion criteria, the participants should be more than 50 years old and practice oriented physical activity at least twice a week. As exclusion criteria the participants could not have any kind of limitation or pain during gait, and could not have any previous history of injuries or diseases that potentially alter their gait pattern. The sample group was composed of 20 physically active participants, 14 females and 6 male (mean age of $67 \pm$

8.56 years old). Subjects of this group had a good score on the Physical Function domain of the QOL SF36 questionnaire (82.33 ± 18.01 marks), meaning that they felt few limitations to perform daily living activities involving physical function.

All participants walked over an 8m walkway, wearing their own shoes at a self selected speed. After a short adaptation each subject walked three times wearing the shoes only, named control condition (CON) and, after that, with each one of the six wedges inside the shoe, in random order.

The wedges were placed in three different plantar foot regions and they were made of polyurethane cushion in six different shapes (Fig.1): two lateral, placed under the 5th metatarsal bone with 1 cm (1L) and 2 cm (2L), two medial placed under the foot medial longitudinal arch with 1.1 cm (1M) and 2.2 cm (2M), and two posterior with 0.9 and 1.9 cm high, placed under the calcaneus bone. Therefore seven conditions were analyzed: CON, 1L, 2L, 1M, 2M, 1P and 2P.

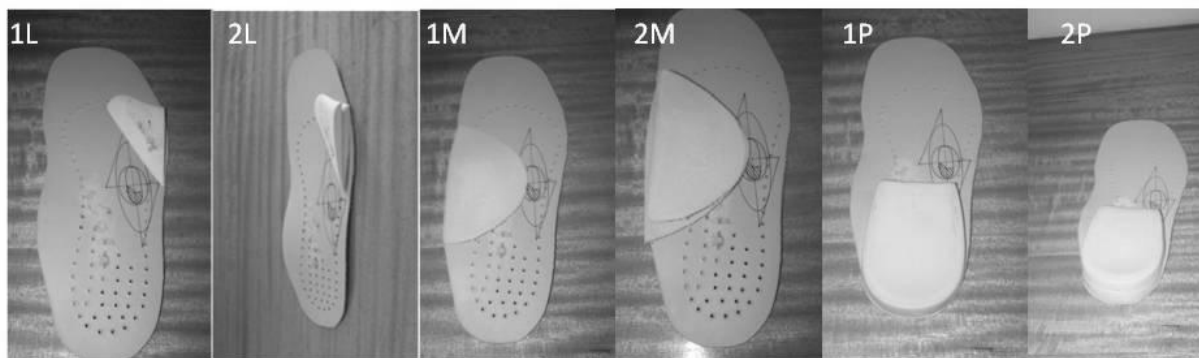


Figure 3.1: The six wedge conditions analysed: two lateral (1L, 2L), two medial (1M, 2M) and two posterior (1P, 2P).

3.2.2 Gait analysis and signal processing

A Simi Motion System® (SIMI Reality Motion Systems, Unterschleissheim, Germany) with four cameras at 50 Hz, placed at the corners of the walkway was used to collect the kinematic data, a piezoelectric force plate (Kistler Instruments AG, Winterthur, Switzerland) at 1000 Hz placed at the middle of the walkway was used to record the ground reaction forces (GRF) and a FootScan pressure plate (RsScan, Olen, Belgium) at

300Hz positioned over the force plate was used to record the centre of pressure (COP) trajectory. The systems were synchronized using a separate unit that, using a manual trigger, was responsible for starting data collection in both devices at the same time.

The sagittal plane joint net moment of Ankle ($Ankle_{SAG}$) and Knee ($Knee_{SAG}$) and the knee moment in frontal plane ($Knee_{FRT}$) were calculated using the inverse dynamics model proposed by Vaughn (1992) with the kinematic data processed after 3D calculation through the solidification procedure described by Cheze et al. (1995). ROM of Ankle ($Ankle_{ROM}$), Knee ($Knee_{ROM}$) and Hip (Hip_{ROM}) in the sagittal plane were obtained from the kinematic analysis. The values of peak net moment of the three waveforms ($PeakAnkle_{SAG}$, $PeakKnee_{SAG}$, $PeakKnee_{FRT}$) and range of motion ($Ankle_{ROM}$, $Knee_{ROM}$, Hip_{ROM}) were retained to represent the classical approach of extracting specific values that is generally used in kinematics and joint moment data. The Principal Component score values were computed for the six waveforms retained for analysis ($Ankle_{SAG}$, $Knee_{SAG}$, $Knee_{FRT}$, $Ankle_{ROM}$, $Knee_{ROM}$ and Hip_{ROM}). Gait speed was obtained from the first derivative of the iliac crest marker displacement.

For kinematic data digitization and reconstruction of images, the *Dvideo v.5.0* (Unicamp, Campinas, Brazil) videogrammetry system was used (2006). The kinetic data was processed in Simi Motion System® (SIMI Reality Motion Systems, Unterschleissheim, Germany) software and the pressure data by FootScan 7 gait 2nd generation (RsScan, Olen, Belgium) software. The treatment of the all data was performed using Matlab® 7.0 (MathWorks, Massachusetts, USA).

To reduce the effect of random noise, the data was filtered using a low pass Butterworth filter. The cutoff frequency was 4Hz in 4th order for the kinetic data and 10Hz in 4th order for the kinematic data. The signals were interpolated and resampled in order to obtain 100 points (variables), providing one variable for each percent of stance phase (SP). In the case of variables that also analyze the swing phase (ROM from ankle, knee and hip), 140 points (variables) were obtained.

3.2.3 Principal Component Analysis

PCA was performed in joint net moment and ROM waveforms according to Deluzio et al. (1997). Briefly, the aim of PCA is to summarize the information contained in 100% of SP, by considering each 1% as one variable (100 variables), and to represent the full

waveform by a smaller number of components that explain most of the variance through linear combinations from those variables (Jolliffe, 2004). Principal Components (PCs) are arranged in decreasing order in such a way that the first PC accounts for most of the variability in the data, and each succeeding component accounts for much of the remaining variability as possible (Daffertshofer et al., 2004). In this study, the number of PCs analyzed was 3.

PCs are an orthogonal transformation which converts p variables $X=x_1, x_2, x_3, \dots, x_p$ (in this case from 0 to 100% of the stance phase) into p new uncorrelated PCs $Z=z_1, z_2, z_3, \dots, z_n$, which are defined by the equation $Z=U^tX$, where U are the eigenvectors of the covariance matrix of X . U_n is calculated by the equation $SU_n=\lambda U_n$ where λ are the p eigenvalues ranked in decreasing order and S is the covariance matrix of X . The load vectors are defined by the equation $Z_n \lambda^{1/2}$ and these vectors are normalized to have Euclidian norm equal to 1. A threshold of ± 0.71 as suggested by Knapp & Comrey (1973) was used to consider a portion of a load vector from one variable as relevant, and then attribute a meaning for this PC. The comparisons between conditions were made only for the PC load vectors that reach the threshold at any instant of the gait cycle.

The PC model was developed based on the gait pattern of the subjects walking in CON condition and then this model was applied to the subjects walking with the wedges conditions. The PC score values (internal product from PC1, PC2 and PC3 to each waveform) for each subject in each condition were retained for analysis, with 3 score values (PC1, PC2 and PC3 scores) for the six waveforms (Ankle_{SAG}, Knee_{SAG}, Knee_{FRT}, Ankle_{ROM}, Knee_{ROM} and Hip_{ROM}), totalizing 18 PC score values to analyze per subject with each wedge.

3.2.4 Statistical Procedures

The normality of the distribution of the PC score values, peak joint net moments (Ankle_{SAG}, Knee_{SAG}, Knee_{VAL}, PeakAnkle_{SAG}, PeakKnee_{SAG}, PeakKnee_{FRT}) and ROM (Ankle_{ROM}, Knee_{ROM} and Hip_{ROM}) was accessed by the Kolmogorov-Smirnov test and the variances homogeneity using *Levene's* test. *One Way* ANOVA and *post hoc* LSD was used for both analysis: in the classical approach, to compare CON and the wedges conditions (1L, 2L, 1M, 2M, 1P and 2P) for the six discrete variables retained (PeakAnkle_{SAG}, PeakKnee_{SAG}, PeakKnee_{FRT}, Ankle_{ROM}, Knee_{ROM}, Hip_{ROM}) and in the PCA

approach, to compare PC score values from CON and the wedges conditions for the 18 variables analyzed (PC1, PC2 and PC3 score values from (Ankle_{SAG}, Knee_{SAG}, Knee_{FRT}, Ankle_{ROM}, Knee_{ROM} and Hip_{ROM})). This statistical procedures were made using SPSS (v.17; SPSS Inc, Chicago, IL) software with a significance level of $\alpha=0.05$.

3.3 Results

All variables presented normal distribution. No differences were found among the mean walking speed in all the conditions tested ($F=2.078$; $p=0.058$).

Considering the classical approach, in which six discrete variables were analyzed (PeakAnkle_{SAG}, PeakKnee_{SAG}, PeakKnee_{FRT}, Knee_{ROM}, Ankle_{ROM}, Hip_{ROM}), only in one of them differences were found between experimental conditions and CON (Table 3.1) where 2L condition presented significant difference from the CON condition in Ankle_{ROM} ($p<0.001$).

Table 3.1: Classical approach results: mean $\pm$ SD for the peak net moment and range of motion (ROM).

		WEDGES						
	Variables	CON	1L	2L	1M	2M	1P	2P
PEAK (Nm/BW)	Ankle _{SAG}	2.5 $\pm$ 2.4	3.0 $\pm$ 2.4	6.0 $\pm$ 1.8	0.0 $\pm$ 1.8	7.0 $\pm$ 1.8	4.0 $\pm$ 1.8	2.5 $\pm$ 2.4
	Knee _{SAG}	2.6 $\pm$ 2.3	2.8 $\pm$ 2.2	2.4 $\pm$ 2.4	2.7 $\pm$ 2.2	2.3 $\pm$ 1.4	0.0 $\pm$ 2.8	4.3 $\pm$ 4.4
	Knee _{FRT}	7.7 $\pm$ 2.6	7.5 $\pm$ 2.3	6.7 $\pm$ 7.0	8.4 $\pm$ 4.0	7.7 $\pm$ 2.7	6.1 $\pm$ 2.8	7.9 $\pm$ 2.2
ROM (deg)	Ankle _{ROM}	40.8 $\pm$ 4.3	40.9 $\pm$ 4.3	53 $\pm$ 5.6*	35.6 $\pm$ 2.7	35 $\pm$ 1.7	41.1 $\pm$ 4.0	39.4 $\pm$ 4.7
	Knee _{ROM}	50.4 $\pm$ 1.8	54.3 $\pm$ 7.0	55.8 $\pm$ 2.5	52.5 $\pm$ 2.3	54.5 $\pm$ 1.9	52.0 $\pm$ 2.7	50.0 $\pm$ 5.4
	Hip _{ROM}	46.5 $\pm$ 1.5	49.7 $\pm$ 0.0	50.8 $\pm$ 2.0	46.7 $\pm$ 1.9	45.6 $\pm$ 1.3	48.6 $\pm$ 5.0	49.2 $\pm$ 2.7

*statistically significant differences from CON group.

Referring to the PCA approach, PC score values were generated for each subject in each condition as shown in Table 2. In five of the six variables analyzed (Ankle_{SAG}, Knee_{FRT}, Ankle_{ROM}, Knee_{ROM} and Hip_{ROM}) the PCA approach was able to identify the wedges influence. Since in the variable Knee_{SAG} the PC model was not able to found the wedges influence, these data were not explored in detail (Table 3.2).

In Ankle_{SAG}, the PC1 load vector is higher than 0.71 (grey area) from 20% until 80% of the stance phase; the PC2 load vector was significant from 82% until 98%; while PC3 did not reach the threshold during the stance phase (Fig. 3.2a). The wedges 1L (p=0.01) and 2L (p<0.001) significantly alter the PC score values in PC1. PC2 score values were not significantly affected by any of the wedges (Table 3.2). Ranking the PC score values from all subjects of all the wedges that showed differences in Ankle_{SAG} PC1 (1L and 2L), the one that presented the highest influence was 2L. Then, to emphasize the different gait patterns with and without the wedge, the waveforms of the lowest (2L) and highest (CON) PC score values condition are presented in Figure 3.2b).

Table 3.2: Classical approach results: mean $\pm$ SD for the peak net moment and range of motion (ROM).

		Wedges							
Variables		CON	1L	2L	1M	2M	1P	2P	
Moment	Ankle _{SAG}	PC1	0.17 ±0.92	-0.13 ±0.71*	-0.56 ±0.38*	0.16 ±0.42*	-0.19 ±0.63	0.01 ±0.64	0.20 ±0.94
		PC2	0.24 ±0.75	0.08 ±1.26	-0.32 ±0.90	0.05 ±0.75	-0.04 ±0.47	0.11 ±0.82	-0.07 ±0.94
		PC3	0.09 ±0.74	-0.14 ±0.75	-0.12 ±1.27	-0.12 ±0.72	0.08 ±0.87	0.01 ±0.64	-0.18 ±0.81
	Knee _{SAG}	PC1	0.17 ±0.73	0.22 ±0.77	-0.39 ±0.92	0.00 ±0.84	-0.09 ±0.71	0.01 ±0.88	0.46 ±1.48
		PC2	0.14 ±0.85	-0.21 ±0.96	0.25 ±1.23	-0.36 ±0.75	-0.15 ±0.58	0.09 ±0.98	0.19 ±1.28
		PC3	0.09 ±0.62	0.20 ±0.76	-0.01 ±1.39	-0.16 ±0.95	-0.28 ±0.44	0.02 ±0.52	0.85 ±0.10
	Knee _{VAL}	PC1	-0.26 ±0.95	0.09 ±0.90	0.29 ±1.23*	-0.44 ±0.76	-0.14 ±0.57	0.17 ±0.67*	-0.04 ±0.61*
		PC2	0.48 ±1.01	-0.65 ±0.91	-0.01 ±1.04	0.14 ±1.4	0.54 ±0.88	0.34 ±0.70	-0.34 ±1.28
		PC3	0.25 ±0.9	0.39 ±1.00	-0.07 ±1.38	-0.74 ±1.2	-0.62 ±1.00	0.06 ±0.93	0.53 ±0.81
Range of Motion	Ankle _{ROM}	PC1	0.31 ±0.74	0.27 ±0.68	-0.73 ±0.93*	0.57 ±0.24	0.52 ±0.42	-0.09 ±0.81	0.58 ±1.26
		PC2	-0.24 ±0.85	1.39 ±0.66*	0.77 ±1.12*	0.72 ±0.99*	-0.01 ±1.26	0.26 ±1.19	0.96 ±0.91*
		PC3	0.13 ±0.99	0.76 ±0.68*	-0.96 ±0.83*	-0.29 ±0.35	-0.31 ±0.66	-0.12 ±1.18	0.20 ±0.65
	Knee _{ROM}	PC1	-0.13 ±0.89	1.02 ±0.83*	0.83 ±0.81*	0.58 ±1.11	0.67 ±1.02	0.48 ±1.02	-1.05 ±1.18
		PC2	-0.13 ±0.86	-0.05 ±0.98	-1.41 ±1.03	0.07 ±1.46	-0.09 ±1.35	-0.04 ±1.23	-0.72 ±1.22
		PC3	0.01 ±1.00	-0.01 ±0.95	-0.49 ±1.36	0.30 ±0.16	0.11 ±0.81	-0.26 ±0.73	-0.19 ±0.96
	Hip _{ROM}	PC1	0.00 ±0.79	-0.30 ±1.17	-0.06 ±1.59	-0.94 ±2.38	-0.45 ±0.81	-0.36 ±0.72	-0.17 ±1.39
		PC2	0.67 ±1.36	-0.74 ±2.36	-1.17 ±1.15*	-2.24 ±1.40*	-0.82 ±1.35*	-0.91 ±1.10*	0.07 2.34
		PC3	0.00 ±0.98	-0.34 ±1.09	-0.40 ±1.37	-0.32 ±1.57	-0.16 ±0.88	-0.04 ±1.10	0.33 ±1.07

*statistically significant differences from CON group.

n.a: not applicable (load vector under the 0.71 threshold).

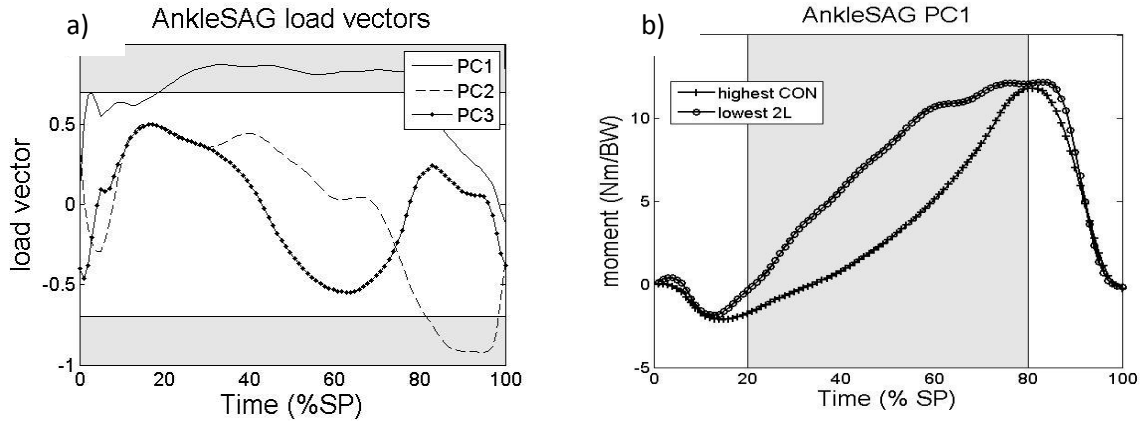


Figure 3.2: Ankle moments in the sagittal plane: a) load vectors for PC1, PC2 and PC3; b) highest and lowest scores in 95% CI. Negative: ankle dorsiflexion moment. Positive: ankle plantarflexor moment. The grey area highlights the 0.71 threshold (Knapp & Comrey 1973).

For the variable $Knee_{FRT}$, PC1 presented a plateau from 10% to 90% and PC2 from 90% to 100% (Fig 3.3a). The conditions that presented significant differences from CON, all in PC1, were 2L ($p=0.01$), 1P ($p<0.001$) and 2P ($p=0.02$), all of them increasing PC score values (Table 3.2). 1P is the one with higher differences. The increasing of the PC score values means that the varus moment is increased, meaning that the wedges tend to reduce the valgus moment when compared to walking with no wedges (Fig. 3.3b).

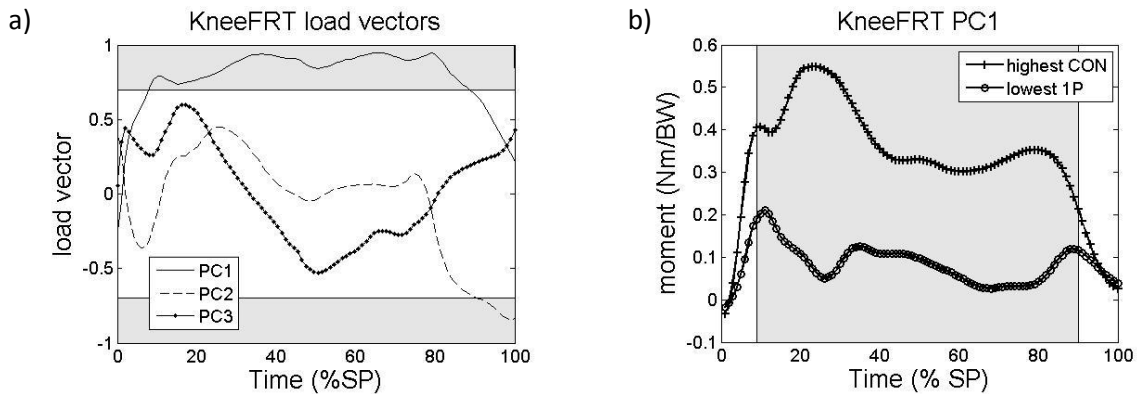


Figure 3.3: Knee frontal moment: a) load vectors for PC1, PC2 and PC3; b) highest and lowest scores in 95% CI. Positive: knee valgus moment. The grey area highlights the 0.71 threshold (Knapp & Comrey 1973)

All of the ROM variables are significantly affected by the wedges. In Ankle_{ROM} the PC1, PC2 and PC3 score values were influenced by the wedges (Table 3.2). PC1 is significantly reduced by the use of 2L ($p<0.001$) which is relevant at the end of SP (80-100%) and at the end of the swing phase (130-140%) (Fig. 3.4a). The plantarflexion is reduced by the use of 2L, emphasized in the grey areas (Fig. 3.4b), in the regions where PC1 is representative (Fig 3.4a). The wedges also significantly decrease PC2 scores (Table 3.2). PC2 shows a significant load vector from 75 until 85% (Fig 3.4a). The wedges tend to decrease the plantarflexion in the region where PC2 load vectors are relevant (Fig. 3.4c). PC3 shows a plateau from the beginning of heel contact (20%) until 60% of SP. The PC score values reduction found by the wedges 2L (Table 3.2) also emphasizes the plantarflexion reduction from 20-60% of SP (Fig 3.4d).

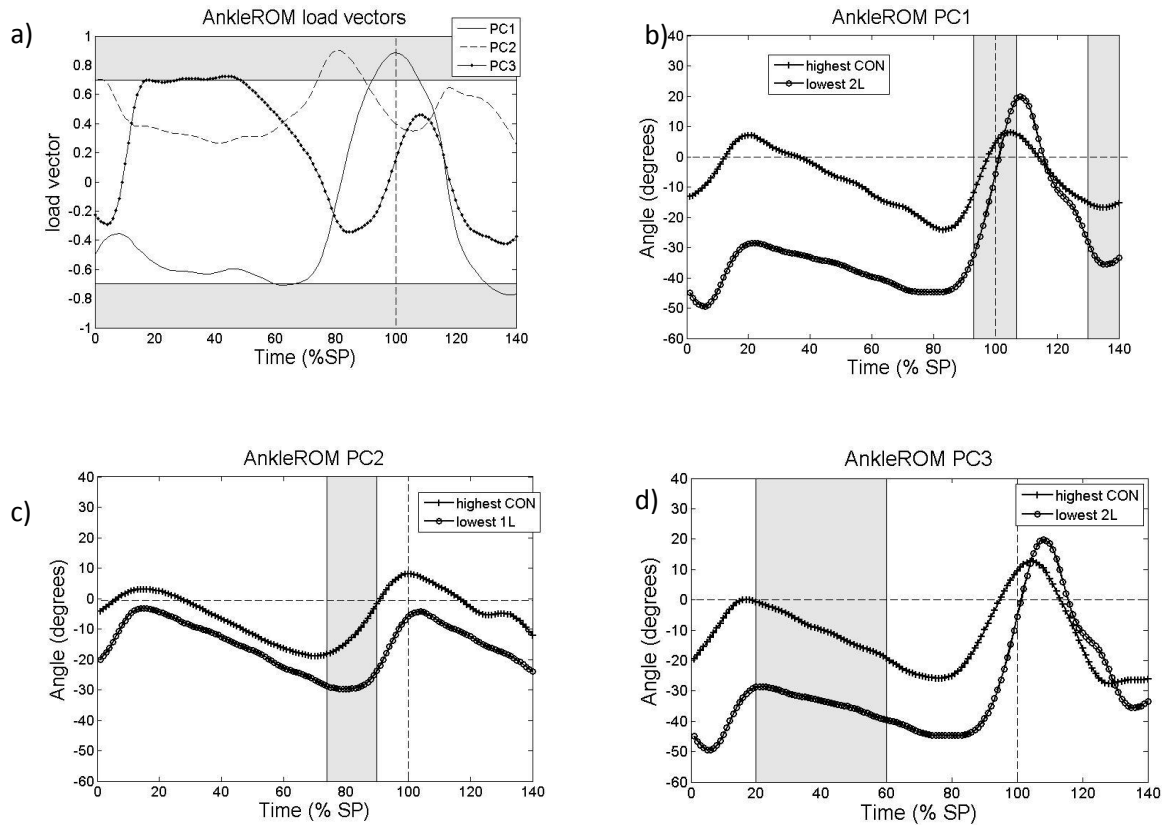


Figure 3.4: Ankle range of motion in the sagittal plane, for the total gait cycle (stance and swing phases): a) load vectors for PC1, PC2 and PC3; b) highest and lowest scores in 95% CI in AnkleROM PC1; c) highest and lowest scores in 95% CI in AnkleROM PC2; d) highest and lowest scores in 95% CI in AnkleROM PC3.

The only PC that is influenced by the wedges in Knee_{ROM} is PC1, when wearing the lateral wedges (Table 2). PC1 reaches the most negative values after 90% of GC, which correspond to the end of SP. The lateral wedges tend to reduce the ROM of the Knee (Fig 5b).

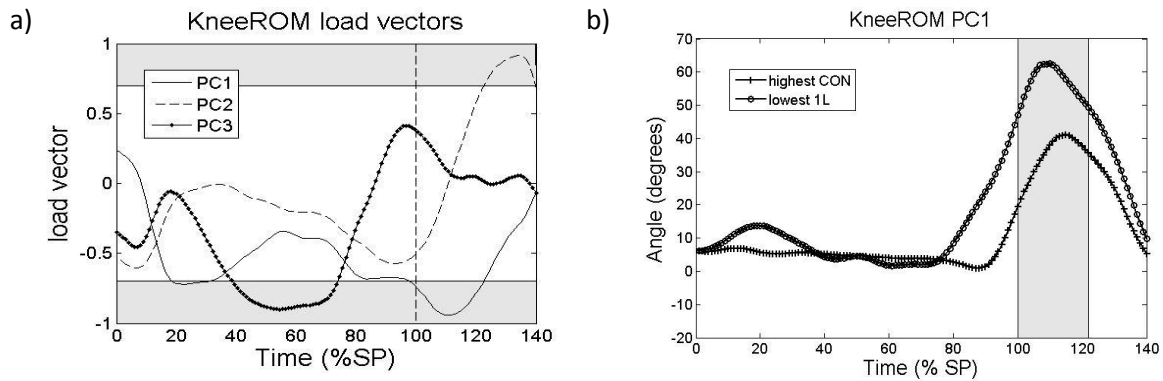


Figure 3.5: Knee range of motion in sagittal plane: a) load vectors for PC1, PC2 and PC3; b) highest and lowest scores in 95% CI. Positive: knee flexion. The grey area highlights the 0.71 threshold (1973).

Hip_{ROM} is the variable that is more influenced by the wedges (2L, $p=0.03$; 1M, $p<0.01$; 2M $p<0.01$ and 1P, $p=0.03$), all of them in PC1 (Table 3.2). The first PC (Fig 3.6a) represents the whole SP (Fig. 3.6a). The wedges tend to decrease hip flexion and to increase hip extension (Fig 3.6b).

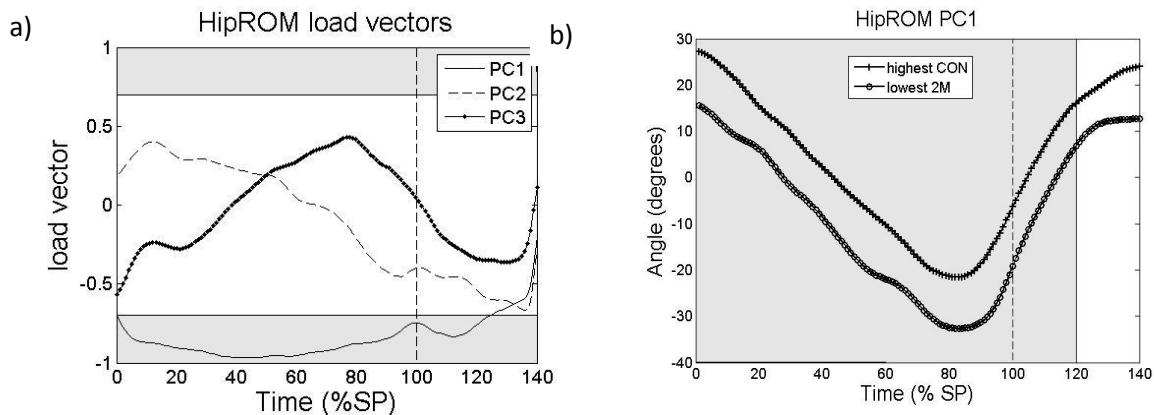


Figure 3.6: Hip range of motion in sagittal plane: a) load vectors for PC1, PC2 and PC3; b) highest and lowest scores in 95% CI. Positive: hip flexion. Negative: hip extension.

3.4 Discussion

The purposes of this study were to apply the PCA analysis in the variables of joint net moment and ROM waveforms to characterize the influence of different wedges on the gait pattern of healthy elderly participants. Furthermore, some discrete parameters representative of net joint moment and joint angles were calculated in order to verify the differences of the results provided for PCA analysis and discrete parameters (being the former the classical approach). The results of the present study corroborate with those of a previous study (Deluzio & Astephen, 2007), which showed that PCA is an interesting method to analyze and find pattern differences in gait data because: a) it reduces the number of variables necessary to represent the whole gait waveform; b) data from the entire gait cycle are considered; c) data reduction results in a set of uncorrelated features that explain most of the variance presented in the data and; d) significant differences that are not found in the discrete parameter analysis were found in PC score comparisons.

The objective in analyzing the waveforms in the light of PCs is first to reduce the number of variables to analyze and then to find a biomechanical meaning for the PCs (Sadeghi et al., 2002). The model presented here retained 3 PCs that accounted for at least 92% of the variance explained in joint moment variables and 86% in ROM variables. In this way, there could be regions of the gait cycle that are not taken into account in this model, namely the regions without any PC load vector above the 0.71 threshold. However, the analysis of PCs with higher index (PC4 to PC10) showed that the variance accounted for each one of these PCs was smaller than 5% and none of them reached the 0.71 threshold. Also, Deluzio et al. (1997) stated that PCs accounting for smaller variances are harder to explain. Even if the analysis of 3 PCs was not able to represent the whole waveform, it usually did represent most of it, while the choice of discrete parameters is often subjective, and chosen parameters can be highly correlated (Olney et al., 1998) and represent a very small portion of the waveform. It is also often difficult to subjectively choose parameters that can adequately characterize the curves; potentially meaningful parameters can easily be overlooked in subjective parameter extraction (Deluzio & Astephen, 2007).

Another limitation to PCA is that this model is generated based in CON data, and then applied to the wedges conditions. In this sense, the variability induced by the wedges conditions could be in a region out of the ones retained for analysis. However, the PCA

model based in CON data provides limits to which experimental conditions should be compared (Deluzio et al., 1997). An advantage in using PCA analysis is the sensitivity of the method. The only variable that showed differences from CON in the classical approach was Ankle_{ROM} (Table 3.1), where 2L conditions decreased the ROM of the Ankle joint. In PCA, 5 of the 6 variables were significantly different from CON. Furthermore, the difference in Ankle_{ROM} found in the classical approach was also apparent when analyzing the PC score values (Table 3.2), while other differences, not observed in the classical analysis, became clear.

The wedges that most influence Ankle_{ROM} were the lateral ones. As PC2 is practically constant along the gait cycle, although not always above the 0.71 threshold, it is expectable that the differences from 2L condition would happen during all of the gait cycle, as found (Figure 3.4c). PC1 and PC3 showed peaks in a very narrow region after toe off (100% and 110%). This could explain why 2L shows the highest differences from CON in both PCs, increasing the plantarflexion angle. During the stance phase, both lateral wedges decrease Ankle_{ROM}, while in the swing phase, plantarflexion angle increases. This could be explained by the higher dorsiflexion position to which the foot is submitted due to the use of the wedges in the forefoot. This increased dorsiflexion position before the toe off phase may cause a higher contraction of the triceps surae muscles by myotactic reflex at toe off and, consequently, to promote a higher ankle plantar flexion. Lee et al. (2009) found that Ankle_{ROM} is also able to differentiate groups (healthy and obese), where the obese group presents a smaller ROM during gait cycle.

The influence of the wedges on the gait is not yet totally clear. Most articles in the literature study the influence of lateral wedges on knee valgus moment, mostly because of its application in patients with knee osteoarthritis. Some of them found differences in the knee frontal moment when wearing lateral wedges (Erhart et al., 2008; Kuroyanagi et al., 2007) or medial wedges (Schmalz et al., 2006), while some found no differences (Kakihana et al., 2005; MacLean et al., 2006). In the present study, the PC1 of Knee_{FRT} explains the total foot contact during SP and indicates that the 2L, 1P and 2P wedges reduce the valgus moment (Figure 3.3b and Table 3.2).

No differences were found in PC score values in Knee_{SAG} data. In contrast, Deluzio et al. (1999) comparing knee moments of healthy subjects and patients submitted to unicompartmental arthroplasty found that the knee moments in the three planes were able

to differentiate the groups. This could be due to the fact that in this study the subjects were all healthy in relation to knee limitations, differently from the patients in the other work (Deluzio et al., 1999), and in the present work, the knee joint in the sagittal plane doesn't show a high variability of motion like the ankle and the hip, and the moment curves in the sagittal plane are also only slightly affected.

The lateral conditions are the most influential, presenting differences from all the variables, except Knee_{SAG} (Table 3.2), which is consistent with other studies (Erhart et al., 2008; Kerrigan et al., 2002; MacLean et al., 2006; Schmalz et al., 2006) even considering that none of them used the PCA techniques to analyze the data. The medial wedges only have influence on Hip_{ROM}. In this study, posterior wedges seem to reduce Knee_{VAL} moment and Hip_{ROM}. Chiu and Shiang (2007) evaluated the cushioning properties of a 2mm thickness posterior insole in healthy subjects and conclude that the use of this insole promotes a higher shock absorption compared to shoes with limited cushioning properties. In a global approach, the whole influence of the wedge in all the joints involved in the skill should be considered.

The influence of the wedges in this study was studied in an acute situation, where the gait pattern was verified after a short adaptation. Nevertheless, differences were found and could be useful to evaluate the influence of wedges in the gait pattern. More studies should be done to analyze the influence of the wedges after a long term adaptation and to determine if the results obtained in a healthy population are similar to those from a special one.

Evidences show that the elderly population is more susceptible to diseases related to bone degeneration and fear of falling (R. N. Kirkwood et al., 2011). Since the wedges seem to alter the pattern of kinetic and kinematic data in a healthy similar population, the use of custom made insoles could be an alternative to help improving their gait pattern. A complementary work was developed to state more clearly how the wedges could be prescribed for a specific population with gait deviations using the analysis of this study (Soares DP, 2012).

3.5 Conclusion

The influence of the wedges seems to be clearer by analyzing through PCA methods than to use discrete parameters of gait individually chosen, where the differences between conditions could be hidden. PCA analysis seems to be a powerful tool to analyze gait parameters considering the whole waveform of the variables involved. The results obtained in a healthy elderly population could be useful to understand the change in the gait pattern and to provide quantitative data supporting wedge prescription for special populations.

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4 GROUND REACTION FORCE COMPONENTS AND CENTER OF PRESSURE DISPLACEMENT FROM TRANSFEMURAL AMPUTEES AND ABLE BODIED SUBJECTS: COMPARISON USING PRINCIPAL COMPONENT ANALYSIS

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ABSTRACT

Transfemoral amputees (TF) have an asymmetrical gait pattern, resulting in an overload of the sound limb. A significant barrier to the clinical use of gait information is the successful reduction and analysis of data. Principal Component Analysis (PCA) has become a method of reducing the amount of data allowing analyzing the whole waveforms in gait analysis. The purpose of this study was to compare the Ground Reaction Force (GRF) components and Center of Pressure (COP) displacement of gait waveforms between TF and able bodied subjects using PCA. Fourteen TF were compared to 20 able bodied subjects. The first three PCs from the 3 GRF components and COP displacements were retained for analysis. Results showed that the variable that most discriminate the TF gait are the single support phase in COPx and COPy and the toe-off in GRF medio lateral. PCA seems to be a powerful tool to analyze the whole waveform, facilitating the identification of TF gait characteristics that differ from asymptomatic individuals. This approach could enable physical therapists to develop strategies to improve the gait pattern and consequently the quality of life of TF patients.

4.1 *Introduction*

Even if almost practically all the recently amputated individuals are affected by an initial depression, this feeling is quickly replaced by a desire of coming back to an active life, and, if possible, to the anterior activities (Christensen et al., 1995). Orthopaedic prosthetic devices are engines that intend to substitute a limb of the human body aiming to allow functionality as close as possible to the original function of the amputated limb. The prosthetic devices available for amputations above the knee induce a pattern of movement away from the normal, with high asymmetry between limbs. Able bodied subjects present less than 10% of asymmetry between limbs during gait, while transfemoral (TF) amputees show more than 23% (Nolan et al., 2003; Rabuffetti et al., 2005), resulting in an overload of the sound limb. 71% from the unilateral amputees report pain in the sound limb and lower back (Kulkarni et al., 1998; Melzer et al., 2001) and they present about 56% higher oxygen consumption than able bodied subjects (Schmalz et al., 2002).

The studies found in the literature that analyzed TF amputees' gait, present comparisons between the sound limb and amputated limb (Nolan et al., 2003), between amputees and a healthy group (Nolan et al., 2003; D. A. Winter & S. E. Sienko, 1988) and also the comparison of different kinds of prosthetics used by the same subject (Schmalz et al., 2002; D. A. Winter & S. E. Sienko, 1988). The results are commonly presented as the analysis of discrete parameters extracted from the kinematic and kinetic curves, generating a huge amount of data, that sometimes is difficult to interpret (Chui & Lusardi, 2010). This approach relies on the definition of discrete parameters that is subjective, and it becomes difficult to extract the same values of all temporal waves, especially in the presence of pathologies (Landry et al., 2007). A significant barrier to the clinical use of gait information is the successful reduction and analysis of data (Chau, 2001).

Deluzio et al. (1997) introduced a novel application of Principal Component Analysis (PCA) to the analysis of kinematic and kinetic data, and since then, PCA has become a method of reducing the amount of data and analyzing the whole waveforms in gait analysis (Muniz, 2009). Some studies in the literature have used PCA in order to state a basis of "healthy" gait patterns, and then to apply this standard to different groups of pathologies like osteoarthritis (Deluzio & Astephen, 2007; Deluzio et al., 1999; R. N. Kirkwood et al., 2011), tibial fracture (Muniz & Nadal, 2009) and Parkinson disease (Muniz

et al., 2010) in order to determine the major differences in walking patterns to facilitate physical therapists' decision making on treatments.

None of the mentioned studies compared the gait of an able bodied group and TF amputees in the light of PCA. This approach may bring new insights about the TF gait providing information which may improve prosthetic models and, consequently, to promote a better gait pattern, pain reduction and improvement in quality of life. In this way, the purpose of this study was to compare the GRF components and COP displacement obtained during gait between TF amputees and able bodied subjects using the PCA technique.

4.2 *Methods*

4.2.1 *Participants*

The control (CON) group was composed by 20 physically active subjects (mean age 67 ± 8.56 years old; mean weight 68.5 ± 6.2 kg). As including criteria, the subjects should be more than 50 years old and practice physical activity regularly. As exclusion criteria, they could not have any kind of limitation or pain during gait. Thirteen TF amputees, 12 male and one female (mean age of 56.7 ± 11.7 years old and mean body mass of 71.4 ± 11.7 kg) were enrolled in the study. All subjects had suffered the amputation more than six years before; the individuals did not show any sign of comorbidity or any pathology other than the amputation. The detailed description of the prosthetic device is shown in Table 4.1.

As a measure of the capacity and physical independence of the participants the questionnaire SF-36 (Ware & Gandek, 1998) v.2 was applied and the physical function field was analyzed. The values of 68 ± 24.9 points to the experimental group and 83 ± 18 points to CON were found, indicating that they felt few limitations to perform activities of the daily living involving physical function

Table 4.1: Subjects and prosthetic device. tra: traumatic; vas: vascular disease. Poli: Policentric; Uni: Uniaxial with friction locker; exo: Uniaxial (exoskeletal). Socket: 1) CAT/CAM suction valve; 2) CAT/CAM with locking pin; 3) Quadrilateral silicone interface with locking pin

subject	age	Years of amputation	Cause	foot	Foot specification	Knee	Knee specification	Socket
1	62	40	tra	Multiaxial	Mutiflex (endolite)	Uni	3R49 (Otto Bock)	2
2	58	36	tra	Articulated	1A13 (Otto Bock)	Uni	3R49 (Otto Bock)	1
3	57	36	tra	Multiaxial	Mutiflex (endolite)	Uni	3R49 (Otto Bock)	1
4	48	25	tra	Fixed	Sach (Otto Bock)	Uni	3R49 (Otto Bock)	2
5	64	50	tra	Multiaxial	Mutiflex (endolite)	Uni	3R49 (Otto Bock)	3
6	36	9	tra	Multiaxial	Mutiflex (endolite)	Uni	3R49 (Otto Bock)	3
7	54	35	tra	Articulated	1A30 (Otto Bock)	Uni	3R49 (Otto Bock)	2
8	54	31	tra	Multiaxial	Mutiflex (endolite)	Uni	3R49 (Otto Bock)	2
9	67	9	vas	Multiaxial	Mutiflex (endolite)	Uni	3R49 (Otto Bock)	1
10	68	9	tra	Fixed	Sach (Otto Bock)	Uni	3R15 (Otto Bock)	3
11	56	25	tra	Articulated	1A30 (Otto Bock)	Exo	Juppa (Otto Bock)	1
12	59	36	tra	Multiaxial	Mutiflex (endolite)	Uni	TK1900 (Ossur)	1

4.2.2 Protocol

The protocol consisted in walking on an 8m walkway, with the subjects wearing their own shoes at a self-selected speed. Previously to the test, the participants were familiarized with the environment, and during data collection they performed, at least, seven steps (three before and three after reaching the force plate). The subjects performed as many trials as necessary to acquire three valid gait cycles with the amputated leg (AL) and three with the sound limb (SL) while three valid trials with the right leg were acquired for the CON group.

4.2.3 Gait analysis and signal processing

A piezoelectric force plate (KistlerTM Instruments AG, Winterthur, Switzerland) at 1000 Hz placed in the middle of the pathway was used to collect the GRF components. The COP trajectory was collected at 300Hz using a pressure plate FootScanTM (RsScan, Olen, Bélgica) with 0.5 m length and 0.4 m wide with 4096 sensors. The pressure plate

was placed on top of the force plate. The kinetic data was collected in the software SIMI™ 7.0 and the pressure data with the software FootScan™ 7 gait 2nd generation. The data processing, filtering, and PCA analysis were performed using MATLAB™ 7.0. All systems were synchronized.

To reduce the effect of random noise, the data was filtered using a Butterworth filter: the kinetic data with cutoff frequency of 4Hz and the COP data with cutoff frequency of 2Hz, both in 4th order. The signals were interpolated and resampled in order to obtain 100 points, providing one point for each percent of the stance phase.

4.2.4 *Principal Component Analysis*

PCA was performed in GRF and COP waveforms according to Deluzio et al. (1997). In summary, the aim of PCA is to summarize the information contained in 100 variables, representing 100% of stance phase in a smaller number of components that explain the greater variance through linear combinations from those variables (Jolliffe, 2004). Principal Components (PCs) are arranged in decreasing order in such a way that the first PC accounts for most of the variability in the data, and each succeeding component accounts for as much of the remaining variability as possible (Daffertshofer et al., 2004). In this study, the number of PCs retained for analysis was always 3.

PCs are an orthogonal transformation which converts p variables $X = x_1, x_2, x_3, \dots, x_p$ (in this case from 0 to 100% stance phase) into p new uncorrelated PCs $Z = z_1, z_2, z_3, \dots, z_p$, which are defined by the equation $Z = U^T X$, where U are the eigenvectors of the covariance matrix of X . U_n is calculated by the equation $S U_n = \lambda U_n$ where λ are the p eigenvalues ranked in decreasing order and S is the covariance matrix of X .

In this work, the PC model was developed based on the gait pattern of the subjects from CON group. This model was afterwards applied to the subjects from TF and PC score values (z_n) for each subject were retained for analysis. The last step was to determine meaningful biomechanical labels for each PC. Knapp (Knapp & Comrey, 1973) suggested a threshold of ± 0.71 to consider a load vector from one variable as relevant, and then to attribute a meaning for this PC.

According to Muniz & Nadal (2009) that analyzed the differences in classification using two, four and six PCs in GRF waveforms, two PCs reached 100% of specificity in

separating the healthy group from the tibial fracture group. Besides, Deluzio et al. (1999) stated that PCs with smaller variances are harder to explain. Considering these facts, 3 PCs were retained in each waveform for analysis (Table 4.2).

4.2.5 Statistical procedures

The mean of the three valid trials of each subject was computed and the statistical procedures were performed with these values. The normality of the data was verified using the *Shapiro-Wilk's* test and the homogeneity of the variances using *Levene's* test. Independent *student t tests* were used to compare CON vs. SL and CON vs. AL for the PCs from the five variables analyzed (GRFvt, GRFml, GRFap, COPx and COPy). The level of significance used was $\alpha=0.05$.

4.3 Results

Analyzing the data presented in Table 4.2, the portions where SL is affected are during initial contact in GRFap (2-4%); heel contact in GRFvt and COPx (7-12%); the beginning of single support in GRFml (12-16%); during single support (30-75%) in GRFvt and COPx; and in late midstance phase (30-55%) in GRFml and COPy. AL gait waveforms are affected during initial contact (2-4%) in GRFml; in heel contact (7-12%) in GRFvt; in single support (30-75%) in GRFvt and COPx; in early midstance (20-40%) in GRFvt and in late midstance (30-55%) in GRFml. Negative and positive antero-posterior peaks were also affected, with the scores being smaller than CON.

Table 4.2: PC1, PC2 and PC3 scores obtained in CON, AL and SL (Mean $\pm$ SD), % of variance explained with 3PCs, portion of the waveform with load vector higher than 0.71 and biomechanical interpretation of this portion in GRFvt, GRFml, GRFap, COPx and COPy.

		CON	AL	SL	%Var	Portion (%)	Interpretation
GRFvt	PC1	0.07 $\pm$ 0.43	-1.61 $\pm$ 0.79*	0.17 $\pm$ 0.50	79.98	20-40	Early midstance
						80-100	heel rise
						35-75	Single support
	PC2	0.33 $\pm$ 0.47	-0.82 $\pm$ 0.53*	-0.28 $\pm$ 0.35*		7-12	Heel contact
	PC3	0.28 $\pm$ 0.35	1.75 $\pm$ 1.21*	-0.74 $\pm$ 1.17*			
GRFml	PC1	0.00 $\pm$ 1.18	1.85 $\pm$ 1.99*	1.67 $\pm$ 1.93*	74.45	30-55	Late midstance
						65-90	heel rise
						12-16	Beginning of single support
	PC2	0.00 $\pm$ 0.90	-0.50 $\pm$ 1.05	-1.38 $\pm$ 0.89*		2-8	Initial contact
	PC3	0.00 $\pm$ 0.90	-1.11 $\pm$ 0.67*	0.08 $\pm$ 0.67			
GRFap	PC1	0.00 $\pm$ 0.89	-2.35 $\pm$ 0.34*	-0.32 $\pm$ 0.86	79.78	15-30	Braking and
						60-98	Propulsive peaks
						5-15	Heel contact
	PC2	0.00 $\pm$ 0.94	0.15 $\pm$ 0.80	-0.48 $\pm$ 1.26		2-4	Initial contact
	PC3	0.00 $\pm$ 1.12	-0.10 $\pm$ 2.04	2.26 $\pm$ 2.29*			
COPx	PC1	-0.52 $\pm$ 2.33	-4.22 $\pm$ 3.42*	-2.73 $\pm$ 2.88*	84.48	30-70	Single support
						8-12	Heel contact
						0-6	Initial contact
	PC2	0.60 $\pm$ 3.40	-3.98 $\pm$ 1.34*	-2.85 $\pm$ 3.90*			
	PC3	0.00 $\pm$ 3.49	1.96 $\pm$ 0.76*	0.25 $\pm$ 2.78			
COPy	PC1	0.57 $\pm$ 2.82	3.65 $\pm$ 5.89	5.44 $\pm$ 3.66*	93.99	25-92	Late midstance
							/heel rise
						-	
	PC2	0.00 $\pm$ 2.78	-1.84 $\pm$ 1.73	3.41 $\pm$ 5.64*		-	
	PC3	0.00 $\pm$ 3.49	0.98 $\pm$ 3.63	0.73 $\pm$ 3.76		-	

*Statistically significant differences from CON group.

Figures 4.1a) to 4.1e) show the first three PCs for all the variables analyzed, GRFvt (Fig. 4.1a), GRFml (Fig. 4.1b), GRFap (Fig. 4.1c), COPx (Fig. 4.1d) and COPy (Fig. 4.1e). The grey area in Figures 4.1 highlights the threshold area of 0.71, where the PCs have a meaningful interpretation (Knapp & Comrey, 1973). A discriminant analysis performed in all the PC models that were able to differentiate groups showed that PC1 COPx, PC1 COPy and PC2 GRFml are most likely to discriminate the groups, classifying correctly 80% of the cases. The component loadings of PC1 COPx are relevant from 30% to 70% representing the single support (Table 4.2, Fig 4.1d); in PC1 COPy from 40% to 90% of SP, corresponding to the late midstance/heel rise (Winter, 1990). GRFml is higher than

0.71 only in a minimal period between 12% and 16% of SP corresponding to the beginning of single support (Fig. 4.1b).

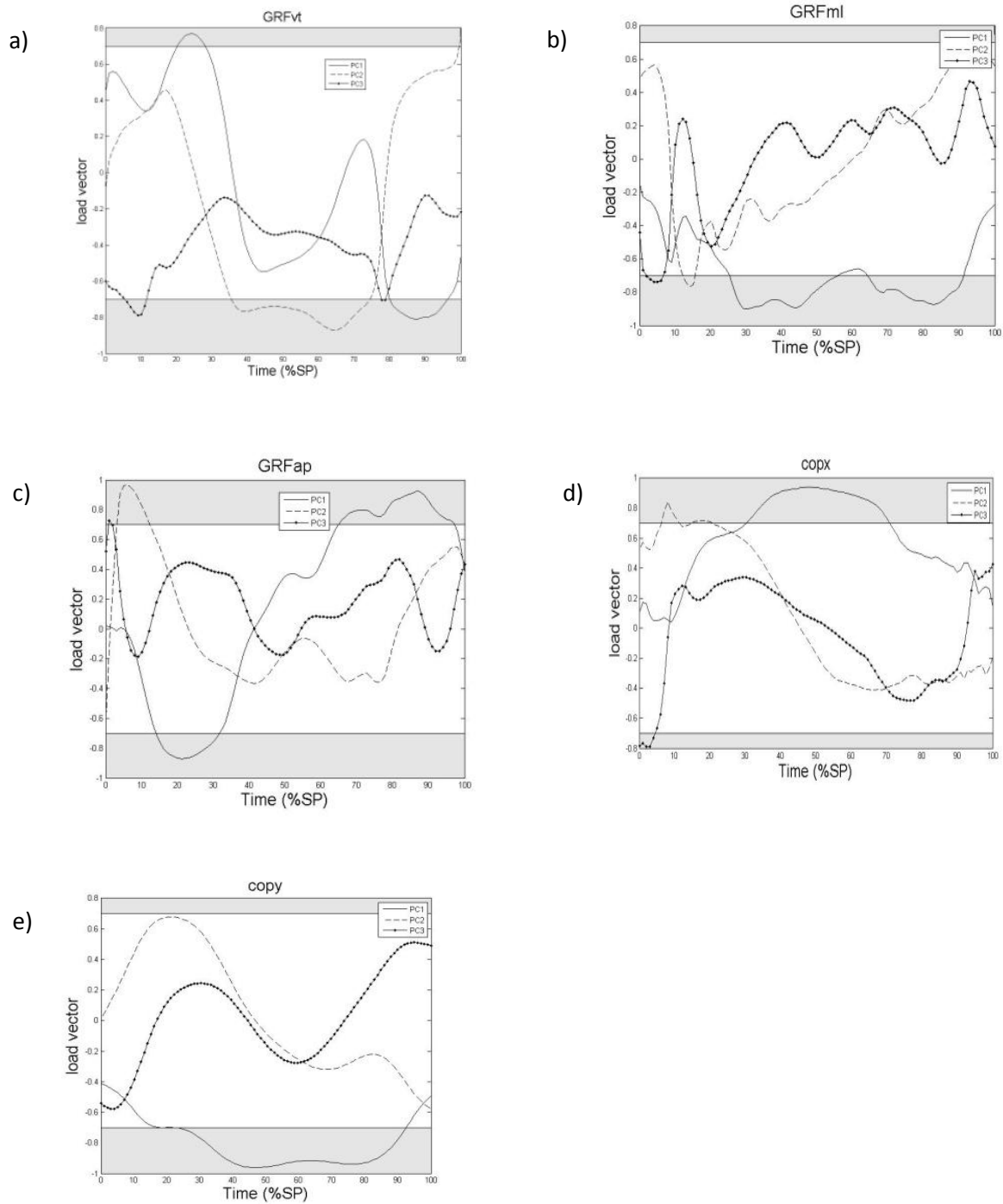
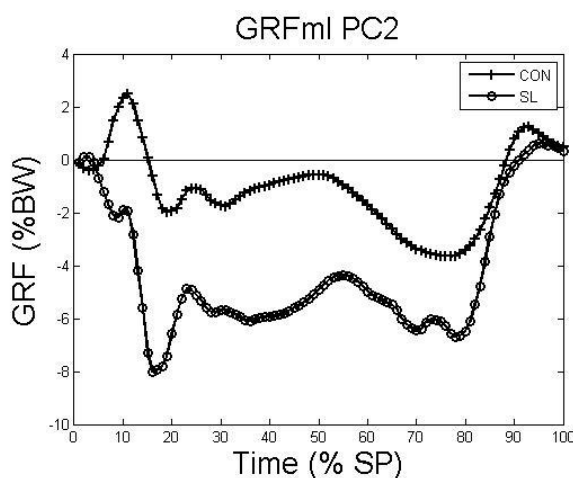
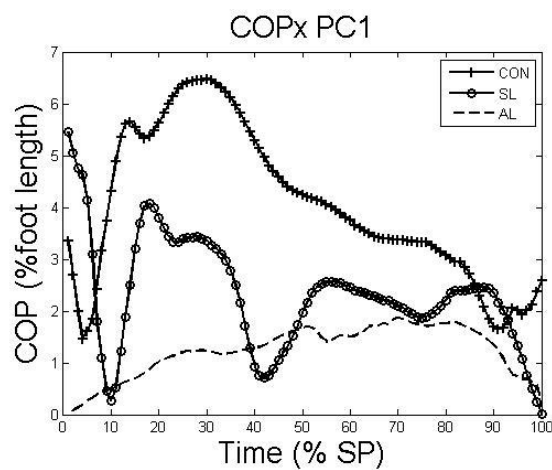


Figure 4.1: PC1, PC2 and PC3 load vectors to GRFvt (a), GRFml (b), GRFap (c), COPx (d) and COpY (e). The grey area indicates the threshold area of 0.71 (Knapp & Comrey, 1973).

COPx PC1 scores for CON are different from AL and SL (Table 4.2). The lateral displacement of COP is higher in CON, intermediary in SL and smaller in AL (Fig 4.2a). The highest and lowest scores of GRFml evidence the high difference between groups at the beginning of single support, from 12% to 16% (Fig 4.2b). The lowest score from SL group in COPy PC1 shows that the subject lowers the whole foot faster than the CON group, having the foot in total contact with the ground before 20% of SP (Fig 4.2c) then keeps the foot in total contact with the ground during the single support phase (midstance), and starts to lift the calcaneus when the contralateral limb already reached the ground, at 80% of SP.



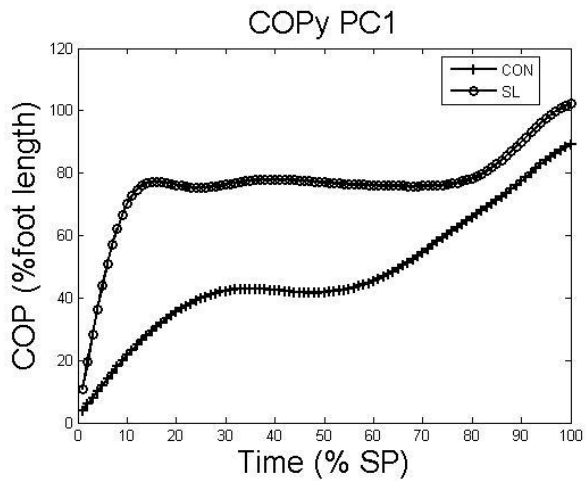


Figure 4.2: Gait waveforms corresponding to highest and lowest PC scores. a) Highest score in CON and lowest scores in AL and SL for COPx PC1; b) Highest scores in CON and lowest in SL for GRFml PC2; c) Highest scores in CON and lowest in SL for COPy PC1.

4.4 Discussion

The purpose of this study was to apply PCA analysis to GRF and COP waveforms to characterize the differences between TF gait patterns and an able bodied group. The results showed that PCA is a powerful method to analyze gait data because: a) it reduces the number of variables necessary to represent the whole gait waveform; b) data from the entire gait cycle are considered; c) data reduction results in a set of uncorrelated features that explain most of the variance presented in the data (Deluzio & Astephen, 2007).

When analyzing the AL gait pattern (Table 4.2, Fig 4.1a, Fig 4.3), GRFvt is affected in the whole waveform: during 7-12% of SP ($PC3 < -0.71$), AL waveform is below CON (Fig. 4.3), then between 20-40% ($PC1 > 0.71$) it shifts from below to above CON due to the slower load acceptance rate, consequently the first peak GRF happens later in the cycle (Fig. 4.3); along the gait cycle AL waveform is below CON ($PC2 > 0.71$), and this could be explained because the amputees don't support all their weight in the prosthetic device by fear of falling (Zmitrewicz et al., 2006). This also leads to the low bone mineral density in AL, due to the unloaded bone characteristics (Sherk et al., 2008). At the end of SP (80-100%) PC1 evidenced that the smaller scores were related to the anticipated support of SL in the ground. In GRFml, the smaller values in the initial contact ($PC3$) and during late

midstance (PC1, and also evidenced in COPx PC1) were due to the lack of control in contacting the ground, making the contralateral limb to perform the swing phase faster to anticipate the ground contact. GRFap braking and propulsive peaks were smaller than in CON. This was also found in other studies, and may be explained by the lack of energy absorption and energy storage properties of the prosthetic device (Fey et al., 2011; Zmitrewicz et al., 2006).

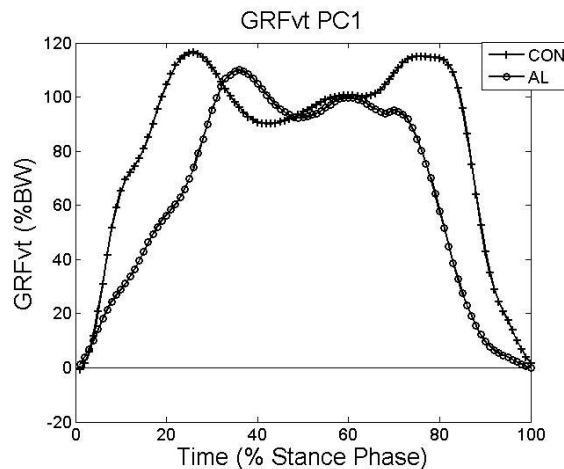


Figure 4.3: GRFvt waveform example: a) Highest score in CON and lowest score in AL for GRFvt PC1;

SL shows higher scores during heel contact and single support for GRFvt (Table 4.2). The heel contact starts earlier due to the lack of control of AL and precocious support of SL (Zmitrewicz et al., 2006). Also, the increased magnitude of GRFvt during the single support (40% of the whole SP) evidences the overload of SL that could explain the high levels of hip osteoarthritis in SL (Pieter et al., 2009). GRFml scores are higher during the beginning of single support, late midstance and heel rise (PC1 and PC2 – Table 4.2). The lack of control of AL difficulties the progression of the limb forward, causing a hip hiking of the remaining limb to lift laterally and progress, consequently the medial-lateral support of SL becomes more medial. This could be related to the overload in the medial compartment of the knee found in TF amputees (Hurwitz et al., 2000; Segal et al., 2006) and high level of knee osteoarthritis (Melzer et al., 2001). GRFap scores are higher in initial contact due to the need of SL in contacting the ground precociously in order to balance the support of AL. This is also evidenced in COPy, where the foot is in total contact with the ground since before 20% of SP (Fig 4.2c).

4.5 Conclusion

PCA seems to be a powerful tool to reduce data dimensionality and to analyze the whole waveform information. The gait analysis of TF amputees using PCA reduces the number of variables, facilitating the identification of gait characteristics that differ from asymptomatic individuals. This approach could enable physical therapists to develop strategies to improve the gait pattern and consequently the quality of life of these patients.

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5 A NEW APPROACH TO PRESCRIBE CUSTOM MADE WEDGES FOR INDIVIDUALS WITH TRANSFEMORAL AMPUTATION USING PRINCIPAL COMPONENT ANALYSIS.

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Abstract

Transfemoral amputees' (TF) gait pattern is asymmetrical, and they could benefit from the use of wedges to make their gait as close as possible to normal. The purpose of this study was to develop a rational and quantitative method to prescribe custom made wedges for the sound limb of TF amputees using Principal Component Analysis (PCA). Wedges were tested and their behavior in able bodied subjects (CON) was described. Using the influence of the wedges in CON, and the gait pattern of each TF individually, wedges were prescribed in order to modify their gait according to the specific effect of each wedge. The variables analyzed were the ground reaction force components and center of pressure displacement. The *Mahalanobis* distance for each variable and the 95% confidence interval (CI) based on CON data was calculated. Results showed that analyzing the *Mahalanobis* distance of the variables, TF subjects improved their gait; the variables that were out of the boundaries of 95% CI of CON, moved inside these boundaries with the use of wedges. The application of wedges to the sound limb of TF amputees improves their gait patterns, thus the PCA approach may help clinicians to prescribe the best device for each patient, and consequently to improve TF patient quality of life.

5.1 *Introduction*

Transfemoral (TF) amputees' gait is notoriously asymmetrical between limbs (Nolan et al., 2003; Rabuffetti et al., 2005). The use of a prosthetic device in one limb leads to a different pattern of gait due to, among other factors, the absence of muscles and the prosthetic device mass properties (Nolan et al., 2003). This asymmetry causes an overload on the intact limb (Zmitrewicz et al., 2006), where high rates of injuries, like knee and hip osteoarthritis (Melzer et al., 2001; Pieter et al., 2009), and low bone density in the hip of the amputated limb (Kulkarni et al., 1998; Sherk et al., 2008) have been reported. The role of therapists is to help TF amputees to reduce this asymmetry and to develop a gait pattern as close as possible to that of the able-bodied subjects (McNealy & Gard, 2008).

One therapeutical approach applied to alter the gait patterns is the insertion of wedged insoles inside the shoe. Their use has been described as a powerful tool for the compensation of small gait deviations. According to Kerrigan et al. (2002), the use of wedged insoles in the individual shoes could influence decisively the quality of their gait. Analyzing the influence of different parameters that constitute an wedged insole such as position, height, material and density, it is possible to build devices best fitting individuals with different gait deviations (Kerrigan et al., 2002).

The assessment of gait parameters in able-bodied subjects is frequently presented in the literature in order to define a normal pattern, due to their ease of access, and because of the possibility to test different interventions in a non-affected group while postulating similar effects on people with gait impairment. Since TF amputees could benefit from the use of wedges owing to their asymmetric pattern of gait (Erhart et al., 2008; Franz et al., 2008; Kuroyanagi et al., 2007; Schmalz et al., 2006), it seems that the study of the effects of wedges on the normal gait may help prescribing the most appropriate wedge for each TF amputee.

Principal Component analysis (PCA) is a method to evaluate data that greatly reduces the number of variables to analyze (Deluzio et al., 1997). This approach applied to gait waveforms allows a better understanding of differences in gait patterns and deviations

from normal walk. Also, it simplifies the evaluation of a clinical intervention because the number of parameters to analyze is smaller (Chau, 2001). The purpose of this study was to develop a rational and quantitative approach based on PCA to the prescription of wedges for the sound limb (SL) of TF amputees.

5.2 *Methods*

5.2.1 *Participants*

The control group (CG) was composed of 20 physically active subjects, 5 male and 15 female (mean age 67 ± 8.56 years old, mean weight 68.5 ± 6.2 kg), and the experimental group had 12 TF amputees, 11 male and one female (mean age 56.7 ± 11.7 years old, mean weight 71.4 ± 11.7 kg) – Table 1. One TF individual was younger than the other subjects (Subject 6, Table 1) but he was included in the analysis because the rationale was that the wedge prescription is individual matter. Both groups had a high score on the Physical Function domain of the QOL SF36 questionnaire (Ware & Gandek, 1998) (82.3 ± 18.0 for CG and 62.8 ± 24.9 for TF), meaning that they felt few limitations to perform activities of the daily living involving physical function.

Table 5.1: Experimental group: subjects and respective prosthetic device features. tra: traumatic; vas: vascular disease. Poli: Policentric; Uni: Uniaxial with friction locker; exo: Uniaxial (exoskeletal). Socket: 1) CAT/CAM suction valve; 2) CAT/CAM with locking pin.

subject	age	Years of amputation	Cause	foot	Foot specification	Knee	Knee specification	Socket
1	62	40	tra	Multiaxial	Mutiflex (endolite)	Uni	3R49 (Otto Bock)	2
2	58	36	tra	Articulated	1A13 (Otto Bock)	Uni	3R49 (Otto Bock)	1
3	57	36	tra	Multiaxial	Mutiflex (endolite)	Uni	3R49 (Otto Bock)	1
4	48	25	tra	Fixed	Sach (Otto Bock)	Uni	3R49 (Otto Bock)	2
5	64	50	tra	Multiaxial	Mutiflex (endolite)	Uni	3R49 (Otto Bock)	3
6	36	9	tra	Multiaxial	Mutiflex (endolite)	Uni	3R49 (Otto Bock)	3
7	54	35	tra	Articulated	1A30 (Otto Bock)	Uni	3R49 (Otto Bock)	2
8	54	31	tra	Multiaxial	Mutiflex (endolite)	Uni	3R49 (Otto Bock)	2
9	67	9	vas	Multiaxial	Mutiflex (endolite)	Uni	3R49 (Otto Bock)	1
10	68	9	tra	Fixed	Sach (Otto Bock)	Uni	3R15 (Otto Bock)	3
11	56	25	tra	Articulated	1A30 (Otto Bock)	Exo	Juppa (Otto Bock)	1
12	59	36	tra	Multiaxial	Mutiflex (endolite)	Uni	TK1900 (Ossur)	1

5.2.2 Instruments

Ground reaction forces and plantar pressure data were recorded using a piezoelectric force plate (Kistler Instruments AG, Winterthur, Switzerland) at 1000Hz, and a pressure plate (FootScan - RsScan, Olen, Belgium) at 300Hz, respectively. Both systems were synchronized using a separate unit equipped with a manual trigger to start simultaneously both systems. All gait data was collected in an 8m walkway, with the pressure plate placed over the force plate.

5.2.3 Protocol

The protocol was divided into four phases. The aim of *Phase I* was to evaluate the influence of the wedges on the CG gait pattern and that of *Phase II* was to analyze the gait pattern of the TF amputees. In *Phase III*, the aim is to prescribe the wedges for TF based on data from *Phases I* and *II*. *Phase IV* aim was to apply the wedges in TF sound limb (SL) prescribed in *Phase III*.

Phase I Protocol - Test of different wedges: the CG walked wearing their own shoes at a self-selected speed. After a short adaptation to the environment, each participant walked three times wearing their shoes (with no wedge), which was referred to as control condition (CON). Thereafter, they walked with four of six wedges selected randomly, inserted into their shoes. Three valid right lower limb trials were recorded for each of the four conditions. The tested wedges were made of polyurethane material: two lateral, placed under the 5th metatarsal head having thicknesses of 1 cm (1L) and 2 cm (2L); two medial, placed under the foot arch having thicknesses of 1.1 cm (1M) and 2.2 cm (2M); and two posterior having thicknesses of 0.9 cm (1P) and 1.9 cm (2P), placed under the *calcaneum* bone. Therefore, in overall, a total of seven conditions were studied: CON, 1L, 2L, 1M, 2M, 1P and 2P.

Phase II Protocol – TF Amputees' gait analysis: the same protocol of gait analysis than Phase I was performed. The amputees were not tested with wedges in this phase and they performed three valid trials with their SL. They wore their own shoes that were prescribed as part of the prosthesis, which were manufactured by the same company.

Phase III Protocol: this phase is described in detail in 5.2.5 (*Wedge prescription*).

Phase IV Protocol – TF Amputees (after wedge prescription): Three participants from the experimental group were tested with all the wedges that were considered to shift their gait pattern closer to that found on the able-bodied participants (see 5.2.5 – *Wedge prescription*). They walked for about five minutes with each wedge for familiarization and the gait protocol from *Phase I* was repeated in which two valid trials were recorded.

5.2.4 *Principal Component Analysis*

For the purpose of this study five waveforms were analyzed: vertical, medial-lateral, and anterior-posterior components of ground reaction forces (GRFvt, GRFml, and GRFap, respectively), as well as the medial-lateral and anterior-posterior displacement of the center of pressure (COPx and COPy, respectively). PCA was performed on the GRF and COP waveforms as per Deluzio et al. (1997). In this study, 3 PCs were retained for further analysis (Jolliffe, 1986).

The PC model was developed based on the gait pattern of the subjects walking in CON condition. This model was then applied to the subjects walking with the wedge

conditions and PC score values for each subject in each condition were retained for analysis. A total of 15 PC score values were analyzed for each subject with each condition: three score values (PC1, PC2 and PC3 scores) for each of the five waveforms (GRFvt, GRFml, GRFap, COPx, and COPy) were calculated.

For *Amputees' gait* analysis (*Phase II*), the PC score values were obtained based on the PC model in CON condition, and then this model was applied to the TF participants. PC score values for SL were retained for analysis.

5.2.5 Wedge Prescription

To prescribe the wedges for each TF individually, some steps were followed. First, the influence of the wedges was verified in CG (*Phase I*), where one way ANOVA and *post hoc* LSD were used to compare the data collected (CON vs. the six wedge conditions) for the three PC score values from the five variables analyzed. The level of significance was set at $\alpha=0.05$.

After that, the *Mahalanobis* distance (T^2) for CG and SL were calculated using the 3 PC score values obtained in each of the 5 waveforms. The CON data was used to calculate the 95% confidence interval (CI) where the normal gait is defined. Then, each TF T^2 measurement was compared to this interval to evaluate how far each subject was from the CON dataset. This parameter represents the distance between each SL PC score, (o_i) and the mean of the PC score values for CON ($\bar{x}$), normalized by the variance of each PC (Muniz & Nadal, 2009).

Equation 5. 1
$$T^2 = [(o_i - \bar{x})' S^{-1} (o_i - \bar{x})]$$

where S^{-1} is the inverse of the covariance matrix of X and $(o_i - \bar{x})'$ is the transpose of the vector $(o_i - \bar{x})$.

When a subject fell outside the CI range, the PC score values were then individually analyzed to determine which wedge could influence his/her gait pattern. This determination employed the results obtained in *Phase I*.

5.3 Results

Considering *Phase I* results, from the five variables analysed, 3 of them (GRFvt, GRFml, and COPx) were influenced by at least one of the six wedges (Table 5.2). GRFvt is the variable that was most influenced by the use of wedges, being significantly affected by four of them: both lateral wedges (1L and 2L) were different from CON in PC2 (1L: $p < 0.001$; 2L: $p = 0.02$) and PC3 (1L: $p < 0.001$; 2L: $p < 0.001$); 1M condition was different in PC2 ($p = 0.02$) and PC3 ($p = 0.04$); and 2M in PC3 ($p < 0.001$); GRFml was different in PC1 in 1L ($p = 0.03$), 1P ($p = 0.01$) and 2P ($p < 0.001$) conditions when compared to CON. COPx PC score values increased in PC1 in the conditions 1M ($p = 0.04$), 2M ($p = 0.01$) and 1P ($p < 0.001$) and decreased in PC2 for the condition 1L ($p = 0.01$) when compared to CON; The GRFap and COPy did not present significant difference between any wedge condition to CON ($p > 0.05$).

Table 5.2 : Phase I results - Wedges influence on CG: mean $\pm$ SD of the PC1, PC2 and PC3 scores from the GRFvt, GRFml, GRFap, COPx and COPy variables.

		CON	1L	2L	1M	2M	1P	2P
GRFvt	PC1	0.07 $\pm$ 0.43	0.12 $\pm$ 0.60	0.29 $\pm$ 0.43	-0.09 $\pm$ 0.39	0.16 $\pm$ 0.49	-0.1 $\pm$ 0.53	-0.09 $\pm$ 0.52
	PC2	-0.33 $\pm$ 0.47	0.44 $\pm$ 0.58*	0.15 $\pm$ 0.48*	0.12 $\pm$ 0.59*	-0.16 $\pm$ 0.27	-0.09 $\pm$ 0.53	-0.16 $\pm$ 0.06
	PC3	-0.28 $\pm$ 0.70	1.01 $\pm$ 0.28*	1.17 $\pm$ 0.16*	0.70 $\pm$ 0.95*	0.86 $\pm$ 1.11*	-0.03 $\pm$ 1.01	0.04 $\pm$ 1.18
GRFml	PC1	0.29 $\pm$ 0.95	-0.62 $\pm$ 0.92*	-0.29 $\pm$ 1.20	-0.24 $\pm$ 1.16	0.01 $\pm$ 0.91	-0.70 $\pm$ 0.68*	-1.38 $\pm$ 0.58*
	PC2	-0.13 $\pm$ 0.77	-0.17 $\pm$ 0.80	0.07 $\pm$ 0.47	-0.02 $\pm$ 0.67	-0.20 $\pm$ 0.42	-0.13 $\pm$ 0.69	-0.43 $\pm$ 0.38
	PC3	0.18 $\pm$ 0.54	-0.33 $\pm$ 0.91	0.27 $\pm$ 0.72	0.12 $\pm$ 0.16	-0.02 $\pm$ 0.55	-0.05 $\pm$ 0.20	0.19 $\pm$ 0.80
GRFap	PC1	0.00 $\pm$ 0.89	0.23 $\pm$ 0.66	0.23 $\pm$ 0.66	0.16 $\pm$ 0.69	0.07 $\pm$ 0.63	-0.28 $\pm$ 0.57	0.03 $\pm$ 0.47
	PC2	-0.29 $\pm$ 0.58	0.26 $\pm$ 0.57	0.05 $\pm$ 0.59	0.14 $\pm$ 1.00	-0.03 $\pm$ 0.94	-0.08 $\pm$ 0.77	-0.06 $\pm$ 0.72
	PC3	0.00 $\pm$ 1.12	0.07 $\pm$ 0.58	-0.19 $\pm$ 0.83	-0.19 $\pm$ 0.40	-0.2 $\pm$ 10.01	0.34 $\pm$ 0.62	-0.29 $\pm$ 0.30
COPx	PC1	-0.15 $\pm$ 2.21	-1.44 $\pm$ 4.55	-0.81 $\pm$ 3.87	-2.70 $\pm$ 1.82*	-3.25 $\pm$ 2.18*	-3.99 $\pm$ 2.12*	-1.81 $\pm$ 2.98
	PC2	0.00 $\pm$ 3.68	4.01 $\pm$ 4.20*	1.68 $\pm$ 4.31	-1.08 $\pm$ 3.42	-1.54 $\pm$ 2.42	-0.97 $\pm$ 3.01	0.05 $\pm$ 3.71
	PC3	0.00 $\pm$ 3.49	-0.21 $\pm$ 3.95	0.91 $\pm$ 3.88	0.11 $\pm$ 2.86	-1.89 $\pm$ 2.01	0.34 $\pm$ 2.23	-1.96 $\pm$ 1.65
COPy	PC1	0.00 $\pm$ 3.18	1.40 $\pm$ 2.63	-0.61 $\pm$ 2.80	-0.92 $\pm$ 2.90	-0.40 $\pm$ 2.47	1.21 $\pm$ 2.53	0.00 $\pm$ 1.86
	PC2	0.00 $\pm$ 2.78	-0.28 $\pm$ 1.54	0.07 $\pm$ 2.02	0.68 $\pm$ 2.78	0.17 $\pm$ 2.88	0.14 $\pm$ 1.75	1.27 $\pm$ 2.21
	PC3	0.75 $\pm$ 3.18	-0.07 $\pm$ 1.37	1.60 $\pm$ 3.44	0.86 $\pm$ 2.58	2.15 $\pm$ 1.33	-0.46 $\pm$ 3.94	-2.49 $\pm$ 2.18

*Statistically significant differences from CON condition ($p < 0.05$).

Considering the TF analysis (*Phase II*), almost all the subjects presented at least one variable with a *Mahalanobis* distance out of the 95% confidence interval range (Table 5.3), with the exception of subject 12. Subjects 11 and 12 have “n.a.” in GRF variables because their data could not be analyzed. The variables most affected were COPy, with 8 and GRFvt with 7 of the 12 participants out of CON range.

Table 5.3: Phase II results: Mahalanobis distance (T^2) for each amputee subject, in the variables studied.

subject	GRFvt	GRFml	GRFap	COPx	COPy
95% CI	15.9-30.1	20.9-34.2	12.8-21.6	18.2-29.0	23.3-40.6
1	3.41*	24.74	21.88*	25.31	62.02*
2	78.93*	42.90*	41.25*	21.06	45.92*
3	38.87*	52.69*	15.60	50.38*	19.52*
4	53.00*	5.94*	11.79*	14.89*	9.98*
5	29.80	32.48	-1.21*	19.79	35.46
6	12.76*	169.00*	21.92*	19.81	163.00*
7	19.82	18.71*	10.11*	28.42	22.98*
8	4.83*	15.19*	15.76	40.98*	10.46*
9	10.68*	29.35	18.29	14.33*	13.69
10	24.85	21.68	12.64	11.65*	30.36
11	n.a.	n.a.	n.a.	20.22	10.38*
12	n.a.	n.a.	n.a.	18.23	27.20

* Out of 95% Confidence Interval determined by CON. n.a.: not applicable.

The wedge prescription was made based on the results obtained in *Phase I* and *Phase II*, and is summarized in Table 5.4. As shown in Table 5.2, COPy and GRFap were not affected by the wedges, thus their PC score values are not presented (Table 5.4). The second column in Table 5.4 summarizes the variables out of the 95% CI range presented in Table 5.3. The next 5 columns present the PC score values for each subject SL that could be influenced by the wedges, as determined in Table 5.2. If a subject had the PC score values inside 95% CI range, the cell is shown with “n.a.”, to indicate it does not apply in this specific case. Columns 8 to 12 present the possible wedges that could influence positively a determined PC score value. If a wedge affects it in an opposite direction or if the variable for a subject was not out of the 95% CI range, it is shown with “n.w.”, to

indicate that no wedge could positively influence this PC score. The last column summarizes all the wedges that could positively improve SL for each TF subject.

Table 5.4: Phase III - Wedge prescription: variables for each TF with Mahalanobis distance (T^2) out of CON range (column 2); individual PC score values in the relevant PCs from those variables (columns 3 to 7); wedges that influence positively the same variables (columns 8 to 12); total possibilities of wedges that could improve amputees' gait (column 13).

TF	variables ($T^2 > 95\%$ CI)	Individual score in PCs (Phase II)					possible wedges per variable (Phase I)					possibilities
		GRFvt PC2	GRFvt PC3	GRFml PC1	COPx PC1	COPx PC2	GRFvt PC2	GRFvt PC3	GRFml PC1	COPx PC1	COPx PC2	
1	GRFvt	0.08	0.03	n.a.	n.a.	n.a.	1L	n.w.	n.w.	n.w.	n.w.	1L
2	GRFvt, GRFml	-1.1	-1.98	3.6	n.a.	n.a.	1L	1L, 2L, 1M, 2M	1L, 1P, 2P	n.w.	n.w.	all
3	GRFvt, GRFml, COPx	0.3	-0.85	4.43	-8.02	-3.44	1L	1L, 2L, 1M, 2M	1L, 1P, 2P	n.w.	1L	all
4	GRFvt, GRFml, COPx	0.78	-0.15	-0.47	-2	-1.82	2L, 1M	1L, 2L, 1M, 2M	n.w.	n.w.	1L	1L, 2L, 1M, 2M
5	∅	n.a.	n.a.	n.a.	n.a.	n.a.	n.w.	n.w.	n.w.	n.w.	n.w.	n.w.
6	GRFvt, GRFml	0.21	2.44	0.43	n.a.	n.a.	1L	n.w.	1L, 1P, 2P	n.w.	n.w.	1L, 1P, 2P
7	GRFvt, GRFml	0.74	-0.47	0.42	n.a.	n.a.	2L, 1M	n.w.	1L, 1P, 2P	n.w.	n.w.	1L, 1P, 2P
8	GRFvt, GRFml, COPx	-0.11	-0.16	-1.63	-7.38	2.05	1L	1L, 2L, 1M, 2M	n.w.	n.w.	n.w.	all
9	GRFvt, COPx	0.37	-0.2	n.a.	-0.44	-0.34	2L, 1M	1L, 2L, 1M, 2M	n.w.	n.w.	1L	1L, 2L, 1M, 2M
10	COPx	n.a.	n.a.	n.a.	0.3	-3.61	n.w.	n.w.	n.w.	1M, 2M, 1P	1L	1L, 1M, 2M, 1P
11	∅	n.a.	n.a.	n.a.	n.a.	n.a.	n.w.	n.w.	n.w.	n.w.	n.w.	n.w.
12	∅	n.a.	n.a.	n.a.	n.a.	n.a.	n.w.	n.w.	n.w.	n.w.	n.w.	n.w.

n.a.: not applicable; n.w.: no possible wedges.

Application of the prescribed wedges: example of use.

Three TF subjects (numbers 5, 7 and 9) were selected from the 12 analyzed to validate the suitability of the proposed wedges, in order to show an example of application in different cases. The first was selected because apparently no wedge could influence his gait; number 7 was selected due to the differences in GRFml and number 9 was selected due to the differences in GRFvt and COPx which were not found in the other two cases.

Subject 5 already had his parameters inside the CON range (Table 5.5). Nevertheless, the application of the wedges tends to approximate the values to the center of the CON CI distribution. The wedges in GRFvt in CG tend to reduce the PC score values, and this is confirmed by the behavior in this subject. In GRFml and COPx the application of the proposed wedges kept the values inside the normal range. For subject 7, the wedges increased the PC score values, but in 1L and 1P the increase was over the desired 95% CI range. For subject 9, GRFvt PC score values increased with all the wedges, but only 1L and 2L were able to keep the Mahalanobis distance inside CON CI range. In COPx, the values increased inside the CON range.

Table 5.5: Phase IV results: Mahalanobis distance (T^2) calculated before wearing the wedges (see Table 5.3) and with the wedges proposed for SL.

subject	variables	SL (before)	wedges prescribed					
			1L	2L	1M	2M	1P	2P
5	GRFvt	29.80✓	23.31✓	22.41✓	11.27	25.93✓	n.a.	n.a.
	GRFml	32.48✓	31.14✓	n.a.	n.a.	n.a.	33.34✓	27.96✓
	COPx	19.79✓	27.45✓	n.a.	n.a.	n.a.	n.a.	23.50✓
7	GRFvt	n.a.	n.a.	n.a.	n.a.	n.a.	n.a.	n.a.
	GRFml	18.71	42.63	n.a.	n.a.	n.a.	323	27.80✓
	COPx	n.a.	n.a.	n.a.	n.a.	n.a.	n.a.	n.a.
9	GRFvt	10.68	19.64✓	18.36✓	14.67	39.09	n.a.	n.a.
	GRFml	n.a.	n.a.	n.a.	n.a.	n.a.	n.a.	n.a.
	COPx	14.33	25.38✓	n.a.	n.a.	n.a.	n.a.	n.a.

✓: Inside normal range of 95% confidence interval of CON (see Table 5.3); n.a.: not applicable.

5.4 Discussion

The purpose of this study was to prescribe wedges for TF amputees relying upon the influence that six different wedges had on the gait pattern of a group of able-bodied individuals. The prescription of the insoles was based on the scores and in the Mahalanobis distance presented by each TF subject for the first three PCs of the biomechanical patterns analyzed (GRFvt, GRFml, GRFap, COPx and COPy). Using the magnitude of these scores and the potential influence that wedges could have, wedges were prescribed to the TF amputees to be worn inside their SL shoes. Not all TF amputees could be helped with wedges; for some amputees, their PC scores revealed gait deviations similar to those expected to be obtained with the use of wedges, while for other amputees,

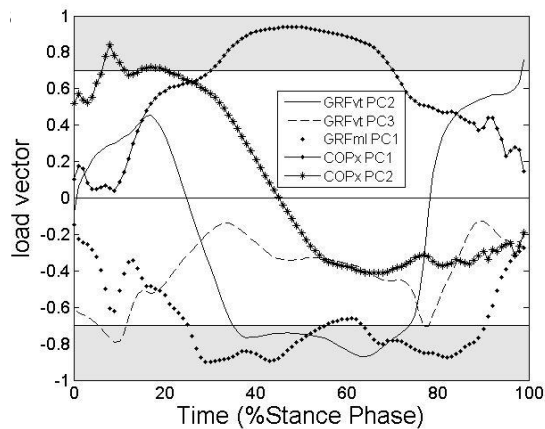
their gait alterations shown to be in variables where the wedges seem not to have influence (Table 5.2). Eight of the twelve TF amputees could have wedges prescribed to alter their gait patterns. GRF patterns from subjects 11 and 12 could not be analyzed, and their analysis was limited to their COP displacement.

For GRF components in CG, the main effects of the wedges are on the vertical and medio-lateral components. GRFvt is the pattern most affected by the use of wedges (Table 5.2). GRFvt PC2 is lower than CON for only 2 of the 10 TF subjects. Previous studies showed that the gait of TF amputees is characterized by an overload of their SL (Nolan & Lees, 2000). In turn this is believed to be causing a high incidence of injuries (Pieter et al., 2009). The purpose of the intervention would be to reduce the scores to unload the SL. As the effect of the wedges 1L, 2L and 1M was to increase GRFvt in PC2, it is likely this would be detrimental to the health of their SL. The smaller GRFvt PC3 score values presented in 5 of the 9 TF amputees (Table 5.4) suggest an earlier heel loading of the sound limb to compensate for the lack of balance provided by their amputated limbs (Figure 5.2a – GRFvt PC3 load vector). Analysis of wedges 1L, 2L, 1M, and 2M indicates that they increase this PC score value. This could imply the sound limb to contact the ground more slowly, as GRFvt PC3 is significant from 8-12%; consequently the amputated limb would have a longer single support phase. This finding should be evaluated by clinicians to determine if this would be beneficial to the patient. The influence of the wedge on this variable is further highlighted by the changes experienced by subject 5 who already had GRFvt inside normal boundaries (Table 5.5).

GRFml differences in CG were found in PC1, in 1L and the posterior conditions, where the three wedges tend to reduce the scores (Table 5.2). Only two of the 10 amputees showed smaller PC score values in GRFml (Subjects 4 and 8 – Table 5.4). Higher PC scores indicate a more lateral migration of the medial-lateral component. The lack of knee active flexion on the amputated limb makes the forward motion more challenging, causing hip abduction to lift AL laterally and progress forward. This provides an explanation for the more lateral support provided by SL. This could also be related to the overload reported in the TF amputees medial knee compartment (Hurwitz et al., 2000; Segal et al., 2006) and the associated high incidence of knee osteoarthritis (Melzer et al., 2001; Pieter et al., 2009). The wedges 1L, 1P and 2P tend to decrease the PC scores and shift the force more medially, helping TF amputees to protect their knee. As can be seen for subject 7, the use of 2P wedge reduces the force peaks (Figure 5.1b).

The medio-lateral displacement of the center of pressure (COPx) in CG is influenced by 1L, both medial wedges and 1P, where 1L tends to increase COPx PC2 while the others tend to decrease COPx PC1. Balmaseda et al. (1988) found that the COPx trajectory is laterally deviated using an Ankle-Foot-Orthosis. Also, Guldemand et al. (2006) found a laterally deviated COP displacement using custom-made foot orthoses.

Four of five amputees who didn't present T^2 values inside the 95% CI, had PC2 scores smaller than those of CG (Table 5.4). As the wedges 1M, 2M and 1P tend to reduce these scores, they would not be useful to them. The other amputee (subject 10) presented PC2 scores higher than CG. As stated before, Erhardt et al. (2008) showed that the COP position is a good predictor of GRF line of action, where a more medial position of COP leads to a greater external knee abduction moment. This would be harmful to the knee since higher abduction moment leads to a higher compression of the medial compartment. This is indicative of a more medially heel contact. 1L wedge tends to displace COPx laterally in this portion of the gait cycle (Figure 5.1a – COPx PC2 load vector and Figure 5.1c), therefore the use of the latter could be beneficial to improve balance at the beginning of SP.



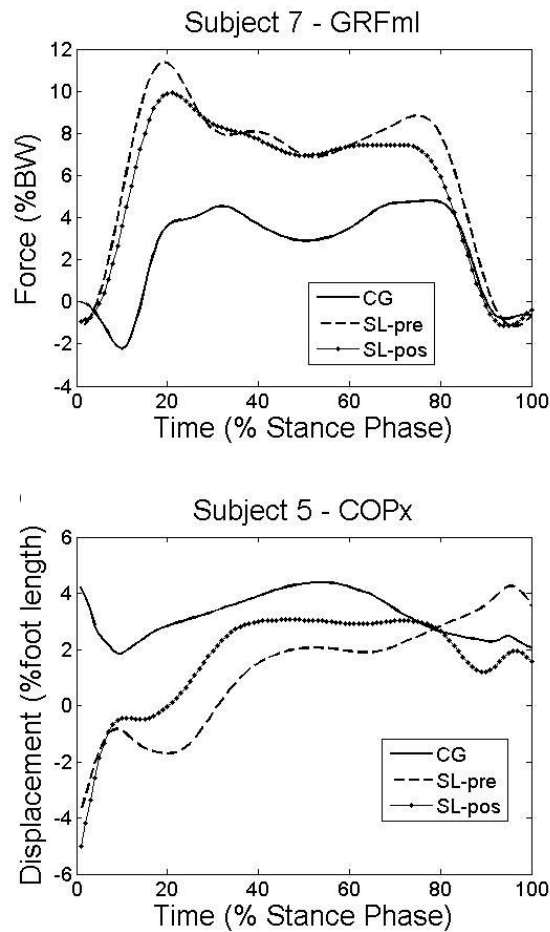


Figure 5.1: a) load vectors from GRFvt PC2, GRFvt PC3, GRFml PC1, COPx PC1 and COPx PC2; the grey area highlights the 0.71 threshold proposed by Knapp & Comrey (1973); b) GRFml waveforms for TF subject before and after intervention; c) COPx waveforms for TF subject before and after intervention.

The gait results using the wedges showed that they could influence positively the participants' gait, as the *Mahalanobis* distance fell within the CON range (Table 5.5). Even subject 5 who already had his parameters within the normal range (Table 5.3), had the *Mahalanobis* distance becoming closer to the central values of CON CI when wearing the wedges. For subject 7, in GRFml, the wedges proposed were 1L, 1P and 2P (Table 5.4 - possibilities), all of them increased the *Mahalanobis* distance as expected, but two of them increased it over the normal range value, so 2P was the best wedge condition. For subject

9, the wedges proposed were 1L for GRFvt and COPx, and 2L, 1M and 2M for GFRvt only (Table 5.4 - possibilities). The three wedges were able to increase the *Mahalanobis* distance, but only 2L showed results approaching the normal range.

The analysis of the subjects who tested the wedges provides evidence for the beneficial influence of wedges in TF gait (Table 5.5), even when the adaptation to the wedge is acute as in this work. More studies should be conducted to evaluate the long term effects of the wedges on gait patterns. Furthermore, the effects of using more than one wedge remain to be elucidated as it could prove to be useful in order to improve more than one gait parameter at a time.

5.5 Conclusion

The application of wedges to the sound limb of TF amputees is successful in improving their gait pattern; it promotes a means to alter the gait variables that are most affected. This could help these individuals to improve their independence and life quality. Our results indicate individual prescription of wedges is critical in order to improve gait successfully. In conclusion, our findings suggest physical therapists and clinicians could prescribe custom made insoles to target patients who suffer from a variety of gait deficiencies by using PCA methods.

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6 ESTIMATION OF THE FORCES GENERATED BY THE THIGH MUSCLES FOR TRANSTIBIAL AMPUTEE GAIT

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ABSTRACT

The forces generated by the muscles with origin on the human femur play a major role in transtibial amputee gait, as they are the most effective of the remaining means that the body can use for propulsion. By estimating the forces generated by the thigh muscles of transtibial amputees, and comparing them to the forces generated by the thigh muscles of normal subjects, it is possible to better estimate the energy output needed from prosthetic devices. The purpose of this paper is to obtain the forces generated by the muscles attached to the human femur of transtibial amputees and compare these with forces obtained from the same muscles in the case of normal subjects. Two transtibial amputees and four normal subjects of similar size to the amputees were investigated. Level ground walking was chosen as the movement to be studied, since it is a common activity that most amputees engage in. Inverse dynamics and a muscle recruitment algorithm (developed by AnyBody Technology®) were used for generating the muscle activation patterns and for computing the muscle forces. The muscle forces were estimated as two sums: one for all posterior muscles, one for the anterior muscles, based on the position of the muscles of the thigh relative to the frontal plane of the human body. The results showed that a significantly higher force is generated by the posterior muscles of the amputees during walking, leading to a general increase of the metabolic costs necessary for one step.

Keywords: Forces; Thigh; Transtibial; Muscles; Gait

6.1 *Introduction*

The forces that the body needs to generate in order to propel itself forward, in the case of a leg with below knee amputation, are generated mostly by the thigh muscles (Sadeghi et al., 2001). Fatigue, discomfort and chronic pain are the main issues that limit the use of most prosthetic devices currently available (Czerniecki, 1996; Ephraim et al., 2005). Of particular importance is the estimation of the compensatory mechanisms that are necessary for body support and forward propulsion due to the loss of the ankle plantar flexors (Centomo et al., 2007; Sadeghi et al., 2001; Sanderson & Martin, 1997). Several studies (Rietman et al., 2002; Ventura et al., 2011) have tried to estimate the advantages

of using energy storage and release prosthetics over using cheaper, less advanced designs. It has been established that the amount of fatigue is dependent on the patient's ability to adapt to the specific prosthetic design used. Therefore, it is clear that the way to improve prosthetic devices lies in understanding the forces that the body has to generate in order to walk properly and reduce the effort necessary for amputee gait. In vivo measurements are a highly useful tool in the estimation of muscle activity. Several authors have successfully used electromyography (EMG) for the estimation of the muscle activity patterns during gait (Isakov et al., 2000; Vaughan CL, 1992; Ventura et al., 2011; D. A. Winter & S. E. Sienko, 1988). However, EMG can only be used as an estimate of the tendency and intensity of muscle group activity, and does not provide information about the forces generated by the muscle groups involved. Recent advances, in the field of biomechanics, have led to the development of highly advanced software packages that are capable of providing data for most of the factors that contribute to human motion. Muscle recruitment algorithms play a great role in the search for obtaining an accurate estimation of the way muscles work together to generate motion (Raikova & Prilutsky, 2001; Rasmussen et al., 2001). By combining muscle recruitment estimation and accurate musculoskeletal models, it is possible to estimate the forces that are needed from the muscles, for the particular motion studied (Fang et al., 2007; Raikova & Prilutsky, 2001).

The development of a numerical musculoskeletal model is a laborious process that involves acquiring data from cadaveric studies and the reconstruction of the desired body structure. Therefore, a highly accurate musculoskeletal model, based on cadaveric data, of the human lower extremity, developed and provided under public domain by the AnyBody® Technology (repository 1.0) was chosen as the basis for this work.

The aim of this study is to modify an existing musculoskeletal model by adding a prosthetic device, and obtain the forces generated by the muscles attached to the human femur by using muscle recruitment algorithms.

6.2 Methods

6.2.1 Subjects

Two unilateral transtibial amputees (amputation on the left side due to vascular complications caused by diabetes) and four non-amputees, control subjects of similar size to the amputees, participated in this study (Table 1). All subjects were free from musculoskeletal disorders and leg pain, were proficient walkers and could walk without the need of assistive devices. The subjects provided informed consent approved by the ethics committee of the rehabilitation centre involved in the study. Each amputee used his own prosthesis (SACH foot) and had at least two years of experience with the prosthetic device. The amputees did not present knee instability. The alignment of each prosthesis was verified prior to testing by a qualified technician.

Table 6.1: Anthropometric and mass data

Subject No.	Amputee (Yes/No)	Height (m)	Mass(Kg)	Prosthetic foot length (m)	Mass of Artificial limb (Kg)	Length of intact tibia (m)	Length of the residual tibia below the knee D_{cut} (m)
1	Yes	1.67	62	0.23	1.08	0.35	0.15
2	No	1.65	64	N/A	N/A	0.36	N/A
3	No	1.60	63	N/A	N/A	0.34	N/A
4	Yes	1.68	78	0.27	1.31	0.38	0.16
5	No	1.70	74	N/A	N/A	0.4	N/A
6	No	1.72	68	N/A	N/A	0.39	N/A

6.2.2 Data collection

The gait data was collected in a motion study laboratory. After a short adaptation, each individual walked three times wearing their own shoes, stepping the right leg over the force plate on an 8m walkway, at a self selected speed. The kinematic data was collected using a Simi Motion SystemTM (SIMI Reality Motion Systems, Unterschleissheim, Germany) of four cameras (operating in a sample frequency of 50Hz) placed on the walls. A BertecTM force plate (model 4060-15, Bertec Corporation, Columbus, USA) operating in a sample frequency of 1000Hz was used to record the ground reaction data. An identical protocol was used for the left leg. The software Acknowledge (Bertec Corporation,

Columbus, USA) was used for the Ground Reaction Force (GRF) data acquisition and the software DvideoTM (UNICAMP, São Paulo, Brazil) was used for the video manipulation and for obtaining the 3D marker positions. The markers were placed in anatomical points according to the “Helen Hayes” marker configuration (Vaughan CL, 1992). Markers were placed on the prosthesis in a matching configuration, with the ankle marker placed at the same level as the maleollus of the intact limb. The force plate data was exported as 3 major force and moment directions.

6.2.3 *Model considerations*

The musculoskeletal human model used for this study is composed of four rigid bodies (representing the lower limb and the pelvis). All the muscles with origin in the tibia and insertion in the foot were removed. The muscles that act on the knee joint, except for the Gastrocnemius muscle, were kept in the model. The hip joint was modelled as a spherical joint with three rotational degrees of freedom (DoF). The knee joint was simulated as a hinge joint with one rotational DoF. The ankle joint had two rotational DoF according to a *cardan* joint (to simulate the SACH foot). A number of 158 Hill Type muscles were attached to the leg in the original model, and 117 muscles remained in the model after the modifications. The data used by AnyBody® Technology for the development of the musculoskeletal model was based on a cadaveric dataset provided by the University of Twente (K. Horsman, 2007).

The selective thigh muscle atrophy present in the case of transtibial amputees was considered, and the muscular properties of the initial model were modified according to the data obtained by Schmalz et al. (2001).

Properly scaled human models were developed for each one of the six participants, based on each individual's anthropometric data (Table 6.1), one for the right leg, one for the left leg, totalling twelve body models. The body fat percentage of each individual was estimated by using the age and sex specific prediction formula for individuals over 15 years of age developed by Deurenberg et al. (1990).

For the transtibial amputee model, the initial musculoskeletal model (GaitUniMiamiTDRightLeg) was modified by dividing the tibia in two separate rigid bodies. The cutting plane was set perpendicular to the longitudinal axis of the tibia and at a set distance (D_{cut}) from the knee joint. The new rigid bodies were articulated with a rigid joint (simulating the contact between the prosthetic pylon and the prosthetic socket) placed at

the intersection between the longitudinal axis of the tibia and the cutting plane. Two local coordinate systems, with axes parallel to those of the local coordinate system of the initial rigid body representing the tibia, were defined for each resulting rigid body. The origins of the new local coordinate systems were placed in the middle of each resulting rigid body, along the axis. The centres of mass for the resulting rigid bodies were defined at the new local coordinate systems. The distance from the knee to the cutting plane was controlled by the length variable l_{cut} . The length variable describes how much of the percentile of the total tibia length l_{Tib} is occupied by the residual limb (i.e. for subject 1, with a residual limb representing 43% of the residual tibia, $l_{cut}=0.43$).

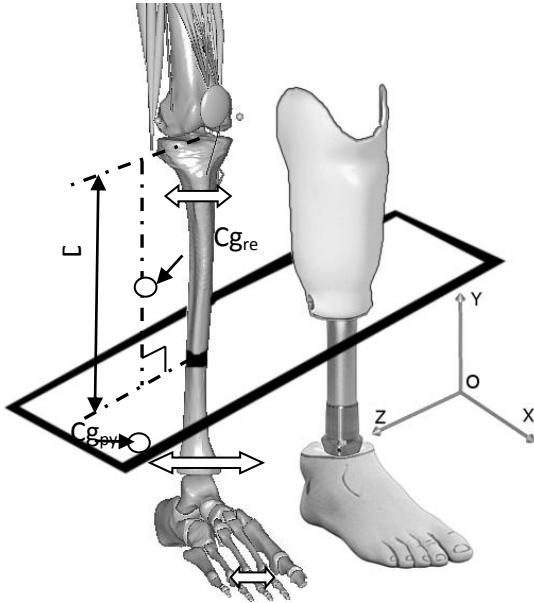


Figure 6.1: The corresponding structural modifications performed on the original model, for the simulation of a transtibial prosthetic device. C_{gpyl} is the centre of mass for the rigid body representing the pylon. C_{gres} is the centre of mass for the rigid body representing the pylon. C_{gres} is the centre of mass for the rigid body representing the socket attached to the residual limb.

The mass of the prosthetic pylon (m_{pyl}) was determined based on the length variable and the set value of 0.31Kg for a 0.42m long pylon. The mass of the residual limb (m_{res}) was determined in a similar manner, as a percentage of the initial mass of the tibia (m_{Tib}).

Equation 6. 1
$$m_{res} = m_{Tib} \times l_{cut}$$

Equation 6. 2
$$m_{pyl} = l_{Tib} (1 - l_{cut}) \frac{0.31}{0.42}$$

The mass of the foot was changed to that of a standard SACH foot (ottobock.com), based on the size of the amputee's intact foot.

The inertia tensors for the residual limb (J^{res}) were calculated considering the residual limb as a cylinder with a radius equal to the initial radius of the initial shank (R_{res}).

Equation 6. 3
$$I_x^{res} = I_z^{res} = \frac{1}{12} m_{res} (3R_{res}^2 + l_{Tib}^2)$$

Equation 6. 4
$$I_y^{res} = \frac{1}{2} m_{res} R_{res}^2$$

The inertia tensors for the pylon (J^{pyl}) were calculated considering the pylon as a tube with an external diameter (D_{ext}) of 0.30m and an internal diameter (D_{int}) of 0.028m.

Equation 6. 5
$$I_x^{pyl} = I_z^{pyl} = \frac{1}{12} m_{pyl} \left[\frac{3}{4} (D_{ext}^2 + D_{int}^2) + [l_{Tib} (1 - l_{cut})]^2 \right]$$

Equation 6. 6
$$I_y^{res} = \frac{1}{8} m_{pyl} [D_{ext}^2 + D_{int}^2]$$

6.2.4 Data analysis

To estimate the forces generated by the thigh muscles, the muscles were grouped in two categories: posterior muscles and anterior muscles (based on the position of the muscles on the thigh relative to the frontal plane of the human body).

The posterior muscle group included the following muscles: Biceps Femoris, Gemellus, Popliteus, Gracilis, Semitendinosus, Semimembranosus, Adductor Magnus, Gluteus Maximus, Quadratus Femoris, Obturator Externus, Obturator Internus, and Piriformis.

The anterior muscle group included the following muscles: Rectus Femoris, Vastus Lateralis, Vastus Medialis, Vastus Intermedius, Sartorius, Tensor Fasciae Latae,

Adductor Longus, Adductor Brevis, Pectineus, Gluteus Medius, Gluteus Minimus and Iliacus.

The muscle forces were calculated using the polynomial muscle recruitment algorithm of AnyBody® software (p=3). The polynomial criterion was chosen considering the limitations of the other criteria available (anybodytech.com) and the fact that it offers reasonable synergy between the muscles and produces results that display similarities with experimental data.

The original un-modified model was used to estimate the muscle forces generated by the thighs of the control group individuals. In addition to the original code, in this case, a routine, that extracted the posterior and anterior muscle forces, was developed.

The metabolic energy consumption of the thigh muscles was calculated for both the amputees and the control group, assuming an efficiency of 25% for concentric muscle work and -120% for eccentric work (using a pre-built feature of the AnyBody® Software). The estimation of the efficiency, as provided by AnyBody® Technology, is in agreement with the ones found in the literature (Bolstad & Ersland, 1978; Hawkins & Molé, 1997).

The metabolic energy consumption of the muscles was calculated considering that $1\text{cal} = 4.184\text{J}$. The energy consumption was calculated considering the metabolic power consumption of the muscle-tendon unit ($P_{met}[\text{W}]$), the time period which corresponds to the readings of power ($t_{power}[\text{s}]$) and the number of readings corresponding to the time period ($N_{readings}$). If the data is normalized by time (% stance phase) then $N_{readings}$ is (usually) 100.

$$\text{Equation 6. 7} \quad E = \frac{P_{met} * t_{power}}{4.184 * N_{readings}}$$

Total energy consumption during stance was determined for the thigh muscles of each leg (E_{stance}). For the control group, inter-subject total energy consumption averages were determined for subjects 2 and 3 ($E_{stance}^{control 23}$), and subjects 5 and 6 respectively ($E_{stance}^{control 56}$).

$$\text{Equation 6. 8} \quad E_{stance} = \sum_{i=1}^{100} E_{stance}^{anterior} + \sum_{i=1}^{100} E_{stance}^{posterior}$$

$$\text{Equation 6. 9} \quad E_{\text{stance}}^{\text{control } 23} = \frac{E_{\text{stance}}^{\text{control } 2} + E_{\text{stance}}^{\text{control } 3}}{2}$$

$$\text{Equation 6. 10} \quad E_{\text{stance}}^{\text{control } 56} = \frac{E_{\text{stance}}^{\text{control } 5} + E_{\text{stance}}^{\text{control } 6}}{2}$$

6.3 Results

6.3.1 Muscle forces prediction

In the case of the first amputee (Table 6.1, subject 1), the posterior forces had maximum value during the first half of stance phase ($F_{\text{posterior}} = 2.17 \text{ BW}$, at 30% of stance phase) and decreased gradually towards single leg support. A second increase was noticed during toe-off ($F_{\text{posterior}} = 1.41 \text{ BW}$, at 75% of stance phase). The anterior forces displayed a similar tendency to the posterior forces, reaching maximum value during the first half of stance phase ($F_{\text{anterior}} = 3.01 \text{ BW}$, at 33% of stance phase). The second peak of the anterior force displayed a local maximum value at the same time ($F_{\text{anterior}} = 2.85 \text{ BW}$, at 75% of stance phase) that the anterior maximum value was obtained (Fig. 6.2a).

A similar behaviour also was noticed in the case of the second amputee (Table 6.1, subject 4). The posterior forces had maximum value during the first half of stance phase ($F_{\text{posterior}} = 1.75 \text{ BW}$, at 46% of stance phase), and a local maximum value during toe off ($F_{\text{posterior}} = 1.13 \text{ BW}$, at 77% of stance phase). The values obtained for the anterior forces had similar tendencies to the values obtained for the posterior forces; a maximum value was obtained during the stance phase ($F_{\text{anterior}} = 2.37 \text{ BW}$, at 46% of stance phase), and a second, local maximum value was obtained during toe off ($F_{\text{anterior}} = 1.65 \text{ BW}$, at 77% of stance phase) (Fig. 6.2 b).

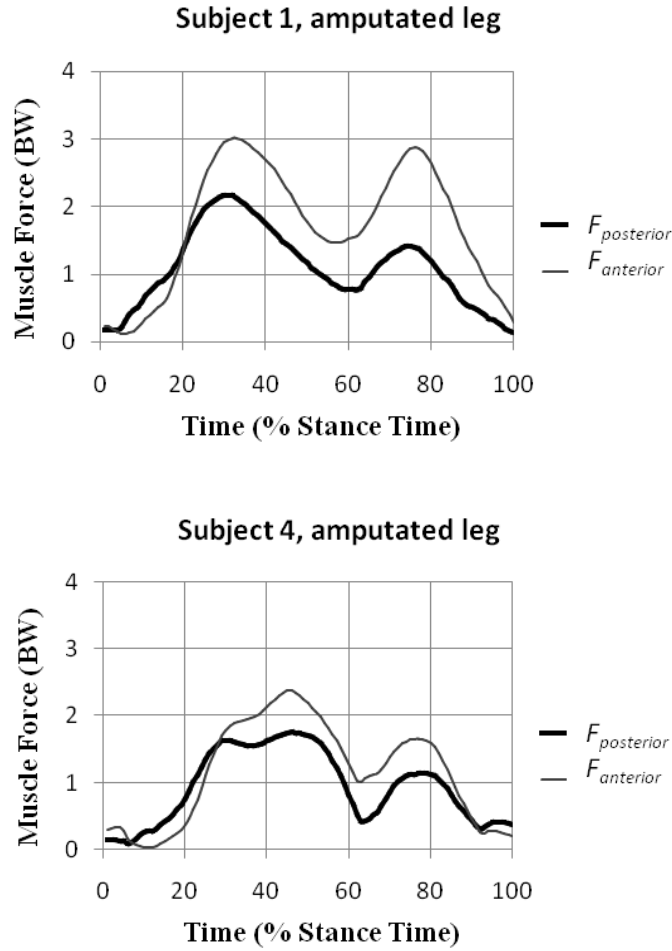


Figure 6.2: Anterior and posterior sums of forces, generated by the muscles attached of the thigh of subject 1 (a) and amputee subject 4 (b), for the leg with amputation.

For the control group, lower values (when comparing to the leg with transtibial amputation), of the force generated by the posterior muscles were obtained. The maximum value of the average muscle force (control group) generated by the posterior muscles was $F_{posterior} = 1.14 \text{ BW}$ (Fig. 6.3a). A lower value for the values of the muscle forces generated was also noticed for the normal leg of the amputees, where a maximum value of $F_{posterior} = 1.02 \text{ BW}$ was found for the average posterior muscle force (Fig. 6.3b). The maximum value for the average muscle force (control group) generated by the anterior muscles was $F_{anterior} = 2.57 \text{ BW}$ (Fig. 6.3a). For the normal leg of the amputees,

the maximum value of the average force generated by the anterior muscles was $F_{anterior} = 2.87 \text{ BW}$ (Fig. 6.3b).

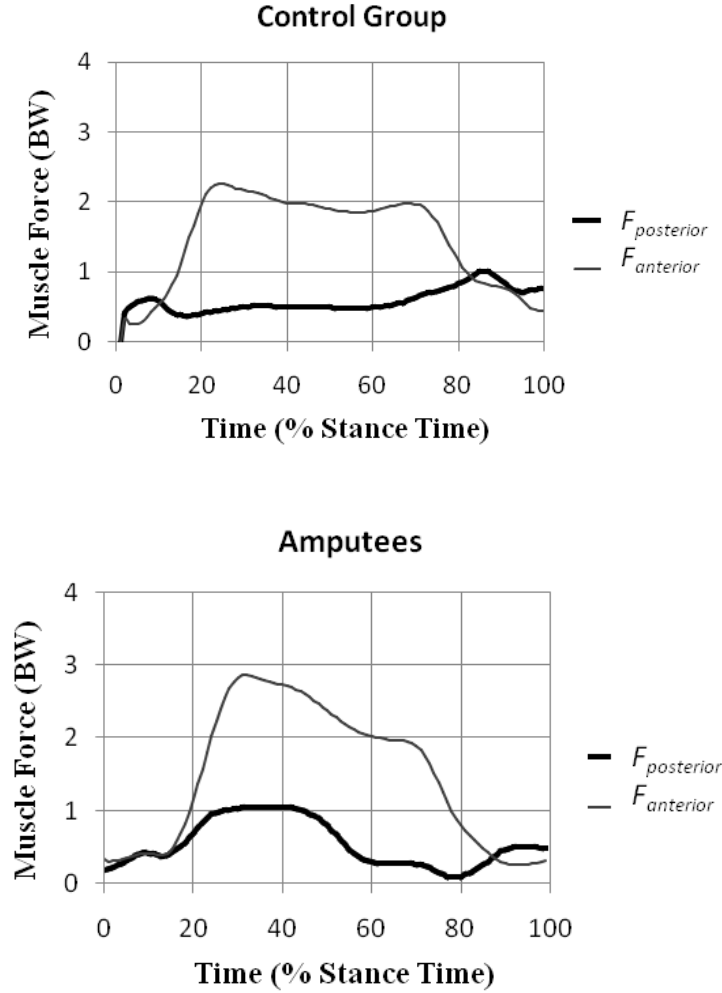


Figure 6.3: Average value of posterior and of anterior muscle force sums generated by the thigh during stance phase for the control group (a) and for the intact leg of the amputees (b).

6.3.2 Muscle energy consumption

The total energy consumption (of the thigh muscles) was $E_{stance}^{amp 1} = 30.49 \text{ cal}$ for subject 1 (Fig. 6.4 a) and $E_{stance}^{amp 4} = 39.78 \text{ cal}$ for subject 4 (Fig. 6.4b). The total energy consumption of the averages from subject 2 and 3 was $E_{stance}^{control 23} = 9.64 \text{ cal}$ (Fig. 6.4a), and that of the averages from subject 5 and 6 was $E_{stance}^{control 56} = 15.04 \text{ cal}$ (Fig. 6.4b).

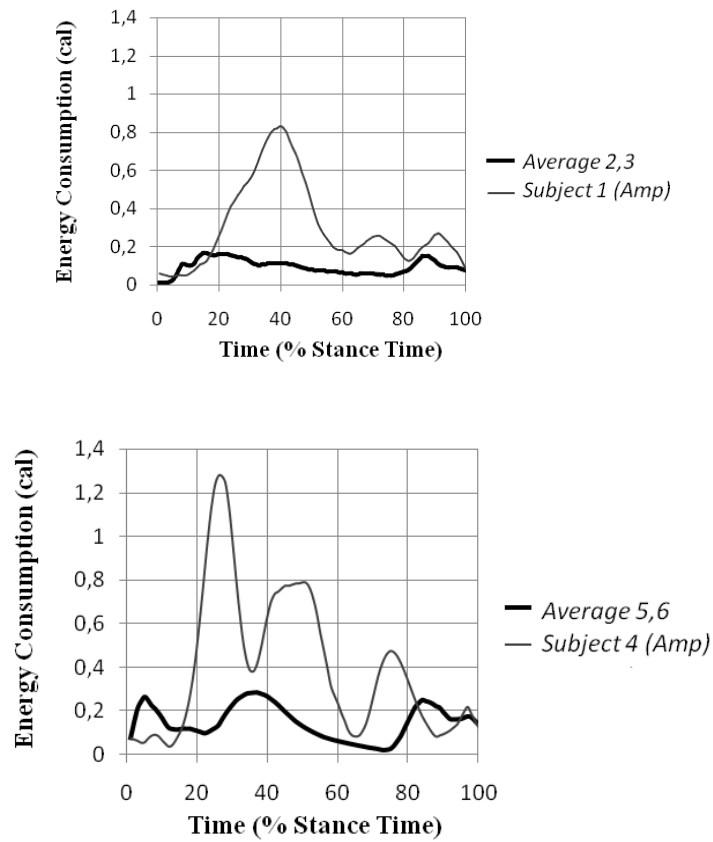


Figure 6.4: Energy consumption necessary for the generation of the muscle forces (for the muscles of the thigh, during stance phase (a) Values from the residual limb of subject 1 overlaid on top of inter-subject average values from the control group (subjects 2, 3). (b) Values from the residual limb of subject 4 overlaid on top of inter-subject average values from the control group (subjects 5, 6).

6.3.3 Resultant contact force

The resultant force at the contact between the pylon and the socket of the prosthetic device had a maximum value of $1\text{BW} \pm 0.01$ for both amputees investigated (Fig. 6.5).

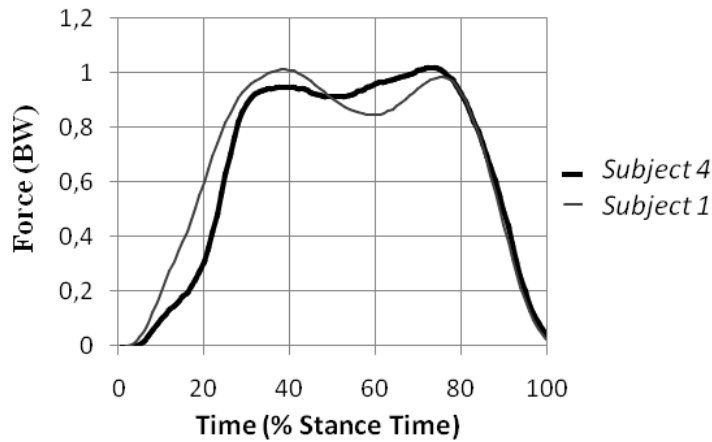


Figure 6.5: :Calculated resultant force at the contact between the pylon and the socket of the prosthetic device (axial force on the socket connector).

6.3.4 Validation

Numerical simulation is one of the best methods available for obtaining muscle forces in a quantitative way. Due to the fact that EMG can have the same tendency as the muscle activity, the comparison of the muscle activity of the muscles in the model with EMG data from literature (Winter and Sienko, 1988), was chosen as a method for validation. The muscular activity, as calculated by AnyBody® software, is defined as the muscle active state in fractions of maximum voluntary contraction (anybodytech.com). The muscular activity (of the thigh muscles) obtained for the amputees was higher than the one obtained for the control group. The activity of the Rectus Femoris muscle (of the residual limbs) was approximately three times larger ($\text{Max. Activity}_{\text{RectusFemoris Amputee}}^{\text{RectusFemoris}} = 0.48$) than that of the control group ($\text{Max. Activity}_{\text{RectusFemoris ControlGroup}}^{\text{RectusFemoris}} = 0.18$). A maximum value four times larger than that obtained for the control group was obtained for the Gluteus Maximus muscle ($\text{Max. Activity}_{\text{GluteusMaximus Amputee}}^{\text{GluteusMaximus}} = 0.27$; $\text{Max. Activity}_{\text{GluteusMaximus ControlGroup}}^{\text{GluteusMaximus}} = 0.07$). The average muscle activity for five muscles (Rectus Femoris, Vastus Lateralis, Semitendinosus, Biceps Femoris and Gluteus Maximus) of the legs with amputation and of the legs of the control group is shown in figure 6.6.

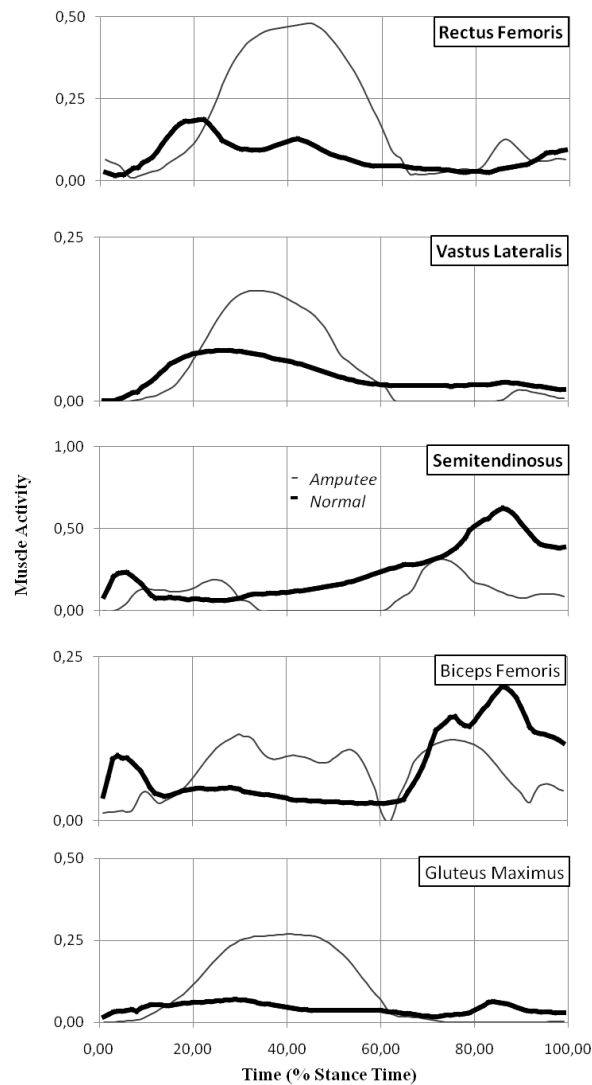


Figure 6.6: Average activity from five residual muscles, for the legs with transtibial amputation, overlaid over the average activity of the same five muscles, for the control group.

6.4 Discussion

In this study, an existing musculoskeletal model was modified in order to determine the forces generated by the muscles attached to the thigh during gait. The distributions of the EMG recorded by Winter and Sienko (1988) and the muscle activity generated by the amputee model showed similar tendencies. For the amputees, the hip extensors had an increased activity during early and mid-stance, when compared to the control group, indicating a compensation mechanism for the missing plantar flexors. The

knee extensors also had a higher activity, indicating an increased need for the control of the knee during the motion (Fig. 6.6). This is consistent with the observations on EMG, for the case of transtibial amputees, of Winter and Sienko (1988).

The forces generated by the muscles positioned (on the thigh) posterior to the frontal plane were visibly lower in the case of the control group, while in the case of the amputees they had similar magnitudes to the forces generated by the muscles positioned anterior to the frontal plane. This is a consequence of the co-contraction of the knee extensors against the hamstrings, a mechanism that is required to achieve balance.

The energy cost necessary for the motion (for the thigh muscles) was greater in the case of the amputees (Fig. 6.4). This is a consequence of the higher forces that the muscles in the posterior group are required to generate. The value of the energy consumption decreases gradually to values similar to those found for the control group during terminal stance.

The resultant contact force between the pylon and the socket is in agreement with values found in the literature (Sanders et al., 1993) and can be useful in studying the behaviour of a prosthetic device in a finite element environment.

The limitations of this study were related to the musculoskeletal model and the inverse dynamics approach. The visco-elastic behaviour of the connection between the prosthesis and the residual limb was neglected (use of a rigid connection). The forces generated by the muscles were dependant on the accuracy of the kinematic and ground reaction recorded signals, the action lines of the muscles and the muscle recruitment algorithm (Raikova & Prilutsky, 2001; Rasmussen et al., 2001). All the obtained muscle activities were proportional to those found using EMG, but displayed some differences. These differences can be attributed to the limitations of the model and to the particularities of the subjects involved in this study.

Considering the limitations of the model, the data reported in this study can be useful in understanding the compensatory mechanisms present at the thigh during gait. A similar model can be used to quantify the benefits of a prosthetic device, by adding the energy storage and return characteristics and observing the effect on the muscle forces obtained.

6.5 Conclusion

The transtibial amputees use specific mechanisms, for body support and movement, to compensate for the limb loss. This research describes a model that attempts to capture the muscular forces involved during transtibial amputee gait and manages to provide data that is consistent to previously reported findings. The data presented reveals the differences between transtibial amputees and control group subjects, in regards of the forces generated by the muscles. An estimate of the increased energy consumption is quantified and presented.

6.6 Acknowledgements

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7 CONCLUSION

The general purpose of this thesis was to develop a biomechanical model for prescribing wedges for TF amputees based on gait data using a PCA approach. The self-selected speed level-walking was assessed during the normal gait and six experimental conditions (wedges varying on shape, height and position). Data from inverse dynamics (lower limb net joint moments), kinematic (lower limb joints range of motion), kinetic (GRF components) and baropodometric (COP displacement) were analyzed by two approaches: the first, named classical approach, discrete values are obtained from the waveforms and, the second, the Principal Component Analysis was applied.

PCA analysis was shown to be better for discriminating gait alterations as consequence of wedges influence than the classical one. By GRF and COP displacement parameters the influence of the wedges on gait pattern was clearly identified. Considering that these variables are less complex to obtain than muscle net moments for example, we decided to prescribe the wedges based on these parameters.

For the wedges prescription, the method proposed based on the PCs and *Mahalanobis* distance from the able bodied participants was adequate for the application to TF subjects. The posterior experimental analysis of the wedges in three TF individuals showed improvements in gait parameters. Our results indicate that individual prescription of wedges may be a useful tool for increasing the quality of the gait. Consequently, it may help to improve the independence and life quality of TF amputees. In conclusion, our findings suggest physical therapists and clinicians could prescribe custom made insoles for TF amputees. Furthermore, we believe that the developed model may be applied for any impairment gait's population.

Future work:

- To analyze the influence of the wedges in a higher number of subjects;
- To use the developed model for verifying the effect of different therapeutic approaches on gait performance;
- To apply the model on different impairment gait's populations.
- To evaluate the wedges influence on gait parameters in a long term adaptation;
- To evaluate the subjects comfort wearing the new devices;

- To compare the wedge prescription by GRF and COP parameters with other biomechanical parameters;
- To analyze the applicability of this protocol by medical doctors and physical therapists in the clinical practice.

8 APPENDIX

Appendix I: Considerations on the 3D Inverse Dynamics model

1) Calculation of the net joint forces and moments:

The net joint forces and moments in the lower limb were obtained using the inverse dynamics technique. This approach is based on the determination of the internal contact forces from the measurement of the contact and non-contact external forces, the limb segment accelerations and masses.

1.1 Model:

The model used in this thesis for the right lower limb is a 3D model based on the model proposed by Vaughn (1992). The general characteristics are:

a) Model segments:

The lower limb was considered as a mechanical system composed by three rigid segments, representing the thigh, calf and foot, connected by the knee and ankle joints. The joints are considered frictionless between the extremities. The rest of the body (head, trunk and upper limbs) are not represented explicitly, only its effect on the hip joint are considered.

b) Markers placement:

The joint positions was estimated by the use of videogrammetry, where external markers were placed in points of interest to allow the prediction of internal anatomical places, like hip ankle and knee joint centers.

The markers were placed in the right limb as follows:

- 1) 5th metatarsal head;
- 2) Calcaneous
- 3) Lateral malleolus;
- 4) Calf lateral wand;
- 5) Knee lateral epicondyle;
- 6) Thigh lateral wand;
- 7) Right Antero Superior Iliac Spine (ASIS)
- 8) Left ASIS
- 9) Sacrum

The anatomical points that this markers represent are named as p..., for example the calcaneus external marker is pCalcaneus.

c) Calculation of the joint centers:

The joint centers were calculated based on the external points, using the following sequence:

1. Select three markers for the segment of interest.
2. Create an orthogonal uvw reference system based on these three markers (Figure 1).
3. Use prediction equations based on anthropometric measurements (Table 1) and the uvw reference system to estimate the joint centre positions.

For the foot, the markers 1, 2, and 3 and the anthropometrical measurements A8 to A11 (Table 1) were used and the joint center of the ankle (pAnkle) and toe (pToe) were calculated as:

$$\begin{aligned} pAnkle &= pLateral\ malleolus + 0.016(A8)uFoot + 0.392(A9)vFoot + 0.478(A11)wFoot \\ pToe &= pLateral\ malleolus + 0.742(A8)uFoot + 1.074(A9)vFoot - 0.187(A10)wFoot \end{aligned}$$

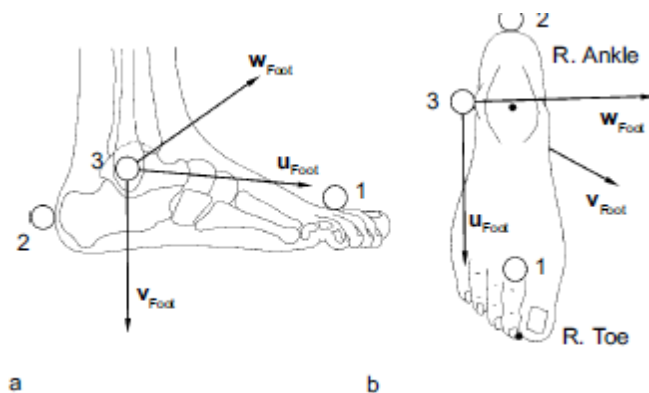
For the knee joint center (pKnee), the markers 3, 4 and 5 and the anthropometrical measurement A7 (Table 1) were used and the joint center was computed as:

$$pKnee = pFemoral\ epicondyle + 0.000(A7)uCalf + 0.000(A7)vCalf + 0.500(A7)wCalf$$

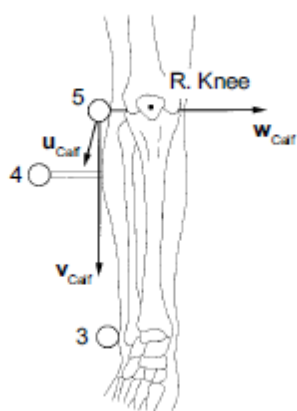
For the hip joint center (pHip) the markers used were 7, 8 and 9 and the anthropometrical measurement A2 (Table 1) as follows:

$$pHip = pSacrum + 0.598(A2)uPelvis + 0.344(A2)vPelvis - 0.29(A2)wPelvis$$

Toe and ankle



Knee



Hip

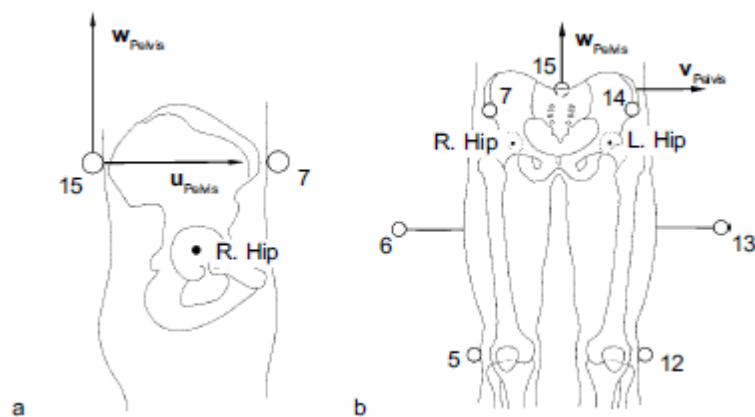


Figure 1: Joint center calculations: determination of the uwv system and position of the external markers. Adapted from Vaughn (1992).

d) Determination of segments masses and moments of inertia:

These parameters are defined according to the protocol proposed by Vaughn (1992), where adjustment coefficients are applied in the measurements proposed by Chandler (1975). The coefficients consider individual differences in shapes and lengths, implying anthropometrical measurements should be taken (Table 1). For the prosthetic limb, the data obtained from Loss (2001), was applied to TF subjects in this study (Tables 1, 2 and 3).

Table 1: Anthropometric Parameters measured in each subject

Parameter	Name
A1	Total body mass
A2	Anterior Superior Iliac Spine (ASIS) breadth
A3	Thigh length
A4	Midthigh circumference
A5	Calf length
A6	Calf circumference
A7	Knee diameter
A8	Foot length
A9	Malleolus height
A10	Malleolus width
A11	Foot breadth
A12	Prosthetic limb weight
A13	Distance from tibial trochanter and knee prosthetic rotation center
A14	Distance from knee and ankle prosthetic rotation centers
A15	Distance from ankle prosthetic rotation centers and end of foot
A16	Socket circumference,
A17	Calf circumference, prosthetic leg
A18	Malleolus height, prosthetic foot
A19	Malleolus width, prosthetic foot
A20	Foot breadth, prosthetic foot

Table 2: mass estimation for each segment

Segment	Mass estimation
Thigh	$(0.1032) * A1 + (12.76) * A3 * A4 * A4 - 1.023$
Calf	$(0.0226) * A1 + (31.33) * A5 * A6 * A6 + 0.016;$
Foot	$(0.0083) * A1 + (254.5) * A8 * A9 * A10 - 0.065;$
Prosthetic thigh	$(0.1032) * A1 + (12.76) * A13 * A16 * A16 - 1.023$
Prosthetic calf	$(0.0226) * A1 + (31.33) * A14 * A17 * A17 + 0.016;$
Prosthetic foot	$(0.0083) * A1 + (254.5) * A15 * A18 * A19 - 0.065;$

Table 3: Moment of inertia for the right leg and prosthetic leg. The masses referred in the Prosthetic leg (Thigh mass, calf mass and foot mass) are taken from Table 2 and refer to the prosthetic limb.

	Segment	Moments of Inertia (I) estimation
Right leg	I_FlxExt.R.Thigh	$= 0.00762 * A1 * (A3 * A3 + 0.076 * A4 * A4) + 0.01153;$
	I_AbdAdd.R.Thigh	$= 0.00726 * A1 * (A3 * A3 + 0.076 * A4 * A4) + 0.01186;$
	I_IntExt.R.Thigh	$= 0.00151 * A1 * A4 * A4 + 0.00305;$
	I_FlxExt.R.Calf	$= 0.00347 * A1 * (A5 * A5 + 0.076 * A6 * A6) + 0.00511;$
	I_AbdAdd.R.Calf	$= 0.00387 * A1 * (A5 * A5 + 0.076 * A6 * A6) + 0.00138;$
	I_IntExt.R.Calf	$= 0.00041 * A1 * A6 * A6 + 0.00012;$
	I_FlxExt.R.Foot	$= 0.00023 * A1 * (4 * A9 * A9 + 3 * A8 * A8) + 0.00022;$
	I_AbdAdd.R.Foot	$= 0.00021 * A1 * (4 * A11 * A11 + 3 * A8 * A8) + 0.00067;$
	I_IntExt.R.Foot	$= 0.00141 * A1 * (A9 * A9 + A11 * A11) - 0.00008;$
Prosthetic leg	I_FlxExt.P.Thigh	$= 1/12 * (\text{thigh mass}) * (A13^2) + 0.076 * (A16^2)$
	I_AbdAdd.P.Thigh	$= 1/12 * (\text{thigh mass}) * (A13^2) + 0.076 * (A16^2)$
	I_IntExt.P.Thigh	$= 1/8 * (\pi^2) * (\text{thigh mass}) * (A16^2)$
	I_FlxExt.P.Calf	$= 1/12 * (\text{calf mass}) * (A14^2) + 0.076 * (A17^2)$
	I_AbdAdd.P.Calf	$= 1/12 * (\text{calf mass}) * (A14^2) + 0.076 * (A17^2)$
	I_IntExt.P.Calf	$= 1/8 * (\pi^2) * (\text{calf mass}) * (A17^2)$
	I_FlxExt.R.Foot	$= 1/12 * (\text{foot mass}) * (A15^2) + 0.076 * (A20^2)$
	I_AbdAdd.R.Foot	$= 1/12 * (\text{foot mass}) * (A15^2) + 0.076 * (A20^2)$
	I_IntExt.R.Foot	$= 1/8 * (\pi^2) * (\text{foot mass})$

e) Solidification process:

As the markers are placed over the skin and, especially the points 4 and 6 that are wands placed in the middle of the segments, the movement associated to the displacement of the skin points leads to errors in measurements that are added to the real displacement of the segment.

To minimize this error, the solidification process proposed by Cheze et al. (1995) was applied. In broad terms, one creates a triangle for each segment at each time T_i , and removes the triangles that are further from the mean triangle. Then one forces the triangles at each time T_i to have the same distance between points as the mean triangle. In this way the segment lengths remain constant along the cycle. This procedure allows a better estimation of the inverse dynamics model.

f) Segments orientation:

This is done by embedding a reference system (xyz) in each segment that will define how each segment is positioned relative to the global reference frame XYZ. The location of each xyz reference frame is at the segments center of mass (CM).

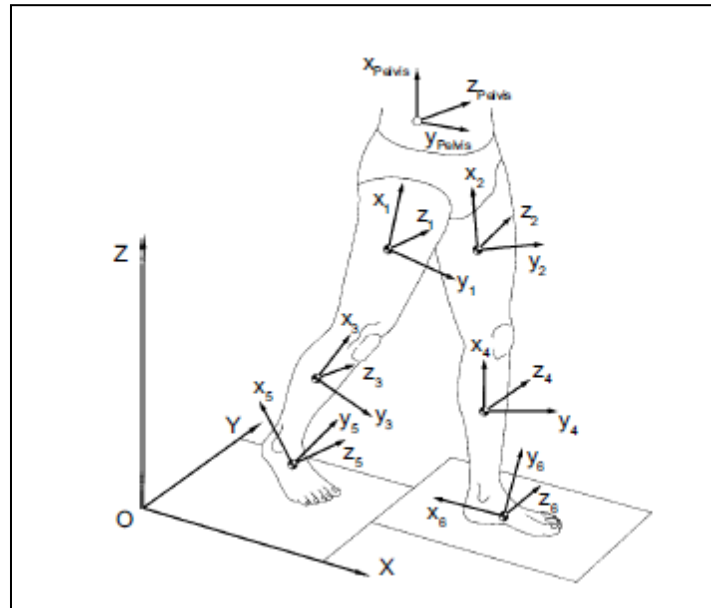


Figure 2: xyz coordinates for each segment

g) Determination of the segments CM:

The CM of each segment is derived by the equations bellow.

$$p_{\text{Thigh.CM}} = p_{\text{Hip}} + 0.39 (p_{\text{Knee}} - p_{\text{Hip}})$$

$$p_{\text{Calf.CM}} = p_{\text{Knee}} + 0.42 (p_{\text{Ankle}} - p_{\text{Knee}})$$

$$p_{\text{Foot.CM}} = p_{\text{Heel}} + 0.44 (p_{\text{Toe}} - p_{\text{Heel}})$$

h) Segments angles:

These parameters are obtained using the methods proposed by Chao (1980) and Grood and Suntay (1983) for defining the anatomical joint angles.

The following conventions are applied to the hip, knee and ankle joints:

k_{Proximal} = flexion/extension axis.

i_{Distal} = internal/external rotation axis.

l_{Joint} = abduction/adduction axis.

k_{Proximal} and i_{Distal} are defined as show in Figures 4a to 4c and l_{Joint} is the internal product from k_{Proximal} and i_{Distal} .

The plane angle definitions are:

α =flexion/extension angle.

β =abduction/adduction angle.

γ =internal/external rotation angle.

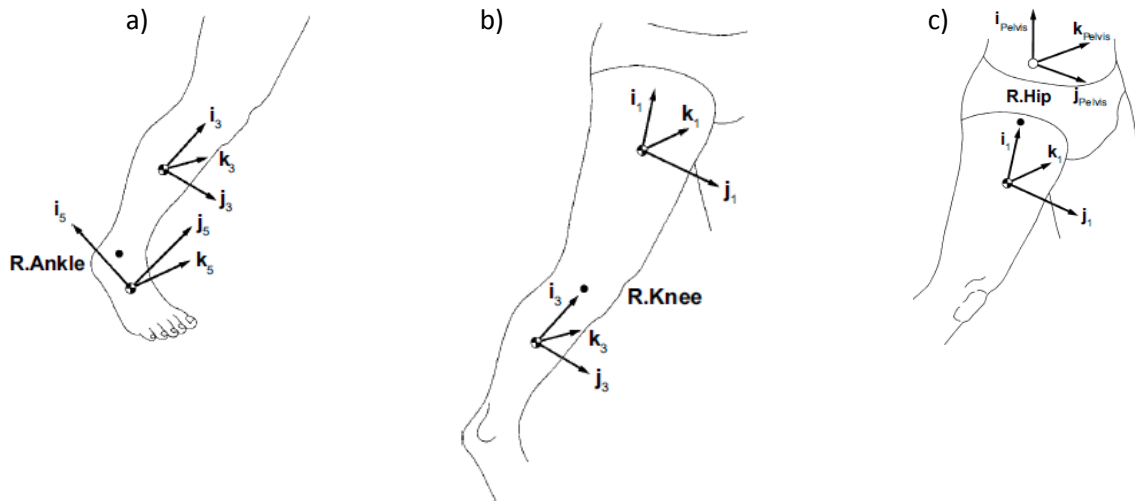


Figure 4: ijk reference systems for the ankle (4a), knee (4b) and hip (4c).

The joints angles are calculated according to the equations:

$$\alpha_{\text{Hip}} = \sin^{-1}[l_{\text{Hip}} \cdot i_{\text{Pelvis}}]$$

$$\beta_{\text{Hip}} = \sin^{-1}[k_{\text{Pelvis}} \cdot i_1]$$

$$\gamma_{\text{Hip}} = -\sin^{-1}[l_{\text{Hip}} \cdot k_1]$$

$$\alpha_{Knee} = -\sin^{-1}[l_{Knee} \cdot i_1]$$

$$\beta_{Knee} = \sin^{-1}[k_1 \cdot i_3]$$

$$\gamma_{Knee} = -\sin^{-1}[l_{Knee} \cdot k_3]$$

$$\alpha_{Ankle} = \sin^{-1}[l_{Ankle} \cdot j_3]$$

$$\beta_{Ankle} = \sin^{-1}[k_3 \cdot i_5]$$

$$\gamma_{Ankle} = -\sin^{-1}[l_{Ankle} \cdot k_5]$$

i) Segment orientation in the global reference frame:

The segment orientations were based in the Euler angle rotations which in turn were based in the formalism of the classical mechanics texts by Synge and Griffith (1959) and Goldstein (1965). They are performed in the following order:

(a) ϕ about the K axis of the global reference frame,

(b) θ about the line of nodes, and

(c) ψ about the k axis of the segment,

where the line of nodes is a unit vector defined as the external product from K and k.

The following rotations are applied to the hip, knee and ankle joints:

$$\phi = \sin^{-1}[(I \times L) \cdot K]$$

$$\theta = \sin^{-1}[(K \times k) \cdot L]$$

$$\psi = \sin^{-1}[(L \times i) \cdot k]$$

j) Segments angular velocities and accelerations:

The segment angular velocities were obtained from the Euler angles as follows:

$$\omega_{segment.x} = \dot{\phi} \sin \theta \sin \varphi + \dot{\theta} \cos \varphi$$

$$\omega_{\text{segment.y}} = \dot{\theta} \sin \theta \sin \varphi + \dot{\phi} \sin \varphi$$

$$\omega_{\text{segment.z}} = \dot{\theta} \cos \theta + \dot{\phi}$$

where the segment angular velocities ω are given relative to the segment-based reference frame xyz, and the dot above the Euler angles indicates the first derivative with respect to time.

The accelerations are obtained using the following equations:

$$\dot{\omega}_{\text{segment.x}} = \ddot{\theta} \sin \theta \sin \varphi + \dot{\theta} \dot{\theta} \cos \theta \sin \varphi + \dot{\theta} \dot{\phi} \sin \theta \cos \varphi + \theta \cos \varphi - \theta \sin \varphi$$

$$\dot{\omega}_{\text{segment.y}} = \ddot{\theta} \sin \theta \cos \varphi + \dot{\theta} \dot{\theta} \cos \theta \cos \varphi + \dot{\theta} \dot{\phi} \sin \theta \sin \varphi + \theta \sin \varphi - \theta \cos \varphi$$

$$\dot{\omega}_{\text{segment.z}} = \ddot{\theta} \cos \theta + \dot{\theta} \dot{\theta} \sin \theta + \ddot{\phi}$$

k) Angular momentum:

The angular momentum is calculated according to Goldstein (1965) and follows the same convention for all the segments:

$$\dot{H}_{\text{Segment.x}} = I_{\text{Segment.IntExt}} \dot{\omega}_{\text{Segment.x}} + (I_{\text{Segment.FlxExt}} - I_{\text{Segment.AbdAdd}}) \omega_{\text{Segment.z}} \omega_{\text{Segment.y}}$$

$$\dot{H}_{\text{Segment.y}} = I_{\text{Segment.AbdAdd}} \dot{\omega}_{\text{Segment.y}} + (I_{\text{Segment.IntExt}} - I_{\text{Segment.FlxExt}}) \omega_{\text{Segment.x}} \omega_{\text{Segment.z}}$$

$$\dot{H}_{\text{Segment.z}} = I_{\text{Segment.FlxExt}} \dot{\omega}_{\text{Segment.z}} + (I_{\text{Segment.AbdAdd}} - I_{\text{Segment.IntExt}}) \omega_{\text{Segment.y}} \omega_{\text{Segment.x}}$$

1.2: Kinetic variables:

The ground reaction forces were obtained using a force plate (Kistler Instruments AG, Winterthur, Switzerland) at 1000Hz, placed in the middle of the walkway. To determine the GRF point of application, the calculation of the center of pressure was obtained using a pressure plate over the force plate, where the medial-lateral and anterior-posterior displacement of the COP were calculated as the weighed mean of the pressure applied.

1.3: Net joint forces and moments calculations:

Each body segment S_i , if idealized as a rigid body, will move according to Newton's mechanical Laws. These principles specify that the S_i movement in a inertial referential R is ruled by the movement equations. The two vectorial equations are:

$$\vec{F} = m \vec{a}$$

$$\vec{M}_{res} = \vec{I} \dot{\vec{H}}$$

where,

$\vec{F}$ force components

$\vec{a}$ acceleration of segment CM

$\vec{M}_{res}$ Moment components

and $\vec{M}_{res}$ is the residual moment from the forces acting on the extremities of the segment:

$$M_{Res,i} = M_{distal} + (p_{Prx,i} \times F_{segment}) + (p_{Dis,i} \times F_{distal})$$

Where $p_{Prx,i}$ and $p_{Dis,i}$ are the proximal and distal moment arms of the $F_{segment}$ forces.

The analysis began with the forces and moments estimation in the foot segment, where the external forces (GRF) acting on the foot are measured with the force plate, then these calculated variables are used as distal forces on the adjacent segment.

1.4) Analysis procedures:

After the data collection, many steps are necessary to the processing and posterior data analysis. For the solution of the equations, programs were developed in MATLAB environment.

- Video process:

- Four cameras were positioned in such a way that each of the 9 external markers placed on the subject were recorded by at least two of the cameras at each time.
- The videos were recorded using SIMI motion system and 4 movies were generated for each trial (one for each camera).

- The videos were imported into the software Dvideo, the external markers digitized and 3D coordinates (xyz) generated into matrices.
- GRF process:
 - The force plate was placed in the middle of the walkway and the six channels (three force components and three moment components) were recorded into SIMI motion system.
 - The force data acquired in electrical units are converted into force units, and a six column matrix was exported.
- COP data:
 - The pressure plate was placed over the force plate and the pressure data from all the sensors was recorded into Footscan gait 2nd generation.
 - The pressure data was used to calculate the COP (x and y) coordinates and exported as a two column matrix.
- MATLAB processing:

Two programs were developed in MATLAB: the first to implement the 3D inverse dynamics model and the second to perform the PCA analysis. The first performed many tasks, with the purpose of generating the variables used to calculate the PCA coefficients:

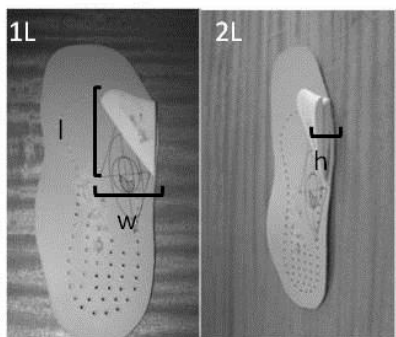
 - Import the three data matrices (3D kinematics, GRF and COP coordinates);
 - Filter the data;
 - Calculate the masses and moments of inertia of the segments;
 - Calculate the joint centers, uvw coordinate systems, make the solidification process, ijk coordinate systems, Euler angles, angular velocities and accelerations;
 - Normalize the GRF components by body weight;
 - Calculate the joint forces and moments;
 - Export the data (waveforms of the three trials in each condition for each subject normalized by step duration) of the variables to be used in the PCA analysis into matrices (ankle and knee net joint moments, ankle, knee and hip range of motion, GRF components, COP displacement);

- Export the discrete data (peaks, range of motion and duration of the three trials in each condition for each subject) of the variables that were compared in Chapter 3 (ankle and knee net joint moments, ankle, knee and hip range of motion);
- Export the data used in Chapter 6 for comparison with ANYBODY data simulation (ankle, knee and hip net joint moments, ankle and knee net joint forces);

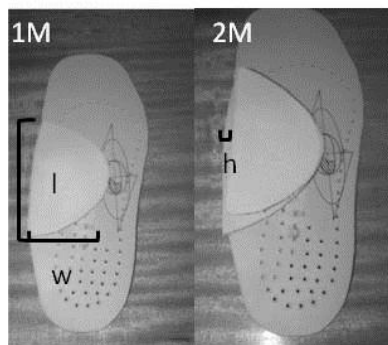
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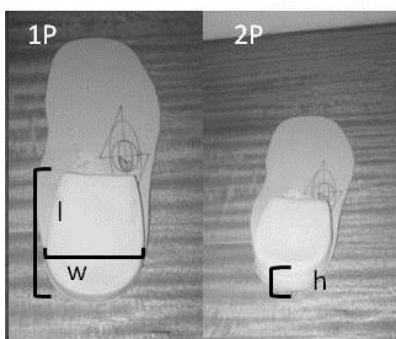
Appendix II: Wedges dimensions



$l = 2\text{cm}$
 $w = 1.8\text{ cm}$
 $h: 1L = 1\text{cm}$
 $2L = 2\text{cm}$



$l = 3\text{cm}$
 $w = 2\text{ cm}$
 $h: 1L = 1.1\text{cm}$
 $2L = 2.2\text{cm}$



$l = 2.5\text{cm}$
 $w = 2.2\text{ cm}$
 $h: 1L = 0.9\text{cm}$
 $2L = 1.8\text{cm}$

Appendix III: COP and GRF analysis in CG

A NEW APPROACH TO PRESCRIBE CUSTOM MADE INSOLES FOR TRANSFEMORAL AMPUTEES USING PRINCIPAL COMPONENT ANALYSIS. PART 1: GRF COMPONENTS AND COP DISPLACEMENT IN HEALTHY GAIT.

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ABSTRACT

Gait pattern alteration is common in the elderly population. Devices like wedges are used in patients with orthopaedic diseases to relief joint pain and to improve gait pattern. Typical gait studies involve an analysis of variables extracted from particular portions of the signal waveforms, in the process generating large amounts of data that sometimes are difficult to interpret. Principal Component Analysis (PCA) is a powerful method used to reduce redundant information and allowing the comparison of the complete waveform. The purpose of this study is to compare the ground reaction force (GRF) components and Center of Pressure (COP) displacement derived from the gait waveforms of a group of healthy elderly subjects wearing different wedges. Twenty subjects walked wearing regular shoes and six wedges conditions varying in different heights and positions. A force platform (1000Hz) and a pressure plate (300Hz) were used to record GRF and COP variables. The results showed that PCA was able to identify the influence of wedges onto GRF and COP components. These results suggest that PCA could be valuable to prescribe wedges to special populations, based on the specific necessity of the each subject.

1 INTRODUCTION

Gait analysis is used to quantify gait disorders and to evaluate patients clinically. In order to establish the normal gait features and to identify abnormalities, ground reaction forces (GRF) and center of pressures (COP) are the most commonly measured biomechanical variables [1]. Most studies focus on the analysis of variables extracted from specific portions of the signal waveforms, thus generating large amounts of data that sometimes are difficult to interpret [2]. This approach relies on the subjective definition of discrete variables. It is difficult to extract these same variables from all temporal signal curves, especially in the presence of pathologies [3]. A significant advance to the clinical use of gait information would be the successful reduction of the amount of generated data [4].

Deluzio et al. [5] introduced a novel application of Principal Component Analysis (PCA) to the analysis of kinematic and kinetic data. PCA has become a common method of reducing the dimensionality and analyzing waveforms in gait analysis [6]. Some studies have successfully used PCA to analyze GRF components in special groups of patients like elderly subjects with fear of falling [7], subjects with lower limb fractures [6] and osteoarthritic patients [8]. No prior studies compared a group of healthy subjects in different walking conditions using the PCA approach.

Gait pattern alteration is common in the elderly population [9]. These alterations make elderly people more susceptible to gait disorders, reduced mobility [10] and falls [7, 11]. Consequently, it is generally accepted that improving gait performance could promote a higher independence and better quality of life for these people. Schmalz et al. [12] showed that devices like wedges are used in patients with orthopedic diseases to relief joint pain and improve gait pattern. Through an analysis of the influence of insoles on different gait variables, it should be possible to apply this knowledge to build insoles addressing the needs of individuals suffering from diseases that affect their gait pattern [13].

The analysis of gait of an elderly healthy population and the understanding of the influence that different wedges have on their gait should provide a useful tool to properly identify and compensate for gait deviations observed in special population. The purpose of this study is to compare GRF components and COP displacements measured in the gait patterns of a group healthy elderly subjects wearing different wedges using the Principal Component Analysis approach.

2 METHODS

2.1 *Participants*

The participant group was composed of 20 physically active subjects (mean age 67 ± 8.56 years old, mean weight 68.5 ± 6.2 kg). To be included in the study, the subjects needed to be over 50 year old and to practice physical activity regularly. Individuals who experienced movement limitations or pain during walking were excluded.

2.2 *Data Collection*

All participants walked over an 8m walkway, wearing their own shoes at a self-selected speed. After a short adaptation, each subject performed three successful trials wearing these shoes, referred as control condition (CON). Thereafter, they walked with four of the six wedges, selected randomly, inserted into their shoes. After the positioning of the wedges inside their shoes by one researcher, the subjects walked for five minutes prior to data collection to adapt to the new conditions.

The six different shape wedges were made of polyurethane material: two lateral, placed under the 5th metatarsal head having thicknesses of 1 cm (1L) and 2 cm (2L), two medial, placed under the foot arch having thicknesses of 1.1 cm (1M) and 2.2 cm (2M), and two posterior having thicknesses of 0.9 and 1.9 cm, placed under the *calcaneum*. Therefore, a total of seven conditions were studied: CON, 1L, 2L, 1M, 2M, 1P and 2P.

2.3 *Gait analysis and signal processing*

A piezoelectric force platform (Kistler Instruments AG, Winterthur, Switzerland) sampled at 1000 Hz was placed in the middle of the pathway to monitor the GFR components (Vertical – GRFvt, Medio-Lateral – GRFml and Antero-Posterior – GRFap). The COP trajectory (Medio- Lateral – COPx and Antero-Posterior – COPy) was collected at 300Hz using a pressure plate FootScan (RsScan, Olen, Bélgica) having a spatial resolution of 2.7 sensors/cm². The pressure plate was placed on top of the force plate. The force signal was collected in the software SIMI 7.0 and the pressure signal with the software FootScan 7 gait 2nd generation. The data processing, filtering, and PCA analysis were performed using MATLAB 7.0. The systems were synchronized by an external unit that allowed starting the systems simultaneously

with a manual trigger.

To reduce the effect of random noise, the data was filtered using 4th order Butterworth filters; the force data with a cutoff frequency of 4Hz and the COP data with a cutoff frequency of 2Hz. The signals were interpolated and resampled to obtain 100 points, providing one point for each percent of the stance phase.

2.4 Statistical procedures

PCA was performed on the GRF and COP waveforms as per Deluzio et al. [14]. In summary, the aim of PCA is to summarize the information contained in 100 variables, representing 100% of the stance phase in a smaller number of components that explain most of the variance through linear combinations of these variables [15]. Principal components (PCs) are arranged in decreasing order in such a way that the first PC accounts for most of the variability in the data, and each subsequent component accounts for as much of the remaining variability as possible [16]. In this study, the number of PCs used was 3.

PCs are an orthogonal transformation which converts p variables $X=x_1, x_2, x_3, \dots, x_p$ (in this case from 0 to 100% of the stance phase) into p new uncorrelated PCs $Z=z_1, z_2, z_3, \dots, z_n$, which are defined by the equation $Z=U^tX$, where U are the eigenvectors of the covariance matrix of X . U_n is calculated by the equation $SU_n=\lambda U_n$ where λ are the p eigenvalues ranked in decreasing order and S is the covariance matrix of X . The load vectors are defined by the equation $Z_n(\lambda^{1/2})$ and these vectors are normalized to vary between (-1,1). Knapp & Comrey [17] suggested a threshold of ± 0.71 to consider a load vector from one variable as relevant, and then to attribute a meaning to this PC.

The PC model was developed based on the gait pattern of the subjects walking in CON condition and then, this model was applied to the wedges conditions. PC score values (obtained by the internal product from PC1, PC2 or PC3 to each waveform) for each subject in each condition were retained for analysis, with 3 score values (PC1, PC2 and PC3 scores) for the five waveforms (GRFvt, GRFml, GRFap, COPx, and COPy), totaling 15 PC score values to analyze per subject per wedge condition.

The normality of the distribution of the data was verified using *Shapiro-Wilk's* test and variances homogeneity using *Levene's* test. *One Way* ANOVA and *post hoc* LSD were used to compare PC score values from CON and from the wedge conditions for the 15 variables

analyzed (PC1, PC2 and PC3 score values from (GRFvt, GRFml, GRFap, COPx, and COPy). The level of significance used was $\alpha=0.05$.

3 RESULTS:

According to Muniz and Nadal [6] that analyzed the differences in group classification using two, four and six PCs from GRF waveforms, two PCs reached 100% of specificity in separating the groups (in their study healthy and tibial fracture subjects). Besides, Deluzio et al. [5] stated that PCs accounting for smaller variances are harder to explain. Thus, 3 PCs were retained in each waveform for analysis (Table 1). The variance explained using the SCREE plot criteria was taken for purpose of comparison [15].

Table 1: Variance explained using the SCREE plot criteria and with 3 PCs.

Variable	number of PCs (SCREE)	variance explained (%)(SCREE)	variance explained (%) 3 PCs
GRFvt	5	92.03	79.98
GRFml	5	86.04	74.45
GRFap	6	94.99	79.78
COPx	4	90.48	84.48
COPy	5	98.86	93.99

No speed differences were found among conditions ($F=2.078$; $p=0.06$). Since GRFap and COPy PC models were not able to differentiate among conditions (Table 2), these data were not explored in further detail. Figures 1 to 3 show the results for GRF and COP analyses: a) The first three PC curves (Z1, Z2 and Z3) and b) the highest score corresponding to a control condition and the score corresponding to the wedge with the highest difference from CON, for a 95% confidence interval of PC scores.

Table 2: PCA results: (mean $\pm$ SD) PC1, PC2 and PC3 scores in GRFvt, GRFml, GRFap, COPx and COPY.

		CON	1L	2L	1M	2M	1P	2P
GRFvt	PC1	0.07 $\pm$ 0.43	0.12 $\pm$ 0.60	0.29 $\pm$ 0.43	-0.09 $\pm$ 0.39	0.16 $\pm$ 0.49	-0.1 $\pm$ 0.53	-0.09 $\pm$ 0.52
	PC2	-0.33 $\pm$ 0.47	0.44 $\pm$ 0.58*	0.15 $\pm$ 0.48*	0.12 $\pm$ 0.59*	-0.16 $\pm$ 0.27	-0.09 $\pm$ 0.53	-0.16 $\pm$ 0.06
	PC3	-0.28 $\pm$ 0.70	1.01 $\pm$ 0.28*	1.17 $\pm$ 0.16*	0.70 $\pm$ 0.95*	0.86 $\pm$ 1.11*	-0.03 $\pm$ 1.01	0.04 $\pm$ 1.18
GRFml	PC1	0.29 $\pm$ 0.95	-0.62 $\pm$ 0.92*	-0.29 $\pm$ 1.20	-0.24 $\pm$ 1.16	0.01 $\pm$ 0.91	-0.70 $\pm$ 0.68*	-1.38 $\pm$ 0.58*
	PC2	-0.13 $\pm$ 0.77	-0.17 $\pm$ 0.80	0.07 $\pm$ 0.47	-0.02 $\pm$ 0.67	-0.20 $\pm$ 0.42	-0.13 $\pm$ 0.69	-0.43 $\pm$ 0.38
	PC3	0.18 $\pm$ 0.54	-0.33 $\pm$ 0.91	0.27 $\pm$ 0.72	0.12 $\pm$ 0.16	-0.02 $\pm$ 0.55	-0.05 $\pm$ 0.20	0.19 $\pm$ 0.80
GRFap	PC1	0.00 $\pm$ 0.89	0.23 $\pm$ 0.66	0.23 $\pm$ 0.66	0.16 $\pm$ 0.69	0.07 $\pm$ 0.63	-0.28 $\pm$ 0.57	0.03 $\pm$ 0.47
	PC2	-0.29 $\pm$ 0.58	0.26 $\pm$ 0.57	0.05 $\pm$ 0.59	0.14 $\pm$ 1.00	-0.03 $\pm$ 0.94	-0.08 $\pm$ 0.77	-0.06 $\pm$ 0.72
	PC3	0.00 $\pm$ 1.12	0.07 $\pm$ 0.58	-0.19 $\pm$ 0.83	-0.19 $\pm$ 0.40	-0.2 $\pm$ 10.01	0.34 $\pm$ 0.62	-0.29 $\pm$ 0.30
COPx	PC1	-0.15 $\pm$ 2.21	-1.44 $\pm$ 4.55	-0.81 $\pm$ 3.87	-2.70 $\pm$ 1.82*	-3.25 $\pm$ 2.18*	-3.99 $\pm$ 2.12*	-1.81 $\pm$ 2.98
	PC2	0.00 $\pm$ 3.68	4.01 $\pm$ 4.20*	1.68 $\pm$ 4.31	-1.08 $\pm$ 3.42	-1.54 $\pm$ 2.42	-0.97 $\pm$ 3.01	0.05 $\pm$ 3.71
	PC3	0.00 $\pm$ 3.49	-0.21 $\pm$ 3.95	0.91 $\pm$ 3.88	0.11 $\pm$ 2.86	-1.89 $\pm$ 2.01	0.34 $\pm$ 2.23	-1.96 $\pm$ 1.65
COPY	PC1	0.00 $\pm$ 3.18	1.40 $\pm$ 2.63	-0.61 $\pm$ 2.80	-0.92 $\pm$ 2.90	-0.40 $\pm$ 2.47	1.21 $\pm$ 2.53	0.00 $\pm$ 1.86
	PC2	0.00 $\pm$ 2.78	-0.28 $\pm$ 1.54	0.07 $\pm$ 2.02	0.68 $\pm$ 2.78	0.17 $\pm$ 2.88	0.14 $\pm$ 1.75	1.27 $\pm$ 2.21
	PC3	0.75 $\pm$ 3.18	-0.07 $\pm$ 1.37	1.60 $\pm$ 3.44	0.86 $\pm$ 2.58	2.15 $\pm$ 1.33	-0.46 $\pm$ 3.94	-2.49 $\pm$ 2.18

*Statistically significant differences from CON group ($p < 0.05$).

GRFvt is the variable that is most influenced by the use of wedges being significantly affected by four of the six conditions, the lateral wedges were different from CON in PC2 (1L: $p=0.00$; 2L: $p=0.02$) and in PC3 (1L: $p<0.001$; 2L $p=0.00$) and the medial conditions were different in PC2 (1M: $p=0.02$) and in PC3 as well (1M: $p=0.04$; 2M: $p=0.00$). The grey areas in Figure 1a highlight the threshold value of 0.71. GRFvt PC2 is significant from 37 to 85% of SP (Fig. 1a), which corresponds to the single support phase. PC3 negative peak is at 8% of SP, which represents the heel contact. Ranking the PC score values of all the wedges that showed differences from CON in PC2 and PC3, the ones that presented the highest influence was 1L in PC2 and 2M in PC3. Then, to emphasize the gait pattern with and without the wedge, the waveforms of 1L (Fig. 1b) and 2M (Fig. 2c) are presented. Differences in the magnitude of the waveforms can be seen in almost all SP, (Fig. 1b), but are more prominent in single support, which is in accordance to PC2 behavior. The behavior of the 2M wedges at the beginning of SP (PC3, see table 2 and Fig 1c) shows that all the wedges affected in PC3 tend to anticipate the beginning of single support (Fig 1c and Table 2).

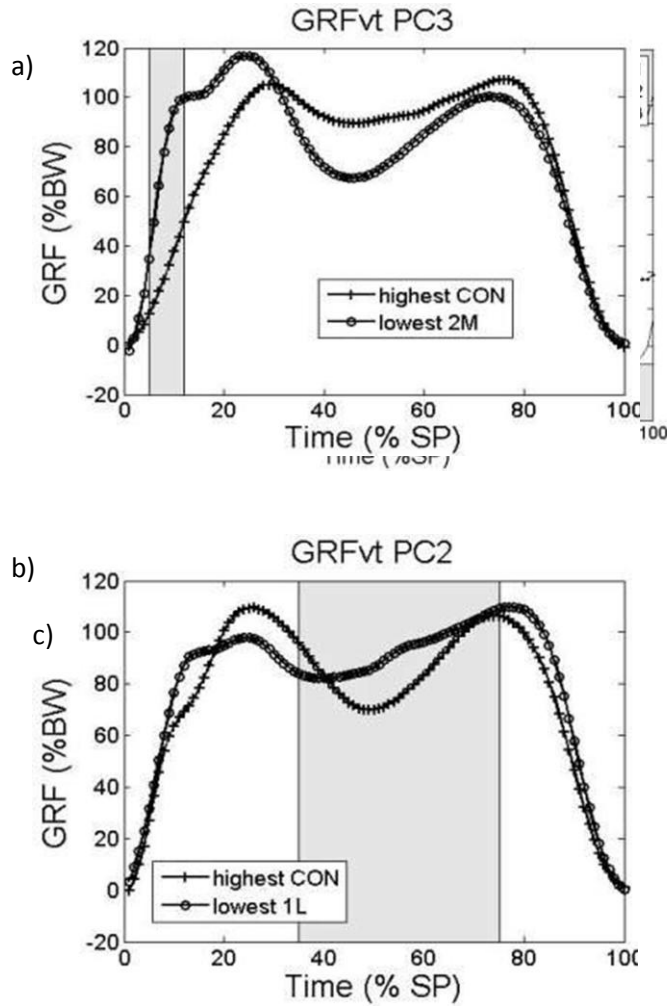


Figure 1: GRFvt analysis; a) PC1, PC2 and PC3 load vectors; b) Highest and lowest scores for PC2; c) Highest and lowest scores for PC3. The grey area highlights the 0.71 threshold [17].

GRFml is different in 1L ($p=0.03$), 1P ($p=0.01$) and 2P ($p=0.00$) for PC1 (Table 2). GRFml load vector for PC1, is significant from 25 to 55% and from 63 to 90% (Fig 2a). The highest and lowest PC1 score values show that the influence of the wedges 1L, 1P and 2P tends to reduce GRFml, in the 25% to 90% interval, where 2P emphasizes the highest differences from CON (Fig 2b).

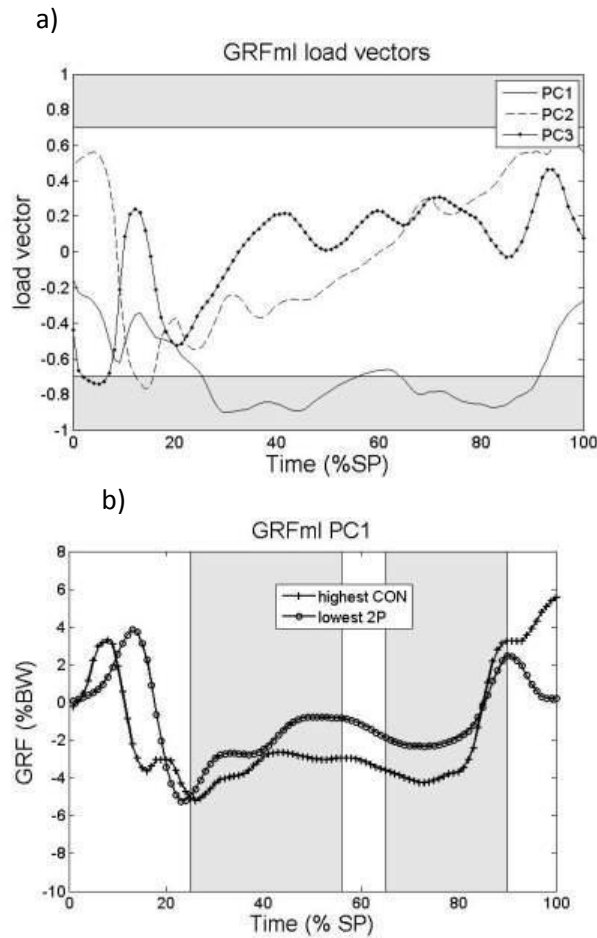


Figure 2: GRFml analysis; a) PC1, PC2 and PC3 load vectors; b) Highest and lowest scores for PC1; The grey area highlights the 0.71 threshold [17].

COPx had differences in PC1 (Table 2) in the conditions 1M ($p=0.04$), 2M ($p=0.01$) and 1P ($p<0.001$) and in PC2 for the condition 1L ($p=0.01$). The load vectors of PC1 COPx are relevant from 30% to 70% representing the single support (Fig 3a) and PC2 is positive from 8-12%, representing heel contact. Figure 3c illustrates that in the beginning of SP the wedge 1L increases the medial-lateral displacement and at the end it reduces the displacement below the control condition.

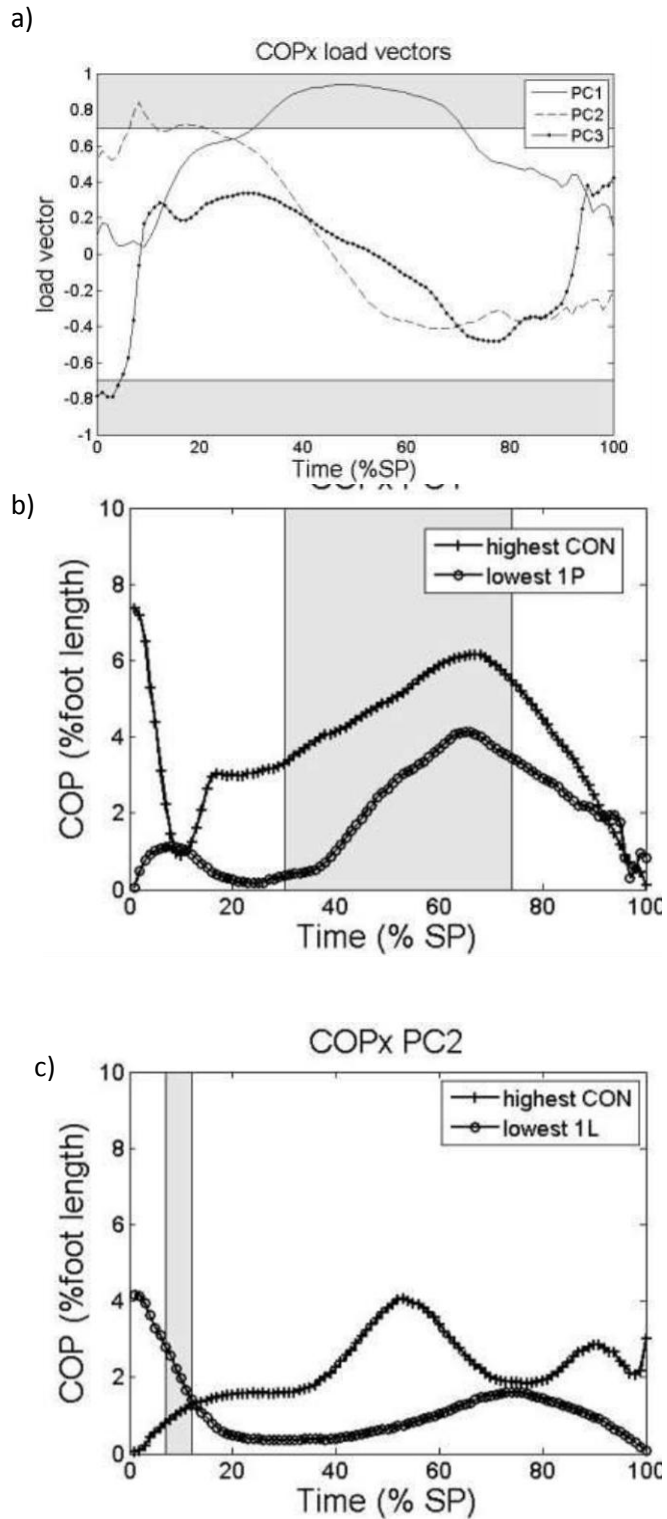


Figure 3: COP analysis; a) PC1, PC2 and PC3 load vectors; b) Highest and lowest scores for PC1;c) Highest and lowest scores for PC2. The grey area highlights the 0.71 threshold [17].

A discriminant analysis performed for all the variables analyzed in this study indicated

that the lateral wedges were the ones that most influenced the gait pattern, in which the GRFvt PC3 was the only variable able to discriminate the groups ($\lambda=0.417$, Fisher coeff=0.672, $p=0.02$). The distribution of the PC scores values in PC2 and PC3 directions emphasizes that the differences are in the direction of PC3 (Fig 4).

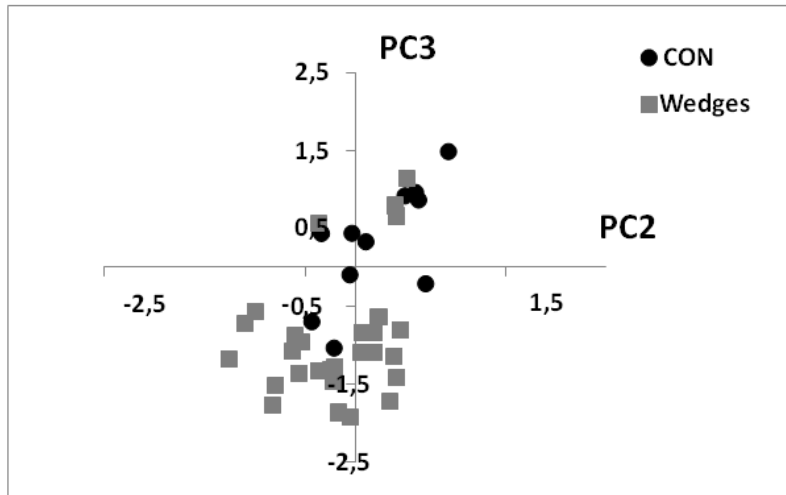


Figure 4: Discriminant analysis: PC2 (horizontal axis) and PC3 (vertical axis) scores for CON and the Wedges different from CON in PC3 (1L, 2L, 1M, 2M). The separation of the conditions from CON is in PC3 direction.

4 DISCUSSION

The purpose of this study was to apply PCA analysis to GRF and COP waveforms to characterize the influence of different wedges in the gait pattern of healthy elderly subjects. The results showed that PCA is a effective method to analyze gait data because: a) it reduces the number of variables necessary to represent the whole waveform; b) data from the entire gait cycle is considered; c) data reduction results in a set of uncorrelated features that explain most of the variance presented in the GRF and COP patterns [18].

To the best of our knowledge, no studies that examined the influence of the wedges in gait have used the PCA technique to analyze the data [12, 13, 19-23]. Studies that apply PCA approach to GRF patterns were not focused on the influence of wedges. Rather they were building a model derived from a control group to compare with patients suffering from different diseases [6, 8, 24].

For GRF components, the main effects of the wedges are on the vertical and medial- lateral components. GRFvt is the pattern most affected by the use of wedges (Table 2). The medial and lateral wedges scores are smaller than CON in PC2 and in PC3.

As PC2 is negative from about 35% till 75%, which coincides with the single support phase, it can be stated that the wedges that alter foot inversion/eversion angle have a high influence on weight acceptance. GRFml differences were found in PC1, in 1L and the posterior conditions. PC1 represents the single support phase, and the differences are illustrated by the highest and lowest scores (Fig 2c).

The medio-lateral displacement of the center of pressure (COPx) is influenced by 1L, both medial wedges and 1P. Balmaseda et al. [25] found that the COPx trajectory is laterally deviated using an Ankle-Foot-Orthosis. Also, Guldemon et al. [26]

found a laterally deviated COP displacement using custom-made foot orthoses. Conversely, Chevalier et al. [27] found no differences in COPx displacement comparing shod and barefoot walking.

Condition 1L is the one that most influenced the behavior of most patterns, reducing the scores from 3 of the 4 (GRFvt, GRFml, COPx) variables that presented differences. This could be due to the fact that this wedge is, according to the subjects, the one less comfortable to walk with.

In a forthcoming paper [28], we will describe a method based on the present results to prescribe custom made wedges to transfemoral amputees. In that work, we apply the PC model to the transfemoral amputees gait waveforms. We looked at the wedges influences and used the information to prescribe a wedge with a “counter-effect” approaching the gait pattern of the CON group.

5 CONCLUSION

PCA analysis is a powerful tool to analyze gait, it takes into consideration the entire waveform of the variables examined. The reduction of data without the loss of information is critical to understand the gait pattern without having to select specific variables to represent the gait cycle. The knowledge gained in a healthy elderly population is useful to understand the normal gait pattern behavior and the influence of wedges.

In turn, it should be helpful to prescribe and apply wedges in special populations.

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Appendix IV: Conference proceedings already published

As mentioned before, the following proceedings were derived from this thesis and the abstracts are presented below:

- Castro M., Soares D.P., Machado L.J.R.; Comparison of vertical GRF obtained from force plate, pressure plate and insole pressure system (2011). Proceedings of the 29th ISBS. Revista Portuguesa de Ciências do Desporto. 11 (2), 849.
- Voinescu M., Soares D.P., Castro M., Mendes E.A., Davidescu A., Machado L.J.R. (2011). A study of moments acting on the tibia during gait in the active elderly population. Proceedings of the 29th ISBS. Revista Portuguesa de Ciências do Desporto. 11 (2), 575.
- Soares D.P., Castro M.P., Sebastião R. A., Thimoteo T., Mendes E.A. (2010). Kinetic Gait Analysis Using Different Wedges in Elderly Population. Annals of the 17th Congress of the European Society of Biomechanics. Edinburgh-Scotland
- Mendes E.A., Soares D.P., Castro M.P., Spence W.D., Correia M.V. (2010). Pressure distribution, by quadrants on shod amputees and controls. Annals of the 17th Congress of the European Society of Biomechanics. Edinburgh – Scotland.
- Castro M.P., Soares D.P., Sebastião R. A., Thimoteo T., Mendes E.A. (2010). Pressure Gait Analysis of Elderly Population Using Different Wedges. Annals of the 17th Congress of the European Society of Biomechanics. Edinburgh – Scotland.
- Castro, M.P., Soares, D.P., Sebastião R. A., Thimoteo T., Mendes E.A., Machado, L.J.R. (2010). Baropodometric analysis of amputees gait: a preliminary study. 13th World Congress of the International Society for Prosthetics and Orthotics and ORTHOPAEDIE + REHA-TECHNIK, Proceedings of 13th World Congress of the International Society for Prosthetics and Orthotics, Leipzig – Germany.
- Soares, D.P., Castro, M.P., Sebastião R.A., Thimoteo T., Mendes E.A., Machado, L.J.R. (2010). The influence of different wedges in elderly gait kinetic parameters. 13th World Congress of the International Society for Prosthetics and Orthotics and ORTHOPAEDIE + REHA-TECHNIK, Proceedings of 13th World Congress of the International Society for Prosthetics and Orthotics, Leipzig – Germany.
- Mendes, E., Soares, D.P., Castro, M.P., Spence, W.D., Correia, M.V. (2010). Analysis of plantar pressure and balance of transfemoral amputees compared with non-amputee subjects, using their shoes. 13th World Congress of the International Society for Prosthetics and Orthotics and ORTHOPAEDIE + REHA-TECHNIK, Proceedings of 13th World Congress of the International Society for Prosthetics and Orthotics, Leipzig – Germany.
- Soares, D.P., Castro, M.P., Ribas, R., Sebastião, R.A., Mendes, E.M., LaFortune, M.A., Machado, L.J.M., (2009). Análise de marcha de indivíduos amputados comparativamente à marcha normal utilizando dinâmica inversa tridimensional; III Congresso Nacional de Biomecânica. Bragança – Portugal. 161-162.

COMPARISON OF VERTICAL GRF OBTAINED FROM FORCE PLATE, PRESSURE PLATE AND INSOLE PRESSURE SYSTEM

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The aim of this study is to compare the vertical component of the ground reaction force (GRF) obtained from the force plate (FP) with those obtained from pressure plate (PP) and insole pressure system (IPS), and to compare the values found between the two pressure systems (PP vs IPS). Twelve subjects walked at a self-selected speed on a 6m walkway, where in the middle there was the FP, and over it, the PP. Simultaneously, the participants used the IPS. The results suggest that there are larger differences between the force values measured by the baropodometric systems when compared to FP, where the baropodometric systems seem to underestimate the force values. Therefore the absolute values recorded by the baropodometric systems should be interpreted very carefully and the comparison of results acquired by different systems should be avoided.

KEY WORDS: plantar pressure, plantar force, kinetic analysis, baropodometry, accuracy.

INTRODUCTION: During gait, loads are transferred between the human body and the ground, starting at the calcaneus and finishing in the forefoot, until toe off (Burnfield et al., 2004). The measurement of this contact forces offers a variety of information about the external loads to which the body is submitted in different situations. The kinetic analysis of human gait comprehend the measurements of forces and pressures (Rosenbaum & Becker, 1997) being the baropodometry by means of a pressure plate (PP) or insoles pressure system (IPS) and extensimetry by means of a force plate (FP) the most used methods. The pressure is calculated using the vertical component of the ground reaction force (GRF), and in this way the pressure sensors are, essentially, force transducers that measure the force acting in a surface of a known area (Cavanagh & Ulbrecht, 1994). The accuracy and repeatability of the absolute values obtained by means of baropodometry have been questioned (Nicolopoulos et al., 2000; Rosenbaum & Becker, 1997; Woodburn & Helliwell, 1996). In other way, the FP provides the most accurate dynamic force measurements (Cobb & Claremont, 1995). Considering this, the purpose of this study is to compare the vertical component of the GRF obtained from the FP with those obtained from PP and IPS, and to compare the values found between the two pressure systems (PP vs IPS).

METHODS: Participants: Twelve subjects participated in this study (7 women and 5 men) with ages between 25 and 35 years old and the body weight between 54 and 81 kg, physically active, without any pain or limitation during gait.

Instruments: A Footscan PP (RsScan, Olen, Belgium) with 0.5 m length and 4096 sensors, where each sensor presents an area of 0.375 cm², making a spatial resolution of 2.7 sensor/cm², operating at a sample frequency of 300Hz; a Pedar IPS (Novel, Munich, Germany) with 99 sensors per insole operating at a sample frequency of 100Hz; and a Bertec FP (model 4060-15, Bertec Corporation, Columbus, USA) operating at a sample frequency of 1000Hz were used. All equipments were calibrated within a period of one year before testing.

Experimental Protocol: The participants walked at a self selected speed in a 6 m walkway, where in the middle there was the FP, and over it, the PP. At the same time, the subjects used the IPS. Therefore, the data from the three systems were recorded simultaneously. The participants should step over the PP with the right foot and the tests were considered valid only when the entire foot was in contact with the plate. Three valid tests for each subject were performed.

Data analysis: For the PP data acquisition was used the Gait Module 2nd Generation software (RsScan, Olen, Belgium); for the IPS the software Pedar-x Data Acquisition (Novel, Munich, Germany); and for the FP the software Acqknowledge (BIOPAC System, California, USA). The pressure data (pressure values of each sensor in each frame) and the force data (Fz in each time instant) were exported and, using the software MATLAB 7.0 (MathWorks, Massachusetts, USA) a program was developed to obtain the force peak values of both pressure systems and FP.

Statistical Analysis: For the comparison of the results between instruments, the protocol proposed by Bland and Altman (1986) was used, where the mean differences between instruments and the confidence interval of the differences were analyzed.

RESULTS: The figures represent the dispersion of the differences and the mean of the differences of the following comparisons: FP vs IPS (Fig. 1), FP vs PP (Fig. 2) and PP vs IPS (Fig. 3). The confidence intervals of the differences between FP vs IPS, FP vs PP and PP vs IPS were, respectively, 40.1 to 510.3 N, 252.4 to 669.7 N and -498.1 to 208.6 N.

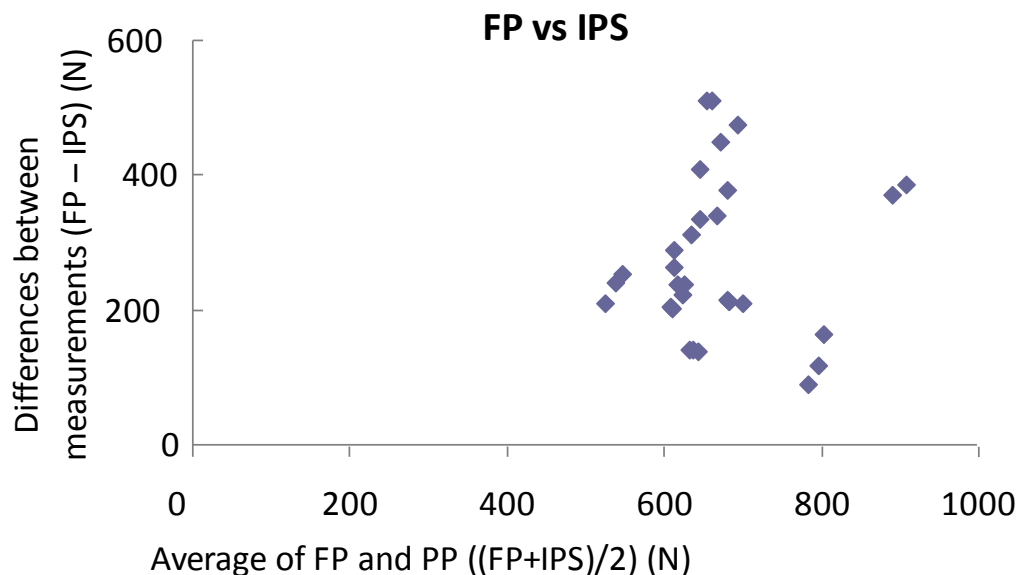


Figure 1: Differences between Force Plate (FP) and Insole Pressure System (IPS).

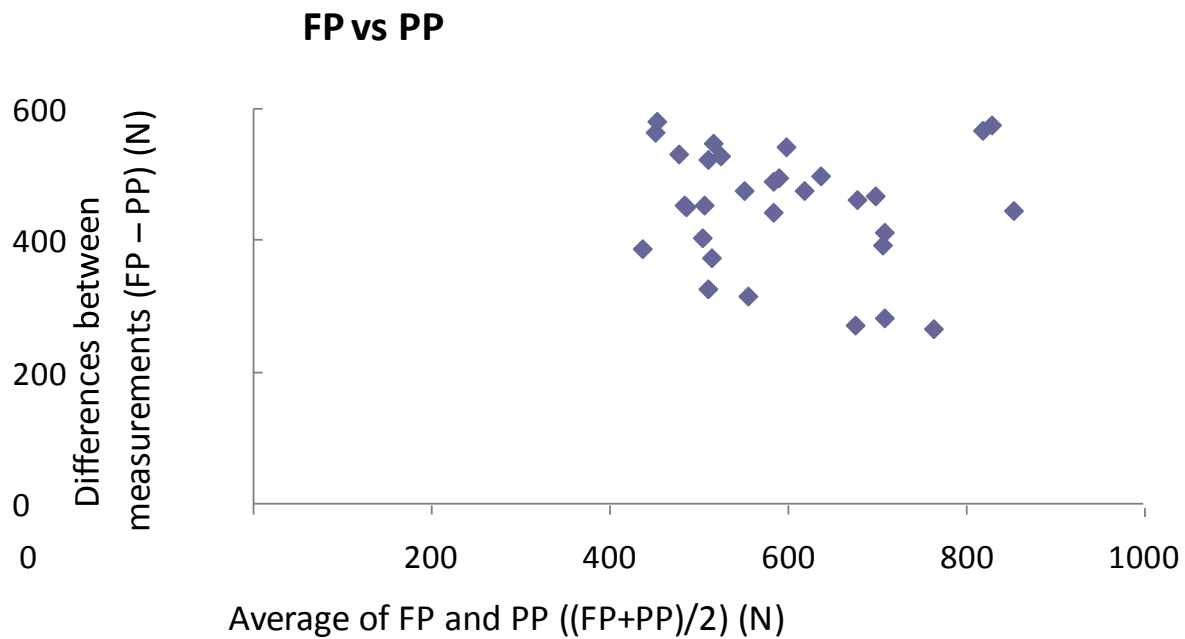


Figure 2: Differences between Force Plate (FP) and Pressure Plate (PP).

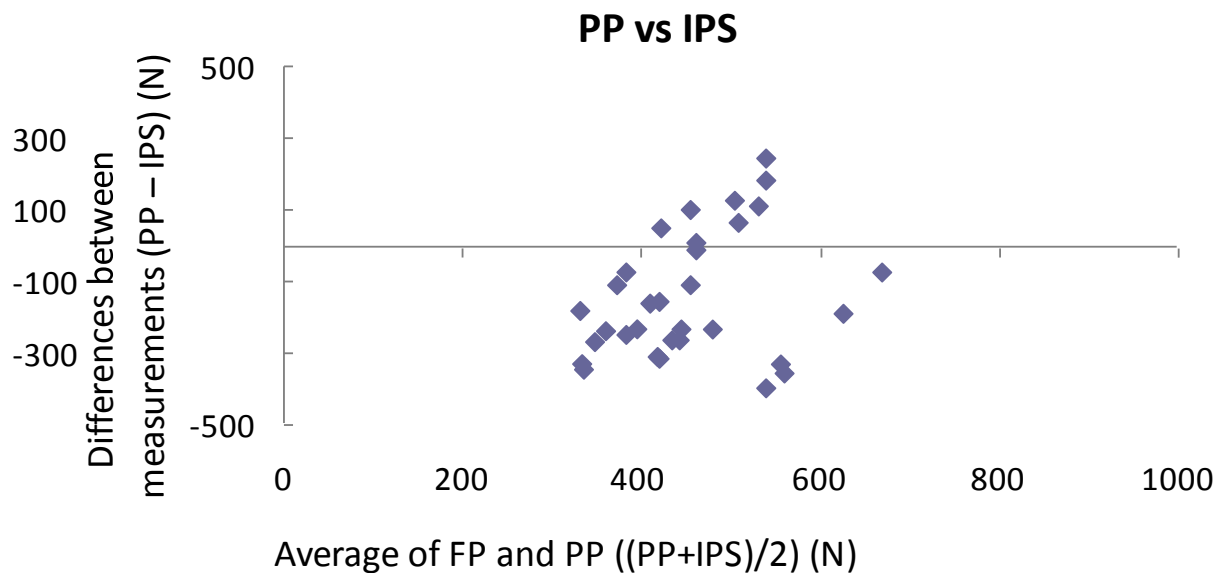


Figure 3: Differences between Pressure Plate (PP) and Insole Pressure System (IPS).

DISCUSSION: The results presented in this study indicate a large difference between the absolute force values recorded by the FP, which is considered the “golden standard” for such measurements (Cobb & Claremont, 1995), when compared to the pressure systems (PP and IPS), where the forces seem to be underestimated in the baropodometric systems. Besides, when the baropodometric systems were compared with each other, larger differences were also found, but not so pronounced as when compared to FP. Even if the

values were normalized by the body weight of the subjects the differences probably are very similar, since the body weight of the participants is the same for all instruments.

A possible explanation for such findings would be the fact that IPS measures the force for each sensor, which is not necessarily the same as the vertical GRF since the angle of the foot influences the angle of the force vector (Barnett et al., 2000). As a result, the force

vectors measured by the IPS are different from the vertical force measured by the FP. As the plates were placed one over the other, they should suffer the contact at the same angle of

the foot; therefore, probably this would not be the real factor responsible for the discrepancy of the data. Another possible explanation for this underestimation that the baropodometric

system presents would be because of a pressure threshold where force and pressure data under this threshold are not recorded (Barnett et al., 2000); this threshold would be used clinically to reduce the noise during the data collection. Even though, during gait cycle, part of the loads on the plantar surface would be under this threshold explaining the constantly

lower force values in the baropodometry when compared to FP.

Other studies reported that the baropodometric systems have a good capacity of providing relative values about the distribution of the force/pressure on the plantar surface, but the

absolute values should be analyzed carefully (Nicolopoulos et al., 2000; Woodburn &

Helliwell, 1996). Considering the comparison of baropodometric systems, a possible imprecision of the insole sensor, generated by changes in temperature inside the shoe is also named as a factor that could promote changes in measurements (Cavanagh & Ulbrecht, 1994). However Low and Dixon (2010), even controlling this factor before their data

collection, they found the same differences described in the literature.

CONCLUSION:

The results presented suggest that there are larger differences between the force values measured by the baropodometric systems when compared to FP, where the baropodometric systems seem to have an underestimation of the force values. Therefore, absolute values recorded by the baropodometric systems should be interpreted very carefully and, if possible, to associate these systems with FP, creating correcting factors which could increase the consistency of these data. Considering the PP and IPS, the analysis of the distribution of the pressure (only relative values) seems more appropriate and the comparison of data collected by different instruments should be avoided. However, we suggest replicating this study with a larger sample size and number of steps to increase the consistency of the results.

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A STUDY OF MOMENTS ACTING ON THE TIBIA DURING GAIT IN THE ACTIVE ELDERLY POPULATION

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The purpose of this study was to estimate the moment acting on three points of the tibia, in order to better understand the loads acting on the shank during gait for the active elderly population. Ten subjects were investigated and walking was chosen as the movement to be studied, since it is a highly common activity in terms of exercising and it is available as a universal solution to improve the quality of life. The data was collected in a motion study laboratory, and inverse dynamics was used to estimate the forces and moments in the joints. An advanced human model was used in this purpose. The results show that in this population, the highest bending moment in the sagittal plane is during late stance phase. It is possible to correlate this data with people with gait disabilities or with implants for bone fixation and help their recovery.

KEY WORDS: active, elderly, tibia, moment.

INTRODUCTION: The most commonly injured bone, by stress fracture, is the tibia (Milner et al., 2006); Milner et al. (2005) stated that tibial injury occurrences may be related to the higher loading of the lower limb. Furthermore there is evidence that some tibial stress fractures are spiral fractures, suggesting that, in addition to vertical and shear forces, moments may be involved in the development of these fractures (Milner et al., 2006). Furthermore, the tibia is exposed to a combination of bending, shearing and torsion simultaneously during activities (Ekenman et al., 1998). The loss of minerals in bones is a serious problem for the elderly people, where the bones become fragile and fractures are more likely; World Health Organisation has estimated that 30% of all women aged over 50 (postmenopausal) have osteoporosis according to a definition of bone mineral density being more than 2.5 standard deviations below the mean for young healthy adult women at any site. Therefore, the understanding of the behaviour of the moments in the tibia of the elderly during common activities is of importance.

To understand the forces in the human body the inverse dynamic has been widely used. Generally, inverse dynamics is performed using simplified models that view each anatomical part of the body as a geometrical shape with a certain mass, the joints of the body being also the joints of the model used in inverse dynamics. While a classical solution offers good results (Vaughan, 1999), the improvement of these models by inputs such as mechanical work of the muscles attached to bones in anatomically correct positions can provide additional and important information about forces acting on the human body, the internal moments on the bones is one example. The internal forces are dependant on the musculoskeletal loads generated by the muscles and tendons attached on the bones while performing various activities. The advanced AnyBody® modelling system is one example of these improved models, being able to capture more accurately the complexity of the musculoskeletal system, which can be very useful in sports science analysis. Therefore, the purpose of this paper is to analyze the moments acting on the tibia during elder people's gait, by using the advanced AnyBody® modelling system.

METHODS: Subjects: Ten healthy adults, 6 female and 4 male were selected as subjects for this study, aged 67 ± 8 years with an average height of 1.64 ± 0.09 m and mass of 66.4 ± 14 kg.

Protocol: The individuals walked on an 8 m walkway, at a self selected speed. After a short adaptation, each subject walked three successful times wearing their own shoes stepping the right leg over the forceplate.

Instruments: A Bertec force plate (model 4060-15, Bertec Corporation, Columbus, USA) operating in a sample frequency of 1000Hz and a video system with four cameras operating in a sample frequency of 50Hz were used. The cameras were placed on the walls with the image taken at the four corners of the force plate, in an angle of 45° with the ground; and a 3D calibrated volume that contemplates the whole area of analysis was filmed before the tests.

Data collection: The software Acknowledge (Bertec Corporation, Columbus, USA) was used for the GRF data acquisition and the software Dvideo® (UNICAMP, São Paulo, Brazil) was used for the video manipulation and for obtaining marker positions;

Data analysis: The video data was processed and the reflective markers placed in anatomical points of the lower right limb, based on the Helen Hayes marker configuration (Vaughan, 1999), were converted into 3D coordinates. The forceplate data was exported as 3 major force and moment directions. The data were synchronized and inverse dynamics was calculated using AnyBody Software®. This paper uses the GaitUniMiamiTDRightLeg model, developed and provided under public domain by the AnyBody® Research Group (anybodytech.com). The TLEM lower extremity model (Horsman, 2007) was chosen. In order to estimate the moments acting on different points of the tibia, several changes were done in the initial model, 3 points of interest from the tibia were chosen, namely at 22% of the length, measured from the knee and towards the ankle, and at 40%, and 60% respectively in the same direction. The rationale for choosing these points was to have a point close to the anatomical middle of the tibia and a point on each side of the initial point, close to the middle of the remaining length. In each point, the tibia was split in two rigid parts, the cut being perpendicular to the axis of the bone, as defined in the leg model (Fig.1). The length of the cut was computed based on the location of the ankle joint and then subtracting the length of the tibia multiplied by the cut coefficient. The two rigid parts were then joined with an additional joint added in the system which was afterwards rigidized. The origins of the two new parts were placed in the middle of the two resulting parts. The masses, inertia matrix, and muscle attachment points were adjusted to suit the length of the rigid tibia parts. The method used is largely based on the work of Wehner et al. (2009).

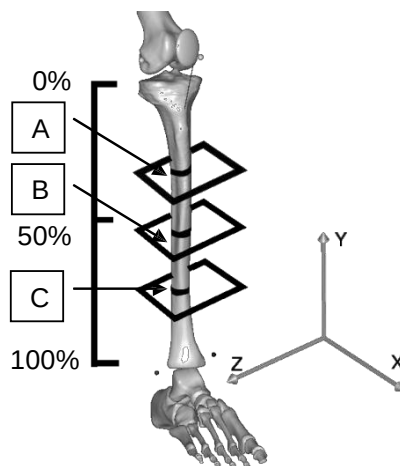


Figure 1: Coordinate system and location of cuts.

For each subject, scaling was considered based on their weight and length of the leg, and an average of 60 time steps/subject were simulated. The study was done on the right leg only. The resultant contact forces in the ankle, knee, and hip were extracted, along with the internal forces in the three cut points. Moments in the three points were also studied.

RESULTS: The average of the extracted resultant forces in the joints showed loading peaks at the beginning and at the end of the stance phase, for all participants. Time was also normalised to total stance time (in percentage) to facilitate the comparison. The average resultant forces maximum value was 3.6 times body weight (BW) for the ankle, 3.8 BW at the knee, and 3.7 BW for the hip (fig.2).

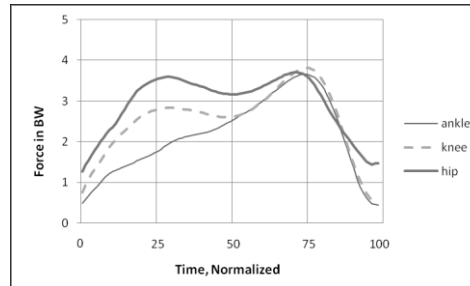


Figure 2: Resultant forces in the ankle, knee and hip joints.

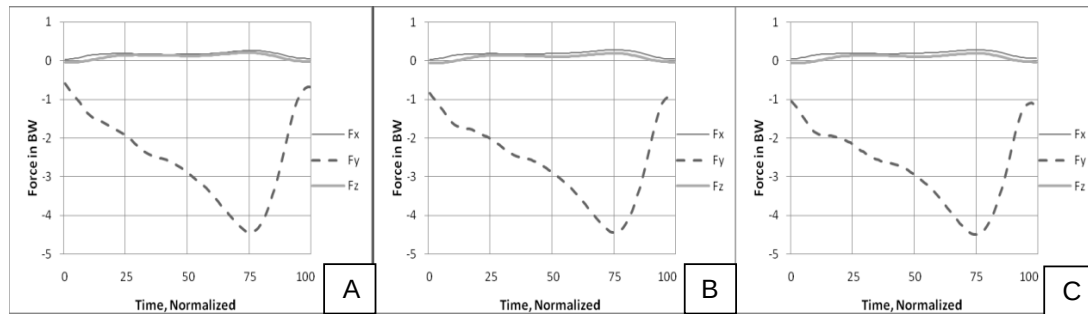


Figure 3: Internal force components at 0,22%(A), 0,40%(B), 0,60%(C) of the tibia from the knee down.

For the rigid joint between the two tibia segments, the maximum average resultant force in each of the 3 points of interest was 4.42 BW, 4.43 BW and 4.49 BW, indicating an increase of the resultant force in the axial direction from the knee to the ankle (Fig.3).

The moments in each section area showed peak values of up to 0.056 BW*m, in the point closest to the knee, but decreased their values towards the ankle, where the maximum value was 0.023 BW*m (Fig.4).

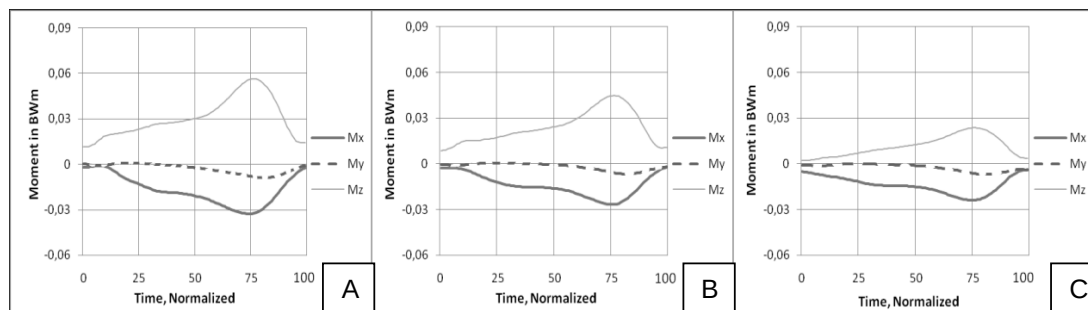


Figure 4: Internal moments at 0,22%(A), 0,40%(B), 0,60%(C) of the tibia from the knee down.

DISCUSSION: This work describes a method to assess stress factors that act on the tibia during gait, and attempts to better understand the internal loads on the tibia during gait. The loads are dependent on the phase of the motion, and also on the location investigated. As expected, the loads increased towards toe off, when the moment in the ankle also increases, in order to facilitate pushing the body forward.

The internal forces in the tibia on the axial direction had a peak of 4.5 BW for this study, with 3 cut points, as opposed to a previous work, where 9 points were used and the peak value was 4.7 BW (Wehner et. al., 2009).

The tendency of the internal moments was to increase towards the knee joint, which can indicate a tendency of fracture in the upper part of the bone if stressed over a long period of time. Also, it must be considered that the people investigated are elderly, even though they live active lifestyles.

The limitations of this study are related to errors in the kinematics which influence the results, and errors induced by force platform signal noise. Also, this study was conducted with only 3 points of interest and, while they are far apart from each other, adding more points into the system could give a better image of the overall load on the tibia.

However, the peak loading values are similar to what has been previously presented, which indicates that while limited, the study does provide useful results.

CONCLUSION: In this study, a method of evaluating moments acting on the tibia has been used on a group of 10 subjects. The data can be used to study loads in the skeleton and simulate the behaviour of the tibia in a finite element environment. With this protocol, it is possible to evaluate different skills and analyse the stress acting on the skeleton in a variety of situations, therefore the model used in the present study can be useful in sports analyses, mainly considering sports that require high loads on the lower limbs. Since the data shown is a mean value of the actual moments and is also normalized by body weight, estimation for similar individuals is possible, and this can be correlated with the loss of minerals and the reduction in bone strength, giving an estimate of the possibility of fracture.

Additionally, it is also possible to use this protocol to compare the pattern of stress obtained in normal individuals with that of people with disabilities and implants for bone fixation. This kind of comparison may help the improvement of this population's recovery.

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Kinetic Gait Analysis Using Different Wedges in Elderly Population

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Introduction

It is well known that in experimental studies, the variable analyzed is frequently tested in individuals with no limitations, called "control group" to see the influence of the device in a normal population.

Some studies showed that devices like wedges are used in patients with orthopedic diseases to relief pain in the joints and improve gait pattern¹. Presupposing that the adaptations are the same, it is wondered if it is possible to apply the results obtained in normal gait to special populations.

The purpose of this study is to analyze the influence of wedges in elderly normal gait and after that, to establish a pattern of alterations caused for different in shoe wedges in elderly population.

Materials and methods

22 subjects (67 years old $\pm$ 8, 56) and physically active (SF36 physical function 82, 33 $\pm$ 18, 01) were asked to walk on a 8 m walkway, passing over a Kistler® force plate wearing their own shoes at a self selected speed. After a short adaptation, each subject walked three times wearing the shoes only and, after that, with each wedge inside the shoe, randomly selected. The wedges were made in three different shapes: one lateral, placed under the 5° metatarsal head (2L: 2cm), one medial placed under the foot arch (2M: 2,2cm), and a posterior (2P: 1,8cm), placed under the calcaneus.

The normality of the data was verified with Shapiro Wilk test and to compare the conditions was used oneway ANOVA test with *posthoc* Tukey. The significant level adopted was $p < 0.05$ and all the analysis were made on the software SPSS 17.0.

The variables analyzed were force peaks in three major components, duration of stance phase and impulse of resultant force.

Results and discussion

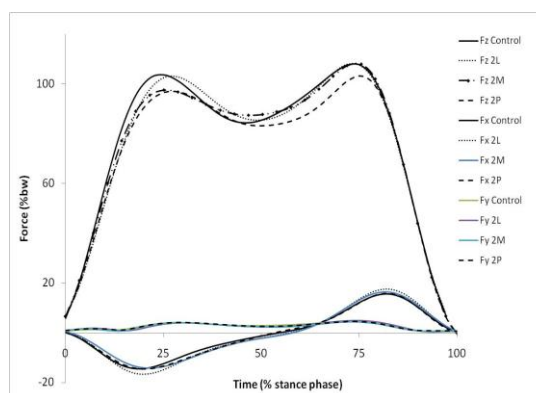


Fig. 1: Force components Fx (antero-posterior), Fy (medio-lateral) e Fz (vertical) in normal gait and in three conditions: 2L, 2M e 2P

The force results showed that there are no statistical differences among any of the conditions for all the variables analyzed. According to², this kind of intervention in a normal population of healthy individuals is not enough to show differences, because the capacity of adaptation is high.

Table 1: Impulse (kN.s) and Duration (ms) of Stance Phase

Duration	SD	Impulse	SD
82.86	12.42	41.64	5.46
75.86	13.02	42.22	7.28
82.54	19.61	43.80	9.39
82.89	7.38	42.34	3.66

SD: standard deviation

Conclusion

The methodology presented in this study doesn't seem to be the most appropriate to establish a pattern for the alterations caused by the wedges. It would be interesting to use some instruments that were able to verify in more details the alterations in gait parameters of this population, or to support the hypothesis that the control group is able to adapt to the different conditions.

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Pressure distribution, by quadrants on shod amputees and controls

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Introduction

Lower limb amputation affects weight distribution, balance and equilibrium ⁽¹⁾. Standing and walking, basic functions for most daily activities are altered, with high impact on the participation and quality of life of amputees. Although prostheses can provide reasonable static support, asymmetry can be observed, frequently, during dynamic locomotion ⁽²⁾. The knowledge of how pressure is distributed over the feet may be very helpful on prosthetic fitting and rehabilitation.

Materials and methods

The pressure and balance data were recorded and analyzed using an RSScan pressure platform and software RSScan Balance 7.7 ®. Data were analyzed with Minitab software version 14.0. Five (5) amputees and 5 controls were recruited and asked to stand comfortably over the platform, during a total of 8,35s. We analyzed the data regarding total area traveled by the COF, force distribution over the two limbs and anterior and posterior quadrants. All the subjects participating in this study where physically active and reported no relevant concern with their health as confirmed by the results on SF36 Scales of Physical Function, General Health and Vitality.

Results and discussion

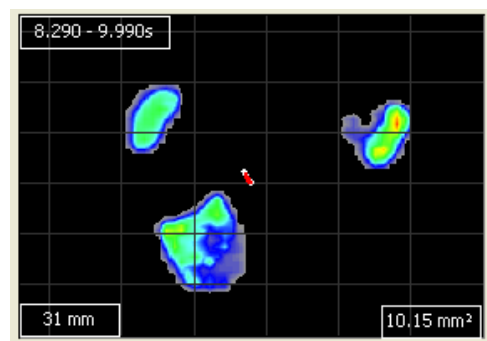
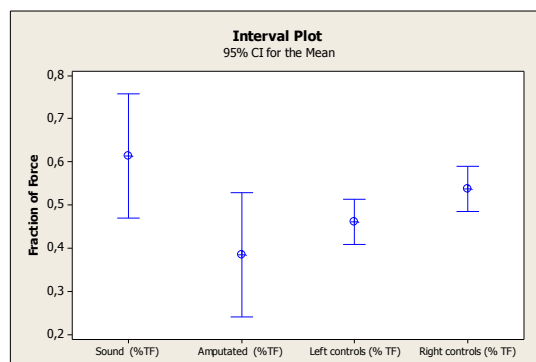


Fig. 1: Example of force distribution of a shod amputee when standing on the platform at 8,3s to 9,9s.

Amputees presented an altered force distribution as observed on Fig. 1. The amputated side has a smaller contact area and the center of force is "dislocated" to the sound side.



No relevant differences were found on left and right support for controls, but a difference between the sound and amputated side for amputees, although not statistically significant (Table.1.1)

Table 1: Paired t-test for amputated and sound and posterior and anterior quadrant of amputated side

T. Test	p	Dif of mean
2,2	0,09	0,228 (SD0,23)

Regarding quadrant distribution, the posterior quadrant on the sound side supports more force than the anterior quadrant, but on the amputated side there is individual variability may due to the prosthetic alignment and component characteristics.

Conclusion

Pressure distribution on amputees is an important data for rehabilitation, and could be used for fine-tuning and monitoring of prosthetic adaptation. However there is a need for further investigation and analysis of a larger sample. The relation between prosthetic alignment and prosthetic component and the differences on force distribution must be investigated

Acknowledgements

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Pressure Gait Analysis of Elderly Population Using Different Wedges

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Introduction

In the literature, there is no consensus about the influence of different wedges in the pattern of normal gait¹. Considering that 11% of the elderly population has problems with mobility², and that the use of insoles in the individuals shoe can influence the quality of his gait³, the gait analysis of the elderly health population is fundamental to constitute a control group to the analysis of the influence of different factors that compose the building of an adequate insole for different necessities.

So, the purpose of this study is to analyze the influence of different wedges in baropodometric parameters in physically active elderly gait.

Materials and methods

22 subjects (67 years old $\pm$ 8, 56) and physically active (SF36 82, 33 $\pm$ 18, 01) walked on a 8 m walkway, passing over a a Footscan® pressure plate at a self selected speed. After a short adaptation, each subject walked three times wearing the shoes only and, after that, with each wedge inside the shoe, randomly selected.

The wedges were made in three different shapes: one lateral, placed under the 5° metatarsal head (2L: 2cm), one medial placed under the foot arch (2M: 2,2cm), and a posterior (2P: 1,8cm), placed under the calcaneus.

The foot was divided in six zones, where two were placed in the fore foot (Z1 under the antero medial zone of the foot and Z2 under the the antero lateral zone of the foot) two in the middle of the foot (Z3 medial zone of midfoot Z4 lateral zone of midfoot) and two in the rear foot (Z5 under the medial zone and Z6 under the lateral zone).

The variables analyzed were peak pressure in each zone and pressure peak total of the interface (sole / floor).

The normality of the data was verified with Shapiro Wilk test and to compare the conditions was used oneway ANOVA test with *posthoc* Tukey. The significant level adopted was $p < 0.05$ and all the analysis were made on the software SPSS 17.0.

Results and discussion

The results showed that Z3 is the zone that is most affected by the wedges, differing all the conditions.

The wedge that affects the pressure most is 2M, where differences were found in, Z3 (2M>2L>2P>Control) and Z6 (2L>2M).

It was not found differences of pressure total peak among the studied conditions.

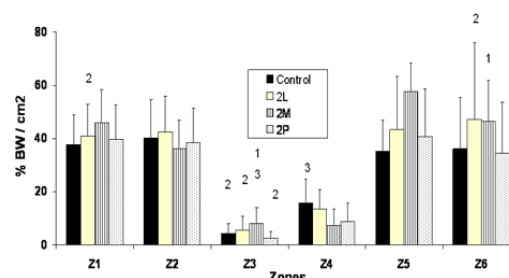


Fig. 1: Peak Pressure in all 6 Zones in all conditions. 1: $\neq$ from 2L; 2: $\neq$ from 2M; 3: $\neq$ from 2P; ($p < 0.05$);

Even with no differences in total pressure among conditions, differences were found in specific zones of the sole, suggesting that is very important the detailing of the analysis to obtain more precise and clinically applicable results.

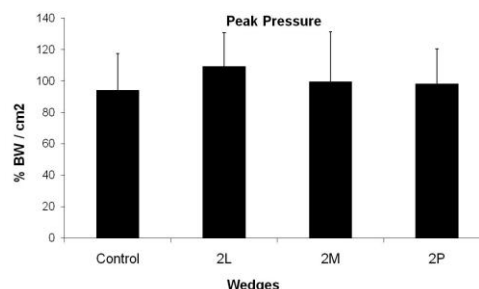


Fig. 2: Peak Pressure in all wedges;

Conclusion

The pressure in the interface sole/floor is susceptible to the wearing of wedges inside the shoe. These results can be used with the purpose of correcting altered patterns of pressure in gait cycle.

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Baropodometric analysis of individuals with lower limb amputations: a preliminary study.

Castro, M.P., Soares, D.P., Sebastião R. A., Thimoteo T., Mendes E.A., Machado, L.J.R.

Introduction

Amputations represent high impact in functional capabilities and quality of life of the individuals. Many studies have been performed with the purpose of understanding such alterations and investigate different ways to reduce the impact of them (Archer et al 2006; Bailey & MacWhannell, 1997; Buckley, 1999; Buckley et al, 1997; Hafner et al, 2002; Han et al, 2003; Hill et al., 1997; Meatherall et al., 2005).

Among these studies, the gait, due to the necessity of this skill for daily life activities has been widely investigated, even though, the studies using baropodometry are rare.

The analysis of plantar pressure can be useful also to the comprehension of the pattern of pressure distribution as to implement therapeutic approaches to improve amputees quality of life.

The purpose of this study is to present the patterns of plantar pressure parameters in amputees gait.

Methods

5 subjects with transfemoral amputation (mean age 67 years old $\pm$ 8, 56) and SF36 physical function 82, 33 $\pm$ 18, 01) were asked to walk on a 8 m walkway, passing over a footscan® pressure plate at a self selected speed. After a short adaptation, each subject walked three times stepping with the right foot and three times with the left foot.

The baropodometric data were recorded using the Footscan 7 Gait 2nd generation software, in a frequency of 300Hz. The foot was divided in six zones, where two were placed in the fore foot (Z1 under the antero medial zone of the foot and Z2 under the antero lateral zone of the foot) two in the middle of the foot (Z3 medial zone of midfoot Z4 lateral zone of midfoot) and two in the rearfoot (Z5 under the medial zone and Z6 under the lateral zone).

The values of zone peaks and impulse were collected using MATLAB routines specifically developed. The data were compared between the amputated leg (AL) and the sound leg (SL). The data were analyzed using SPSS software version 17.0.

Results

The results showed that there are significant differences between the peak pressure in zones Z1 and Z3 and in total pressure when comparing AL and SL. This difference in Z1 can be due to the fact that many amputees don't support the forefoot in stance phase and have no capacity for propulsion, and in Z3 because the prosthesis doesn't allow to transfer the weight properly from the calcaneus to forefoot passing through the medial region.

When comparing different zones in the same foot, there are significant differences in AL, in many comparisons, differently from SL, as showed in table 1. These differences can evidence the abnormal pressure distribution in the prosthetic foot.

Conclusion

According to the results, there are significant differences between the amputated leg and the sound leg. These results are in agreement with the literature and, even though they are preliminary, show that the methodology is useful to analyze amputees gait and can be used to improve rehabilitation protocols to apply in the amputees.

Amputee Leg						Sound leg					
(I) Factor	(J) Factor	Mean Difference (I-J)	Sig.	95% Confidence Interval		(I) Factor	(J) Factor	Mean Difference (I-J)	Sig.	95% Confidence Interval	
				Lower Bound	Upper Bound					Lower Bound	Upper Bound
1	2	-6,421	,647	-18,63	5,79	1	2	9,263	,307	-3,70	22,26
	3	18,740	,000	6,53	30,95	1	3	21,141	,000	8,16	34,12
	4	8,680	,327	-3,63	20,79	1	4	17,543	,002	4,56	30,52
	5	-3,856	,941	-16,06	8,35	1	5	10,428	,190	-2,55	23,41
	6	-18,659	,000	-30,87	-6,45	1	6	3,028	,984	-9,95	16,01
2	1	6,421	,647	-5,79	18,63	2	1	-9,263	,307	-22,26	3,70
	3	25,161	,000	12,95	37,37	2	3	11,858	,094	-1,12	24,84
	4	15,001	,007	2,79	27,21	2	4	8,261	,440	-4,72	21,24
	5	2,566	,990	-9,64	14,77	2	5	1,146	1,000	-11,83	14,13
	6	-12,238	,049	-24,45	-.03	2	6	-6,255	,727	-19,23	6,72
3	1	-18,740	,000	-30,95	-6,53	3	1	-21,141	,000	-34,12	-8,16
	2	-25,161	,000	-37,37	-12,95	3	2	-11,858	,094	-24,84	1,12
	4	-10,160	,160	-22,37	2,05	3	4	-3,598	,966	-16,58	9,38
	5	-22,596	,000	-34,80	-10,38	3	5	-10,713	,167	-23,69	2,27
	6	-37,399	,000	-49,61	-25,19	3	6	-18,113	,001	-31,09	-5,13
4	1	-8,680	,327	-20,79	3,63	4	1	-17,543	,002	-30,52	-4,56
	2	-15,001	,007	-27,21	-2,79	4	2	-8,261	,440	-21,24	4,72
	3	-10,160	,160	-2,05	22,37	4	3	3,598	,966	-9,38	16,58
	5	-12,436	,043	-24,64	-.23	4	5	-7,115	,606	-20,09	5,86
	6	-27,239	,000	-39,45	-15,03	4	6	-14,516	,019	-27,50	-1,54
5	1	3,856	,941	-8,35	16,06	5	1	-10,428	,190	-23,41	2,55
	2	-2,566	,990	-14,77	9,64	5	2	-1,146	1,000	-14,13	11,83
	3	22,596	,000	10,38	34,80	5	3	10,713	,167	-2,27	23,69
	4	12,436	,043	-.23	24,64	5	4	7,115	,606	-5,86	20,09
	6	-14,803	,008	-27,01	-2,59	5	6	-7,401	,564	-20,38	5,58
6	1	18,659	,000	6,45	30,87	6	1	-3,028	,984	-16,01	9,95
	2	12,238	,049	-.03	24,45	6	2	6,255	,727	-6,72	19,23
	3	37,399	,000	25,19	49,61	6	3	18,113	,001	5,13	31,09
	4	27,239	,000	15,03	39,45	6	4	14,516	,019	1,54	27,50
	5	14,803	,008	2,59	27,01	6	5	7,401	,564	-5,58	20,38

The influence of different wedges in elderly gait kinetic parameters

Soares, DP, Mendes EA, Castro MP, Sebastião RA, Machado LJR,

SUMMARY:

The purpose of this study was to investigate the influence of different wedges in elderly gait kinetics. Six kinds of wedges were used, in a sample of 17 subjects. Each subject walked over the force plate wearing each one of the wedges. The results showed that the influence is individual.

INTRODUCTION:

The use of insoles as an auxiliar in small gait deviations is a common practice. From the analysis of the influence of the different parameters that constitute an insole, it is possible to apply these results to build insoles adequate to individuals with different diseases (Kerrigan et al, 2002). In the literature, there is no consensus about the influence of different wedges in the pattern of normal gait (Van Gheluwe et al, 2002). Considering that 11% of the elderly population has problems with mobility (Guccione et al, 1990), and that the use of insoles in the individuals shoe can influence decisively the quality of his gait (Kerrigan et al, 2002), the gait analysis of the elderly health population is fundamental to constitute a control group to the analysis of the influence of different factors that constitute the building of an adequate insole for different necessities. So, the purpose of this study is to analyze the influence of different wedges in physically active elderly gait.

METHODS:

19 subjects (mean age 67 years old $\pm$ 8, 56) and physically active (SF36 physical function 82, 33 $\pm$ 18, 01) were asked to walk on a 8 m walkway, passing over a Kistler® force plate wearing their own shoes at a self selected speed. After a short adaptation, each subject walked three times wearing the shoes only and, after that, with each wedge inside the shoe, randomly selected.

The wedges were made of poliuretano cushion in six different shapes: two lateral, placed under the 5ª metatarsal head (1L: 1cm) and (2L: 2cm), two medial placed under the foot arch (1M: 1,1 cm) and (2M: 2,2cm), and two posterior (1P: 0,9cm) and (2P: 1,8cm), placed under the calcaneus.

The kinetic data were recorded using the Simi Motion System, in a frequency of 1000hz. The values of Fx, Fy and Fz peaks, duration of stance phase and resultant force impulse were collected using MATLAB routines specifically developed. The data were analyzed using SPSS software version 17.0.

RESULTS:

Differently from other studies, the results showed that there are no significant differences in the kinetic parameters analyzed. This can be due to the fact that there is a great variability in some data, and that the subjects are well trained and can adapt their gait to the new situation. Analyzing the data individually, it is clear that each subject has a different adaptation to the wedges, in cases increasing the values, and in other cases decreasing, showing that the

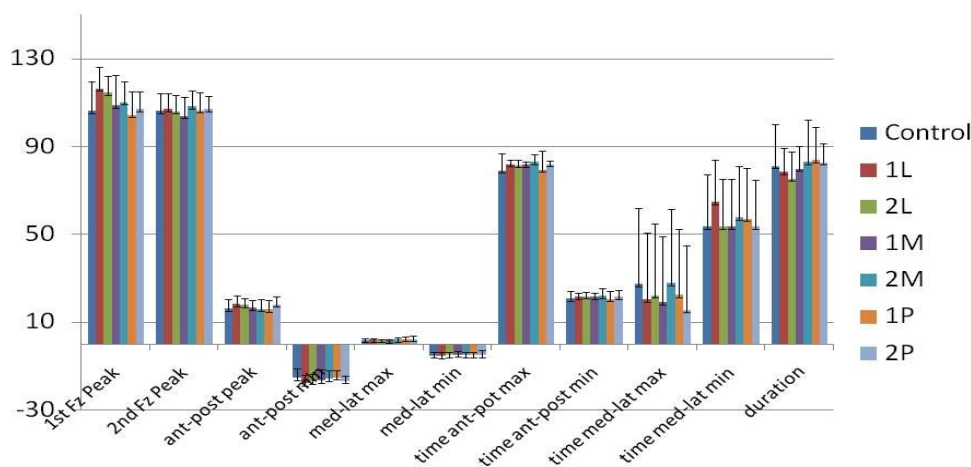
adaptation is an individual process. Kinematic and padoborometric parameters should be analyzed to show a better overview on the influence of the wedges in gait parameters. Other possible reason is that they used their own shoes to perform the trials, then the wedge inside the shoe is not variable enough to show differences.

CONCLUSION:

According to the results, it is possible to conclude that adapation to different wedges is an individual process, and depends not only on the height and position of the wedge, but also to the individual capability to adaptation.

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Mean and SD of all data. Maximal values of Fz, Fx and Fy (First Fz Peak, Second Fz Peak, ant post max, med lat max), and respective time. Minimal values of Fx and Fy (ant post min, med lat min) and duration of stance phase (in seconds). The force values are normalized by BW and time values by total stance phase

Analysis of plantar pressure and balance of transfemoral amputees compared with non-amputee subjects, using their shoes.

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INTRODUCTION:

Lower limb amputation affects daily living, as most of our daily tasks involve standing and moving around the environment. Lower limb amputation as permanent disabling conditions leads to a permanent loss in locomotion and mobility. Posture, a key component of all perception action system, serves to maintain bodily orientation and can be considered as a primary support for the exploration of the environment serving as a mechanical support for action. (1) Prosthetic fitting and rehabilitation contributes to restore the ability of standing and walking in such conditions. In amputees, weight distribution over the feet on standing and walking is altered, and balance and equilibrium are affected. Prostheses can provide good static support, but asymmetry can be observed, frequently, during dynamic locomotion. (2) The knowledge on balance and pressure distribution on amputees and comparison with non amputated subjects may be very helpful on prosthetic fitting and rehabilitation.

METHODS:

9 transfemoral amputees were selected from the population attending CRPG and 18 healthy/active subjects were recruited from FADEUP. The 9 amputees 53 years old (± 16 , 23), physically active- SF36 physical function 62, 78 (± 24 , 89) and 18 non amputees 67 years old (± 8 , 56), physically active - SF36 physical function 82, 33 (± 18 , 01) walked on a 8m walkway, passing over a pressure plate RSScan ® and a force plate Kistler® wearing their own shoes at a self-selected speed. After adaptation, each subject walked six times, three randomly selected, for a three times collection of right and left foot. Kinetic data were recorded with Simi Motion System® (1000 Hz). Fx, Fy and Fz peaks and time, duration of stance and resultant impulse were analyzed using MATLAB®. The pressure and balance data were recorded and analyzed using software package Gait Scientific 3D® and Balance®. Statistics with Minitab software package version 14.0. Due to the normality of the data, parametric tests were used.

RESULTS:

All the subjects participating in this study were physically active and reported no relevant concern with their health as confirmed by the results on SF36 Scales of Physical Function, General Health and Vitality Table 1. Amputees had used prosthesis for at least 2 years and reported to be well adapted to the prosthesis currently in use.

Table 1 - Results on QOL scales Physical Function, General Health and Vitality.
In regards to balance data: total COF traveled way, $\bar{x}$ and $\bar{y}$ values were analyzed. The results are displayed on the graphs and tables bellow.

Subject group	Physical Function	General Health	Vitality
Amputees	62, 78 ($\pm$ 24, 89)	68, 89 ($\pm$ 23, 95)	65,56 ($\pm$ 18, 28)
Non amputees	82, 33 ($\pm$ 18, 01)	59, 64 ($\pm$ 22, 91)	70, 36 ($\pm$ 15, 38)

Balance data

Subject group	COF total traveled way (y)	COF total traveled way (x)
Amputees 508	147 ($\pm$ 70,6)	121,3 ($\pm$ 64,8)
Non amputees	154,1 ($\pm$ 166)	43,2 ($\pm$ 71,9)
	T-value 4,36 pvalue 0,002	T-value 2,82 pvalue 0,01

Regarding pressure, force data and stance time the results are summarized bellow

	Peak Force (%BW)	Peak time (%stance)
Amputees	101 ($\pm$ 6,8)	46, 24 ($\pm$ 19, 27)
Non amputees	103 ($\pm$ 7,3)	57,44 ($\pm$ 7,5)

CONCLUSION:

Prosthesis are usually aligned for a specific shoes, therefore the end result is always a combination of both components.

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ANÁLISE DE MARCHA DE INDIVÍDUOS AMPUTADOS COMPARATIVAMENTE À MARCHA NORMAL UTILIZANDO DINÂMICA INVERSA TRIDIMENSIONAL

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1 INTRODUÇÃO

A determinação das forças e momentos articulares nos membros inferiores têm sido alvo de muitos estudos. Desde o início da década passada, a dinâmica inversa têm sido uma das técnicas mais utilizadas para este fim, considerando-se que não é invasiva e permite a análise de diferentes movimentos como caminhada, corrida e movimentos desportivos. Próteses Ortopédicas são aparelhos e/ou equipamentos que venham substituir partes do corpo humano. As próteses existentes ainda apresentam padrões de movimento muito distantes de um padrão normal, com elevada assimetria entre os membros (Nolan *et al*, 2001) e um custo metabólico maior (Schmalz *et al*, 2002). O estudo dos parâmetros obtidos através da dinâmica inversa permite a avaliação do mecanismo que está associado um determinado movimento bem como dos esforços musculares envolvidos. Desta forma, a utilização da dinâmica inversa pode auxiliar na construção de próteses que proporcionem cada vez mais uma melhoria na qualidade de vida desta população. Dos trabalhos apresentados na literatura até o presente momento, poucos apresentam uma análise tridimensional para avaliar este grupo de indivíduos, o que reforça ainda mais a importância da implementação deste tipo de metodologia.

Sendo assim, o objectivo do presente trabalho é apresentar uma metodologia de comparação entre parâmetros da marcha de indivíduos normais e indivíduos amputados. Para a implementação da metodologia foram utilizadas quatro câmaras sincronizadas com uma frequência de amostragem de 50hz, uma plataforma de forças kistler a 1000Hz. O sistema de aquisição de dados foi o SIMI 7.0; O modelo antropométrico foi o proposto por Vaughn C. (1999) e os dados foram integrados através de rotinas implementadas em ambiente MATLAB.

Para exemplificação da metodologia utilizada foram analisados a marcha de dois sujeitos normais, dois amputados transfemorais e dois transtibiais. Os indivíduos foram solicitados a caminhar sobre a plataforma de forças a uma velocidade seleccionada por eles próprios. Foram analisados duas passadas com cada pé para cada um dos sujeitos.

Foram comparados os valores de Força Articular, Momento Muscular e Amplitude Articular de joelho e tornozelo. As comparações realizadas foram da perna amputada com a perna remanescente e da perna amputada com a marcha dos indivíduos normais.

Os resultados mostram que a metodologia parece adequada à avaliação da marcha humana bem como para a comparação entre indivíduos amputados e indivíduos normais. Os valores obtidos são semelhantes aos apresentados na literatura e as comparações também apresentam resultados semelhantes.

Conclui-se que a avaliação da marcha de amputados utilizando tal metodologia pode vir a ser bastante útil para o melhoramento do alinhamento das próteses e da qualidade de vida dos sujeitos amputados.

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Appendix V: Further results on wedges prescription

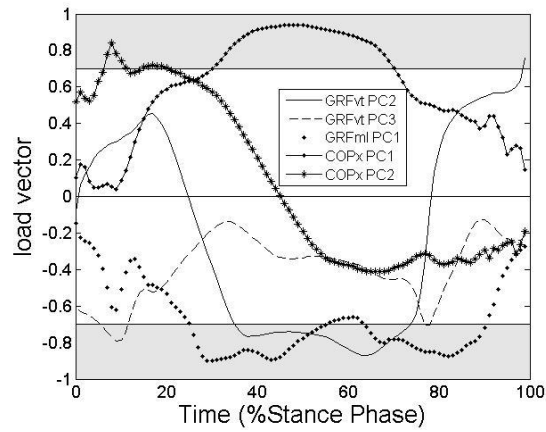
The results presented in Chapter 5 show the prescription of the wedges for 3 TF and some examples of waveforms were shown. In this Appendix, the effects of the wedges in some other waveforms are presented. Table 5.5, reproduced below, summarizes the influence of the wedges in each TF. Also, Figure 5.1, reproduced below, displays the regions where the PCs are relevant in the wedges influence.

Figures V.1 a) and b) show the results using the wedges 1L and 2L for GRFvt and 1L and 2P for GRFml. In Table 5.5 one saw that these wedges reduce the Mahalanobis distance, making it closer to CG 95% CI. GRFvt is affected in PC3 and the load vector is relevant from 8-12% of the stance phase (Figure 5.1). GRFml is relevant from 30-50% and 70-90%. In both situations, the wedges were able to improve TF gait waveforms taking them closer to CG. COPx from subject 5 was already presented in Chapter 5.

Reproduction of Table 5.5: Phase IV results: Mahalanobis distance (T^2) calculated before wearing the wedges (see table 5.3) and with the wedges proposed for SL.

subject	variables	SL (before)	wedges prescribed					
			1L	2L	1M	2M	1P	2P
5	GRFvt	29.80✓	23.31✓	22.41✓	11.27	25.93✓	n.a.	n.a.
	GRFml	32.48✓	31.14✓	n.a.	n.a.	n.a.	33.34✓	27.96✓
	COPx	19.79✓	27.45✓	n.a.	n.a.	n.a.	n.a.	23.50✓
7	GRFvt	n.a.	n.a.	n.a.	n.a.	n.a.	n.a.	n.a.
	GRFml	18.71	42.63	n.a.	n.a.	n.a.	323	27.80✓
	COPx	n.a.	n.a.	n.a.	n.a.	n.a.	n.a.	n.a.
9	GRFvt	10.68	19.64✓	18.36✓	14.67	39.09	n.a.	n.a.
	GRFml	n.a.	n.a.	n.a.	n.a.	n.a.	n.a.	n.a.
	COPx	14.33	25.38✓	n.a.	n.a.	n.a.	n.a.	n.a.

✓: Inside normal range of 95% confidence interval of CON (see Table 5.3); n.a.: not applicable.



Reproduction of Figure 8.1: a) load vectors from GRFvt PC2, GRFvt PC3, GRFml PC1, COPx PC1 and COPx PC2; the grey area highlights the 0.71 threshold proposed by Knapp & Comrey (1973);

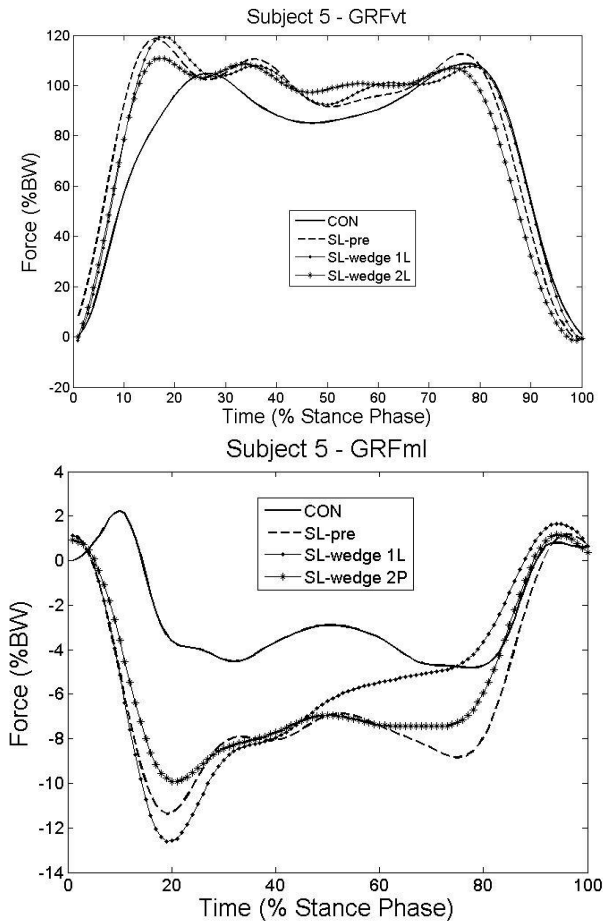


Figure V.1: Subject 5 influence of the wedges prescribed: CON group mean waveform, SL before the prescription and with two of the wedges prescribed, for GRFvt and GRFml waveforms.

In relation to Subject 7, a “bad” example is shown. The *Mahalanobis* distance for GRFml was outside the the 95% CI (Table 5.5). Figure V.2 shows that using the 1L wedge, the waveform is further out from CON mean waveform, implying this wedge “underperformed” and worsened the gait pattern. The remaining results of Subject 7 from GRFvt were considered as “not applicable” and those from COP were already shown in Chapter 5.

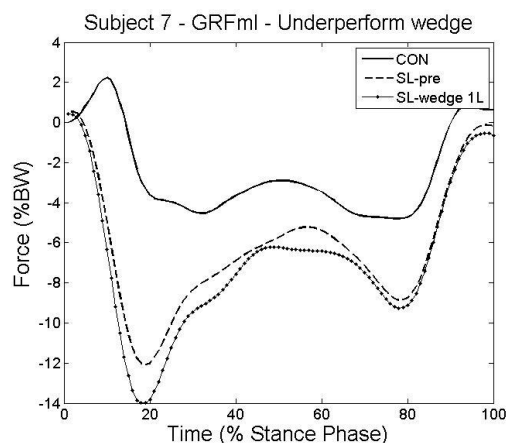
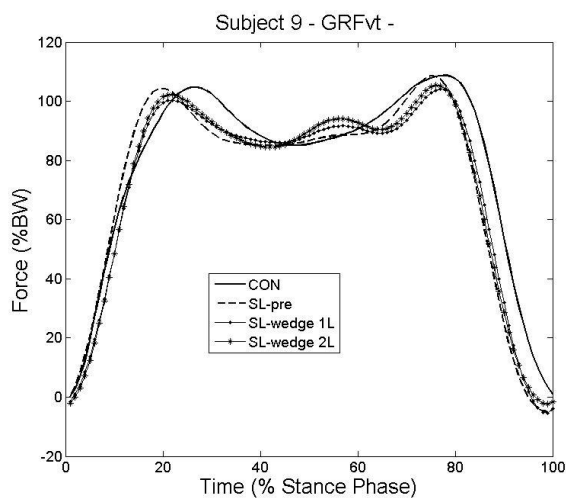


Figure V.2: Underperform wedge: GRFml waveform before and after the wedge prescription

Subject 9 also confirms the results, showing that the wedges prescribed (1L and 2L for GRFvt and 1L for COPx) improve the gait patterns, where the waveforms become more similar to CON (Figure V.3).



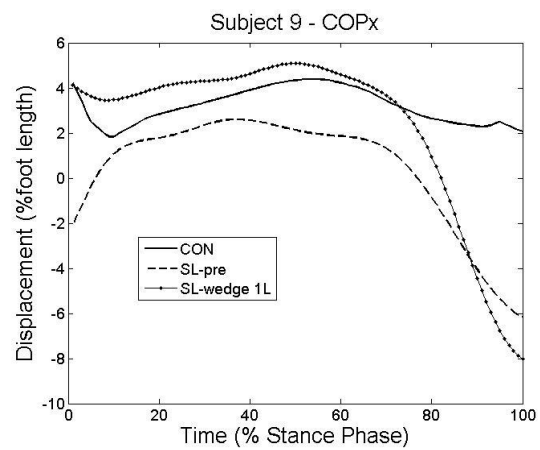


Figure V.3: Subject 9 waveforms before and after the wedges prescription.

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