



*BIOMECHANICS OF THE LOADED GAIT (BACKPACKERS &
OBESSE PEOPLE), TESTING OF PRESSURE RELIEF INSOLES
AND RELIABILITY OF A NEW GAIT ANALYSIS DEVICE*

Academic dissertation submitted with the purpose of obtaining a doctoral degree in Sports Sciences according to the Decree-Law nº. 74/2006 March, 24th. The thesis supervisors are Prof. Dr. João Paulo Vilas-Boas and Prof. Dr. Leandro José Rodrigues Machado

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Resumo

O padrão de caminhada de pessoas submetidas a sobrecarga tanto de maneira ocasional (p. ex. mochileiros) como permanente (p. ex. pessoas obesas), apresenta alterações nos aspectos biomecânicos quando comparado ao padrão de caminhada de pessoas sem sobrepeso. Possivelmente, tais alterações contribuem para os altos índices de lesões músculo-esqueléticas encontrados nestas populações. Assim, o aprofundamento acerca das características das forças a que tais populações estão submetidas durante a caminhada pode auxiliar no desenvolvimento de medidas preventivas e no estabelecimento de programas de atividade física mais seguros e eficientes. De entre os diversos dispositivos usados para melhor distribuir as pressões plantares, as palmilhas mostram-se ser opções apropriadas para este fim. Porém, pouco se sabe sobre o desenvolvimento e validade de tais dispositivos às populações submetidas a carga (ocasional e permanente) durante a caminhada. Um outro tópico importante para maior compreensão acerca da influência da carga na biomecânica da caminhada, seria explorar dispositivos que permitissem realizar análises de maneira contínua ao longo do dia-a-dia. Neste sentido, esta tese apresenta os seguintes objetivos: (i) analisar a influência da sobrecarga e da velocidade nas forças de reação do solo e pressões plantares durante a caminhada; (ii) desenvolver e testar experimentalmente palmilhas para melhor distribuir as pressões plantares para populações submetidas a sobrecarga; e (iii) verificar o rigor e a repetibilidade de um novo dispositivo para análise da caminhada (WalkinSense®). Para o objetivo “i”, foram selecionados 77 participantes, para o objetivo “ii” 30 participantes; e para o objetivo “iii”, 40 participantes. Os objetivos “i” e “ii” foram contemplados por meio de uma plataforma de forças Bertec® (Bertec Corporation) e um sistema de análise de pressões plantares F-scan® (TekScan); enquanto que para o objetivo “iii” dois dispositivos WalkinSense®, o sistema Pedar® (Novel) e uma prensa de calibração Trublu® (Novel) foram usados. Foram identificadas alterações tanto em magnitude quanto no comportamento das forças de reação do solo e das pressões plantares nos mochileiros e nas pessoas obesas comparados aos participantes sem sobrepeso. Tais parâmetros mostraram-se ser relevantemente influenciados pelas alterações da velocidade da caminhada. Foram desenvolvidas duas palmilhas para a caminhada sobrecarregada. A avaliação experimental das mesmas indicou que uma delas foi eficiente no sentido de minimizar os picos de pressão plantar durante a caminhada. O dispositivo analisado mostrou-se rigoroso e com excelente repetibilidade.

Abstract

The gait pattern of people either occasionally (i.e. backpackers) or permanently (i.e. obese people) loaded shows biomechanical changes compared to normal-weight subjects. Possibly, these changes contribute to the high levels of musculoskeletal injuries describe for these populations. Thus, a better knowledge about the features of the forces which these loaded populations receive during walking might be helpful for developing preventive measures and to establish exercise protocols safer and more efficient. Among the various devices used to improve the plantar pressure distribution while walking, the insoles have been shown to be powerful. However, there are scarce information about the development and experimental testing of these kinds of gait aids for people under occasional or permanent load. Since the most devices for gait analysis need a laboratorial setup, the ecological validity of these analyses is compromised. Devices which allow the biomechanical analysis during daily life activities, such as walking, seem to be important. Therefore, this PhD thesis has the following aims: (i) to analyze the influence of the load (occasional and permanent) and the speed gait on the ground reaction forces (GRF) and plantar pressure distribution while walking; (ii) to verify the influence of two pressure relief insoles developed for loaded population on the GRF and plantar pressure peaks during occasional and permanent loaded gait; and (iii) to verify the accuracy and repeatability of a new gait analysis device (WalkinSense® - Tomorrow Options). For accomplishing the aim “i”, 77 adult participants (60 normal-weight and 17 obese); for the aim “ii”, 30 participants (20 normal-weight and 10 obese); and for the aim “iii” 40 normal-weight individuals were included. For the aims “i” and “ii” a force plate Bertec® (Bertec Corporation), an in-shoe pressure system F-scan® (TeckScan) were used; whereas for the aim “iii”, two WalkinSense® devices, an in-shoe pressure system Pedar® (Novel) and a calibration bench Trublu® (Novel) were used. Alteration in the magnitude and pattern of the GRF and plantar pressures were identified for both backpackers and obese people compared to the normal-weight participants while walking. Also, these parameters were shown to be influenced by changing gait speed. Two pressure relief insoles were developed for loaded population. The experimental tests indicated that one of them was powerful in order to relieve the pressure peaks during loaded gait. The plantar pressure parameters of the WalkinSense® were found to be repeatable and accurate.

List of Abbreviations

Abs	absolute
BMI	body mass index
BpG	backpack's group
CG	control group
CI	confidence interval
Ct	central
Fap1	braking peak
Fap1 _{imp}	braking impulse
Fap2	propulsive peak
Fap2 _{imp}	propulsive impulse
FEM	finite element model
FF	forefoot
Fml	medial-lateral peak
Fml _{imp}	medial-lateral impulse
Fz1	load acceptance peak
Fz1 _{imp}	load acceptance impulse
Fz2	thrust peak
Fz2 _{imp}	thrust impulse
Fz _{Min}	minimum between peaks
GRF	ground reaction forces
G _{Toe}	great toe
Hlx	hállux
ICC	intraclass correlation coefficient
Lat	lateral
Med	medial

MF	midfoot
Norm	normalized
NW	normal-weight
OG	group of obese people
ORIGINAL _{COND}	original condition
P	significant level
P_{Integral}	pressure-time integral
P_k	peak
$P_k A_{P_B}$	braking peak
$P_k A_{P_P}$	propulsive peak
P_{Mean}	mean pressure
P_{Peak}	peak pressure
P_{Time}	peak pressure time
RF	rearfoot
SD	standard deviation
SHOE-ONLY _{COND}	shoe-only condition
SLS1	stress-less shoe insole 1
SLS2	stress-less shoe insole 2
Time	time of the event
TMVt	thrust maximum
Toes	2 nd , 3 rd , 4 th and 5 th toes
UNLOADED _{COND}	unloaded condition
V_{t_i}	impact peak
$V_{t_{\text{Min}}}$	minimum between the peaks
η_p^2	partial Eta square

CHAPTER 1

GENERAL INTRODUCTION

In terms of load carriage, the backpack is often the favorite option, either by the conviction that it is a very comfortable solution, or because it is considered a healthy and practical way. In addition to the hikers, other groups also use backpacks, among them: students fill backpacks with books and stationery, postmen with mails, while soldiers load them with tents and supplies (Al-Khabbaz et al., 2008).

The backpack seems to be an appropriate way to carry load because the load is positioned close to the body's center of gravity while maintaining stability (Chansirinukor et al., 2001; Datta & Ramanathan, 1971; Legg, 1985). Studies analyzing physiological aspects of load carriage indicate that the energy cost increases progressively with increases in backpack load (Knapik et al., 1996). This is possibly related to biomechanical changes: such as those found in the temporal (Birrell & Haslam, 2009, 2010; Simpson et al., 2012), kinetic (Birrell & Haslam, 2010; Birrell et al., 2007; Goh et al., 1998), kinematics (Attwells et al., 2006; Birrell & Haslam, 2010; Birrell et al., 2007; Chansirinukor et al., 2001; Hong & Cheung, 2003; Simpson et al., 2012), and electromyographic (Al-Khabbaz et al., 2008; Ghori & Luckwill, 1985; Harman et al., 1992; Simpson et al., 2011a) aspects of the gait. These alterations are likely to contribute to the high level of discomfort and injury associated with this condition (Birrell & Haslam, 2009; Grimmer & Williams, 2000; Knapik et al., 1996; Negrini et al., 1999; Simpson et al., 2011b; Skaggs et al., 2006).

Beyond this group (backpackers), there is another one that it is worryingly emerging in the industrialized societies in the XXI century: the obese people (Au & Low, 2004). This condition is associated with metabolic disease and diabetes, increasingly sedentary lifestyle, changing in dietary habits, the increase of leisure time, and offer of quick solutions of (poor quality) food; the obesity has grown exponentially worldwide, and it becomes a common situation in Europe, including Portugal (Do Carmo et al., 2008). This condition gives rise to a social group also characterized by "carrying a load" coupled to the body; however, in this case the load is carried permanently.

Obesity is related to a range of disabling musculoskeletal conditions in adults (Anandacoomarasamy et al., 2007). The repetitive overload in obese people while walking has been implicated in the predisposition to pathological gait patterns, loss of mobility and subsequent progression of disability (Messier et al., 1996), as well as higher risk of hip and knee osteoarthritis (Felson, 1990; Hochberg et al., 1995; Ko et al., 2010), increase of the likelihood of foot ulceration (Vela et al., 1998), and heel pain (Prichasuk, 1994).

The role of physical exercise for the treatment and prevention of obesity has been increasingly highlighted. Walking is one of the most often prescribed exercises for controlling and treating obesity. However, this activity can be critical in biomechanical terms (Browning & Kram, 2007). The walking velocity is one important factor that causes influence on the biomechanical parameters. Nevertheless, the walking speed usually is prescribed only regarding physiological criteria, which are undoubtedly important, but disregard or ignore the different mechanical loads on the different walking speeds can be risky for populations that are already in a condition far from the natural one. Just a few studies (Browning & Kram, 2007; DeVita & Hortobágyi, 2003) investigated the influence of walking velocity on kinetic parameters during walking of obese adults and, to the best of our knowledge, none of them assessed the plantar pressures. The knowledge of the influence of speed on the biomechanical aspects of loaded gait might be helpful to preserve the integrity of the musculoskeletal system.

The knowledge of the distribution of forces on the foot along the stance phase seems to be essential to detecting overloaded regions. The evaluation of the plantar pressures allows assessing the function of the ankle or foot while walking, and other functional activities, as they are responsible for providing the support and flexibility needed to sustain the weight transfer (Cavanagh & Ulbrecht, 1994). On the other hand, baropodometric systems do not provide any information regarding the shear forces. The analysis of the ground reaction forces (GRF) provides global information about the vertical and shear stress forces during gait, whereas the plantar pressure analysis identifies the distribution of the vertical GRF over the plantar foot surface. The combination of both analyses may provide more detailed information about specific features of forces acting on the foot while walking. Such information recorded simultaneously could provide a better comprehension of the biomechanical strategies of the locomotor system when loaded. It could be helpful in order to develop strategies, such as insole or gait training, for minimizing the consequences that the occasional or permanent loads impose on the musculoskeletal system.

Foot orthoses is a general term to describe a broad range of devices including heel lifts, lateral/medial wedges, or insoles (custom-made or prefabricated) (Chevalier & Chockalingam, 2012). These devices have been shown to be effective for managing many foot problems (Bonanno et al., 2011; Colagiuri et al., 1995; Cronkwright et al., 2011; Lynch et al., 1998; Sasaki & Yasuda, 1987). They can reduce and redistribute plantar foot pressure and subsequently avoid or decrease foot pain (Burns et al., 2007). However, the exact mechanisms by which foot orthoses work are yet to be fully

understood (Chevalier & Chockalingam, 2012), and the biomechanical effect of these devices is far from the simplistic model often proposed in a clinical context (Nester et al., 2003). Moreover, there is a need to establish the most suitable shoes/foot orthoses across the clinical populations (Rao et al., 2012). The development and respective experimental tests of insoles designed specifically for these potential harmful populations might be helpful to understand the mechanisms of foot orthoses action, to perform physical exercise more safely and to prevent injuries.

In the field of physical activity monitoring, there is a constant shift in the paradigm for the assessment of human movement and performance. Empowered by the fast paced development of portable and wearable technology, research in this field can now take place in real life scenarios, under everyday and long term conditions, as opposed to short term, laboratory or otherwise controlled experiments. This trend towards the use of wearable monitoring and recording equipment, seamlessly attached to the human body, allows effortless data capturing without disturbance or discomfort to the subject under observation (Pantelopoulos & Bourbakis, 2010). However, for these equipments to be widely accepted as research or clinical tools, they have to be validated against well known and established methods and instruments (Bland & Altman, 1986). The WalkinSense® is one such type of equipment designed for activity monitoring, combined with plantar pressure evaluation and analysis of temporal gait parameters. By means of this device, the loaded gait could be assessed in more realistic conditions, such as every day walking.

Considering the mentioned aspects of the loaded gait the main aims of this study were:

- To describe and compare the GRF, plantar pressures, and temporal parameters in terms of absolute and normalized (by the body weight) values of normal-weight and occasional loaded participants (backpackers);
- To describe and compare the GRF, plantar pressures, and temporal parameters in terms of absolute and normalized values of normal-weight and permanent loaded participants (obese people);
- To analyze the influence of gait speed and obesity on the GRF and plantar pressure parameters during walking;

- To verify the influence of two pressure relief insoles designed for loaded population on GRF and plantar pressure peaks during walking of subjects under occasional or permanent load conditions;
- To verify the accuracy and repeatability of a new gait analysis device: WalkinSense®.

As the present PhD thesis adopted the Scandinavian Model structure, every aim was developed and showed by a manuscript structure (chapters two to six). All the manuscripts were either published or have been submitted and are under review. Some preliminary investigations and further researches related to the main aims of this thesis were also carried out. These studies were considered supplementary information and were included as appendixes. Their aims were:

- To compare the vertical GRF captured by a force plate with those obtained by a pressure plate and an in-shoe pressure system;
- To analyze the influence of a backpack on human gait: a preliminary study;
- To compare the GRF among normal-weight, backpackers and obese young adults during self-selected walking speed;
- To assess the speed influence on the kinetic aspects of backpacker's gait;
- To perform a preliminary validation of the WalkinSense® device, under controlled conditions and evaluate the repeatability of its temporal-spatial gait parameters, as well as the accuracy of its distance measure with respect to ground truth data.

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CHAPTER 2: Original Research

**GROUND REACTION FORCES AND PLANTAR
PRESSURE DISTRIBUTION DURING OCCASIONAL
LOADED GAIT**

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Abstract

This study compared the ground reaction forces (GRF) and plantar pressures between unloaded and occasional loaded gait. The GRF and plantar pressures of 60 participants were recorded during unloaded gait and occasional loaded gait (wearing a backpack that raised their body mass index to 30); this load criterion was adopted because is considered potentially harmful in permanent loaded gait (obese people). The results indicate an overall increase (absolute values) of GRF and plantar pressures during occasional loaded gait ($p < 0.05$); also, higher normalized (by total weight) values in the medial midfoot and toes, and lower values in the lateral rearfoot region were observed. During loaded gait the magnitude of the vertical GRF (impact and thrust maximum) decreased and the shear forces increased more than did the proportion of the load (normalized values). These data suggest a different pattern of GRF and plantar pressure distribution during occasional loaded compared to unloaded gait.

Keywords: Backpack; Ground reaction forces; Loaded gait; Load carriage; Plantar pressure.

2.1. Introduction

The backpack seems to be an appropriate way to carry load because the load is positioned close to the body's center of gravity while maintaining stability (Chansirinukor et al., 2001; Datta & Ramanathan, 1971; Legg, 1985). It has been widely used for different purposes: students fill backpacks with books and stationery, while hikers and soldiers load them with tents and supplies (Al-Khabbaz et al., 2008). Studies analyzing physiological aspects of load carriage indicate that the energy cost increases progressively with increases in backpack load (Knapik et al., 1996). This is possibly related to changes in the biomechanical aspects of gait. Kinetic analyses found increases in magnitude of vertical and anterior-posterior ground reaction forces (GFR) (Birrell & Haslam, 2010; Birrell et al., 2007; Simpson et al., 2012) and in peak lumbosacral forces (Goh et al., 1998). Kinematic analyses indicate increases in knee range of motion (Attwells et al., 2006; Birrell & Haslam, 2010; Birrell et al., 2007; Simpson et al., 2012) and hip flexion (Attwells et al., 2006; Birrell & Haslam, 2010), while hip abduction and rotation decreases (Birrell & Haslam, 2010). An increased forward lean of the trunk and forward position of the head also was found (Attwells et al., 2006; Chansirinukor et al., 2001; Hong & Cheung, 2003). A longer double support time and duration of stance phase (Birrell & Haslam, 2010) as well as decreased step length (Birrell & Haslam, 2009; Simpson et al., 2012) compared to no load gait were evidenced. Increasing load also has been shown to alter lower limb muscle activity (Ghori & Luckwill, 1985; Harman et al., 1992; Simpson et al., 2011a) and rectus abdominal muscle activity (Al-Khabbaz et al., 2008).

The mentioned alterations possibly contribute to a significant association between the backpack weight and occurrence of back pain (Grimmer & Williams, 2000; Skaggs et al., 2006). Simpson et al. (2011b) found loads of 20, 30 and 40% of the body weight inducing significant changes in posture, self-reported exertion and shoulder discomfort in female hikers. Johnson et al. (1995) showed that as load increased, fatigue and muscle discomfort intensified, and alertness and feelings of well-being diminished in military personnel during road marches. Negrini and Carabalona (1999) found 65.7% of school children felt that carrying a backpack causes fatigue. A significant relationship was found between fatigue and back pain (Negrini et al., 1999). The higher muscular tensions necessary to sustain these charges have also been associated with injury, muscle strain and joint problems (Birrell & Haslam, 2009). Rucksack palsy is another injury related to load carriage (Knapik et al., 1996).

It is common the occurrence of lower limb injuries as a consequence of backpack usage. Carrying heavy loads such as 20 kg seems to contribute to second metatarsal stress fractures (Arndt et al., 2002) and may play a role as a co-factor in plantar fasciitis onset (Wearing et al., 2006). Analyzing strenuous conditions such as walking long distances (20 km) with heavy loads (45 kg), the incidence of metatarsalgia and knee pain were found from 3.3% to 20% and 0.6e15%, respectively (Knapik et al., 1992). The most common load-carriage-related injury is foot blisters (Cooper, 1981; Knapik et al., 1992). Their development is believed to be a consequence of increasing pressure on skin and the creation of more friction between the foot and shoe through higher propulsive and braking forces (Kinoshita, 1985; Knapik et al., 1992). As the abnormal force application over the plantar surface of the foot may be an important factor in the development of many of the mentioned injuries (Arndt et al., 2002; Knapik et al., 1996; Wearing et al., 2006), the knowledge of the GRF and plantar pressure distribution over the foot may help to better understand these pathological conditions as well as to prevent and treat them.

The influence of backpack load on plantar pressure distribution was assessed by Rodrigues et al. (2008) and Pau et al. (2011) , who analyzed school children during quiet stance upward position. The former study did not find influence of load (5, 10 and 15% of the body weight) on plantar force distribution, whereas the latter found higher plantar peak pressures in midfoot and rearfoot regions (20 to 30%) while children carried their own backpacks (not a controlled load). Regarding the influence of carriage load on pressure distribution along the plantar surface on dynamic conditions such as walking, little is known. The previous studies have been interested only in the GRF. The GRF analysis does not provide information on where the forces are acting on the foot. The combined analysis of the horizontal and vertical GRF and pressure data (distribution of the vertical GRF along the plantar surface) provide more detailed information about characteristics of the forces acting on the human body. Therefore, the aim of the present study was to compare the GRF and plantar pressure parameters between unloaded and loaded gait. These data may help in developing strategies, such as special insole or gait training, to minimize the impact that load carriage seems to have on the locomotor system.

2.2. Methods

Participants

A convenience sample of 60 (30 males, 30 females) sport science students (mean age of 23.0 ± 3.7 years old, mean height of 1.68 ± 0.10 m and mean body mass of 67.8 ± 11.2 kg) participated in this study. All were physically active and had body mass indexes (BMIs) lower than 25. Participants were excluded from this study if they showed any traumatic-orthopedic dysfunction or difficulty with independent gait. This research was approved by a local ethics committee and all participants freely signed an informed consent form, based on the Helsinki declaration, which explained the purpose and the procedures of the study.

Apparatus

Bertec force plate model 4060-15, operating at 1000 Hz, an amplifier signals system model AM 6300 (Bertec Corporation, Columbus, Ohio, USA), and a Biopac analog-digital converter (BIOPAC System, California, USA) were used to capture GRF. F-Scan in-shoe pressure system (TekScan, South Boston, USA) operating at 300 Hz with about 960 pressure cells, 3.9 sensors / cm^2 and a 0.18 mm thick insole sensor were used to capture plantar pressure data. Three digital cameras were used for visual inspection, if necessary.

Tasks and procedures

The participants underwent three phases: preparation, familiarization and testing. In the first phase the procedures that would be performed were explained to the participants and their weight and height were recorded. For each participant, the amount of additional weight needed to raise the BMI to 30 was calculated, and then a backpack was filled with corresponding amount of sand and fixed at the central area of the back. The loads placed inside the backpack ranged from 14.1 to 30.1 kg (mean load 20.3 ± 4.4 kg). For the school children population, 10 to 15% of the body mass is considered the load limit for the backpack in order to prevent impairment (Lindstrom-Hazel, 2009). Based on changes in muscle activity, posture and self-reported exertion and discomfort, a load limit of 30% of the body mass was suggested for female recreational hikers (Simpson et al., 2011a, 2011b). The I Class Obesity (BMI > 30) is a well documented risk factor for traumatic-orthopedic injuries being considered as possible threshold for such dysfunctions (Ko et al., 2010; WHO, 2000). The traditional methods of load normalization are body mass percentage and fixed load approach.

Even with the differences in body mass distribution between obese people and backpackers, we have opted to use the threshold established for the permanent loaded population (obese) in order to assess the effect a harmful load applied occasionally presents on the musculoskeletal system.

A cuff unit measuring 98 x 64 x 29 mm with Velcro straps was attached on the lateral malleolus region of both legs of each participant: a 9.25 mm cable linked the cuff to the VersaTek hub (F-Scan system), which was connected to a computer. The cable did not cause any restriction of the gait. A pair of thin socks and, aiming to minimize the effects of different soles, neutral shoes (ballet sneakers) with sensor insoles inside was provided for every participant. During the familiarization, the participants walked freely (without backpack) over a 6 m walkway with a force plate embedded in the middle. The researcher identified the starting position for the participant so the right foot would hit the force plate without altering the gait. In the last phase, the participants performed three valid trials without backpack (unload condition, which was called control group – CG) and three with backpack (loaded condition, which was called backpacker's group – BpG). They walked looking forward with a self-selected speed and performed, at least two steps before and after reaching the plate. The trials were considered valid when the subjects reached the plate with the whole foot over it without altering their gait pattern.

Data Analysis

The Acknowledge software (BIOPAC System, California, USA) was used to acquire the GRF. The F-Scan Research 6.33 software (TekScan, South Boston, USA) was used to acquire the plantar pressure data. The GRF and plantar pressure data (values of each sensor in each frame) were exported to Matlab 7.0 software (MathWorks, Massachusetts, USA). A program was developed for processing and calculation of the analyzed variables.

All force and pressure variables were shown in absolute values and normalized by the total weight (body mass for CG and body mass plus backpack mass for the BpG), while all time variables were normalized by the stance phase. The systems were synchronized by an external trigger that started them together.

The dependent variables from the GRF data were calculated for absolute ($_{Abs}$) and normalized ($_{Norm}$) values and time ($_{Time}$), respectively, for the following events:

- Impact peak ($PkVt_{t_{Abs}}$, $PkVt_{t_{Norm}}$ and $PkVt_{t_{Time}}$): the highest value of the vertical GRF at the first half of the stance phase (first peak);

- Thrust maximum ($TMV_{t_{Abs}}$, $TMV_{t_{Norm}}$ and $TMV_{t_{Time}}$): the highest value of the vertical GRF found at the second half of the stance phase (second peak);
- Minimum between the peaks ($V_{t_{Min_Abs}}$, $V_{t_{Min_Norm}}$ and $V_{t_{Min_Time}}$): minimum value of the vertical GRF between the PkV_{t_i} and TMV_{t_i} ;
- Braking peak ($PkAP_{B_Abs}$, $PkAP_{B_Norm}$ and $PkAP_{B_Time}$): the highest value (negative) of the anterior-posterior GRF at the first half of the stance phase;
- Propulsive peak ($PkAP_{P_Abs}$, $PkAP_{P_Norm}$ and $PkAP_{P_Time}$): the highest value (positive) of the anterior-posterior GRF found at the second half of the stance;
- Medial-lateral peak ($PkML_{Abs}$, $PkML_{Norm}$ and $PkML_{Time}$): the highest value of the medial-lateral GRF during the stance phase.

Considering in-shoe pressure data, first the program divided the foot into 10 regions as proposed and adapted from previous studies (Cavanagh & Ulbrecht, 1994; Gurney et al., 2008). The boundary between the rearfoot (RF) and midfoot (MF) was located at 73% of the foot length (from toes to heel). The RF was divided into three equal parts (33% each). The boundary between the MF and forefoot (FF) was located at 45% along the foot length. The MF was divided into two equal parts (50% each). The FF was divided into three regions: 30% medial (first metatarsal region), 25% central (second metatarsal region) and 45% lateral (lateral metatarsals region). The other two regions were the Hallux (Hlx) and lesser toes (Toes) (2nd, 3rd, 4th and 5th toes). The sensor peak (Pk), which was defined as the sensor that presented the highest pressure value, and the time of its occurrence were calculated for every region. The Pk data were calculated to absolute (_{Abs}) and normalized (_{Norm}) values. Thus, the follow dependent variables were calculated: medial RF ($PkRF_{Med_Abs}$, $PkRF_{Med_Norm}$ and $PkRF_{Med_Time}$); central RF ($PkRF_{Ct_Abs}$, $PkRF_{Ct_Norm}$ and $PkRF_{Ct_Time}$); lateral RF ($PkRF_{Lat_Abs}$, $PkRF_{Lat_Norm}$ and $PkRF_{Lat_Time}$); medial MF ($PkMF_{Med_Abs}$, $PkMF_{Med_Norm}$ and $PkMF_{Med_Time}$); lateral MF ($PkMF_{Lat_Abs}$, $PkMF_{Lat_Norm}$ and $PkMF_{Lat_Time}$); medial FF ($PkFF_{Med_Abs}$, $PkFF_{Med_Norm}$ and $PkFF_{Med_Time}$); central FF ($PkFF_{Ct_Abs}$, $PkFF_{Ct_Norm}$ and $PkFF_{Ct_Time}$); lateral FF ($PkFF_{Lat_Abs}$, $PkFF_{Lat_Norm}$ and $PkRF_{Lat_Time}$); hallux ($PkHlx_{Abs}$, $PkHlx_{Norm}$ and $PkHlx_{Time}$); and lesser toes ($PkToes_{Abs}$, $PkToes_{Norm}$ and $PkToes_{Time}$). The initial and final double limb stance (as percentage of stance phase) was calculated as well. The program automatically divided the plantar regions: all divisions were checked by two trained researchers and, if necessary (eventually), corrected manually.

The in-shoe pressure system presents good information about relative distribution of plantar forces while their absolute values have been questioned (Nicolopoulos et al., 2000; Rosenbaum & Becker, 1997; Woodburn & Helliwell, 1996).

The force plate is considered the most accurate dynamic measurements of force (Cobb & Claremont, 1995): thus, the force plate was used to calibrate (post-test) the plantar pressure data test by test.

Statistical Analysis

The intra-individual repeatability for the variables $PkFF_{ct_Abs}$, $PkRF_{ct_Abs}$, $PkVt_{l_Abs}$ and duration of stance phase was verified by means of intra-class correlation coefficient (ICC). The mean of the three trials of each subject was computed and all the statistical procedures were performed with these mean values. The normality of the data was verified by the Kolmogorov-Smirnov test and the homogeneity of the variances using Levene's test. Nine out of the 102 sets of value calculated (48 for each group) did not show normal distribution ($PkHlx_{Abs}$ in both groups, $PkRF_{lat_Abs}$, $PkRF_{ct_Abs}$ and $PkRF_{med_Anb}$ in CG, $PkRF_{med_Abs}$ and $PkToes_{Norm}$ in BpG, and $PkML$ in both groups). The natural logarithmic transformation was done for these variables and the transformed values were used in inferential statistics tests. To compare the variables between the groups (CG vs. BpG), the paired Student's t-test was used. The significance level was $\alpha = 0.05$. The statistical procedures were made using SPSS software (v.17; SPSS Inc, Chicago, IL, USA).

2.3. Results

An excellent data repeatability was found. The variables $PkFF_{ct_Abs}$, $PkRF_{ct_Abs}$, $PkVt_{l_Abs}$, and duration of stance phase showed ICC of 0.98 ($CI_{95\%}$ 0.97 – 0.99), 0.97 ($CI_{95\%}$ 0.95 – 0.98), 0.86 ($CI_{95\%}$ 0.78 – 0.91) and 0.94 ($CI_{95\%}$ 0.90 – 0.96), respectively.

The duration of the stance phase and the initial double limb stance were longer during BpG gait compared to CG, while the final double limb stance did not show statistical differences (Table 1).

Table 1. Mean, standard deviation (SD) and significant level (p) of time variables.

Variables	Control Group		Backpacker's Group		p
	Mean	SD	Mean	SD	
Duration of stance phase (s)	0.787	0.064	0.813	0.069	0.005
Initial double limb stance (% stance phase)	22.969	4.616	24.836	5.086	0.003
Final double limb stance (% stance phase)	25.577	5.362	26.667	4.306	0.124

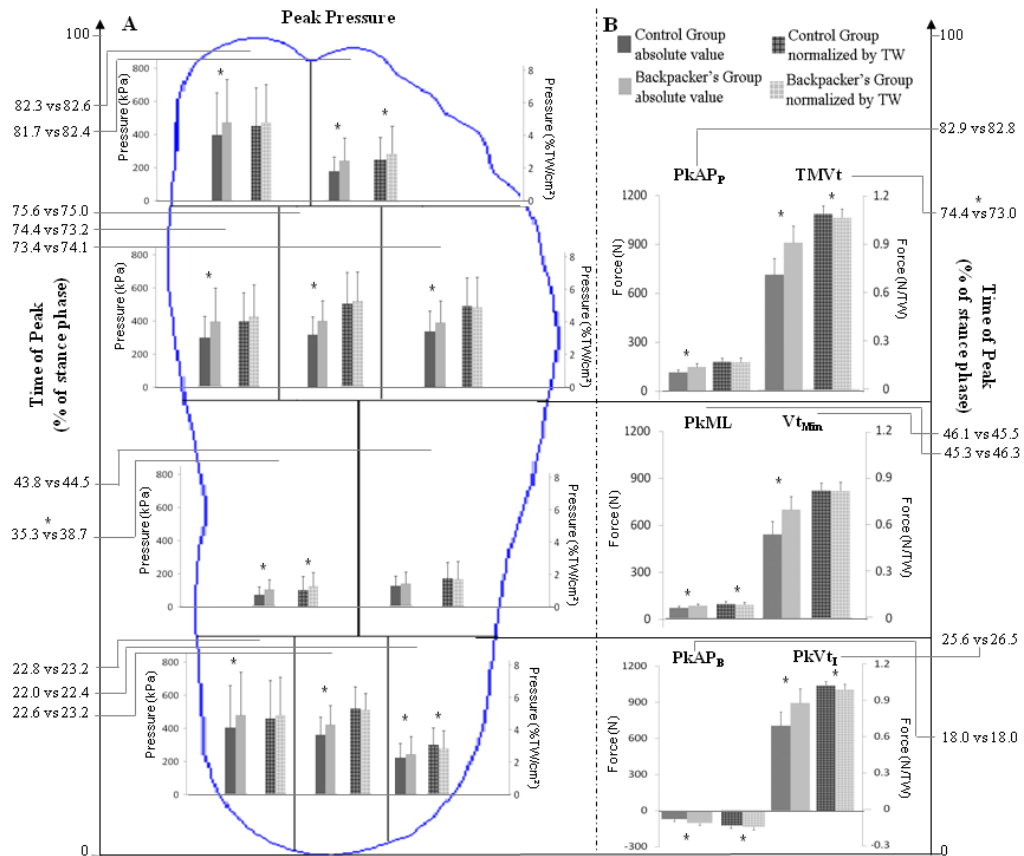


Figure 1A. Peak pressure (absolute and normalized values) of each foot's region and respective time. **B.** Force (absolute and normalized values) and respective time of the main events of ground reaction force (GRF). PkVt_I - impact peak of GRF vertical component; Vt_{Min} - minimum between the peaks of GRF vertical component; TMVt - thrust maximum of GRF vertical component; PkAP_B – braking peak GRF anterior-posterior component; PkAP_P - propulsive peak GRF anterior-posterior component; TW – total weight (in control group is equal to body weight and in backpacker group equal body weight plus backpack weight); Y axis represents time of the events to control group (first value) and backpacker group (second value). * - statistical difference with $p < 0.05$.

In the BpG, except for the MF, nine out of 10 plantar regions showed significantly larger absolute pressure values compared to CG (Figure 1A). The larger sensor peak magnitudes in BpG occurred in HLx, RF_{Ct} and FF_{Ct} with values of 471.99 ± 260.56 kPa, 419.00 ± 117.25 kPa and 403.26 ± 121.01 kPa, respectively. In the CG they occurred in HLx, RF_{Ct} and FF_{Lat} with values of 397.39 ± 255.05 kPa, 356.72 ± 108.20 kPa and 335.41 ± 124.15 kPa, respectively. Considering the normalized values, the BpG presented larger values in PkToes_{Norm} and MF_{Med}_{Norm} while lower magnitudes in PkRF_{Lat}_{Norm} compared to CG (Figure 1A). The largest absolute differences occurred

in FF_{Med} , RF_{Med} and RF_{Ct} , and the largest normalized differences occurred in Toes, FF_{Med} Hlx with the BpG showing always higher values compared to CG (Table 2).

In all GRF events the BpG presented significantly larger absolute forces compared to CG (Figure 1B and Table 2). Considering normalized values, BpG presented higher normalized values for $PkAP_{B_Norm}$ and lower values for $PkVt_{L_Norm}$, $PkML_{Norm}$ and $TMVt_{Norm}$ compared to CG (Figure 1 and Table 2). In BpG $PkMF_{Med_Time}$ occurred later and $TMVt_{Time}$ earlier compared to CG. No differences were found for the other time variables (Figure 1B).

Table 2. Mean, standard deviation (SD), confidence interval and significant level (p) of the differences between CG and BpG for all force and pressure variables.

Variables	ABSOLUTE DATA					NORMALIZED DATA				
	Mean	SD	Confidence Interval		p	Mean	SD	Confidence Interval		p
<i>Force</i>	(N)		Lower	Upper		(N/TW)		Lower	Upper	
$PkVt_i$	-177.262	74.480	-196.846	-157.679	< 0.001	0.032	0.048	0.018	0.045	< 0.001
Vt_{Min}	-159.510	56.006	-174.105	-144.915	< 0.001	0.003	0.049	-0.009	0.016	0.586
$TMVt$	-197.264	77.401	-217.259	-177.269	< 0.001	0.025	0.051	0.011	0.038	0.001
$PkAP_B$	39.221	20.419	33.946	44.496	< 0.001	0.014	0.024	0.007	0.020	< 0.001
$PkAP_P$	-32.577	18.204	-37.498	-27.655	< 0.001	0.002	0.023	-0.004	0.008	0.549
$PkML$	-14.182	11.744	-17.327	-11.037	< 0.001	0.007	0.014	0.003	0.011	0.001
<i>Pressure</i>	(kPa)					(%TW/cm ²)				
$PkRF_{Med}$	-88.775	78.721	-110.915	-66.634	< 0.001	-0.123	0.880	-0.376	0.129	0.331
$PkR_{F_{Ct}}$	-62.284	79.487	-84.413	-40.155	< 0.001	0.107	0.982	-0.172	0.387	0.443
$PkRF_{Lat}$	-21.046	61.709	-37.889	-4.202	0.015	0.204	0.662	0.018	0.390	0.032
$PkMF_{Med}$	-32.183	35.782	-41.950	-22.417	< 0.001	-0.215	0.488	-0.345	-0.086	0.002
$PkMF_{Lat}$	-14.166	52.876	-29.691	1.359	0.073	0.063	0.686	-0.136	0.262	0.529
$PkFF_{Med}$	-97.372	149.339	-138.535	-56.209	< 0.001	-0.276	1.721	-0.750	0.198	0.248
$PkFF_{Ct}$	-85.274	76.039	-107.600	-62.948	< 0.001	-0.183	1.048	-0.469	0.104	0.206
$PkFF_{Lat}$	-55.269	9.987	-80.843	-29.695	< 0.001	0.095	1.468	-0.314	0.504	0.642
$PkHlx$	-74.604	167.317	-119.411	-29.796	0.002	-0.241	1.616	-0.695	0.214	0.293
$PkToes$	-62.868	87.000	-87.088	-38.647	< 0.001	-0.363	1.164	-0.669	-0.057	0.021

The acronym of the variables can be seen in session *Data Analysis* in Methods. TW – total weight. Negative values indicate that the BpG presented larger magnitudes than CG; only in $PkAP_B$ variable the interpretation is different, where positive values indicate that the BpG presented larger magnitudes than CG.

2.4. Discussion

The present study investigated the influence of occasional load in the GRF and plantar pressure parameters during gait. Other studies have already reported higher GRF during load carriage compared to CG (Birrell et al., 2007; Chow et al., 2005; Harman et al., 2000; Simpson et al., 2012), which corroborates our results. On the other hand, analyses of the GRF normalized by the total weight (body mass plus backpack mass) and the in-shoe plantar pressure while walking occasionally loaded

are scarce in literature. The approach employed in this study allowed us to determine the amount of load applied on the foot (absolute values) and changes in the gait pattern (normalized values), as well as the distribution of the forces on the plantar surface of the foot (pressure data) while walking carrying load.

Rearfoot Region

We found an increase in impact forces (absolute values) in BpG. Similar results have already been shown (Birrell & Haslam, 2010; Birrell et al., 2007; Harman et al., 2000; Tilbury-Davis & Hooper, 1999). Birrell et al. (2007), analyzing loads of 8, 16, 24, 32 and 42 kg, and Tilbury-Davis and Hooper (1999), analyzing loads of 20 and 40 kg military backpacks, found a proportional increase in vertical and anterior-posterior GRF with the increased load. In contrary, Simpson et al. (2012), analyzing loads of 20, 30 and 40% of body weight, found similar values of GRF parameters between 30 and 40% of the body weight, indicating an attenuation of the force progression with female hikers walking with 40% of their body weight. In the present study, the GRF normalized data also indicate alteration in the gait pattern during backpackers' gait. The different populations among the studies may be the reason for the differences (recreational female hikers (Simpson et al., 2012) and untrained young adults (present study) vs. military (Birrell et al., 2007; Tilbury-Davis & Hooper, 1999)) in adaptation on GRF parameters during load carriage.

On the other hand, the PkAP_B was larger (absolute and normalized values). It may indicate that the braking forces are potentiated (increased more than backpack mass) during occasional loaded gait. Birrell & Haslam (2010) also found higher absolute braking forces with load carriage (military). We did not find other studies analyzing braking forces normalized by the total weight to compare to our results. The anterior-posterior force helps slow the body down during the initial part of the gait cycle (Birrell & Haslam, 2010). Its increase seems to be related to blister development (Knapik et al., 1997). Birrell & Haslam (2010) suggested that load carriage increases the pressure on the skin and causes more movement between the foot and the shoe through higher propulsive and braking forces, thus increasing the risk of blister. The absolute pressure increases in all RF regions (medial, central and lateral) in the present study supports this notion. This relation (PkAP_B and RF pressure increase) may be one of the mechanisms that contribute to blister development, which is the most common injury related to load carriage (Cooper, 1981; Knapik et al., 1992).

Midfoot Region

The medial MF was specially needed during BpG. Larger absolute and normalized values were found compared to CG. It may be read as an adaptation in the gait pattern as a result of load carriage. For the lateral MF, the opposite behavior was found. Similar absolute and normalized peak pressures were shown between BpG and CG. It may indicate a protective strategy for relieving the pressure on the lateral MF by putting more load on the lower loaded regions (medial MF).

Regarding $V_{t_{\min}}$, the BpG presented larger magnitudes of absolute and similar magnitudes of normalized data. It indicates that there was no alteration in gait pattern as a result of carrying a backpack. Birrell et al. (2007), analyzing military hikers, and Simpson et al. (2012) analyzing female hikers, found an increase in medial-lateral impulse as compared to normal gait (no load condition). These results can be related to decrease of stability (Birrell et al., 2007). In the present study, analyzing other variables related to medial-lateral axis (PkML), we also found a increase in medial-lateral forces while loaded walking (absolute values). However, the normalized values indicated that this increase in PkML is not proportional to the weight of the load. Therefore, even when load carriage seems to be a lower stable condition (Birrell et al., 2007; Simpson et al., 2012; absolute data from this study), some gait adaptation may be developed in order to reduce this instability as indicated by the normalized PkML.

Forefoot Region

Considering the pressures and GRF acting at the FF, as expected all the variables ($TMV_{t_{\text{Abs}}}$, $PkAP_{P_{\text{Abs}}}$, $PkHlx_{\text{Abs}}$, $PkToes_{\text{Abs}}$, $PkFF_{\text{Med}_{\text{Abs}}}$, $PkFF_{\text{Ct}_{\text{Abs}}}$ and $PkFF_{\text{Lat}_{\text{Abs}}}$) showed larger magnitudes in BpG compared to CG. The medial FF was the region that presented the highest increase in the pressure when a backpack was used (97.4 kPa, $CI_{95\%}$ 138.5 to 56.2), while the lowest increases occurred in lateral FF (55.3 kPa, $CI_{95\%}$ 80.8 to 29.7). It indicates a higher recruitment of the medial region to support load carriage. By normalizing the data we expected that there were no differences between groups. However, in the PkToes, the values were larger in the BpG. It suggests that during occasional loaded gait the toes region was more needed than in the unloaded gait. BpG also showed lower $TMV_{t_{\text{Norm}}}$. Possibly this has occurred because the backpack promotes an increase in forward lean due to the posterior location of the center of mass during gait (Birrell & Haslam, 2010). Thus, the forces required to advance the body from the mid-stance to toe-off were reduced as a

consequence of the decrease in the passive movement of the body (Birrell & Haslam, 2010).

Time Variables

A longer duration of stance phase and initial double stance was found in BpG as compared to CG. Female hikers wearing a backpack with 30 to 40% of their body weight and military personnel carrying loads of 8, 16, 24, 32 and 40 kg also showed longer duration of the stance phase compared to the no load condition (Birrell et al., 2007; Simpson et al., 2012). Singh & Koh (2009), Hong & Brueggemann (2000) and Chow et al. (2005) found in primary school students carrying a backpack with between 10 and 20% of their body weight a larger initial double stance compared to no loaded gait. Even with older participants carrying heavier backpacks (32.2% of the body mass, $CI_{95\%}$ 29.5 to 34.8), our data corroborates with theirs. One possible explanation for this behavior is that walking with a backpack raises the combined center of mass of the backpack and body in posterior and superior fashion. It induces postural imbalance for static and dynamic conditions (Hong & Brueggemann, 2000; Singh & Koh, 2009). The longer double stance may be an attempt to minimize the duration of unsteady single-limb stance (Hong & Brueggemann, 2000). This mechanism brings down the combined center of mass, providing a counter effect to stabilize the gait (Singh & Koh, 2009). The $PkMF_{Med_Time}$ occurred later and $TMV_{t_{Time}}$ earlier in BpG than CG. The increase of the initial double stance may promote this delay in $PkMF_{Med}$, while the posterior shifting of the center of mass (Birrell & Haslam, 2010) may be responsible for the alterations in $TMV_{t_{Time}}$.

For the adult population, a load limit is not well established. A varying range of heavy loads are carried by different populations. The total load masses carried by soldiers average 40 kg: in some situations they could be required to carry loads of up to 76 kg (Reynolds et al., 1999). Korean beverage workers usually carry approximately 53.4 kg (ranging from 20 to 80 kg) while carrying backpacks (Chung et al., 2005). Tourist trekkers in New Zealand carry backpacks with up to 29% of their body weight for five or more consecutive hours over distances of 11 or more kilometers per day (Lobb, 2004). In all of these populations a high injury incidence was described. Recently, a well-grounded load limit of 30% of body mass was established for female recreational hikers (Simpson et al., 2011a, 2011b). In our study, we adopted a BMI of 30 kg/m² as load normalization criterion. The 95% confidence interval of the applied load in our study was 29.5 to 35.8 % of body mass. The load adopted in our study, even while based on other criterion, was just slightly higher than the load limit

previously proposed (Simpson et al., 2011a, 2011b). It reinforces that the load selected in our study was successful in putting a potentially harmful load on the musculoskeletal system.

Some possible limitations in this study should be considered. First, the backpack load used was not the same for all participants; we could have normalized the load using either percentage of body mass or a fixed load: however, since the locomotor system of people with BMI ≥ 30 is considered more susceptible to injuries (Ko et al., 2010; WHO, 2000), we preferred to use the BMI of 30 as load criterion in order to promote a harmful load. It seems to us that this was an effective way of doing so. Second, the gait speed adopted in the present study was the one with which the subjects felt more comfortable (self-selected), and such behavior can influence the characteristics of the force. We opted for the self-selected speed in order to prevent disturbances in the gait pattern and ensure normal walking (Cavanagh & Ulbrecht, 1994; Hennig & Rosenbaum, 1991). Thus, we analyzed the unloaded and loaded self-selected gait (which we considered a more realistic condition), which does not mean that the participants walked at the same speed using both gaits. Small variations in walking speed are not critical for peak pressure measurements (Taylor et al., 2004). Our subjective analysis during the data collection indicates that the speed of the two conditions was similar. Finally, the pressure analysis considered only the vertical forces: therefore, we do not know about the distribution of the shear forces. As far as we know, there are very restricted devices that are able to perform this kind of analysis.

2.5. Conclusions

In conclusion, we observed an overall increase in the GRF and plantar pressure parameters, as well as alterations in gait pattern during occasional loaded gait (BpG) as compared to CG. The medial MF and Toes were the most used regions during occasional loaded gait while the lateral RF was used less. Regarding the other regions, the increase seemed to be proportional to the weight of the backpack (higher absolute values in BpG and similar normalized values than CG). A protective behavior in BpG was evidenced by the diminished magnitude of impact and propulsive forces. On the other hand, the shear forces increased more than the proportion of the load, which may mean higher susceptibility to blister development. Further investigation assessing the effects of training or different materials (shoe, insole, socks, etc.) on the GRF and plantar pressures in occasionally loaded people (students, hikers, military personnel,

etc.) may be important in improving the capacity of the musculoskeletal system to handle potential harmful conditions.

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CHAPTER 3: Original Research

**IN-SHOE PLANTAR PRESSURES AND GROUND
REACTION FORCES DURING OBESE ADULTS'
OVERGROUND WALKING**

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Abstract

Purpose: Since walking is highly recommended for prevention and treatment of obesity, and some of its biomechanical aspects are not clearly understood for obese people, we compared the absolute and normalized ground reaction forces (GRF), plantar pressures and temporal parameters of normal-weight and obese participants during overground walking. *Method:* A force plate and an in-shoe pressure system were used to record GRF, plantar pressures (foot divided in 10 regions), and temporal parameters of 17 obese and 17 gender-matched normal-weight adults while walking. *Results:* With high effect sizes, the obese showed higher absolute medial-lateral and vertical GRF, and pressure peaks in the central rearfoot, lateral midfoot, lateral and central forefoot. However, when we analyzed normalized (scaled to body weight) data, the obese participants showed lower vertical and anterior-posterior GRF, and lower pressure peaks in the medial rearfoot and hallux, but the lateral forefoot peaks continued greater compared to normal-weight participants. Time of occurrence of medial-lateral GRF and pressure peaks in the midfoot occurred later in obese individuals. *Conclusions:* The obese participants adapted their gait pattern to minimize the consequences of the higher vertical and propulsive GRF in their musculoskeletal system. However, they were not able to improve their balance as indicated by medial-lateral GRF. The obese participants showed higher absolute pressure peaks in four out of 10 foot regions. Furthermore, the normalized data suggest that the lateral forefoot in obese adults was loaded more than did the proportion of their extra weight, while the hallux and medial rearfoot were seemingly protected.

Keywords: Biomechanics; Gait; Locomotion; Obesity.

3.1. Introduction

Obesity is defined as a condition of excessive fat accumulation in adipose tissue and is a global epidemic disease (Gutin, 2013; Hill & Peters, 1998; Lai et al., 2008). To combat obesity, the most cited approach is to combine exercise and a dietary intervention (Hill & Peters, 1998). Walking is highly recommended and popular for prevention and treatment of obesity (Browning & Kram, 2007). Although this activity might be critical in terms of biomechanical loading on the musculoskeletal system (Browning & Kram, 2007). Obesity is associated with a range of disabling musculoskeletal conditions in adults (Anandacoomarasamy et al., 2007). The repetitive overload during obese people's gait has been related with the predisposition to pathological gait patterns, loss of mobility and subsequent progression of disability (Messier et al., 1996), as well as higher risk of hip and knee osteoarthritis (Felson, 1990; Hochberg et al., 1995; Ko et al., 2010), increase of the likelihood of foot ulceration (Vela et al., 1998) and heel pain (Prichasuk, 1994). Thus, more attention must be given to the physical/mechanical consequences of repetitive overload, mainly in the lower limbs, in order to provide support in the areas of prevention, treatment, and control of obesity (Hills et al., 2002).

The analyses of the three components (horizontal and vertical) of the ground reaction forces (GRF), and plantar pressures can provide useful information about the influence of overweight on the musculoskeletal system (Birtane & Tuna, 2004; Hills et al., 2002; Hills et al., 2001; Messier et al., 1996). Higher absolute GRF in healthy obese (no pathology other than obesity) than in normal-weight individuals (Browning & Kram, 2007), and positive correlation between body mass index (BMI) and absolute GRF components (anterior-posterior, medial-lateral, and vertical) in older obese adults with osteoarthritis (Messier et al., 1996) were already described. However, a contradiction is observed in normalized (relative to body weight – BW) GRF data: one article refers similar horizontal components (anterior-posterior and medial-lateral) and lower vertical GRF (Browning & Kram, 2007), while another observed higher anterior-posterior propulsive force and similar vertical GRF (Lai et al., 2008). Thus, while the absolute GRF values clearly indicate an overall overloading during obese people's gait, the normalized ones suggest some alterations on gait patterns which are not clear.

The assessment of plantar pressure distribution represents an important clinical tool for understanding the structural and functional implications of obesity (Filippin et al., 2007). The decrease in plantar pressure peaks is considered important for susceptible populations like obese people to avoid and treat injuries (Pérez-Soriano et

al., 2011). Significant positive correlations were found between plantar pressures and pain ratings (Hodge et al., 1999). Two studies addressed the plantar pressure analysis in adult obese population. One study (Hills et al., 2001) found higher absolute pressure peaks in almost all foot regions; while in the other (Birtane & Tuna, 2004), the obese individuals showed higher absolute pressure peaks only in the midfoot compared to their normal-weight peers. In both studies (Birtane & Tuna, 2004; Hills et al., 2001), the participants were assessed barefoot; the midfoot and rearfoot were considered as one region; and only absolute data were analyzed. Besides the conflicting results between the studies (Birtane & Tuna, 2004; Hills et al., 2001), there are scarce information regarding plantar pressures during obese people's walking. Issues such as in-shoe plantar pressure analysis as well as its pattern, and a more detailed approach for the midfoot and rearfoot would be interesting to be investigated.

A better understanding on the biomechanical features of obese people during common activities of daily living, such as walking, would be important to identify the characteristics of movement-related difficulties and possible pathogenesis of the musculoskeletal impairments associated to obesity (Wearing et al., 2006). Therefore, our aim was to compare the magnitude (absolute values) and gait pattern (normalized by BW values) of GRF, in-shoe plantar pressure peaks, and temporal parameters between obese and normal-weight adult participants while walking. We hypothesized that higher absolute GRF and pressure peaks will be observed in the obese participants compared to their normal-weight peers; that similar pattern of GRF and plantar pressures will be found between groups; and that there will be differences in the temporal gait parameters between groups.

3.2. Methods

This is a cross-sectional study with a convenience sample. This project was approved by the local ethics committee and all participants freely signed an informed written consent form based on the Helsinki declaration.

Participants

We selected two groups of participants: people with BMIs between 20 and 25 were included in the normal-weight group (labeled as NW), and participants with BMIs above 30 were included in the group of obese people (labeled as obese group – OG).

The participants were excluded if they showed any traumatic-orthopedic impairment or difficulty with independent gait. For the OG, there were 12 male participants (mean age of 37.00 ± 6.06 years old, height of 1.75 ± 0.04 m, body mass of 111.20 ± 10.51 kg and BMI of 36.23 ± 3.54 kg/m²) and five female participants (mean age of 36.40 ± 6.02 years old, height of 1.55 ± 0.06 m, body mass of 96.08 ± 10.52 kg and BMI of 40.21 ± 5.87 kg/m²). For the CG, there were also 12 male participants (mean age of 27.42 ± 3.09 years old, height of 1.74 ± 0.05 m, body mass of 71.98 ± 4.68 kg and BMI of 23.73 ± 1.14 kg/m²) and five female participants (mean age of 27.4 ± 1.34 years old, height of 1.60 ± 0.05 m, body mass of 52.92 ± 6.43 kg and BMI of 20.67 ± 1.81 kg/m²).

Instruments and Data Acquisition

We used to record GRF a Bertec force plate (model 4060-15, Bertec Corporation, Columbus, OH, USA) operating at 1000 Hz, and the Acknowledge software (BIOPAC System, Goleta, CA, USA). To record in-shoe plantar pressures, we used a F-Scan system (TekScan, South Boston, MA, USA) operating at 300 Hz with 0.18 mm thick insole sensor, and the F-Scan Research 6.33 software (TekScan, South Boston, MA, USA). Three digital video camera recorders, Sony (model DCR-HC62E, Sony Corporation, Tokyo, Japan) operating at 50 Hz, and the Dvideo v.5.0 software (Unicamp, Campinas, SP, Brazil)(Figuroa et al., 2003) were used to capture, synchronize, digitalize and reconstruct the images. We used an external trigger to start the force plate and in-shoe plantar pressure system simultaneously.

Tasks and procedures

First, we explained all the procedures of the study to the participants and after their weight and height were recorded. We gave every of the participants fitted black shorts and one reflective marker with a diameter of 1.2 cm was placed with adhesive tape at the right great trochanter of the femur. A cuff unit measuring 98 x 64 x 29 mm was attached on the lateral malleolus region of both legs of each participant, and a 9.25 mm cable linked the cuff to the VersaTek hub (F-Scan system). The cable did not cause any restriction for walking. Each participant received a pair of thin socks and neutral shoes (ballet sneakers) with sensor insoles inside. Second, the participants familiarized themselves with the trial by walking freely with a comfortable speed (self-selected speed) over a 6 m walkway with the force plate embedded in the middle. One of the researchers identified the starting position for the participants walking at their self-selected speed to hit the force plate without altering their gait pattern. The participants performed three trials in which, at least, two steps before and after

reaching the force plate were performed. We used the third step to further analysis and then we avoided the effects of acceleration (Macfarlane & Looney, 2008).

Data Analysis

We exported the data from the force plate (three GRF components) and in-shoe pressure system (values of each sensor in each frame) to Matlab 7.0 software (MathWorks, Natick, MA, USA) and developed a program to process and calculate the variables. We calculated the following GRF parameters:

- Fz1 (load acceptance peak): first peak from the vertical GRF;
- Fz1_{imp} (load acceptance impulse): the impulse calculated from the beginning of the stance phase to the minimum between the two vertical GRF peaks;
- Fz2 (thrust peak): second peak from the vertical GRF;
- Fz2_{imp} (thrust impulse): impulse from the minimum between the vertical GRF peaks to the end of the stance phase;
- Fap1 (braking peak): first (negative) peak from the anterior-posterior GRF;
- Fap1_{imp} (braking impulse): impulse calculated from the beginning of the stance phase to the middle zero (negative phase) from the anterior-posterior GRF;
- Fap2 (propulsive peak): second peak (positive) of the anterior-posterior GRF;
- Fap2_{imp} (propulsive impulse): impulse from the middle zero to the toe off from the anterior-posterior GRF;
- Fml (medial-lateral peak): positive peak from the medial-lateral GRF;
- Fml_{imp} (medial-lateral impulse): impulse from the beginning to the end of the stance phase of the medial-lateral GRF.

We also calculated the stance phase duration, time of occurrence of the GRF peaks, and the walking speed, which was considered as the first time derivate of the great trochanter reflective marker position. For the in-shoe plantar pressure data treatment, first the program divided the foot into 10 regions: hallux, distal phalanges, medial, central and lateral forefoot; medial and lateral midfoot; and medial, central and lateral rearfoot, as used in another study (Castro et al., 2013). The program automatically divided the foot, and the regions were checked by two trained researchers, who, if necessary, corrected manually this procedure. The program calculated the plantar pressure peaks, which were considered as the highest pressure sensor value during the third step, and their time of occurrence for each region. We used the vertical GRF to calibrate the plantar pressure data trial-by-trial, as suggested by Castro et al. (2013). All data (GRF and pressure peaks) were showed as absolute

and normalized (scaled to BW) values. The time of occurrence of the peak events were normalized by the stance phase.

Statistical analysis

We arbitrarily chose some variables to verify the intra-individual repeatability of the three trials. For this, we calculated the intraclass correlation coefficient (ICC) to the stance phase duration, Fz1, time of occurrence of Fz1, Fz1_{Imp}, and for the pressure peaks in three regions (hallux, central forefoot, and central rearfoot). We computed the mean of the three repetitions of each participant and all statistical procedures were performed with these mean values. We used eight MANOVAs with the groups (NW and OG) as between-subjects factor, and the (1) absolute and (2) normalized GRF peak parameters (Fz1, Fz2, Fap1, Fap2, and Fml), (3) absolute and (4) normalized GRF impulse parameters (Fz1_{Imp}, Fz2_{Imp}, Fap1_{Imp}, Fap2_{Imp}, and Fml_{Imp}) and (5) temporal parameters (stance phase duration, speed, time of occurrence of Fz1, Fz2, Fap1, Fap2 and Fml), (6) absolute and (7) normalized pressure peaks (10 foot regions), and (8) their time of occurrence as dependent measures. Whenever a statistical significant difference was found, the Fisher's Least Significant Difference was calculated. Considering MANOVA assumptions, the data were found to be normal as indicated by the Shapiro-Wilk test ($p > 0.05$), and the sphericity verified by the Mauchly's test, was held. We used the partial Eta square (η_p^2) to measure the effect sizes considering that an η_p^2 of 0.01 was small, of 0.06 was medium, and higher than 0.14 was large (Stevens, 2002). We used the Statistica® v.8 software (Statsoft®, Tulsa, USA) and an α value set at 0.05 to perform the statistical analyses.

3.3. Results

We found excellent data repeatability. The GRF parameters duration of stance phase, Fz1, Fz1_{Time}, and Fz1_{Imp} displayed ICCs between 0.94 and 0.99. While the pressure peaks in the hallux, central forefoot and central rearfoot regions showed ICCs of 0.93, 0.96 and 0.91, respectively.

Differences between OG and NW were found in absolute GRF peaks ($F(4, 128) = 79.637$; $p < 0.001$; $\eta_p^2 = 0.71$), and absolute GRF impulses ($F(4, 128) = 38.465$; $p < 0.001$; $\eta_p^2 = 0.55$) with large effect sizes were found. The OG showed higher values for both vertical and medial-lateral peaks (Fz1, Fz2, and Fml) and impulses (Fz1_{Imp}, Fz2

$_{imp}$, and $F_{ml_{imp}}$), whereas for the anterior-posterior GRF (F_{ap1} , F_{ap2} , $F_{ap1_{imp}}$, $F_{ap2_{imp}}$) similar values were found (Table 1 and Figures 1a, 1b and 1c). For the normalized GRF, differences were found between OG and NW for the GRF peaks ($F(4, 128) = 6.03$; $p < 0.001$; $\eta^2 = 0.16$) while for the normalized GRF impulses similar values were found ($F(4, 128) = 1.232$; $p = 0.30$; $\eta^2 = 0.04$), large and small effect sizes were found, respectively. The OG presented lower values for both vertical GRF peaks ($Fz1$ and $Fz2$) and for the F_{ap2} compared to NW (Table 1 and Figure 1d, 1e and 1f).

Table 1. Ground reaction forces in normal-weight (NW) and obese group (OG) during walking.

Variable	Unit	ABSOLUTE DATA			NORMALIZED DATA			
		NW Mean±SD	OG Mean±SD	p- Value	Unit	NW Mean±SD	OG Mean±SD	p- Value
$Fz1$	N	684.9±118.3	1053.0±124.5	<0.001	N/BW	1.04±0.04	1.01±0.05	0.002
$Fz2$	N	715.0±100.8	1098.0±117.2	<0.001	N/BW	1.10±0.05	1.05±0.05	<0.001
F_{ap1}	N	-97.0±20.9	-135.4±33.3	0.140	N/BW	-0.15±0.02	-0.13±0.03	0.141
F_{ap2}	N	120.9±20.3	166.7±25.1	0.079	NBW	0.19±0.02	0.16±0.02	0.036
F_{ml}	N	66.7±14.5	121.1±21.1	0.037	N/BW	0.10±0.02	0.12±0.02	0.274
$Fz1_{imp}$	N.s	165.9±29.3	295.3±48.9	<0.001	(N/BW).s	0.26±0.04	0.28±0.03	0.300
$Fz2_{imp}$	N.s	213.7±55.1	340.7±66.2	<0.001	(N/BW).s	0.32±0.05	0.33±0.05	0.300
$F_{ap1_{imp}}$	N.s	-18.2±4.5	-28.5±5.0	0.363	(N/BW).s	-0.03±0.00	-0.03±0.00	0.300
$F_{ap2_{imp}}$	N.s	18.2±3.3	28.8±3.9	0.353	(NBW).s	0.03±0.00	0.03±0.00	0.300
$F_{ml_{imp}}$	N.s	29.4±8.0	56.2±12.6	0.019	(N/BW).s	0.05±0.01	0.05±0.01	0.300

$Fz1$ - load acceptance peak; $Fz1_{imp}$ - load acceptance impulse; $Fz2$ - thrust peak; $Fz2_{imp}$ - thrust impulse; F_{ap1} - braking peak; $F_{ap1_{imp}}$ - braking impulse; F_{ap2} - propulsive peak; $F_{ap2_{imp}}$ -propulsive impulse; F_{ml} - medial-lateral peak; $F_{ml_{imp}}$ - medial-lateral impulse.

Differences between groups were also found in the plantar pressure peaks for both absolute ($F(9, 288) = 10.040$; $p < 0.001$; $\eta_p^2 = 0.24$) and normalized ($F(4, 288) = 8.222$; $p = p < 0.001$; $\eta_p^2 = 0.20$) values (Figure 2). Large effect sizes were also observed. The absolute pressure peaks were higher for the OG in the central and lateral forefoot, medial midfoot, and central rearfoot regions, while they were lower in the hallux compared to the NW (Figure 2a). The normalized pressure peaks were higher in the lateral forefoot and lower for the hallux ($p < 0.001$) and medial rearfoot ($p = 0.001$) regions for the OG compared to NW (Figure 2b).

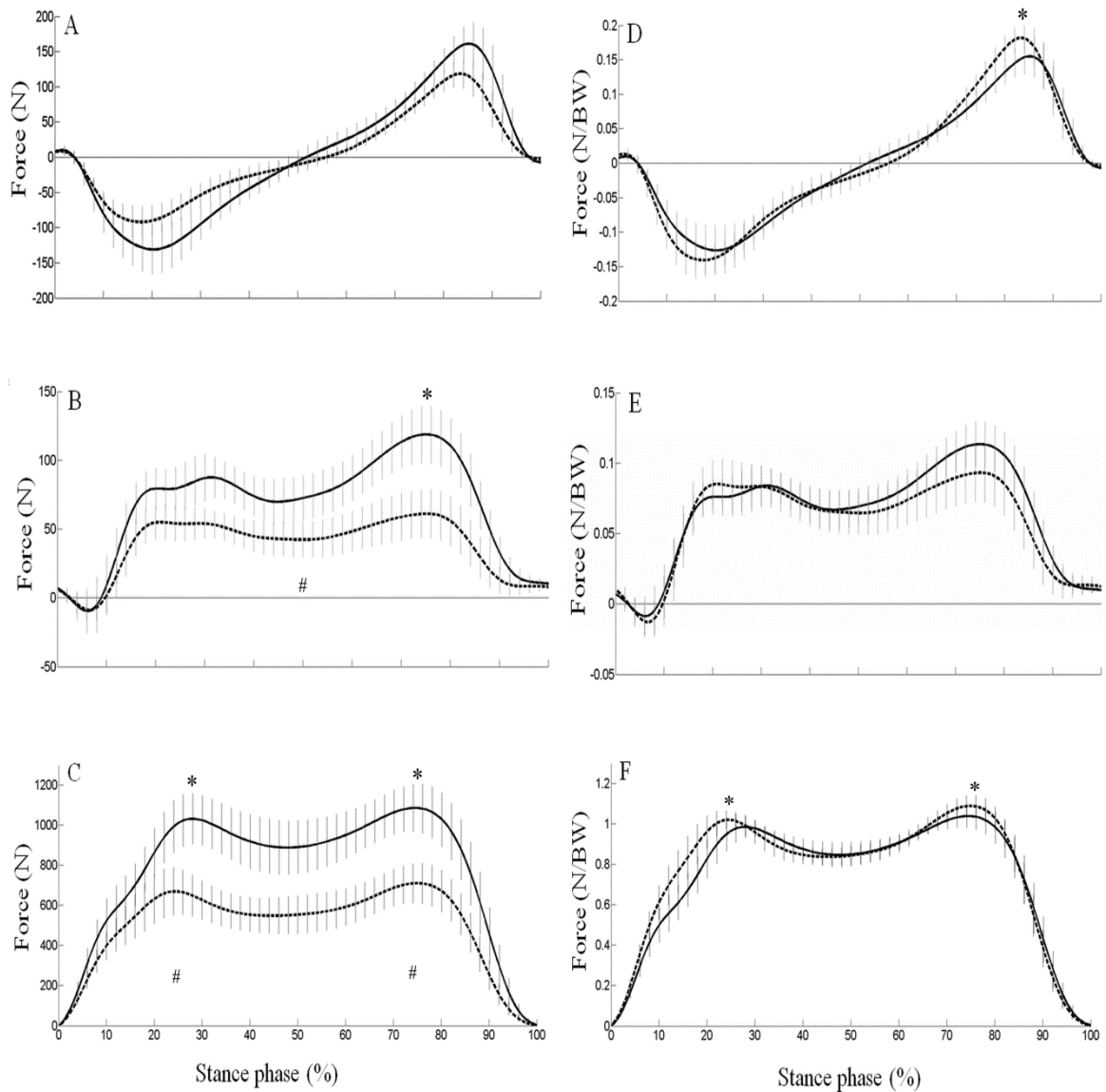


Figure 1. Comparison of ground reaction force (GRF) parameters between normal-weight and obese subjects. Absolute mean anteroposterior GRF (A); absolute mean medial-lateral GRF (B); absolute mean vertical GRF (C); normalized mean anteroposterior GRF (D); normalized mean medial-lateral GRF (E); normalized mean vertical GRF.

Bold lines represent obese group and dotted lines represent normal-weight group.

Error bars represent the standard deviation.

* $P < 0.05$ for peak variables.

$P < 0.05$ for impulse variables.

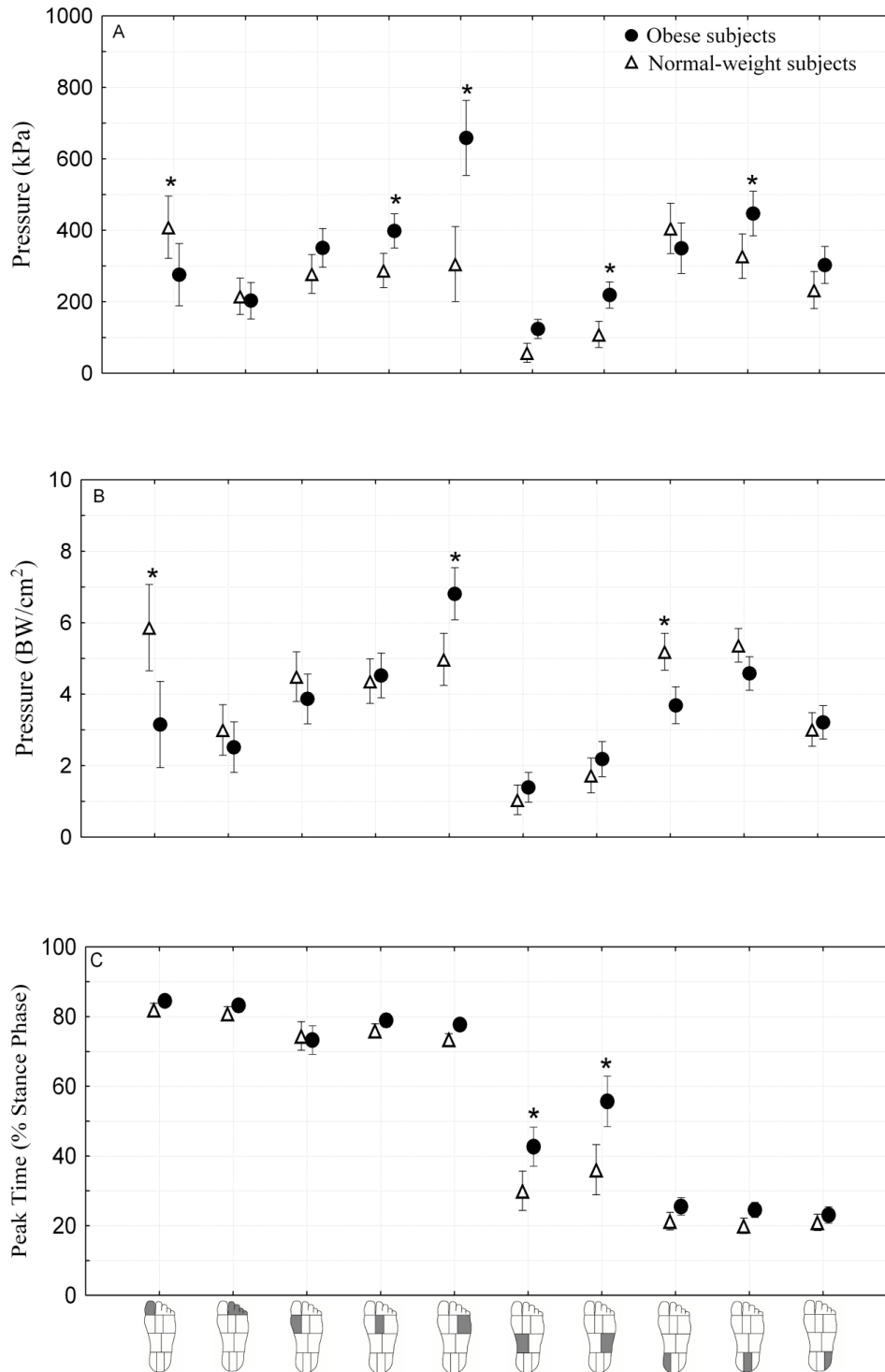


Figure 2. Comparison of pressure parameters between normal-weight and obese subjects as function of foot region: absolute mean pressure peaks (A); normalized mean pressure peaks (B); mean time of occurrence for the pressure peaks (C). Error bars represent the 95% confidence interval. * $P < 0.05$.

Differences were found in the time of occurrence of GRF peaks ($F(6, 192) = 12.090$; $p < 0.001$; $\eta_p^2 = 0.27$) and time of occurrence of pressure peaks ($F(9, 288) =$

6.566; $p < 0.001$; $\eta_p^2 = 0.17$), both displaying large effect sizes. The time of occurrence for the Fml and pressure peaks in the medial and lateral midfoot regions were larger for the OG than in NW. The stance phase duration, speed and time of occurrence for the GRF peaks (Fz1, Fz2, Fap1, and Fap2) and pressure peaks in the medial, central and lateral rearfoot, medial, central and lateral forefoot, hallux, and distal phalanges were similar between groups (Table 2 and Figure 2c).

Table 2. Temporal variable.

Variable	Normal-weight	Obese Group	P-value
	Mean±SD	Mean±SD	
Stance phase duration (s)	0.76±0.06	0.82±0.08	0.980
Speed (m/s)	1.13±0.10	0.98±0.14	0.950
Fz1 _{Time} (% Stance Phase)	24.59±2.76	28.63±3.71	0.102
Fz2 _{Time} (% Stance Phase)	74.64±2.06	73.69±5.15	0.700
Fap1 _{Time} (% Stance Phase)	18.01±2.84	20.46±2.66	0.319
Fap2 _{Time} (% Stance Phase)	83.65±1.23	85.17±1.95	0.538
Fml _{Time} (% Stance Phase)	50.95±24.94	74.13±5.13	<0.001

_{Time} – time of occurrence of Fz1, Fz2, Fap1, Fap2, and Fml.

3.4. Discussion

We compared the absolute and normalized GRF and plantar pressure peaks, and some temporal parameters between normal-weight and obese participants during self-selected overground level walking. We almost fully satisfied our first hypothesis as higher absolute GRF (medial-lateral and vertical components), and absolute pressure peaks in four out of the 10 foot regions were observed in the obese participants; however, we did not expected similar absolute anterior-posterior GRF, and pressure peaks in five regions between groups, as well as higher pressure peaks in the hallux for the normal-weight participants. Our second hypothesis was mainly not confirmed as in the normalized GRF (anterior-posterior and vertical components), and in three foot regions we observed differences between the groups. We partially confirmed our third hypothesis: the medial-lateral GRF peaks, and the pressure peaks in the midfoot were different between groups; on the other hand, 14 out of 17 temporal variables were similar between OG and NW. The aforementioned differences between groups were not only statistically significant but also provide relevant information about the influence of obesity on the magnitude and pattern of GRF, plantar pressure, and temporal parameters, as large effect sizes were observed.

At the beginning of the stance phase (first 30% of stance time), the obese participants showed a different behavior in the vertical GRF compared to their normal-weight peers. Corroborating with our data, higher absolute Fz1 (Browning and Kram (2007): 1080 ± 65 vs. 676 ± 32 N; Messier et al. (1996): 968 ± 21 vs. 756 ± 17 N), and lower normalized Fz1 (1.00 ± 0.01 vs. 1.03 ± 0.01 N/BW) for obese compared to non-obese individuals (Browning & Kram, 2007) were already showed. This behavior suggests an adaptation in obese gait pattern to relieve the consequences of their extra BW on the musculoskeletal system. Thus, the obese people did present higher vertical loads as expected, which reflects their increased BW. However, and interestingly, they seemingly altered their gait pattern in order to decrease the effect of this overload (excessive body mass) on their bodies. We did not find differences between groups in terms of absolute and normalized braking forces (Fap1 and Fap_{imp}). Contrary to our results, higher values of absolute Fap1 (152 ± 10 vs. 91 ± 5 N) were already reported (Browning & Kram, 2007), whereas data regarding the normalized values corroborated with our findings (Browning & Kram, 2007). These contradictions observed in some variables between the studies might be occurred as a consequence of the different protocols adopted: walking overground at a self-selected speed (present study) versus walking on a treadmill at a controlled speed of 1 m/s (Browning & Kram, 2007). Messier et al. (1996) found a significant positive correlation between BMI and absolute Fap1 and Fap_{imp} in obese osteoarthritis subjects. Our data suggest, for people without musculoskeletal impairment, no differences in braking forces between NW and OG. Thus, the braking forces may play a relevant role for discriminating the gait of obese people with or without physical impairment.

As expected, the pressure events occurring at the beginning of the stance were the rearfoot pressure peaks. Hills et al. (2001) found higher values for male obese compared to normal-weight participants (391 vs. 335 kPa), and similar ones for females (375 vs. 358 kPa). In contrast, Birtane and Tuna (2004) found no differences in rearfoot pressure peaks between obese and lean people (210 vs. 193 kPa). Differently from the previous studies (Birtane & Tuna, 2004; Hills et al., 2001), we divided the rearfoot into three regions. In the medial and lateral rearfoot, similar absolute pressure peaks were displayed between the groups, while in the central region the OG showed higher values than the NW (447 vs. 328 kPa). The normalized data indicated similar pressure in the central and lateral rearfoot, and lower values in the medial rearfoot region for obese compared to their normal-weight counterparts. Plantar fasciitis is a common musculoskeletal disorder which is observed in 11% to 15% of adults and is characterized by pain in the inferomedial aspect of the heel (Mendonça et al., 2013;

Thomas et al., 2010). Obesity is considered a risk factor for such disorder and mechanical overload is believed to be its most common cause (Thomas et al., 2010). Therefore, we suppose that the decreased normalized medial rearfoot peaks found in the OG from this study might be occurred as a protective adaptation of the gait pattern to avoid overloading this region, which is considered the most susceptible one in the rearfoot (Thomas et al., 2010).

Previous studies described higher pressure peaks in the midfoot for obese (141 vs. 99 kPa and 135 vs. 46 kPa) compared to normal-weight people (Birtane & Tuna, 2004; Hills et al., 2001). In these studies (Birtane & Tuna, 2004; Hills et al., 2001) the midfoot was analyzed as one region. Therefore, a direct comparison with our findings is not valid as we divided the midfoot into medial and lateral regions. Nevertheless, our data reveal higher values in the lateral midfoot for the obese than the non-obese participants (218 vs. 108 kPa). However, this behavior is not observed in the medial midfoot region, where similar values were found (125 vs. 57 kPa).

The medial longitudinal arch in prepubescent obese children are collapsed (Riddiford-Harland et al., 2000). This collapse could promote an increased contact area of the medial midfoot region, and then compensate the increased forces resulting in similar pressure values. In this sense, we would expect that the normalized medial midfoot pressures were lower in this region. However, they were not. Nyska et al. (1997) analyzed the influence that a backpack with 20 and 40 kg had on the plantar pressures of normal-weight participants and concluded that the human foot adapts itself under loading condition by maintaining the medial longitudinal arch. These adaptations involved to shift the plantar loads to the central and medial forefoot (Nyska et al., 1997). Our data support this maintenance of the medial longitudinal arch function in adult obese individuals. Moreover, we observed an adaptation that shifted the plantar pressures to the lateral midfoot and lateral forefoot regions.

Analyzing the end of the stance phase (from 70 to 100% of stance phase), positive correlation between BMI and absolute Fz2 and lower normalized Fz2 between obese and non-obese participants were described (Lai et al., 2008; Messier et al., 1996). These results are in agreement with ours, and reinforce the theory of a protective adaptation of the gait pattern in terms of vertical forces during obese people's walking. Regarding the medial-lateral forces (Fml and Fml_{imp}), we found increased absolute values but similar normalized ones. These results are in agreement with previous studies (Browning & Kram, 2007; Lai et al., 2008; Messier et al., 1996). As the increase of this component had been linked with a decrease of stability (Birrell

et al., 2007), obese individuals while walking are seemingly more unstable compared to normal-weight people. Differently than the protective adaptation evidenced for the vertical GRF, we observed no adaptation in the gait pattern in obese individuals to improve their balance, as similar normalized medial-lateral (peak and impulse) GRF were found.

The end of the stance phase was the period in which the highest pressure peaks and differences between groups occurred. The pressure peaks in the lateral forefoot for the OG reached 659 kPa while in their normal-weight counterparts were 305 kPa. The OG also showed higher values in the central forefoot. In the medial forefoot and distal phalanges regions similar values were observed between groups, while lower values in the hallux for OG were found. Birtane and Tuna (2004) found no differences for the absolute pressure peaks in the hallux and forefoot regions. In contrast, Hills et al. (2001) found higher absolute values for all regions for obese individuals. Possibly, these differences among the studies might be occurred as a consequence of the different levels of obesity assessed (Birtane and Tuna (2004): 32.2 kg/m²; Hills et al. (2001): \approx 38.8 kg/m²; our study: 37.4 kg/m²). Another possible cause of the differences between our study and the mentioned ones (Birtane & Tuna, 2004; Hills et al., 2001) might be the instruments used: an in-shoe pressure system versus pressure plates. In terms gait pattern of plantar pressures during walking, we could not compare our findings with others as the aforementioned studies did not show normalized data. We observed that even escalating the data by BW the differences between the lateral forefoot and hallux regions between groups continued. These results indicate that the lateral forefoot in the obese participants was loaded more than did the magnitude of their extra BW, while the hallux seemed to be protected.

The times of occurrence were later in the Fml and midfoot (medial and lateral) pressure peaks for the OG compared to NW. This can be explained by the increased calcaneal fat pad characteristic in obese people (Mirrashed et al., 2004), which might have promoted a delay in shifting the forces from the rearfoot to the midfoot. In the current study, no differences in the duration of the stance phase and gait speed between groups while walking at their preferred speed were found. Dufec et al. (2012) also found similar self-selected walking speed between normal-weight (1.25 m/s) and obese adolescents (1.17 m/s). On the other hand, Hulens et al. (2003) used the 6-minute walk test, and verified that normal-weight (BMI < 26 kg/m²), overweight/obese (BMI between 27.5 and 35 kg/m²) and morbidly obese women (BMI > 35 kg/m²) have differences in walking speeds. The authors (Hulens et al., 2003) found a decreased

speed as the BW increased (2.00 m/s vs. 1.64 m/s vs. 1.50 m/s). Spyropoulos et al. (1991) found a lower preferred walking speed in obese compared to non-obese men (1.09 m/s vs. 1.64 m/s). The walking speed from both OG and NW were slower in our study compared to the mentioned studies (Hulens et al., 2003; Spyropoulos et al., 1991). One possible explanation for that might be that the 6-minute walk test is a longer test (Hulens et al., 2003) compared with that one used in the present study, which was performed in a laboratory over a 6 m walkway. Regarding the latter study (Spyropoulos et al., 1991), the main differences was in the walking speed in their normal-weight participants, however the authors (Spyropoulos et al., 1991) did not provide any information about them for comparison with our normal-weight participants. Since walking speed can influence the kinetic parameters of the gait, and the natural gait pattern can be altered by a controlled speed (Hennig & Rosenbaum, 1991), we believe that the differences between groups found in our study were neither related to the walking speed nor with alterations on the gait pattern as a consequence of a controlled speed.

A high degree of linear dependence was found among the most common plantar pressure parameters (pressure peak, mean pressure and pressure-time integral) (Keijsers et al., 2010). Therefore, we decided to present just one parameter (pressure peaks) to avoid redundant information. Different magnitudes of pressure peaks between studies might occur as a consequence of the plantar peak calculation (Keijsers et al., 2010). Previous studies (Birtane & Tuna, 2004; Hills et al., 2001) did not describe how their pressure peaks were calculated. In our study, we used the sensor peak approach instead of the regional peak approach, as the latter aggregates data from multiple sensors into a single regional value compromising the individual sensor information (Keijsers et al., 2010). The sensor peak approach provides more reliable information and leverages the high resolution of our in-shoe pressure system by analyzing the sensors individually (Keijsers et al., 2010).

This study showed some limitations; namely the distribution between men and women among the participants was not homogenous. However, there are some studies that have found no differences between gender in pressure parameters for normal-weight (Hills et al., 2001; Putti et al., 2010) and obese people (Hills et al., 2001). Also we did not examine the foot structure and the posture of the participants, and these features could influence the plantar pressure parameters (Razeghi & Batt, 2002).

In conclusion, obese adults showed in all set of parameters (GRF, plantar pressure and temporal parameters) differences in magnitudes (absolute) and in gait pattern (normalized data). The obese participants displayed an altered gait pattern to minimize the consequences that their increased vertical and propulsive forces could have in their musculoskeletal system. However, they were not able to improve their balance, as similar normalized medial-lateral GRF were observed between groups. Higher pressure peaks were found in the central and lateral forefoot, lateral midfoot and central rearfoot regions. The lateral forefoot was the most loaded region, while the hallux and medial rearfoot regions appeared to be protected during obese people's walking. It would be interesting that future studies assessed the influence of different approaches, such as therapeutic relief-insoles or shoes, training of the intrinsic foot muscles, as well as those conditions such as fatigue or incline levels of ground in the GRF and plantar pressure parameters on obese adults' walking.

What does this paper add?

To our knowledge, this is the first study that simultaneously assessed both magnitude and gait pattern of ground reaction forces (GRF) and in-shoe plantar pressures during obese adults' overground walking. We identified not only higher magnitudes of GRF as expected, but also alterations in the gait pattern: the obese subjects showed decreased normalized vertical (load acceptance and trust maximum phases), and horizontal forces (propulsive anterior-posterior). Thus, as a consequence of obesity, there are some strategies in the musculoskeletal system to minimize the joint contact forces and the shear stress while they walked at a self-selected speed. Regarding the plantar pressures, the obese participants showed higher magnitudes of pressure peaks in the central (rearfoot and forefoot) and lateral (midfoot and forefoot) plantar foot regions. Therefore, to prescribe safe exercise routines and avoid foot-related injuries, these regions should be carefully and frequently checked. When we analyzed normalized data, the lateral forefoot continued showing higher pressure peaks, whereas the medial rearfoot and hallux appeared to be protected as lower pressure peaks were observed. We found the highest pressure peaks in the lateral forefoot. These values were more than 200 kPa higher than all other regions indicating that this region needs special care. Clinicians and trainers should pursue pressure-relieving interventions to improve the plantar pressure distribution in obese adults.

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CHAPTER 4: Original Research

**SPEED AND OBESITY INFLUENCE THE GAIT
PATTERN OF GROUND REACTION FORCES AND
PLANTAR PRESSURES**

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Abstract

Background: Walking is highly recommended as a technique for losing weight. To prevent lower limb injuries and to prescribe safer exercise routines, it is important to understand the effects of obesity and speed on biomechanical gait parameters. Thus, we analyzed the influence that walking speed and obesity had on the pattern of ground reaction force (GRF) and plantar pressure gait parameters. *Methods:* We assessed the GRF and in-shoe plantar pressures in 17 obese adults (body mass index “BMI”: $37.4 \pm 4.6 \text{ kg/m}^2$) and 17 gender-matched, normal-weight participants (BMI: $22.8 \pm 1.7 \text{ kg/m}^2$) during slow (70 steps/minute) and fast (120 steps/minute) overground walking. *Findings:* There was no interaction between speed and obesity in influencing the gait pattern. However, obesity influenced the GRF impulses and plantar pressure peaks with medium effect sizes, and speed influenced the GRF peaks and impulses (with high effect sizes) and plantar pressure peaks (with medium effect size). All GRF peaks were measured to be higher, and vertical and medial-lateral GRF impulses to be lower when participants walked fast. During the fast walking condition, higher pressure peaks were observed in the rearfoot, hallux, and distal phalanges than during the slow gait. *Interpretations:* Both walking speed and obesity independently influenced the biomechanical gait pattern. Walking fast appeared to have a more aggressive effect on the musculoskeletal system (higher GRF peaks) and used more of the rearfoot and hallux. In obesity, there were adaptations in the gait pattern in which the load acceptance phase was more loaded, requiring special care.

Keywords: Baropodometry; Force plate; In-shoe pressure system; Locomotion; Normal-weight; Obesity; Walking.

4.1. Introduction

Obesity can endanger the integrity of the lower limbs and feet by causing higher weight-bearing forces (Hills et al., 2002). Obese subjects show higher risk of having knee and hip osteoarthritis (Hochberg et al., 1995; Ko et al., 2010), plantar foot ulceration (Vela et al., 1998) and heel pain (Prichasuk, 1994) than non-obese people. Walking is highly recommended as a technique to control and lose weight (Browning & Kram, 2007). Changing walking speed influenced the magnitude and pattern (values scaled to the subjects' body weight) of the ground reaction forces (GRF) (Chiu & Wang, 2007; Chung & Wang, 2010; Goble et al., 2003; Jordan et al., 2007) and plantar pressures (Rosenbaum et al., 1994) in normal-weight subjects. However, how walking speed and obesity is not clear.

To our knowledge, only one study investigated the influence of walking speed on the GRF during the gait of obese subjects. Browning and Kram (2007) assessed 10 obese adults and 10 normal-weight subjects as they walked on a treadmill at different speeds. They found that as the speed increased, the absolute anterior-posterior, medial-lateral, and vertical GRF peaks also increased. On the other hand, there were no differences found between obese and normal-weight subjects when the data were scaled to body weight (BW) (Browning & Kram, 2007). However, it is important to recognize that differences in GRF peaks were observed between overground walking and walking on a treadmill (Riley et al., 2007). Two other studies assessed the influence of obesity in absolute plantar pressures while participants walked barefoot at a self-selected (Birtane & Tuna, 2004) and slow walking speed (Hills et al., 2001). Nevertheless, there is no information about the roles that speed walking and obesity have on influencing the pattern of GRF and in-shoe plantar pressures while overground walking.

Evaluating the plantar pressures is useful for enhancing the comprehension of the foot structure and function (Filippin et al., 2007). Whereas this kind of analysis informs the amount of vertical force applied to each region of the plantar surface of the foot, the GRF analysis provides information not only about the "overall" vertical forces acting in the body, but also about the shear forces. Thus, the association of in-shoe plantar pressures and GRF analyses might provide relevant insights for understanding how speed and obesity interact to influence the kinetics aspects of the gait. This information could be used to prevent lower limb and foot-related injuries, and also to prescribe safer exercises for the obese population.

Therefore, the aim of this study was to analyze the influence of walking speed and obesity on the gait pattern (data scaled to the BW) of the GRF and plantar pressures. We hypothesized the following: that walking speed and obesity will interact in influencing the gait pattern; that obesity will not influence the pattern of the GRF peaks, but will influence the GRF impulses and plantar pressure peaks, resulting in the obese participants showing higher GRF impulses and lower plantar pressure peaks than the normal-weight participants; and that during the fast walking compared to the slow walking condition, we will observe higher GRF peaks and plantar pressure peaks, and lower GRF impulses for both normal-weight and obese participants.

4.2. Methods

Participants

We selected two groups of participants for the study: a normal-weight group (people with body mass indexes—BMIs—between 20 and 25) and an obese group (subjects with BMIs above 30). Participants with any traumatic-orthopedic impairment or difficulty of independent gait were excluded. For the obese participants, there were 12 males (mean age of 37.00 ± 6.06 years old; height of 1.75 ± 0.04 m; body mass of 111.20 ± 10.51 kg; and BMI of 36.23 ± 3.54 kg/m²) and five females (mean age of 36.40 ± 6.02 years old; height of 1.55 ± 0.06 m; body mass of 96.08 ± 10.52 kg; and BMI of 40.21 ± 5.87 kg/m²). For the normal-weight participants, there were also 12 males (mean age of 27.42 ± 3.09 years old; height of 1.74 ± 0.05 m; body mass of 71.98 ± 4.68 kg; and BMI of 23.73 ± 1.14 kg/m²) and five females (mean age of 27.40 ± 1.34 years old; height of 1.60 ± 0.05 m; body mass of 52.92 ± 6.43 kg; and BMI of 20.67 ± 1.81 kg/m²).

Instruments and data acquisition

To record the GRF, we used a Bertec force plate (model 4060-15, Bertec Corporation, Columbus, OH, USA), operating at 1000 Hz, and the Acknowledge software (BIOPAC System, Goleta, CA, USA). We recorded the plantar pressure parameters using an F-Scan in-shoe pressure system (TekScan, South Boston, USA) operating at 300 Hz with about 960 pressure cells with 0.18 mm thick insole sensor, and the F-Scan Research 6.33 software (TekScan, South Boston, USA). To control the cadence of the gait, we used a metronome (Wittner Maelzel Metronome, Germany). We recorded the gait speed by videogrammetry, using three digital video camera

recorders and system Dvideo v.5.0 (Unicamp, Campinas, Brazil) (Figuroa et al., 2003) to capture, synchronize, digitalize and reconstruct the images. We used an external trigger to synchronize the force plate and the in-shoe plantar pressure system by starting them simultaneously.

Procedures

First, we explained all of the procedures of the study to the participants and then we recorded their weight and height. We gave each of the participants fitted black shorts and one reflective marker with a diameter of 1.2 cm, which was placed with adhesive tape at the right great trochanter of the femur. We placed the F-Scan system on the participants without causing any restriction to the gait. All participants then received neutral shoes (ballet sneakers) with the sensor insoles already inside.

Second, the participants familiarized themselves with the test by walking freely over a 6 m walkway. The force plate was embedded in the middle of the walkway. Afterwards, they became familiar with walking at 70 steps per minute (labeled as the slow condition) and 120 steps per minute (labeled as the fast condition). Finally, the participants performed three valid trials for each condition in which they took, at least, two steps before and after reaching the force plate.

Data Analysis

We exported the data from the force plate (three components of the GRF), videogrammetry and in-shoe pressure system (values of each sensor in each frame) to Matlab 7.0 software (MathWorks, Massachusetts, USA) and developed a program to compute and process the relevant variables. The gait speed was calculated by the first time derivate of the great trochanter reflective marker horizontal anterior-posterior position. Considering the GRF data, 10 dependent variables were calculated:

- Duration of the stance phase;
- Fz1 (load acceptance peak): first peak of the vertical GRF;
- Fz1_{imp} (load acceptance impulse): vertical GRF impulse from the beginning of the stance phase to the minimum between the two peaks;
- Fz2 (thrust peak): second peak from the vertical GRF;
- Fz2_{imp} (thrust impulse): vertical GRF impulse from the minimum between the peaks to the end of stance phase;
- Fap1 (braking peak): first (negative) peak from the anterior-posterior GRF;

- $Fap1_{imp}$ (braking impulse): an anterior-posterior GRF impulse from the beginning of the stance phase to the middle zero;
- $Fap2$ (propulsive peak): second (positive) peak of the anterior-posterior GRF;
- $Fap2_{imp}$ (propulsive impulse): anterior-posterior GRF impulse from the middle zero to the end of stance phase;
- Fml_{imp} (medial-lateral impulse): positive medial-lateral GRF impulse from all stance phase.

Regarding the in-shoe pressure system data, the program automatically divided the footprint into 10 plantar foot regions, as previously proposed (Castro et al., 2013): hallux, distal phalanges, medial, central and lateral forefoot; medial and lateral midfoot; and medial, central and lateral rearfoot. One of the researchers of the current study verified this procedure and, eventually, the boundaries between the foot regions were manually corrected. For each of the 10 plantar foot regions the program calculated its corresponding plantar pressure peak, which was considered the highest pressure value shown on the sensor during the stance phase when the participant stepped on the force plate. We used the force plate to calibrate the plantar pressure data as previously proposed (Castro et al., 2013).

In order to assess the gait pattern, the data were scaled to the participants' BW. Thus, the unit of the GRF peaks was "N/BW," the unit of the GRF impulses was "(N/BW).s," and the plantar pressure peaks was "BW/cm²."

Statistical analysis

We used the intraclass correlation coefficient (ICC) to verify the intra-individual repeatability for the variables—duration of the stance phase, $Fz1$, $Fz2$, and pressure peak (all regions together)—for the slow and fast conditions. We computed the mean of each of the participants' three repetitions and then used these mean values to perform all statistical procedures. To analyze the influence of speed and obesity on the dependent measures of GRF peaks ($Fz1$, $Fz2$, $Fap1$ and $Fap2$), GRF impulses ($Fz1_{imp}$, $Fz2_{imp}$, $Fap1_{imp}$, $Fap2_{imp}$ and Fml_{imp}), and plantar pressure peaks (in the 10 foot regions), we conducted three repeated measures MANOVAs with the conditions (slow and fast) as the within-subjects factor, and the BMI (normal-weight and obesity) as the between-subjects factor. For the duration of the stance phase and gait speed, we used a repeated measures MANOVA with speed as the within-subjects factor. We used the partial Eta square (η_p^2) to measure the effect sizes considering that an η_p^2 of 0.01 or less was small, of 0.06 was medium, and of 0.14 or more was large (Stevens, 2002).

Statistical analysis was performed using the Statistica® v.8 software (Statsoft®, Tulsa, USA) with a α value set at 0.05.

4.3. Results

We found a good-to-excellent data repeatability. During the slow condition the variables duration of the stance phase, Fz1, Fz2, and pressure peaks showed ICCs of 0.81 (CI_{95%} 0.57 – 0.92), 0.91 (CI_{95%} 0.80 – 0.97), 0.91 (CI_{95%} 0.79 – 0.96) and 0.92 (CI_{95%} 0.89 – 0.94), respectively; whereas during the fast condition they were 0.91 (CI_{95%} 0.79 – 0.96), 0.81 (CI_{95%} 0.57 – 0.92), 0.91 (CI_{95%} 0.79 – 0.96), 0.95 (CI_{95%} 0.93 – 0.96), respectively.

Gait speed displayed no difference between groups during slow (obese participants: 0.71 ± 0.08 m/s; normal-weight participants: 0.74 ± 0.05) and fast (obese participants: 1.31 ± 0.15 m/s; normal-weight participants: 1.34 ± 0.11 m/s) conditions. However, the speed was significantly different between conditions (slow vs. fast) ($p < 0.001$). This indicates that the speed was adequately controlled between groups and conditions. Likewise, the duration of the stance phase was similar between groups during the slow (obese participants: 1.10 ± 0.07 s; normal-weight participants: 1.11 ± 0.06 s) and fast (obese participants: 0.70 ± 0.04 s; normal-weight participants: 0.67 ± 0.02 s) conditions.

There was no interaction between speed and BMI in the GRF peaks ($F(3, 96) = 1.56$, $p = 0.21$; $\eta^2 = 0.046$); and there was no significant effect of BMI in the pattern of GRF peaks ($F(3, 96) = 0.64$, $p = 0.59$; $\eta^2 = 0.020$). But we observed a significant effect of speed in the GRF peaks ($F(3, 96) = 105.90$, $p < 0.001$; $\eta^2 = 0.770$). All the GRF peaks (Fz1, Fz2, Fap1 and Fap2) were higher (for Fap2 in module) during the fast gait compared to the slow condition (Table 1).

Table 1. Normalized ground reaction force peaks during slow and fast conditions.

Variables	Obese participants		Normal-weight participants		Main Effects (speed)
	Slow condition	Fast condition	Slow condition	Fast condition	
	Mean (SD)	Mean(SD)	Mean (SD)	Mean(SD)	
Fz1 (N/BW)	1.01 (0.03)	1.12 (0.07)	1.00 (0.02)	1.14 (0.09)	p<0.001
Fz2 (N/BW)	1.01 (0.02)	1.11 (0.04)	1.02 (0.01)	1.13 (0.06)	p<0.001
Fap1 (N/BW)	-0.10 (0.02)	-0.18 (0.04)	-0.10 (0.02)	-0.19 (0.04)	p<0.001
Fap2 (N/BW)	0.11 (0.02)	0.21 (0.02)	0.13 (0.02)	0.21 (0.03)	p<0.001

Fz1, load acceptance peak; Fz2, thrust peak; Fap1, braking peak; Fap2, propulsive peak.

There was no interaction between speed and BMI in the GRF impulses ($F(4, 128) = 0.44, p = 0.78; \eta^2 = 0.014$). However, there were significant effects of both speed ($F(4, 128) = 137.83, p < 0.001; \eta^2 = 0.812$) and BMI in the GRF impulses ($F(4, 128) = 5.22, p < 0.001; \eta^2 = 0.140$). In the slow condition $Fz1_{imp}$, $Fz2_{imp}$, and Fml_{imp} were higher than in the fast condition ($p < 0.01$ for all), while similar values ($p > 0.05$) were observed in the $Fap1_{imp}$ and $Fap2_{imp}$ between conditions (Figure 1a). The obese participants showed an increased $Fz1_{imp}$ and decreased $Fz2_{imp}$ compared to their normal-weight counterparts ($p < 0.01$); there were no effects of BMI ($p > 0.05$) on the $Fap1_{imp}$ and $Fap2_{imp}$ and Fml_{imp} (Figure 1b).

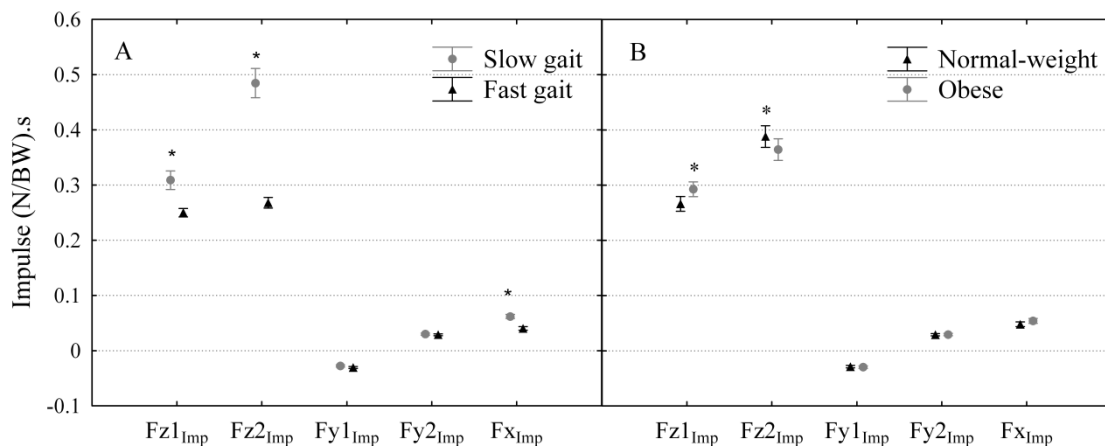


Figure 1. Main effects of (A) walking speed and (B) obesity in the GRF impulses. $Fz1_{imp}$, load acceptance impulse; $Fz2_{imp}$, thrust impulse; $Fap1_{imp}$, braking impulse; $Fap2_{imp}$, propulsive impulse; Fml_{imp} , medial-lateral impulse. The circle and triangle represent the mean and the error bars the 95% confidence interval. * $p < 0.01$ differences between either walking speeds or groups.

There was also no interaction between speed and BMI in the pressure peaks ($F(9, 288) = 0.958, p = 0.474; \eta^2 = 0.029$). But the speed ($F(9, 288) = 3.99, p < 0.001; \eta^2 = 0.111$; power = 100 %) and BMI ($F(9, 288) = 4.78, p < 0.001; \eta^2 = 0.130$) showed significant effects in the plantar pressure peaks during the gait (Figure 2). During the fast condition, we observed higher pressure peaks in the regions of the medial ($p < 0.001$) and central rearfoot ($p < 0.001$), hallux ($p = 0.02$) and distal phalanges ($p = 0.03$), compared to the slow gait; while in the other regions we found similar values ($p > 0.05$) between conditions (Figure 1a). The normal-weight participants showed higher pressure peak values in the great toe ($p < 0.001$) and medial rearfoot ($p = 0.03$) regions compared to the obese participants; in the other eight regions no significant differences ($p > 0.05$) were observed (Figure 2b).

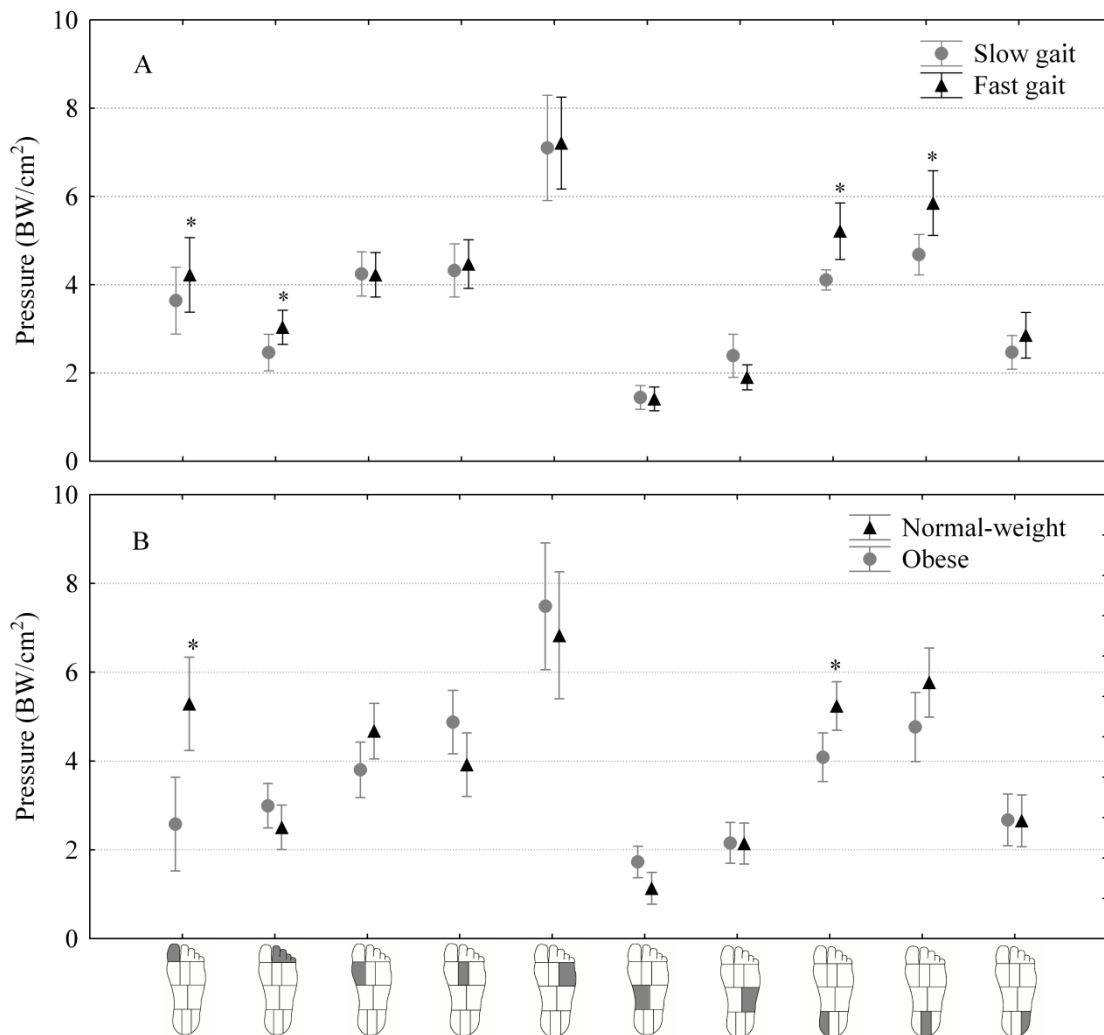


Figure 2. Main effects of (A) walking speed and (B) obesity in the plantar pressure peaks. The circle and triangle represent the mean and the error bars the 95% confidence interval. * $p < 0.05$ differences between either walking speeds or groups.

4.4. Discussion

This study analyzed the influence of walking speed and obesity on the kinetic aspects of the gait pattern. Our first hypothesis was not confirmed, as there were no interactions found between walking speed and BMI in the pattern of the GRF peaks, GRF impulses, and plantar pressure peaks. Our second hypothesis was partially satisfied; there was no influence of obesity in the pattern of the GRF peaks between groups, and we observed higher GRF impulses ($Fz1_{imp}$) and lower plantar pressure peaks (medial rearfoot and great toe) in the obese compared to the normal-weight participants. However, we did not expect any lower GRF impulses in the obese participants as observed in $Fz2_{imp}$ and similar pressure peaks in most of the foot regions. We almost fully satisfied our third hypothesis: during the fast condition, all GRF peaks and pressure peaks in four out of the 10 foot regions (medial and central rearfoot, and great and little toes) showed higher values, and lower GRF impulses ($Fz1_{imp}$, $Fz2_{imp}$, and Fml_{imp}) compared to the slow condition. The influence of either walking speed or BMI on the gait pattern was not only significant, but also appeared to have practical/clinical relevance as we observed medium to large effect sizes.

Browning and Kram (2007) found increased GRF peaks in faster walking speeds. They (Browning & Kram, 2007) found an increase of 7% in $Fz1$ and 83% in $Fap1$ when obese adults changed their gait speed from 0.75 to 1.25 m/s. Similar alterations were found in the present study: in the fast condition (1.31 m/s), the obese participants showed 11% and 80% higher $Fz1$ and $Fap1$, respectively, compared to the slow condition (0.71 m/s). Our results also suggest similar behavior for the GRF propulsion variables ($Fz2$ and $Fap2$), which increased 10% and 91% in the fast condition. Other studies that investigated the speed influence on normal-weight subjects corroborate with these findings (Jordan et al., 2007; Orendurff et al., 2008). Regarding the GRF impulses, we observed lower vertical and medial-lateral GRF during the fast condition, irrespective of the groups. In addition, the obese participants showed different vertical GRF impulses compared to their normal-weight peers irrespective of walking speed.

Increase of mechanical stress is associated with the development of osteoarthritis (Piscoya et al., 2005). The vertical GRF peaks and impulses might provide relevant information about the joint contact forces that may be responsible for developing the condition. In addition, high magnitudes of vertical GRF represent a continuous inability to absorb the BW load while walking (Simpson et al., 2012) and has been considered a major risk factor in overuse injuries (Birrell et al., 2007).

Evidently, the obesity condition would provide an increase of the absolute vertical GRF; on the other hand, the adaptations in the gait pattern as the consequences of obesity and walking speed are not so clear. Our data indicate no differences in GRF peaks between normal-weight and obese participants. This means that obese subjects do present higher magnitudes of vertical GRF peaks as previously evidenced (Browning & Kram, 2007); however, we added the notion that there were no differences in the gait pattern of the GRF peaks. On the other hand, we noted that the pattern of GRF impulses was influenced by obesity. The obese participants showed a higher $Fz1_{imp}$ and lower $Fz2_{imp}$ compared to non-obese people. Thus, the amount of vertical load received by obese participants at the load acceptance phase is higher than the proportion of their excess weight. This is of special importance as this phase of the gait cycle is related with the pathogenesis of knee osteoarthritis (Asthephen & Deluzio, 2008).

When we analyzed the influence of walking speed on the GRF gait pattern, our data indicated higher vertical GRF peaks in the fast condition. On the other hand, the vertical GRF impulses decreased as speed increased. Therefore, it is not conclusive which condition would be more aggressive to the musculoskeletal system in terms of increasing joint contact forces. Other variables seemingly related to foot damage (blister development) are those from anterior-posterior GRF (Knapik et al., 1997). Similar impulse values ($Fap1_{imp}$ and $Fap2_{imp}$) were found between the conditions, and higher peak values ($Fap1$ and $Fap2$) were found in the fast condition. Thus, our data support that foot injury might be more likely to develop while walking fast. The medial-lateral GRF impulse had been used to assess gait stability, in which higher values have been linked to a decrease in balance (Birrell et al., 2007). Thus, our data would suggest that the gait while walking fast is more stable. This might have occurred as a consequence of the longer stance phase, which requires a higher degree of active regulation (Jordan et al., 2007). A study designed to assess gait stability by the analysis of stride-to-stride fluctuations of the gait cycle supports this notion (Jordan et al., 2007); whereas another study assessing kinematic variability of joint angles found the opposite (lower stability in walking fast) (England & Granata, 2007). Thus, the influence of walking speed on gait stability should be further investigated and our interpretation of the Fml_{imp} should be carefully read, as the use of this variable for assessing gait stability is not well established.

Higher pressure peak values were described for the rearfoot (when analyzed as one region) of normal-weight individuals walking fast (Rosenbaum et al., 1994). Our

data partially corroborate with these findings. We have analyzed the rearfoot as three regions (medial, central and lateral). Thus, we do agree that the pressure peaks increase in the medial and central rearfoot regions; however, in the lateral rearfoot, no differences were found between slow and fast conditions. One explanation for these results may be that, as previously suggested (Rosenbaum et al., 1994), at higher speeds there is an increased eversion of the rearfoot and, as a consequence, the applied load during the load acceptance phase shifts medially. Therefore, the medial and central rearfoot are more likely to be overloaded, while changing the walking velocity did not influence the lateral region.

Walking speed did not influence the pressure peak in the midfoot (medial and lateral) in obese and normal-weight subjects. Data from normal-weight people corroborate with these findings (Pataky et al., 2008). During the stance phase, the raised foot arch tended to be flat as a consequence of the body load; this event possibly unwind the windlass mechanism (Hicks, 1954). With the increase of walking speed, the GRF peaks also increased and this might have contributed to the flattening of the arch as well. Thus, with the increase of BW and walking speed, the collapse of the plantar aponeurosis could be expected. Pre-pubescent obese children have significantly flatter feet than their normal-weight counterparts (Dowling et al., 2001). However, it is not clear if the excessive weight bearing is responsible for the longitudinal arch collapse, or whether these alterations remain in the obese adult population (Hills et al., 2002). As similar pressure peaks in the medial and lateral midfoot were found between conditions and between groups, we suggest that, despite a permanent overload condition, obese individuals do have the longitudinal arch function preserved.

Regarding the forefoot region, our data suggest that the pressure peak along the forefoot (medial, central and lateral) is similar between slow and fast walking, and that there was a similar pattern of forefoot pressure peaks between normal-weight and obese subjects. A study comparing normal-weight young adults walking at 0.88 m/s and 1.33 m/s also did not find differences in the pressure peaks between speeds. On the contrary, Rosenbaum et al. (1994) assessed normal-weight participants walking at 0.8 m/s and at 1.7 m/s and found increased pressure peaks in the medial and central forefoot at the highest speed. In the present study the normal-weight participants walked at 0.71 m/s and 1.31 m/s. These results may suggest that there is no difference in forefoot peak pressure when the speed gait changes less than 0.6 m/s. After that, it is possible that higher pressure peaks would be observed in the medial and central

forefoot regions. It is important to highlight the lateral forefoot as the region in which the highest pressure peaks occurred ($\approx 7 \text{ BW/cm}^2$) for both groups. Neither walking speed nor BMI was able to influence these results. On the other hand, the hallux and distal phalanges were more needed during the fast condition; whereas higher pressure peaks in the hallux were observed in the non-obese compared to the obese participants. This suggests that obese people alter their gait pattern in order to protect the hallux.

This study presented some limitations. The distribution between men and women among the participants was not homogenous. However, no differences between gender in pressure parameters for normal-weight and obese people was evidenced (Hills et al., 2001). Only the right lower limbs of our participants were assessed; however, similar GRF values have been shown between limbs (Seeley et al., 2008). We used a metronome, which provides a sound stimulus of cadence, in order to control the walking speed. However, since we observed no differences between groups for both slow and fast walking—and a short and similar 95% confidence interval was found for both groups in the slow (normal-weight: 0.72 to 0.77 m/s; obese: 0.67 to 0.75 m/s) and fast condition (normal-weight: 1.29 to 1.40 m/s; obese: 1.23 to 1.39 m/s)—it appears that using the metronome was successful in controlling the walking speed. In addition, we chose the metronome because of its ease of use; it is also accessible as an application in mobile phones. Thus, it can be used outside the laboratory to monitor walking speed during exercise and, indirectly, to provide a notion about the GRF and plantar pressures. Finally, another possible limitation is that differences in preferred (self-selected) walking speed have been observed between normal-weight and obese subjects (Hulens et al., 2003). Therefore, it is plausible to assume that in our study both groups walked at different percentages of their preferred speed. However, we used this approach because we aimed to verify the effect of obesity in the same speed gait; if we had used the percentage of the preferred walking speed for calculating the slow and fast conditions, it would have been difficult to differentiate the causes of alterations in gait pattern (speed or obesity).

In conclusion, there were no interactions between walking speed and obesity in influencing the gait pattern of GRF and plantar pressure. But we observed that walking speed and obesity independently influenced the gait pattern of these biomechanical parameters. Walking fast appeared to be more aggressive to the musculoskeletal system as we observed higher GRF peaks and pressure peaks in the medial and central rearfoot, and hallux regions. Meanwhile, obesity changed the pattern of GRF vertical impulses, in which higher values were found in the load acceptance phase and

lower values in the thrust phase. Therefore, obese people walking fast might be more likely to develop lower limb and plantar foot injuries mainly at the load acceptance phase.

4.5. References

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CHAPTER 5: Original Research

**INFLUENCE OF PRESSURE RELIEF INSOLES
DEVELOPED FOR LOADED GAIT (BACKPACKERS
AND OBESE PEOPLE) ON PLANTAR PRESSURE
DISTRIBUTION AND GROUND REACTION FORCES**

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Abstract

When the musculoskeletal system is occasionally or permanently loaded it seems to be more likely to get injured. Insoles could be helpful for managing these potential harmful conditions. The aims of this study were: (i) to experimentally verify the influence of two pressure-relieve insoles developed for loaded population on the ground reaction forces (GRF) and plantar pressure peaks during backpackers and obese' adults gait; and (ii) to compare the GRF and plantar pressure peaks among normal-weight, backpackers, and obese people. Based on GRF and plantar pressure data, and on simulations with a finite element model two pressure-relieve insoles were manufactured: flat cork-based insole with (i) corkgel in the rearfoot and forefoot (SLS1) and (ii) with poron foam in the great toe and lateral forefoot (SLS2). The GRF and in-shoe plantar pressure data were recorded from 21 normal-weight/backpackers (age of 25.81 ± 2.47 yrs; BMI of 21.56 ± 3.65 kg/m²) and 10 obese subjects (age of 35.60 ± 4.90 yrs; BMI of 36.50 ± 4.51 kg/m²) during gait under four conditions: only shoes (without insole), shoes (original insole), SLS1, and SLS2. The normal-weight, backpackers and obese participants showed differences in GRF and plantar pressure parameters. The SLS1 did not influence the GRF, but did influence positively the pressure peaks for both backpackers and obese participants; it seemed to be the most appropriate insole to loaded gait population (mainly for the obese adults). The SLS2 was able to decrease the Fz1; however, it did not show positive influence on plantar pressure distribution.

Keywords: Orthoses; Hikers; Obesity; Walking.

5.1. Introduction

The musculoskeletal system is often either permanently loaded, as in obese people, or occasionally loaded as in walkers that wear weighted backpacks (backpackers). In both cases (labeled as loaded populations), alterations in the biomechanical parameters of gait, such as in plantar pressure distribution (Castro et al., 2013; Hills et al., 2001) and ground reaction forces (GRF) (Birrell & Haslam, 2010; Birrell et al., 2007; Browning & Kram, 2007; Castro et al., 2013; Messier et al., 1996; Simpson et al., 2012) has been shown. Possibly, these biomechanical alterations may contribute to the higher incidence of low back pain (Grimmer & Williams, 2000; Skaggs et al., 2006), higher perceived exertion and shoulder discomfort (Simpson et al., 2011), second metatarsal stress fractures (Arndt et al., 2002), muscle strain (Birrell & Haslam, 2009), joint problems (Birrell & Haslam, 2009), and foot blisters (Knapik et al., 1992) found in backpackers; and the loss of mobility (Messier et al., 1996), higher risk of hip and knee osteoarthritis (Felson, 1990; Hochberg et al., 1995; Ko et al., 2010), foot ulceration (Vela et al., 1998), and heel pain (Prichasuk & Subhadrabandhu, 1994) described in obese people.

Foot orthoses is a general term to describe a broad range of devices including heel lifts, lateral/medial wedges, or insoles (custom-made or prefabricated) (Chevalier & Chockalingam, 2012). These devices have been shown to be effective for managing many foot problems (Bonanno et al., 2011; Colagiuri et al., 1995; Cronkwright et al., 2011; Lynch et al., 1998; Sasaki & Yasuda, 1987). They can reduce and redistribute plantar foot pressure and subsequently avoid or decrease foot pain (Burns et al., 2007). However, the exact mechanisms by which foot orthoses work are yet to be fully understood (Chevalier & Chockalingam, 2012), and the biomechanical effect of these devices is far from the simplistic model often proposed in a clinical context (Nester et al., 2003); also, there is a need to establish the most suitable shoe/foot orthoses across clinical or high-risk populations (Rao et al., 2012).

The comprehension of how the forces are distributed on the foot along the stance phase seems to be essential to detecting overloaded regions. The evaluation of the plantar pressures allows assessing the function of the ankle or foot while walking and other functional activities, as the foot and ankle are responsible for providing support and flexibility while weight transferring (Cavanagh & Ulbrecht, 1994). On the other hand, the plantar pressure systems do not provide any information regarding the shear forces. The analysis of the GRF provides global information about the vertical and shear stress forces during gait, whereas the plantar pressure analysis identifies the

distribution of the vertical GRF over the plantar foot surface. The combination of both analyses offer more detailed information about specific features of forces acting on the foot during gait (Castro et al., 2013).

The knowledge of the different adaptations of the body when submitted to occasional or permanent load, and the development and testing of insoles developed specifically for these potential harmful situations may be helpful to further understand the mechanisms of foot orthoses to make physical exercise safe and to prevent injury. The primary aim of this study was to verify the influence of two pressure-relief insoles developed for loaded populations on GRF parameters and plantar pressure peaks during backpackers and obese' adults gait. The secondary aim was to compare the GRF and plantar pressures among normal-weight, backpackers and obese participants.

5.2. Methods

The participants were between 18 and 45 years old, had body mass index (BMI) either lower than 25 or higher than 30 and did not have any traumatic-orthopedic impairment or difficulty with independent gait. Twenty-one participants (10 men and 11 women; age = 25.81 ± 2.47 yrs; body mass = 63.62 ± 6.96 kg; height = 1.68 ± 0.07 m; and BMI = 21.56 ± 3.65 kg/m²) were selected as the normal-weight group. These participants wore loaded backpacks and they were also considered as backpackers group. Ten participants (five men and five women; ages = 35.60 ± 4.90 yrs; body mass = 101.80 ± 20.31 kg; height = 1.66 ± 0.10 m; and BMI = 36.50 ± 4.51 kg/m²) formed the obese group. This project was approved by the local ethical committee and all participants freely signed an informed consent term, based on Helsinki's declaration.

Instruments and Data Acquisition

A Bertec force plate, model 4060-15 (Bertec Corporation, Columbus, USA), operating at 1000 Hz, and the Acknowledge software (BIOPAC System, California, USA) were used to capture GRF. The F-Scan in-shoe pressure system (TekScan, South Boston, USA) operating at 300 Hz with about 960 pressure cells (depending on the size of the shoe) with 0.18 mm thick insole sensor, and the F-Scan Research 6.33 software (TekScan, South Boston, USA) were used to capture plantar pressure data.

Insoles

Peak pressure-relieving was the rationale for the development of the insoles. Thus, based on the analysis of previous in-shoe plantar pressure data from backpackers (Castro et al., 2013) and obese subjects (non published data) during gait, as well as on simulations with a finite element model (FEM) which was adapted from a previous study (Pinto et al., 2011), the features of the insoles were selected for manufacturing and experimental testing.

Finite Element Model (FEM) of the Foot and Insole

From computed tomography medical images of one of the obese participants (male, with body mass of 121 kg) was obtained a single file with the corresponding mesh clouds, using Mimics® v9.1 software (Materialise, Belgium). These mesh cloud were exported as STL files to Solidworks® 2009 (Dassault Systèmes SolidWorks Corporation, Massachusetts, USA), where they were edited and improved to generate a solid part for each 3D object (further information about the FEM in appendix 1).

For the insole construction model, a combined method of optical techniques (laser scanning method) and CAD model adjustments were used to obtain its geometry. Each obtained 3D insole model, with geometrical and material adjustments, was computationally tested at midstance gait cycle by a static simulation with the insole between the foot and the ground in the Ansys Workbench Platform® v.11 (Ansys, Pennsylvania, USA). The data were analyzed qualitatively (based on pressure distribution along the plantar surface) and quantitatively (based on values extracted from eight foot regions – appendix 1).

After analyzing the FEM static simulation data, and the experimental gait data from backpackers (Castro et al., 2013) and obese subjects, two insoles were considered the most appropriate to reduce the pressure peaks. Thus, they were selected for manufacturing by a specialized company of shoe components: 3DCork Lda (Passos de Brandão, Portugal).

Both insoles were full-length, dual-density prefabricated (not customized), and had similar geometry (Figure 1). They were labeled as stress-less shoe insole 1 (SLS1): flat insole made of cork (Young's Modulus = 1060 kPa) with corkgel A30 (Young's Modulus = 7.5 kPa) in the forefoot and rearfoot regions (Figure 1a); and as stress-less shoe insole 2 (SLS2): made of cork with poron foam (Young's Modulus = 63 kPa) in the great toe and lateral forefoot regions (Figure 1b). Unisex casual shoes with

rubber sole (sneaker) with cutting and lining leather (Eject Shoes, Felgueiras, Portugal – Figure 1c), and its original insole (unisex flat insole made of polystyrene and leather lining) were used. The original insoles were removed from the shoes when testing the manufactured insoles.

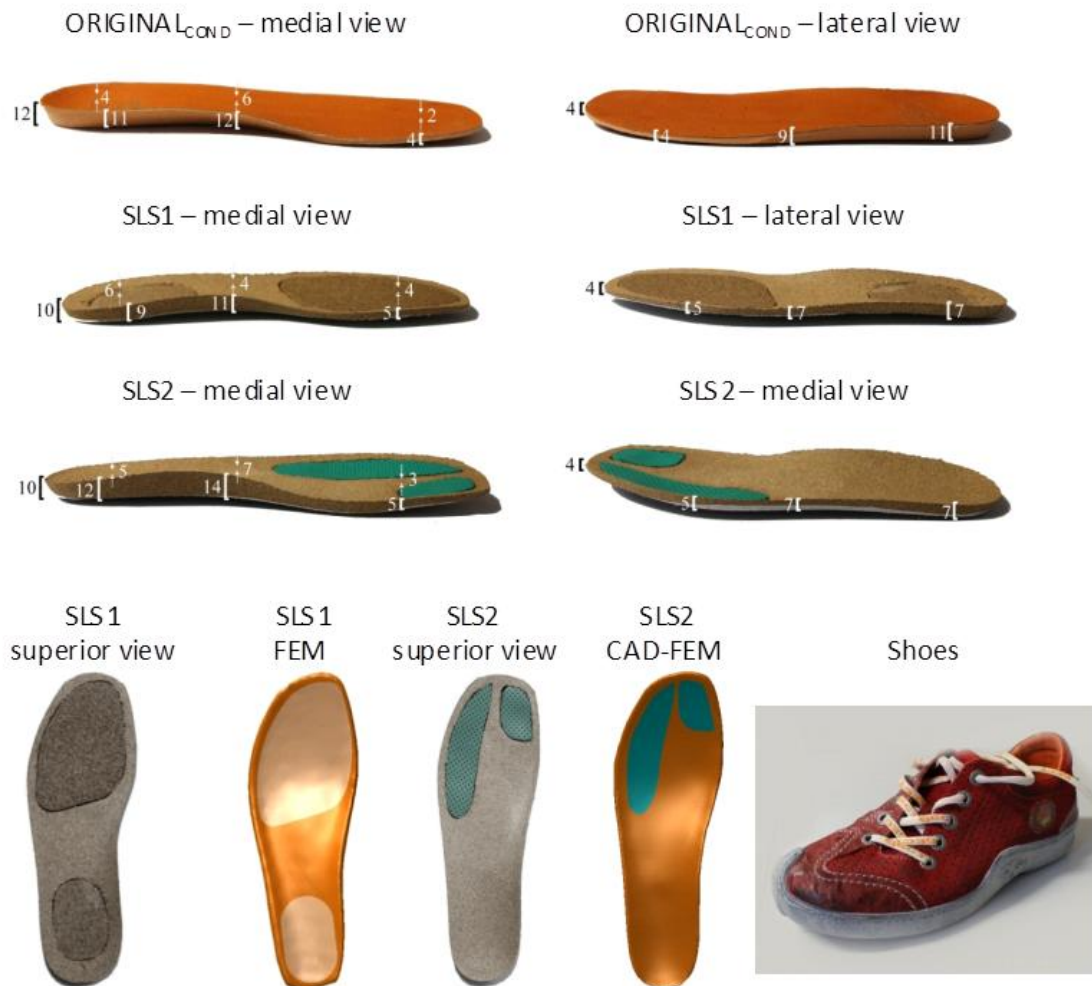


Figure 1. Insoles and shoes used in the study. The numbers in the lateral and medial view figures are the thickness in millimeters. SLS1 and SLS2 – manufactured insoles; ORIGINAL_{LCOND} – shoes’ original insole; CAD- used to generate the finite element mesh.

Tasks and procedures

The participants’ body mass and height were recorded. A cuff unit (VersaTek hub, F-Scan system) measuring 98 x 64 x 29 mm with Velcro strap up was attached on the lateral malleolus region of both legs of each participant. The participants were given a pair of thin socks and a sneaker with the sensor-pressure insole inside. This sneaker was selected due to its regular flat sole and a wide internal space. The participants

familiarized with the setup by walking at a pace of 100 steps per minute, controlled by a metronome (Wittner Maelzel Metronome, Germany), over a 6 m walkway in which a force plate was embedded in the middle. For the testing, the obese participants performed three valid right foot trials in four conditions:

- Shoe-only condition (SHOE-ONLY_{COND}): wearing the sneakers without any insole;
- Original condition (ORIGINAL_{COND}): wearing the sneaker with its original insole;
- SLS1: wearing the sneaker with the SLS1;
- SLS2: wearing the sneaker with the SLS2.

The normal-weight group performed three valid right foot trials wearing the sneakers without any insole, which was labeled as unloaded condition (UNLOADED_{COND}). After that, the mass to raise their BMI to 30 was calculated, and a backpack was filled with sand and fixed at the central area of the participants' back (making them as backpackers group). The mass of the backpack ranged from 19.04 to 25.61 kg (mean load 22.26 ± 1.44 kg). This load criterion was used in order to promote a potential harmful occasional load for the musculoskeletal system (Castro et al., 2013). Then they performed three right foot trial with the same four conditions as the obese participants. The order of the conditions was randomized.

Data Analysis

The data from the force plate (three GRF components) and the in-shoe pressure system (values of each sensor in each frame) were exported to Matlab 7.0 software (MathWorks, Massachusetts, USA) and a program was developed for data processing and calculation of the variables.

Considering the absolute GRF data, the followed parameters were calculated:

- Duration of the stance phase;
- Fz1 (load acceptance peak): first peak from the vertical GRF;
- Fz2 (minimum between peaks): minimum value between the two peaks from vertical GRF.
- Fz3 (thrust peak): second peak from the vertical GRF;
- Fap1 (braking peak): first (negative) peak from the anterior-posterior GRF;
- Fap2 (propulsive peak): second peak (positive) of the anterior-posterior GRF;
- Fml (medial-lateral peak): peak from the medial-lateral GRF.

Regarding the pressure data, the program calculated the in-shoe pressure peak (sensor that showed the highest pressure value during the stance) for 10 foot regions: great toe, little toes, medial, central and lateral forefoot; medial and lateral midfoot; and medial, central and lateral rearfoot, as previously proposed (Castro et al., 2013).

Statistical Analysis

Some variables were chosen arbitrarily to verify the intra-individual repeatability of the three trials. For this, the intra-class correlation coefficient (ICC) was calculated to the stance phase duration, Fz1, Fz3, and for the pressure peaks in the lateral forefoot and central rearfoot.

The mean of the three repetitions of each participant was computed and all the statistical procedures were performed with these mean values. To verify the influence of the insoles on the gait pattern, a repeated measures MANOVA was conducted with the groups (normal-weight/backpackers and obese participants) as between group factor, conditions (UNLOADED_{COND}, SHOE-ONLY_{COND}, ORIGINAL_{COND}, SLS1, and SLS2) as within group factor, and the set of variables (GRF: duration of stance phase, Fz1, Fz2, Fz3, Fap1, Fap2, and Fml; or pressure peaks from the 10 regions) as dependent measures. The partial Eta square (η_p^2) was used to measure the effect sizes considering that an η_p^2 of 0.01 was considered small, of 0.06 medium, and above 0.14 large (Stevens, 2002). The statistical analyses were performed using the Statistica® v.8 software (Statsoft®, Tulsa, USA) with an α value set as 0.05.

5.3. Results

Good to excellent data repeatability was found. All variables showed ICC higher than 0.89.

Differences in GRF among groups

ANOVA interaction among conditions, groups and GRF variables ($F(24, 696) = 25.498$; $p < 0.001$; $\eta_p^2 = 0.468$) with a large effect size was found. Similar stance phase

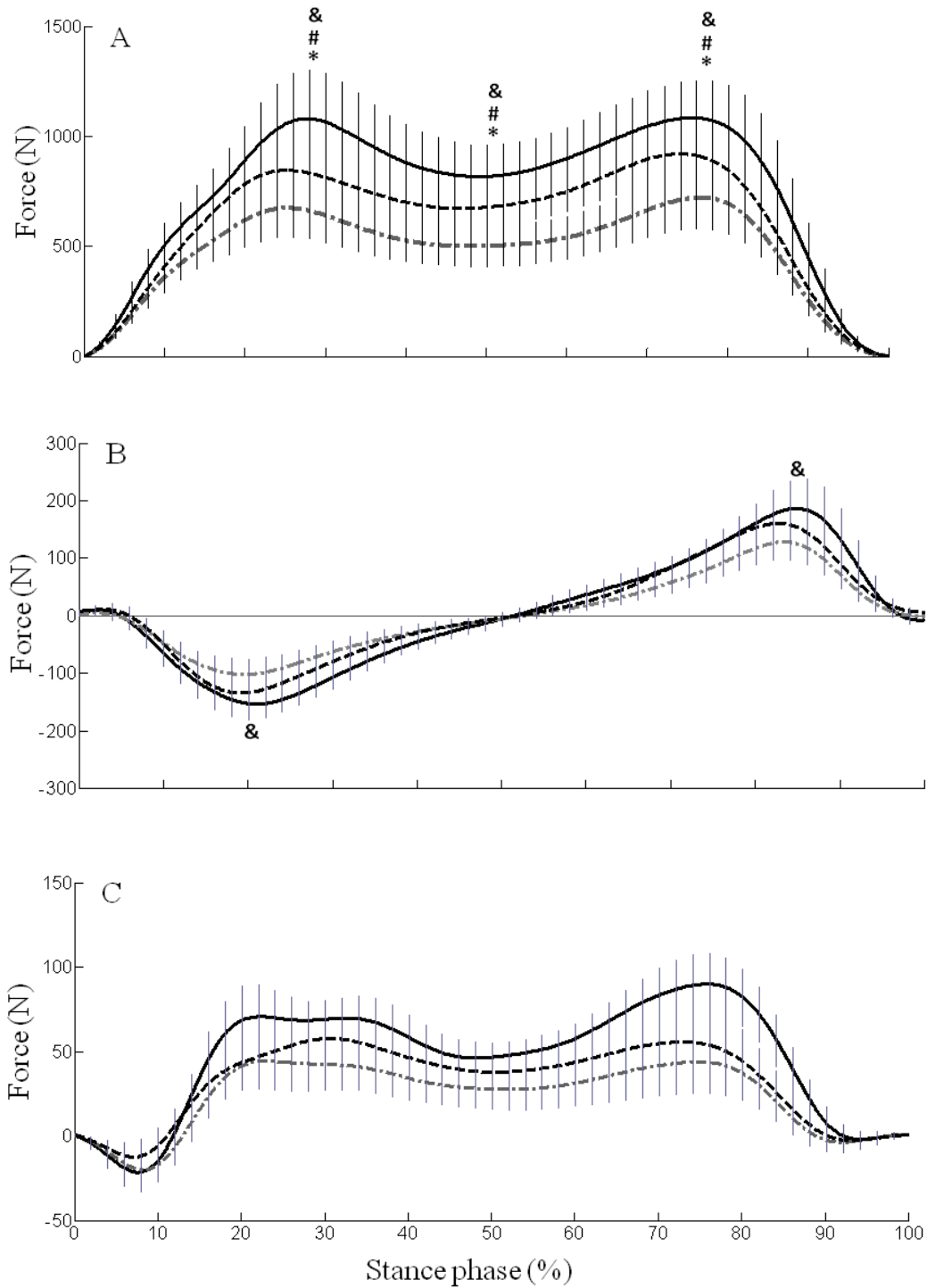


Figure 2. Ground reaction forces (GRF). (A) Vertical GRF; (B) Anterior-posterior GRF; (C) Medial-lateral GRF during normal-weight (UNLOADED_{COND} – dash-dotted gray line), backpackers (SHOE-ONLY_{COND} - dashed black line), and obese participants (SHOE-ONLY_{COND} - continuous black line) during gait. Significant differences ($p < 0.05$) between obese and normal weight (*), between obese and backpackers (#), and between backpackers and normal-weight participants (&).

durations ($p > 0.05$) among normal-weight (0.74 ± 0.03 s), backpackers (0.78 ± 0.03 s), and obese participants (0.77 ± 0.05 s) were found. The obese participants showed higher Fz1, Fz_{Min} and Fz2 compared to backpackers and normal-weight participants, and higher Fap1 and Fap2 than normal-weight participants. The backpackers showed higher Fz1, Fz_{Min}, Fz2, Fap1 and Fap2 than normal-weight subjects. Similar Fml were found among the groups (Figure 2).

Influence of insoles on the GRF

Differences among conditions were found in GRF parameters for both backpackers and obese participants. Obese individuals presented lower Fz1 and Fz_{Min} in SLS2 than in the other conditions. Backpackers also showed lower Fz1 in SLS2 than in ORIGINAL_{COND} and SLS1, and lower Fz_{Min} in ORIGINAL_{COND} than in the other conditions. Similar values for Fz2, Fap1, Fap2 and Fml among conditions during obese and backpackers' gait were found (Table 1).

Table 1. Ground Reaction Forces during backpackers and obese' gait in different conditions.

Variables	SHOE-ONLY _{COND}	ORIGINAL _{COND}	SLS 1	SLS 2	
	Mean (SD)	Mean (SD)	Mean (SD)	Mean (SD)	
Backpackers	Fz1 (N)	860.0 (118.6)	868.0 (113.7)*	863.3 (118.5) [#]	845.2 (105.5) ^{*#}
	Fz _{Min} (N)	664.4 (78.9)*	646.0 (64.3) ^{*#&}	669.0 (76.2) [#]	665.5 (78.1) ^{&}
	Fz2 (N)	928.35 (123.0)	931.4 (114.6)	928.4 (121.2)	919.7 (103.3)
	Fap1 (N)	-141.5 (11.9)	-145.5 (27.6)	-140.7 (23.9)	-145.6 (33.4)
	Fap2(N)	157.8 (29.6)	159.0 (26.4)	153.7 (30.3)	151.1 (24.1)
	Fml (N)	63.0 (14.1)	64.7 (13.1)	61.9 (13.7)	58.8 (10.1)
Obese	Fz1 (N)	1076.4 (226.0)*	1074.8 (233.8) [#]	1078.4 (227.3) ^{&}	1049.8 (227.1) ^{*#&}
	Fz _{Min} (N)	777.1 (164.9)*	784.1 (167.9) [#]	777.8 (171.8) ^{&}	749.3 (132.2) ^{*#&}
	Fz2 (N)	1052.2 (197.0)	1049.4 (202.4)	1060.0 (195.4)	1034.7 (187.0)
	Fap1 (N)	-159.2 (24.1)	-166.8 (25.5)	-166.8 (26.2)	-165.0 (25.7)
	Fap2 (N)	186.2 (50.7)	184.6 (46.8)	183.6 (42.7)	184.6 (44.1)
	Fml (N)	91.5 (18.3)	94.9 (17.8)	95.2 (19.3)	94.0 (20.1)

SHOE-ONLY_{COND} – Shoe-only condition; ORIGINAL_{COND} - Original condition; SLS1 – stress-less shoe insole 1; SLS2 - stress-less shoe insole 2.

* # & - equal symbol between conditions indicate significant difference with $p < 0.05$.

Differences in plantar pressure peaks among groups

ANOVA interaction among conditions, groups and pressure peaks ($F(36, 1044) = 2.138$; $p < 0.001$; $\eta_p^2 = 0.068$) with a medium effect size was found. The obese participants showed higher pressure peaks in the central and lateral forefoot compared to normal-weight (UNLOADED_{COND}) and backpackers; and higher values in the medial forefoot, lateral midfoot and lateral rearfoot than normal-weight participants. The backpackers showed higher pressure peaks in the great toe, little toes, medial and central forefoot, and medial and central rearfoot than walking without a backpack (Figure 3).

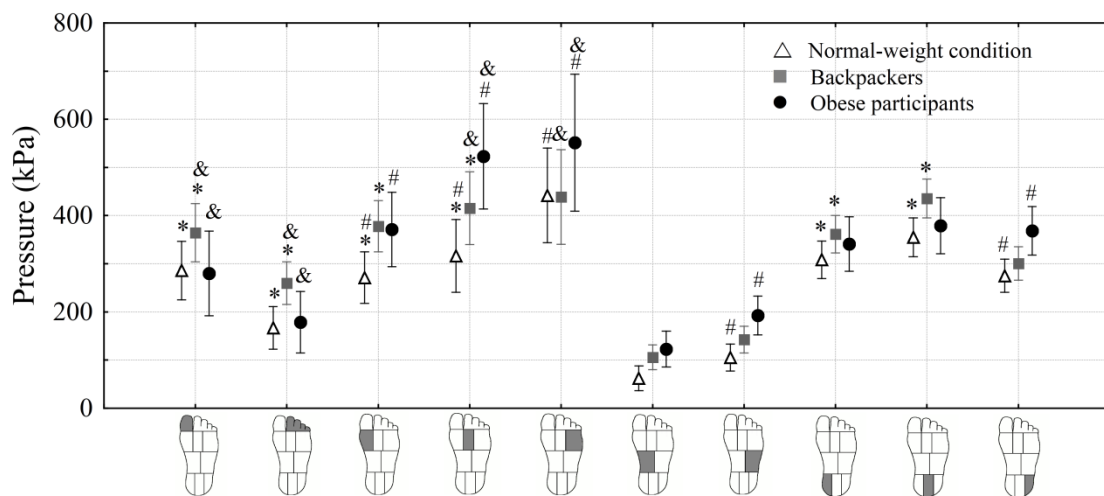


Figure 3. Mean and 95% confidence interval of pressure peaks for the normal-weight condition, backpackers and obese participants (SHOE-ONLY_{COND}). Equal symbols (* # &) between groups indicate significant differences ($p < 0.05$).

Influence of insoles on plantar pressure peaks

Significant differences among conditions were found in backpackers' gait. In the great toe and medial forefoot regions the SHOE-ONLY_{COND} showed lower pressure peaks than the other conditions (ORIGINAL_{COND}, SLS1 and SLS2). The SLS1 showed lower values in the little toes and medial midfoot regions and higher value in the medial rearfoot compared to SHOE-ONLY_{COND} (Figure 4a).

Differences among conditions in pressure peaks were also found for the obese participants. The SLS1 showed higher pressure peaks in the great toe, the SLS2 in the medial forefoot, and the SHOE-ONLY_{COND} in the central forefoot compared to the other conditions. For the lateral forefoot and midfoot the pressure peaks in the SLS1 were

lower than in the SHOE-ONLY_{COND}. Similar pressure peaks in the rearfoot were found among conditions during obese' gait (Figure 4b).

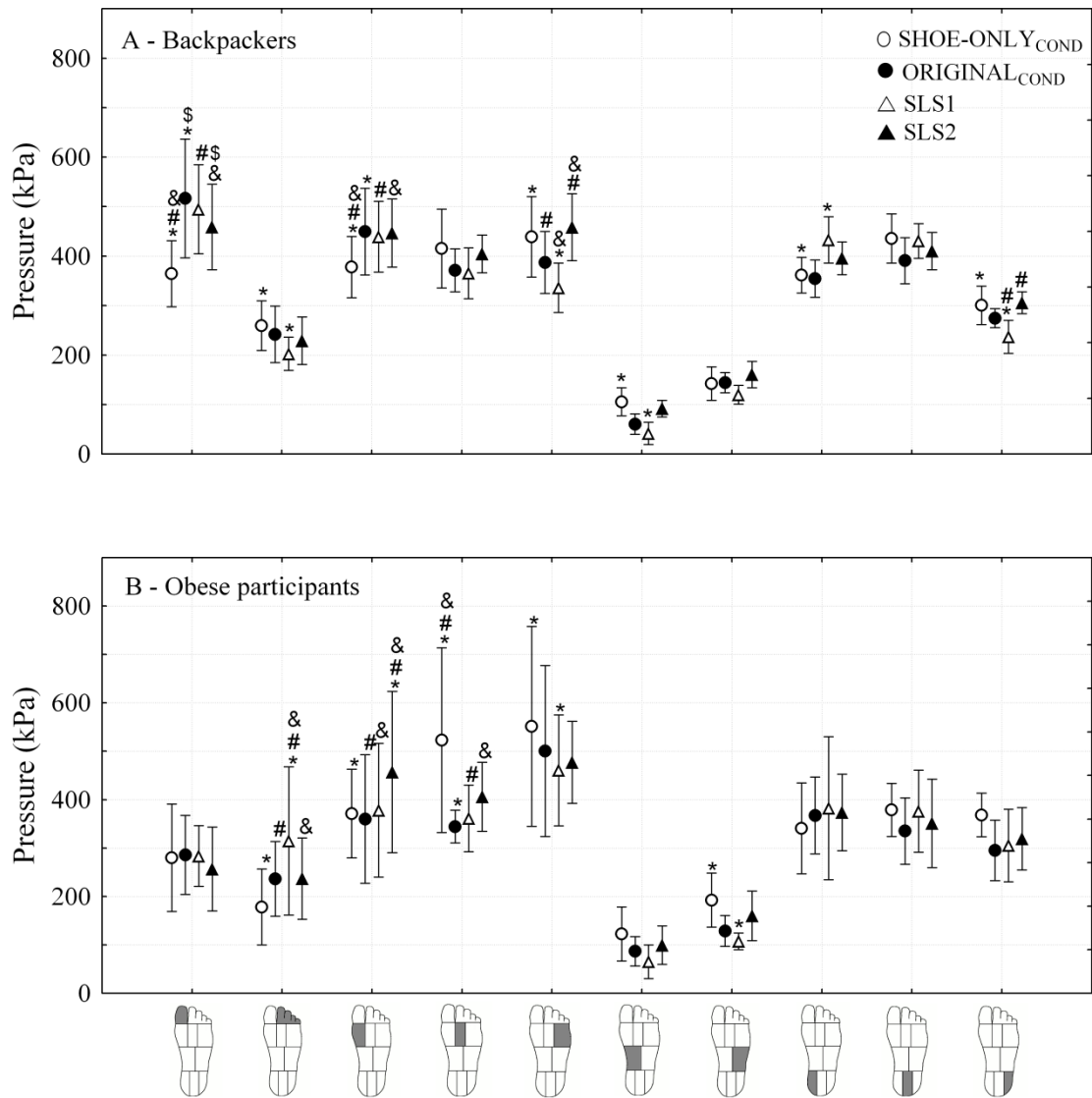


Figure 4. Mean and 95% confidence interval for (A) backpackers and (B) obese while walking in different conditions. Equal symbols (*, #, &, \$) between conditions indicate significant changes ($p < 0.05$).

5.4. Discussion

The aims of this study were to compare the plantar pressure peaks and GRF among normal-weight, backpackers and obese participants, and to verify the influence of two pressure-relief insoles on these parameters during gait. The differences found among the groups (normal-weight, backpackers, and obese participants) and among

conditions (SHOE-ONLY_{COND}, ORIGINAL_{COND}, SLS1, and SLS2) for the backpackers and obese individuals were not only statistically significant, but also appear to have a practical relevance as medium to large effect sizes were found.

Differences in GRF among groups

All analyzed events of the vertical GRF (Fz1, Fz2, and Fz3) suggest a load-dependent behavior. As expected, the obese participants (mean weight: 102 kg) showed higher values than the backpackers (86 kg) and normal-weight (64 kg) group, and the backpackers showed higher values than the normal-weight group. Other studies comparing backpackers (Birrell & Haslam, 2010; Birrell et al., 2007; Castro et al., 2013; Simpson et al., 2012) and obese subjects (Browning & Kram, 2007; Messier et al., 1996) with normal-weight individuals corroborate with these findings. Considering the anterior-posterior GRF and being aware that higher magnitudes in this component have been related to blister development (Knapik et al., 1997). Our data suggest the obese participants may have developed gait pattern adaptations for preventing this kind of injury; while the backpackers seem to be more likely to develop blisters compared to normal-weight group, as evidenced by their higher anterior-posterior GRF peaks. These findings are in accordance with other studies (Birrell & Haslam, 2010; Castro et al., 2013; Simpson et al., 2012). In terms of medial-lateral GRF, no differences were found among the groups. As the medial-lateral GRF is related with gait stability (Birrell et al., 2007), these results were surprising because other studies assessing the GRF component in backpackers (Birrell et al., 2007; Castro et al., 2013; Simpson et al., 2012) or obese people (Browning & Kram, 2007), and kinematic studies (Lai et al., 2008; Qu, 2013) found greater instability in the loaded population' gait. We assessed the gait during over-ground level walking under a controlled pace. The aforementioned studies assessed the gait either with a self-selected speed (Birrell et al., 2007; Castro et al., 2013; Qu, 2013; Simpson et al., 2012), or with a controlled speed on a treadmill (Browning & Kram, 2007). These methodological criteria might cause these differences between the current study and others. Overall, our data support similar stability conditions among normal-weight group, backpackers and obese people walking over-ground at a controlled pace showed.

Differences in plantar pressures among groups

Only few studies have investigated the plantar pressure distribution during loaded gait (Birtane & Tuna, 2004; Castro et al., 2013; Hills et al., 2001). A previous study (Castro et al., 2013) found higher pressure peaks in nine out of ten foot regions

in backpackers compared to a normal-weight condition (only the lateral midfoot showed similar values). In the present study higher values were observed in six out of ten regions. This differences may occurred because of the different shoes used in the studies: casual shoes with rubber sole (present study) versus ballet sneakers (Castro et al., 2013).

The obese participants presented an overall increased in the pressure peaks while walking compared to the normal-weight subjects. Two studies (Birtane & Tuna, 2004; Hills et al., 2001) assessed the plantar pressures in obese adults; in both of them, the rearfoot and midfoot were each considered as one region. Higher (Hills et al., 2001) and similar (Birtane & Tuna, 2004) values were found in the rearfoot, and the midfoot showed higher values (Birtane & Tuna, 2004; Hills et al., 2001) when compared to lean individuals. The present study partially agrees with them. We find higher values for the lateral region in both areas (rearfoot and midfoot), whereas similar values were found for the medial and central areas. The present study and a previous one (Hills et al., 2001) found increased pressure peaks in the three forefoot regions in obese participants compared to normal-weight peers. In contrary, another study (Birtane & Tuna, 2004) did not find any difference between these populations. The different levels of obesity may be the cause of these differences among the studies: 36.5 kg/m^2 (present study), 38.8 kg/m^2 (Hills et al., 2001), and 32.2 kg/m^2 (Birtane & Tuna, 2004).

When both loaded gait groups were compared (obese vs. backpackers), different pressure peaks were found. These findings suggest that the load distribution (obese: abdominal – men; gluteofemoral – women vs. backpackers: mid back) and the duration of loading (permanent vs. occasional) may play an important role on the plantar pressure distribution pattern.

Influence of SLS1 on gait

This insole did not influence the GRF parameters, but it did influence the pressure peaks. This condition caused different effects in the groups. In the backpackers: the rearfoot pressure in SLS1 shifted from the lateral to the medial region. This insole also showed an interesting effect in the lateral forefoot, in which it decreased the peaks compared to the other conditions. However, compared to the shoe-only condition, higher values were found in the great toe and medial forefoot. In the obese participants: it seems that the SLS1 is the most appropriated insole as it displayed the most consistent pressure-relieving in all forefoot (region in which the

highest peaks were found) and lateral midfoot regions with this orthosis. Thus, a homogenous application of lower-density material (in the rearfoot and forefoot regions) seems to have a positive effect for relieving pressure peaks during loaded gait.

Influence of SLS2 on gait

The SLS2 promoted attenuation in Fz1 compared to the other conditions for both backpackers and obese participants. Thus, this insole appears to play a relevant role in helping the heel in shock absorption. One possible explanation is the different material applied in the rearfoot region. In this case, the higher density of SLS2 may be more effective in load-acceptance attenuation when compared to the lower-density material of SLS1. However, a positive influence of this condition was not identified in the plantar pressure peaks for both groups: no alterations in the rearfoot, midfoot, and great toe regions were found; whereas, in the forefoot, the SLS2 increased the pressure peaks in the lateral region for the backpackers and in the medial region for the obese participants.

Limitations of this study

The findings of this study should be read considering several limitations. First, only the immediate biomechanical effects of the insoles were investigated; thus, the results may not reflect the long term changes such as the degradation of the insole material and the acclimatizing of the participants. Second, the participants wore standardized shoes which may not be representative of what they typically wear; however, as the shoes can influence the GRF and pressure parameters, this approach was adopted in order to decrease other sources of influence on the data. Third, the FEM is a static model, analyzing the midstance gait, and it was based on only one of the participant's morphology; nevertheless, we also analyzed experimental data of both populations (backpackers and obese) for establishing the insoles' features.

Conclusions

The SLS1 did not influence the GRF parameters but did influence positively the pressure peaks; it seemed to be the most appropriate insole for loaded gait population (mainly for the obese). The SLS2 helped the heel in shock absorption during the heel strike (lowered Fz1); however it did not show any positive influence on the plantar pressure distribution. Several differences in GRF and plantar pressure peaks were found among the normal-weight group, backpackers and obese participants.

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5.6. Supplementary Data

Finite Element Model (FEM) simulation of the Foot

The FEM used in this study was adapted from a previous study (Pinto et al., 2011). From medical images recorded with computed tomography (CT) the DICOM files of one of the obese participants (male with body mass of 121 kg), were used to obtain a single file with the corresponding point clouds, using Mimics® v9.1 (Materialise, Belgium) software (Marques et al., 2008) (**Erro! Fonte de referência não encontrada.**). By using different masks, according to its tissue density, one could separate point clouds of two groups: bone structure (that includes bones and cartilage) and soft tissue (ligaments, muscles, skin, tendons, fat pad).

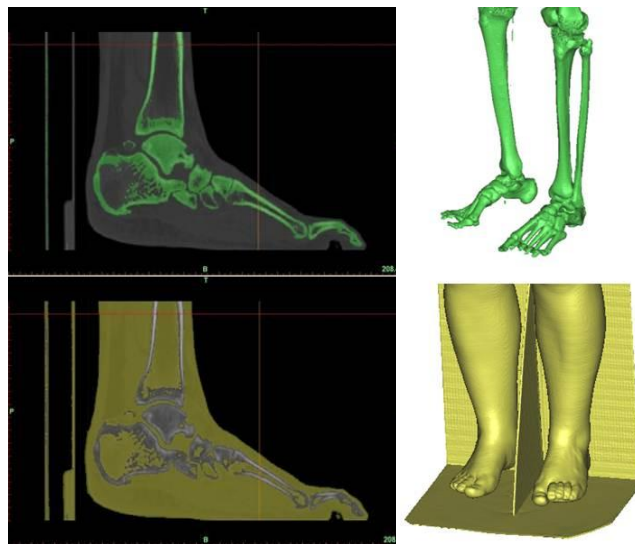


Figure 1. Computed tomography scan of individual foot (left) and mesh clouds from Mimics® (right).

These point clouds were exported as STL files to Solidworks® 2009 (Dassault Systèmes SolidWorks Corporation, Massachusetts, USA), where they were edited and improved to generate a solid file for each 3D object. As bone structure is a complex object and it is difficult to generate surfaces along all the structure. So, the bone structure was anatomically divided into five parts: (i) tibia and fibula; (ii) calcaneus and talus; (iii) cuboid, cuneiforms and navicular; (iv) all the metatarsals, and (v) all the phalanges. Then, solid files for all objects were generated and combined, resulting in a

unified bone structure, which was assembled with the soft tissue and a rigid support was added to simulate the ground.

With Ansys Workbench Platform® v.11 (Ansys, Pennsylvania, USA) the midstance gait cycle stage from stance phase was statically simulated. It was assumed that bone structure and soft tissue were bonded in the corresponding contact surfaces and edges, and five springs were added to the bone structure in order to simulate the most important tendons in the plantar fascia. The ground was vertically recessed, with no longitudinal expansion and rigid, while the tibia and fibula were fixed (Figure 2).

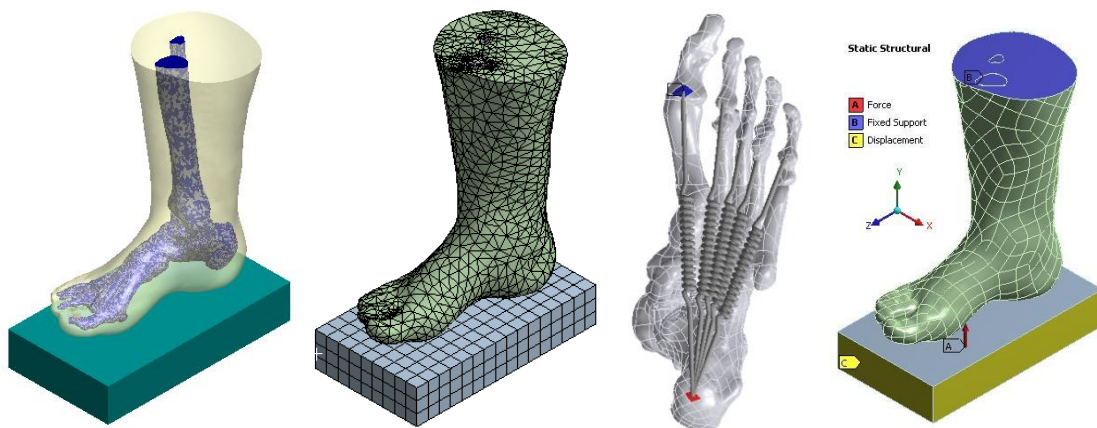


Figure 2. 3D foot model after being processed in Solidworks®, foot mesh, springs to simulate tendons in the plantar fascia and foot mesh for static simulation in Ansys® (from left to right).

Bone structure, soft tissue and tendons were considered linearly elastic, isotropic and homogeneous (Cheung et al., 2004). For bone, Young's modulus and Poisson's ratio, were calculated by weighting cortical and trabecular bone, resulting in a value of 7300 MPa and 0.3, respectively; for soft tissues the values were 0.15 MPa and 0.45, respectively; and for the plantar fascia the Young's modulus used was 350 MPa (Cheung et al., 2005). A 605 N vertical load was applied on the ground, in order to get the displacement and stress distributions, attending to the individual body mass. Note that, changing the applied load magnitudes, the results would change at the same proportion. Thus, the distribution of the applied load was the focus of the analyses.

It was shown that it is possible to determine the displacement, internal stress, strain and pressure distribution all over the foot, and maximum stress peaks were found mostly on the metatarsal heads, on the calcaneus, and on the contact between tibia and fibula (Erro! Fonte de referência não encontrada.). For plantar pressure, the

maximum peak was 1.556 kPa on the calcaneus zone (Erro! Fonte de referência não encontrada.).

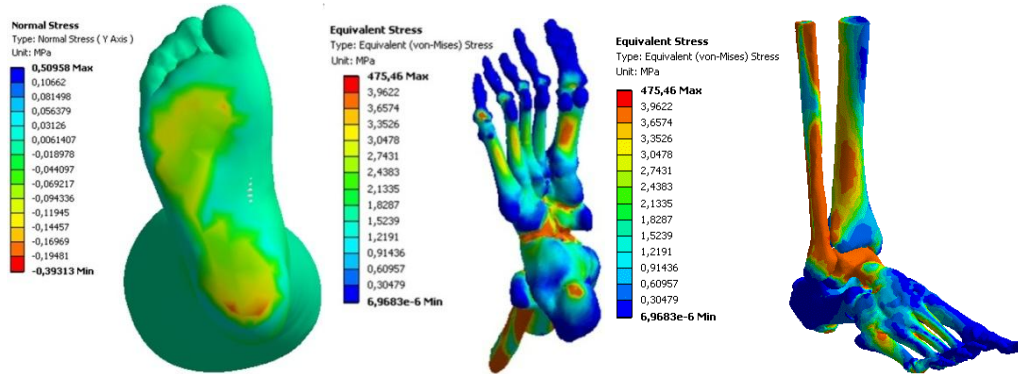


Figure 3. Results for pressure and stress distribution of the obese foot.

However to get accurate simulation results, foot geometry and foot-ground contact have to be very well defined, which is difficult to achieve, due to foot alignment and its position in the computed tomography scan and geometrical separation of the foot with the ground in the model. Thus, it is pertinent to refer that a qualitative analysis was performed based on these results. So, attending to this approach, a mesh of a foot with an insole model was reproduced using the same method, including an insole between the foot and the ground.

Finite Element Model of the Insole

For the construction of the insole model it was used a combined method of optical techniques to obtain its geometry from the point cloud, namely: the laser scanning method and the CAD model adjustment (Figure 4).

Firstly, with the laser scanning method one obtained, through point clouds, the real shape of some original insoles already existing in Portuguese shoe market (Figure 4), and with these files a 3D CAD model of each one was reproduced in Solidworks®.

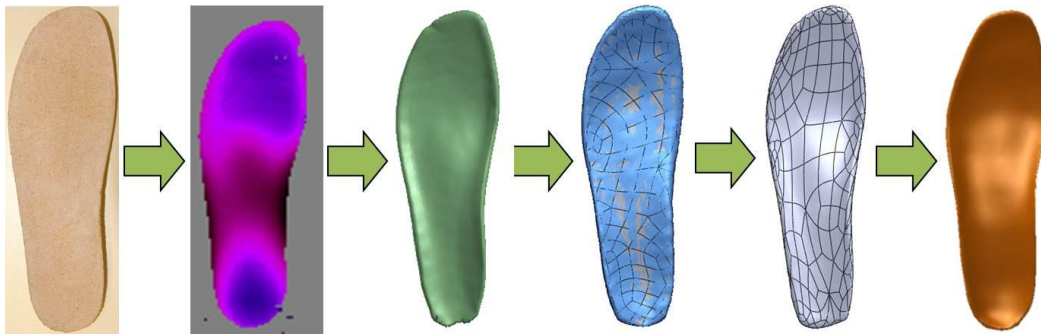


Figure 4. 3D original insole model processing in Solidworks®.

Analyzing FEM static simulation results from the midstance gait cycle stage for the obese foot, it was possible to geometrically adjust in Solidworks® some of the less functional aspects of each one of these insoles, mainly its mechanical properties and geometry, to enhance stress and pressure distribution, i.e., reducing its maximum peak values and homogenizing its distribution, appropriate to an overweight load. Based on previous data from a population occasionally loaded (Castro et al., 2013) and permanent loaded (non published data), geometrical adjustments and material changes were developed in each insole. The insoles were cork-based and have included areas of cork gel, latex foam, and poron foam (**Erro! Fonte de referência não encontrada.**). Hence, several insole geometries and materials were reproduced in 3D CAD models.

Table 1. Young modulus of materials tested.

Material	Young Modulus (kPa)
Cork	1060
Poron Foam (PV)	63
Latex Foam (L-30)	82
Corkgel A15	22
Corkgel A30	7.5

Each obtained 3D insole model with geometrical and material adjustments were tested in the FEM static simulation for the midstance gait cycle stage for its effects on the foot, with the insole between the foot and the ground (Figure 5).

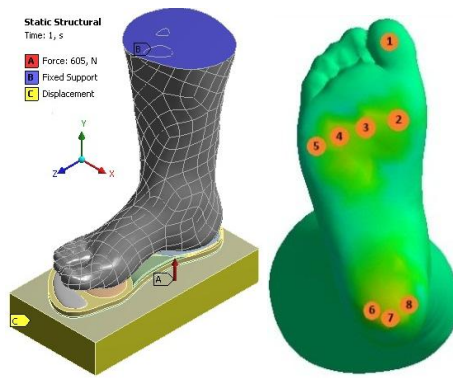


Figure 5. FEM static simulation for the midstance gait cycle of the foot with the insole (left). Positioning of the 8 regions analyzed quantitatively (right).

Figure 6 shows some of the computationally tested models for several shapes with latex foam, poron foam and corkgel that presented the best results of all the insoles models in reducing pressure and homogenizing its distribution. Besides the subjective analysis of the FEM results, eight specific regions on the foot surface were chosen for quantitative analysis (Figure 5). From all geometries and composition insoles tested in the FEM static simulation, the SLS1 and SLS2 (Figure 7) were the most appropriate to reduce the maximum stress and pressure peaks, as well as to homogenize the pressure distribution. The insoles described in Figure 7 are cork based, with corkgel A30 in the forefoot and rearfoot for the SLS1 and poron foam for SLS2 (green zones).

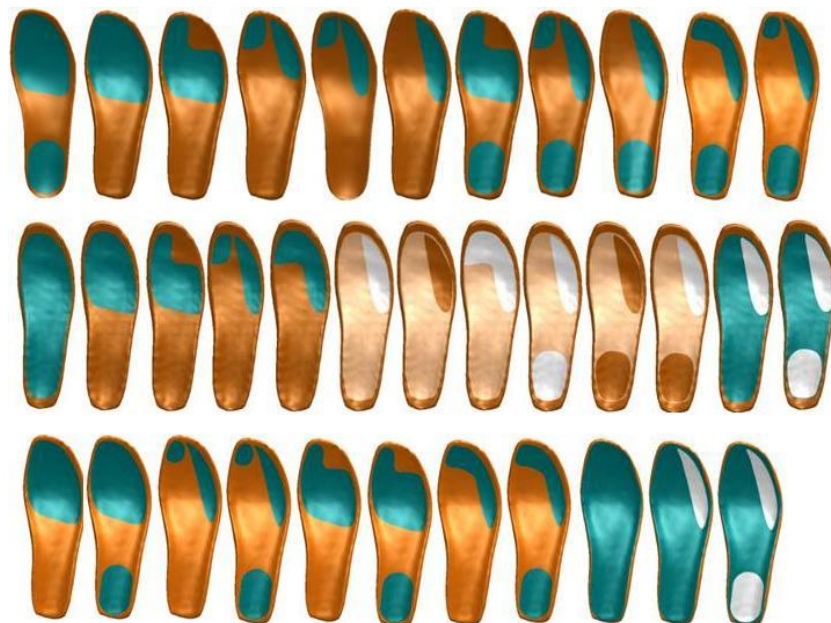


Figure 6. Some of the tested 3D insole models with geometrical and material adjustments.

In these FEM simulations the errors in stress and pressure values, due to the geometry of the foot, the foot-ground alignment and the undifferentiated soft tissues, could be overestimated. However, within the same insole FEM simulation, changes in the material that cause changes in the pressure values are accurate, allowing a precise comparative analysis of the pressure results within the same geometry. This means that the material changes could in fact promote enhanced stress and pressure distribution, by reducing its maximum peaks values and homogenizing its distribution.

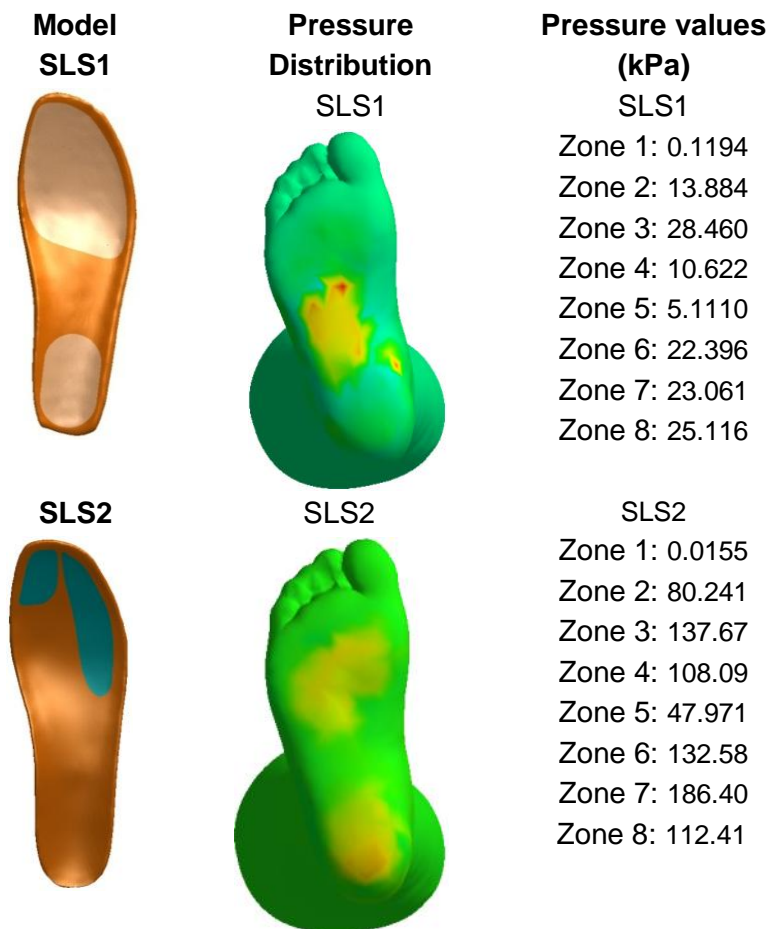


Figure 7. Finite Element Model static simulation: pressure distribution along the plantar surface of the foot and pressure values in the eight foot zones for SLS 1 and SLS2.

Based on these results, the SLS1 and SLS2 were manufactured to be tested experimentally.

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CHAPTER 6: Original Research

**ACCURACY, REPEATABILITY AND NORMAL
VALUES OF A NEW GAIT ANALYSIS DEVICE:
WALKINSENSE®**

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Abstract

WalkinSense® is a new device designed for activity monitoring that provides in-shoe plantar pressure and spatial-temporal measurements during gait. Thus, the aim of this study was to measure accuracy, repeatability and plantar pressure gait values observed in the normal foot obtained by the WalkinSense® system. A bench experiment with ten levels of pressure selected (from 0 to 492kPa) was used to compare the Walkinsense® to the Trublu® calibration device. Afterwards, a dynamic test was carried out overlapping the WalkinSense® and the Pedar® insoles in 40 healthy participants during walking. The Pressure Peak, Pressure Peak Time, Pressure-time Integral and Mean Pressure at eight foot regions were calculated. The bench experiment showed a high Intraclass Correlation Coefficient (ICC = 0.999) within and between-WalkinSense® devices and between the WalkinSense® and the Trublu® devices. The percentage differences were lower than 9 % for nine out of the ten loads applied. The dynamic test indicated an excellent overall (all regions together) within-trial and between-trial repeatability for the four dependent variables assessed. The regional within-trial and between-trial repeatability were good to excellent in 88% of the data and fair in 11%. The overall ICCs (WalkinSense® vs. Pedar®) were higher than 0.95 and the regional ICCs were higher than 0.83 for all variables. The regional percentage differences (WalkinSense® vs. Pedar®) ranged from -31 % to 10 %. The WalkinSense® was found to be repeatable and accurate, and normal plantar pressure gait values collected by it were provided.

Keywords: Plantar pressure; Validation; Reliability; Walking; In-shoe.

6.1. Introduction

In-shoe plantar pressure systems have been widely used by researchers and clinicians in the field of clinical rehabilitation (Bus et al., 2011; Cavanagh & Ulbrecht, 1994), ergonomics (Castro et al., 2013; Hsiao et al., 2002) and sport activities (Fourchet et al., 2012; Tessutti et al., 2010). Such systems allow monitoring the pressure in the interface between the plantar surface of the foot and the insole of a shoe during either static or dynamic activities. Although the operating principle behind each in-shoe system is generally the same, they use different technologies that make them more or less appropriate to certain specific tasks. Particularly, the sensors/insoles collect and send the pressure values to a hub which is usually attached to the lateral malleolus or pelvic girdle. Then, the data are recorded in a memory card or transferred in real-time to a computer by cable, Bluetooth or other wireless means for further analysis. In any case, prior to data collection, a standardized calibration is needed.

The need of a laboratorial setup together with a limited operating time precludes long-time in-shoe plantar pressure measurements in real life circumstances (Hurkmans, Bussmann, Benda, et al., 2006; Hurkmans, Bussmann, Selles, et al., 2006) or during sport activities. Another issue that hinders the use of in-shoe plantar pressure systems for walking analysis, is the influence of the gait speed on plantar pressure parameters (Kernozek et al., 1996). Therefore, to perform complex analysis like comparisons between pathological or healthy populations or between different interventions, other instruments have been used simultaneously to the in-shoe plantar pressure systems to monitor speed gait (Burnfield et al., 2004; Chung & Wang, 2011) increasing the complexity of the experimental setup.

Reliable measures must be ensured before starting to use a new device. To this purpose, validation studies against reliable, gold standard instruments are claimed (Bland & Altman, 1986). In order to verify the reliability of a device, accuracy and repeatability analyses are needed. Accuracy is defined as the difference between the value of a known quantity and the value measured by the device; while repeatability is the difference between two or more measurements performed by the same instrument under the same testing conditions. One of the most used in-shoe plantar pressure devices by clinicians and researchers is the Pedar® in-shoe system (Novel GmbH, Munich, Germany). This system has shown an excellent between-trial (Ramanathan et al., 2010) and between-day (Kernozek et al., 1996; Murphy et al., 2005) repeatability, and it was shown to be accurate (Hsiao et al., 2002).

WalkinSense® (Tomorrow Options SA, Porto, Portugal) is a device designed for activity monitoring that provides plantar pressure and spatial-temporal measurements during gait and running. Gait speed, traveled distance, stride length and frequency, and plantar pressure parameters are some of the measurements provided by this system. Through eight removable force sensing piezoresistors, the WalkinSense® allows recording gait parameters for several days of activity without the need of a standardized calibration. To our knowledge, no previous studies assessed the accuracy and the repeatability of this new device. Finding usefulness in the field of lower limb prophylaxis and rehabilitation, the WalkinSense® system needs to be validated for clinical and sportive usage. Thus, the main purpose of this study was to verify the accuracy and repeatability of the WalkinSense® system. Moreover, normal values for the plantar pressure parameters obtained by the WalkinSense® were provided.

6.2. Methods

Participants

Forty volunteering university students (20 males and 20 females, with mean age of 21.6 ± 3.4 , weight 67.2 ± 11.6 kg, height 170.6 ± 0.9 cm) were recruited as a convenience sample. All subjects were healthy and capable to ambulate independently. The exclusion criteria were any pain or difficulties on independent gait, disabilities that could affect natural gait (musculoskeletal, visual or hearing impairments), and necessity of walking aids. This study was approved by the local ethical committee, and all participants gave their written consent to participate on the study after being informed about the study procedures.

Equipments

The WalkinSense® (weight: 68 g, length: 78 mm, width: 48 mm and thickness: 18 mm) is a CE Mark class I electronic medical device designed to dynamically monitor human lower limbs activity (Figure 1). The device contains a micro electro-mechanical system (MEMS) triaxial accelerometer and one gyroscope, and is connected to a net of eight force sensing piezoresistors (weight: 5 g; size: 1.8 cm^2) for foot pressure measurements that can be freely positioned under or over any insole. This device operates at a sampling frequency of 100Hz in two modes: offline mode, where data is

stored to a SD memory card, and in real-time mode, by communicating with a PC through Bluetooth technology.



Figure 1. WalkinSense® and Pedar® attached on one of the participants during data collection.

For the static experiments, the Trublu® calibration device (Novel GmbH, Munich, Germany) was used. It allows testing applied loads from 0 to 600 kPa.

For the dynamic experiments, the Pedar® system (weight: 400 g, length: 150 mm, width: 100 mm and thickness: 40 mm) was used. The Pedar® records in-shoe plantar pressures through 99 capacitive pressure-sensitive sensors with an area of $\cong 1.5 \text{ cm}^2$ (depending on the size of the insole) and a sampling frequency of 100 Hz.

Data collection

Data collection was carried out into two phases: first, a gold standard bench testing comparison by using the Trublu® calibration device and the WalkinSense®; second, a dynamic experimental test comparing WankinSense® with Pedar® during gait.

Bench experiment (Trublu®)

During the bench testing two WalkinSense® devices were assessed. Each one of the eight sensors of the WalkinSense® nets were positioned under an insole with 2 mm of thickness (provided by the manufacturer) in correspondence to a random sensor and attached by adhesive Velcro straps. Then, the insoles were positioned into the Trublu®, in which ten levels of pressure were sequentially applied during 10 seconds

on each one: 0.00, 23.54, 49.03, 73.55, 99.05, 146.12, 199.07, 294.20, 391.29 and 492.29 kPa. Afterwards, the WalkinSense® and Pedar® were assessed together to check if the two systems work well jointly (Figure 2).

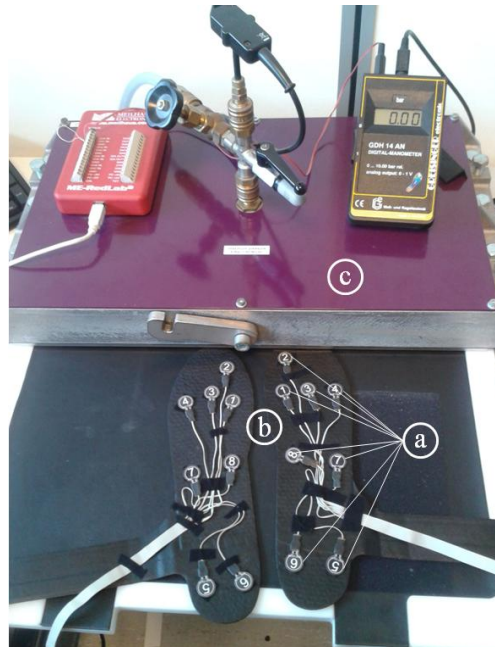


Figure 2. Experimental set of the bench experiment. (a) Position of the eight WalkinSense® sensors on the (b) Pedar® insoles prior to insertion in the (c) Trublu® calibration device.

Dynamic experiment

The dynamic experimental protocol consisted in recording the plantar pressure during gait simultaneously with the WalkinSense® and Pedar® overlapped. Before data collection the Pedar® insoles were checked by the Trublu® calibration device in order to verify the performance of all sensors. The eight sensors of the WalkinSense® were positioned under the Pedar® insole (Figure 3) in correspondence to the main reference foot areas, as proposed and adapted from previous studies (Castro et al., 2013; Cavanagh & Ulbrecht, 1994). The centroid for positioning each of the WalkinSense® sensors on the Pedar® insole was manually identified by pressing a stick with the same area of the sensor on the insole in correspondence to the selected regions. The activity of the Pedar® sensors was controlled on a computer screen and when only the aimed sensor was active, the region was marked. Afterwards, the WalkinSense® sensors were attached to the insole using adhesive Velcro straps. This

procedure was repeated for all pairs of Pedar® insoles. The insoles were put into a neutral pair of shoes (ballet sneaker). Then, the participants stood in the upright position and their weight and height were recorded by a force plate (Bertec Corporation, Columbus, Ohio, USA) and a stadiometer (Seca, Birmingham, United Kingdom). The participants familiarized themselves with the experimental setup by walking freely over a 12 meters walkway at a pace of 100 steps per minute marked by electronic metronome software (Metronome Beat, Andy Stone). Following the familiarization, participants performed a variable number of trials and three valid ones were used for further analysis. In each trial, about 12 steps were recorded and only the central four (two with each foot) were used in the statistical analysis.

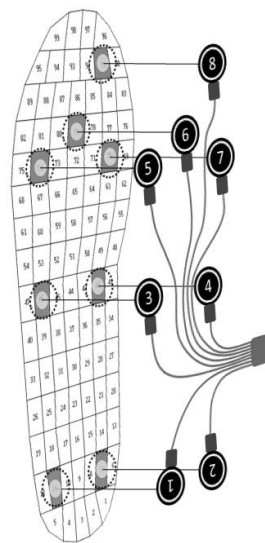


Figure 3. Position of the WalkinSense® sensors at the Pedar® insole.

Data Analysis

Data from Pedar® were recorded by the Pedar-X® software (Novel GmbH, Munich, Germany) and that from WalkinSense® using the WalkinSense® software (Tomorrow Options SA, Porto, Portugal). The sensor pressure values from both systems were exported and then analyzed by Matlab® 7.0 software (MathWorks, Massachusetts, USA) through an appropriate program for data processing and variable calculation.

Bench experiment

At each applied load level, the data of the central second (100 central samples) from the two WalkingSense® devices were analyzed.

Dynamic experiment

The following anatomical regions were studied: great toe (G_{Toe}); medial, central and lateral forefoot (FF_{Med} , FF_{Ct} and FF_{Lat} , respectively); medial and lateral midfoot (MF_{Med} and MF_{Lat} , respectively); and medial and lateral rearfoot (RF_{Med} and RF_{Lat} , respectively). For each sensor of the WalkinSense® and the respective sensor from Pedar®, four dependent variables were calculated: peak pressure (P_{Peak} , in kPa), defined as the highest value displayed by the sensor along the stance phase; peak pressure time (P_{Time} , in % of the stance phase), defined as the instant correspondent to the P_{Peak} ; mean pressure (P_{Mean} , in kPa), defined as the mean pressure during the stance phase; and pressure-time integral ($P_{Integral}$, kPa·s), defined as the integral along the stance phase.

The gait analysis used the mid-gait method as it represents well the normal walk (McPoil et al., 1999) and three trials were performed in order to provide a consistent mean (van der Leeden et al., 2004). Subjects wore standardize shoes and adopted a controlled gait cadence since footwear and walking speed have shown to influence plantar pressures during gait (Burnfield et al., 2004). Gender differences were not considered as a previous article reported no gender influence on P_{Peak} and $P_{Integral}$ parameters (Putti et al., 2010).

Statistical Analysis

Statistic analysis was performed using the SPSS® statistics v.20 software (IBM SPSS, Chicago, USA) and Statistica® v.8 software (Statsoft®, Tulsa, USA). We considered $ICC \leq 0.69$ as poor, $0.79 - 0.70$ as fair, $0.89 - 0.80$ as good and ≥ 0.90 as excellent (Youdas et al., 1991). The 95% confidence intervals ($CI_{95\%}$) were calculated with the ICC and the absolute and percentage differences aiming to verify the uncertainty of them (Deschamps et al., 2009).

Bench experiments

Repeatability

The overall (all regions together), regional within-net (sensor vs. sensor from the same WalkinSense® net) and between-nets (8 sensors vs. 8 sensors from two different WalkinSense® nets) repeatability were verified by the Two-Way Mixed Model (Type: consistency) Intraclass Correlation Coefficient (ICC).

Accuracy

The relation (accuracy) between the applied load (Trublu®) and WalkinSense® was verified by the (i) ICC, (ii) Person correlation coefficient and (iii) absolute (Trublu® – WalkinSense®) and percentage $[(\text{Trublu}^{\circledR} - \text{WalkinSense}^{\circledR}) \times 100 / \text{Trublu}^{\circledR}]$ differences analyses. Negative values indicate the WalkinSense® showing values lower than the Trublu®, while positive values indicate the WalkinSense® showing values higher than the Trublu®.

Dynamic experiments

Repeatability

The overall and regional within-trial (first right step vs. second right step; and first left step vs. second left step) and between-trial (four steps from the first trial vs. second trial vs. third trial) repeatability were verified by the ICC.

Accuracy

The relation (accuracy) between the WalkinSense® and Pedar® records was verified by the (i) ICC (overall and regional) and (ii) overall and regional absolute (Pedar® - WalkinSense®) and percentage $[(\text{Pedar}^{\circledR} - \text{WalkinSense}^{\circledR}) \times 100 / \text{Pedar}^{\circledR}]$ differences analyses.

Normal values

The descriptive statistics included mean and standard deviation for each parameter of the WalkinSense® and Pedar® for each plantar region.

6.3. Results

Bench Experiments

Repeatability

The overall within-net ICC was 0.999 (CI_{95%} 0.998 to 0.999) and the overall between-net ICC was 0.993 (CI_{95%} 0.990 to 0.995).

Accuracy

Excellent ICC and high level of correlation between the applied loads (Trublu®) and WalkinSense® records were found (Figure 4). The absolute differences ranged

from -7.88 to 12.81 kPa along the different applied loads. The percentage differences were lower than 9 % for nine out of the ten load applied. At the first load applied (25.54 kPa) the WalkinSense® showed pressure values 33.48 % lower than the bench. The heaviest applied loads (> 294 kPa) showed the smallest differences (< 2%) (Figure 4).

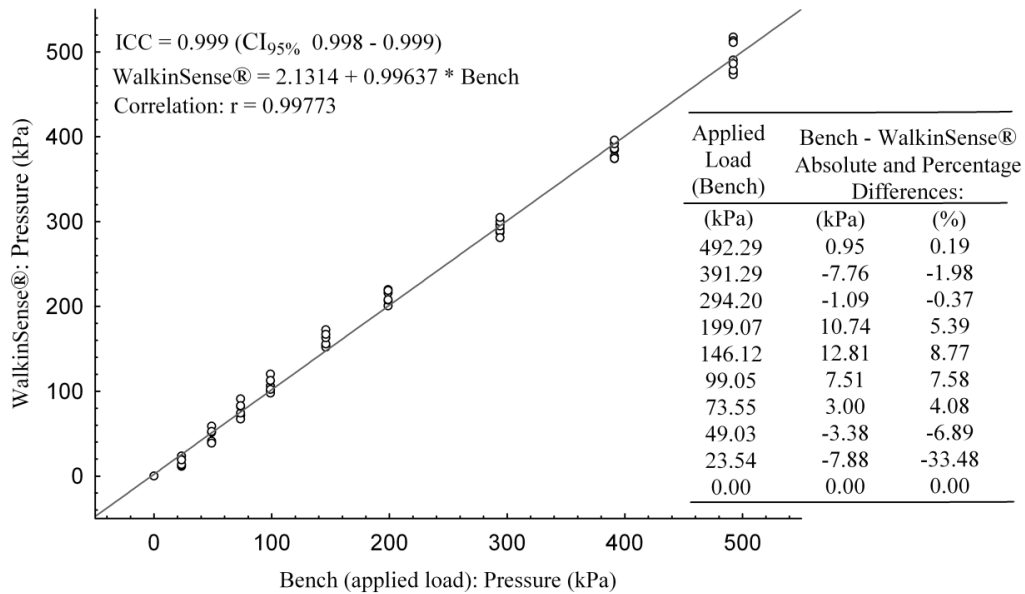


Figure 4. Bench test: relation between the applied loads and WalkinSense® records at the 10 levels of load. ICC – intraclass correlation coefficient. At the second and third column of the table: positive values indicate greater values in the WalkinSense® and negative values indicate lower values in the WalkinSense®.

Dynamic Experiments

Repeatability

Excellent overall within-trial and between-trial repeatability was found in the four dependent variables. All of the range of the ICC CI_{95%} were smaller than 0.02 (Table 1).

Table 1. WalkinSense® within and between-trial Intraclass Correlation Coefficients (ICC) for all measurements (all regions together) during gait.

Variables	Within-trial			Between-trial		
	ICC	CI _{95%}	ICC	CI _{95%}	ICC	CI _{95%}
Peak Pressure	0.972	0.969	0.975	0.979	0.977	0.981
Peak Pressure Time	0.987	0.986	0.988	0.915	0.904	0.925
Mean Pressure	0.940	0.933	0.946	0.993	0.992	0.994
Pressure-time Integral	0.938	0.931	0.944	0.965	0.960	0.969

The regional within-trial and between-trial ICCs for the P_{Peak} were excellent in four (RF_{Lat} , RF_{Med} , FF_{Ct} and G_{Toe}), good in two (FF_{Lat} and FF_{Med}) and fair in one (MF_{Lat}) out of the seven analyzed regions (Table 2). For the P_{Time} the regional within-trial ICCs were excellent, good and poor in one region each and fair in four regions; and the between-trial ICCs were good in five regions, excellent (G_{Toe}) and fair (MF_{Lat}) in one region each. All within-trial and between-trial ICCs for $P_{Integral}$ and P_{Mean} were good or excellent (Table 2).

Table 2. WalkinSense® within and between-trial Intraclass Correlation Coefficients (ICC) for each foot region during gait.

Variable	Region	Within-trial			Between-trial		
		ICC	CI _{95%}		ICC	CI _{95%}	
Peak Pressure	RF_{Lat}	0.978	0.971	0.983	0.971	0.961	0.979
	RF_{Med}	0.964	0.953	0.973	0.961	0.948	0.972
	MF_{Lat}	0.763	0.688	0.821	0.794	0.713	0.855
	MF_{Med}	&	&	&	&	&	&
	FF_{Lat}	0.868	0.826	0.900	0.896	0.858	0.926
	FF_{Ct}	0.949	0.932	0.961	0.956	0.940	0.968
	FF_{Med}	0.834	0.779	0.875	0.858	0.802	0.900
	G_{Toe}	0.928	0.905	0.946	0.966	0.954	0.976
Peak PressureTime	RF_{Lat}	0.718	0.630	0.785	0.859	0.808	0.897
	RF_{Med}	0.763	0.684	0.823	0.826	0.758	0.877
	MF_{Lat}	0.665	0.554	0.749	0.763	0.669	0.834
	MF_{Med}	&	&	&	&	&	&
	FF_{Lat}	0.873	0.831	0.904	0.857	0.801	0.900
	FF_{Ct}	0.786	0.717	0.838	0.808	0.736	0.863
	FF_{Med}	0.748	0.664	0.811	0.843	0.780	0.891
	G_{Toe}	0.900	0.867	0.924	0.946	0.926	0.962
Pressure-time Integral	RF_{Lat}	0.959	0.946	0.969	0.960	0.946	0.971
	RF_{Med}	0.908	0.879	0.930	0.928	0.901	0.948
	MF_{Lat}	0.848	0.798	0.885	0.873	0.825	0.910
	MF_{Med}	&	&	&	&	&	&
	FF_{Lat}	0.890	0.855	0.917	0.912	0.879	0.937
	FF_{Ct}	0.960	0.948	0.970	0.965	0.953	0.975
	FF_{Med}	0.810	0.746	0.857	0.835	0.769	0.885
	G_{Toe}	0.882	0.843	0.911	0.915	0.882	0.939
Mean Pressure	RF_{Lat}	0.965	0.955	0.974	0.961	0.947	0.972
	RF_{Med}	0.903	0.871	0.926	0.928	0.902	0.949
	MF_{Lat}	0.817	0.758	0.862	0.867	0.814	0.906
	MF_{Med}	&	&	&	&	&	&
	FF_{Lat}	0.884	0.847	0.912	0.915	0.883	0.939
	FF_{Ct}	0.959	0.947	0.969	0.962	0.948	0.973
	FF_{Med}	0.816	0.755	0.862	0.840	0.774	0.889
	G_{Toe}	0.890	0.853	0.917	0.916	0.885	0.941

& – as the midfoot region was loaded only in \cong 5% of the trials, the percentage difference and ICC were not calculated for this region.

Accuracy

Analyzing all regions together (overall) the WalkinSense® showed lower values for P_{Peak} (6.2 %), $P_{Integral}$ (14.1 %) and P_{Mean} (13.2 %) compared to the Pedar®. The P_{Peak} occurred slightly later (3.3 %) in the WalkinSense® compared to the Pedar® (Table 3). The overall ICCs (WalkinSense® vs. Pedar®) were higher than 0.95 for all variables (Table 3).

Table 3. Percentage difference and Intraclass Correlation Coefficient (ICC) between Pedar® and WalkinSense®.

	Percentage Differences (%)			ICC	CI _{95%}	
	Mean	CI _{95%}	CI _{95%}		CI _{95%}	CI _{95%}
Peak Pressure	-6.23	-6.84	-5.62	0.973	0.971	0.975
Peak Pressure Time	3.36	3.11	3.61	0.997	0.997	0.997
Pressure-time Integral	-14.08	-14.93	-13.24	0.955	0.952	0.958
Mean Pressure	-13.22	-14.09	-12.35	0.956	0.953	0.959

CI_{95%} – 95% confidence interval

The regional percentage differences between the systems for P_{Peak} ranged from 3.6 % at the RF_{Med} to 16.4 % at the MF_{Lat}. The percentage differences for the P_{Time} were similar among the regions ranging from 2.3 to 4.3 %. In the $P_{Integral}$ and P_{Mean} the highest differences were observed at the MF_{Lat} (\cong 30 % for both $P_{Integral}$ and P_{Mean}) and the lowest differences in the RF_{Med} (7.7 % for $P_{Integral}$ and 6.3 % for P_{Mean}) (Table 4). In 7 out of the 8 regions lower values were observed in the WalkinSense® (negative percentage differences) for the P_{Time} , $P_{Integral}$ and P_{Mean} variables. Only at the FF_{Ct}, the WalkinSense® showed higher values than Pedar®. The range of the CI_{95%} of the percentage differences for the P_{Peak} was \cong 3.5 %, for the P_{Time} it was \cong 1.5 %, for the $P_{Integral}$ and for the P_{Mean} it was \cong 4.5 % (Table 4).

Twenty six out of the 28 regional ICCs (Pedar® and WalkinSense®) were greater than 0.9 with the 95% confidence interval ranging between 0.2 and 0.35. The remaining two ICCs (out of the 28) were 0.86 and 0.83 for the P_{Peak} at the RF_{Lat} and 0.84 for the P_{Time} at the FF_{Ct}.

Table 4. Percentage difference and Intraclass Correlation Coefficient (ICC) between Pedar® and WalkinSense® for each foot region.

Variable	Region	Percentage Differences (%)			ICC	CI _{95%}	
		Mean	CI _{95%}	CI _{95%}		CI _{95%}	CI _{95%}
Peak Pressure	RF _{Lat}	-10.8	-12.4	-9.1	0.859	0.827	0.885
	RF _{Med}	-3.6	-5.0	-2.2	0.958	0.949	0.965
	MF _{Lat}	-16.4	-18.1	-14.6	0.963	0.954	0.971
	MF _{Med}	&	&	&	&	&	&
	FF _{Lat}	-15.2	-16.7	-13.7	0.936	0.922	0.947
	FF _{Ct}	-5.4	-6.9	-3.9	0.953	0.942	0.961
	FF _{Med}	10.4	8.5	12.4	0.945	0.932	0.956
	G _{Toe}	-7.9	-9.8	-6.1	0.953	0.942	0.961
Peak Pressure Time	RF _{Lat}	4.3	3.1	5.5	0.974	0.967	0.980
	RF _{Med}	3.4	2.2	4.5	0.982	0.977	0.985
	MF _{Lat}	3.6	2.9	4.3	0.983	0.980	0.986
	MF _{Med}	&	&	&	&	&	&
	FF _{Lat}	3.7	3.3	4.1	0.964	0.956	0.971
	FF _{Ct}	2.3	1.9	2.7	0.839	0.803	0.868
	FF _{Med}	3.0	2.6	3.5	0.944	0.931	0.954
	G _{Toe}	3.5	3.1	3.8	0.938	0.923	0.949
Pressure-time Integral	RF _{Lat}	-18.4	-20.7	-16.2	0.907	0.888	0.923
	RF _{Med}	-7.7	-9.6	-5.8	0.949	0.937	0.958
	MF _{Lat}	-31.2	-33.7	-28.8	0.937	0.923	0.948
	MF _{Med}	&	&	&	&	&	&
	FF _{Lat}	-24.7	-26.6	-22.8	0.915	0.897	0.929
	FF _{Ct}	8.4	6.4	10.4	0.923	0.907	0.937
	FF _{Med}	-12.2	-13.8	-10.6	0.961	0.952	0.968
	G _{Toe}	-14.0	-15.8	-12.2	0.959	0.950	0.966
Mean Pressure	RF _{Lat}	-17.5	-19.8	-15.2	0.915	0.897	0.930
	RF _{Med}	-6.3	-8.1	-4.4	0.947	0.936	0.957
	MF _{Lat}	-30.1	-32.6	-27.7	0.942	0.927	0.953
	MF _{Med}	&	&	&	&	&	&
	FF _{Lat}	-23.7	-25.6	-21.8	0.913	0.895	0.929
	FF _{Ct}	9.8	7.8	11.8	0.949	0.937	0.959
	FF _{Med}	-11.0	-12.6	-9.4	0.959	0.950	0.967
	G _{Toe}	-13.3	-15.1	-11.4	0.970	0.963	0.975

RF_{Lat} – lateral rearfoot; RF_{Med} – medial rearfoot; MF_{Lat} – lateral midfoot; MF_{Med} – medial midfoot; FF_{Lat} – lateral forefoot; FF_{Ct} – central forefoot; MF_{Med} – medial forefoot; G_{Toe} – great toe; CI_{95%} – 95% confidence interval. & – as the midfoot region was loaded only in \cong 5% of the trials, the percentage difference and ICC were not calculated for this region.

Normal values

Both systems showed the highest and lowest P_{Peak} and $P_{Integral}$ and P_{Mean} values at the FF_{Ct} and MF_{Med}, respectively. Also, the earliest and latest P_{Time} occurred at the same regions in both systems (RF_{Lat} and G_{Toe}, respectively) (Table 5).

In most trials of both systems, there was no pressure at the MF_{Med}. In 472 out of 480 stance phases analyzed (four stance phases x three trails x 40 participants) of the WalkinSense® and 436 out of 480 of the Pedar® the MF_{Med} was not loaded. The four highest values for P_{Peak} (WalkinSense®: 42.2, 32.4, 16.7 and 12.7 kPa; Pedar®: 77.5, 62.5, 57.5 and 52.2 kPa), P_{Integral} (WalkinSense®: 6.2, 4.8, 2.5 and 0.8 kPa.s; Pedar®: 15.6, 13.0, 12.7 and 9.1 kPa.s) and P_{Mean} (WalkinSense®: 10.9, 9.7, 4.3 and 1.4 kPa; Pedar®: 21.7, 19.1, 18.1 and 13.2 kPa) occurred at the same trials in both systems.

Table 5. Mean and standard deviation (SD) for the WalkinSense® and Pedar®.

Variable	Region	Pedar®		WalkinSense®		Variable	Region	Pedar®		WalkinSense®	
		Mean	SD	Mean	SD			Mean	SD	Mean	SD
Peak Pressure (kPa)	RF _{Lat}	291.0	73.0	257.2	71.6	Peak Pressure Time (%Stance)	RF _{Lat}	16.2	5.1	16.9	5.4
	RF _{Med}	285.3	95.8	274.7	96.3		RF _{Med}	17.8	6.5	18.4	6.8
	MF _{Lat}	80.8	35.0	67.8	34.2		MF _{Lat}	48.2	11.5	49.6	11.3
	MF _{Med}	0.8	6.4	0.3	2.7		MF _{Med}	26.7	8.7	33.5	13.3
	FF _{Lat}	250.0	86.7	211.9	80.8		FF _{Lat}	71.0	7.4	73.6	7.4
	FF _{Ct}	302.9	104.1	330.7	110.8		FF _{Ct}	76.9	4.1	78.6	3.6
	FF _{Med}	257.0	88.8	242.4	86.9		FF _{Med}	72.3	6.7	74.3	6.4
G _{Toe}	247.0	121.7	224.2	111.9	G _{Toe}	80.6	6.0	83.3	5.6		
Pressure-time Integral (kPa.s)	RF _{Lat}	70.5	26.6	58.1	27.8	Mean Pressure (kPa)	RF _{Lat}	94.5	33.0	79.1	36.3
	RF _{Med}	70.7	27.0	65.6	28.9		RF _{Med}	96.2	37.1	90.4	39.5
	MF _{Lat}	25.8	10.6	18.3	11.0		MF _{Lat}	34.5	14.4	24.7	14.3
	MF _{Med}	0.4	1.6	0.0	0.4		MF _{Med}	0.5	2.3	0.1	0.7
	FF _{Lat}	82.5	26.0	63.5	27.9		FF _{Lat}	111.5	33.2	86.5	36.1
	FF _{Ct}	84.1	30.7	90.7	35.2		FF _{Ct}	112.9	39.4	123.2	44.9
	FF _{Med}	78.4	28.3	69.9	30.3		FF _{Med}	106.1	37.4	95.7	40.4
G _{Toe}	57.7	32.9	50.4	32.0	G _{Toe}	78.3	43.5	69.0	42.8		

RF_{Lat} – lateral rearfoot; RF_{Med} – medial rearfoot; MF_{Lat} – lateral midfoot; MF_{Med} – medial midfoot; FF_{Lat} – lateral forefoot; FF_{Ct} – central forefoot; MF_{Med} – medial forefoot; G_{Toe} – great toe.

6.4. Discussion

The present study aimed to verify the repeatability and accuracy of the plantar pressure parameters of the WalkinSense® system during gait. For this purpose, two experiments were carried out: a static experiment, in which the WalkinSense® records were compared to a gold standard Bench test (Trublu®); and a dynamic experiment, in which the WalkinSense® was compared to one of the most used in-shoe plantar pressure systems (Pedar®) during gait.

Bench experiments

Our first aim was to assess repeatability of the eight WalkinSense® sensors from a single net and the repeatability of a couple of nets. Secondly, we wanted to

verify the accuracy of the measurements when compared to a gold-standard device. Results reported an excellent repeatability ($ICC > 0.999$) of the measurements for both the single and the couple of nets, together with an excellent overall repeatability and a high Person correlation coefficient between the WalkinSense® and TruBlu® systems. However, further analysis reported that the absolute and the percentage differences varied with the applied load. In fact, the highest percentage differences, -33 % and -7 % were observed at the lowest pressures, 24 and 49 kPa applied. Hsiao et al. (2002) reported similar results in a previous study where the accuracy of Pedar® and F-scan® (TekScan, South Boston, USA) systems were analyzed by bench tests. They found a low accuracy at the lowest pressures in both systems and a gradual reduction of the percentage difference at the higher loads. Pedar® showed percentage differences between -57.2 % and 1.3 % when pressures between 12 and 59 kPa were applied; F-scan® reported a similar trend with percentage differences between 19.4 and 27.9 % for loads between 5 and 41 kPa (Hsiao et al., 2002). On the other hand, in our study we observed the lowest percentage differences, equal to 0.4 and 0.2%, at the highest load magnitudes ($\cong 300$ and 500 kPa). In the same way, when Pedar® and F-scan® systems were loaded with 300 and 500 kPa by Hsiao et al. (2002), low differences, equal to 5.2 and 3.6% and 1.2 and -11%, respectively, were observed. As the thickness of the contact surface in which the sensors are placed decreases their sensitivity at low-pressure ranges (from 10 to 80 kPa) (Sumiya et al., 1998), the insole in which the WalkinSense® sensors were placed on, that had 2 mm of thickness, may play an important role to explain these higher percentage differences found at the lowest applied pressures.

Dynamic experiments

Excellent overall within and between-trial repeatability were found for all dependent variables (P_{Peak} , P_{Time} , P_{Mean} and $P_{Integral}$) in this study. However, the ICCs varied among regions. For both P_{Peak} and $P_{Integral}$, six out of the seven regions (disregarding the MF_{Med} which was not analyzed) reported a good to excellent repeatability while only one, the MF_{Lat} , reported fair values. Also in a study of Kernozek et al. (1996), where the repeatability of the Pedar® system was assessed by the coefficient of variation, different results were obtained among the regions. As in our study, the authors (Kernozek et al., 1996) also found the midfoot (which was considered as a unique region) as one of the least repeatable regions. In another study where the between-day repeatability of a pressure plate (EMED) was assessed in ten foot regions, Gurney et al. (2008) reported that P_{Peak} repeatability was poor to fair in

four regions and good to excellent in six, while the P_{Integral} repeatability was poor to fair in three regions and good to excellent in the remaining seven (Gurney et al., 2008).

In our study, overall high degrees of agreement (ICCs > 0.95) were found comparing gait parameters of WalkinSense® and Pedar®. The overall percentage differences indicate the WalkinSense® showed lower P_{Peak} ($\cong 6\%$), P_{Integral} ($\cong 14\%$) and P_{Mean} ($\cong 13\%$), and the P_{Time} slightly later ($\cong 3\%$) compared to Pedar®. The regional ICCs between the systems were excellent for almost all regions. When considering the regional percentage differences, P_{Integral} and P_{Mean} reported the highest differences: of $\cong 30\%$ in the MF_{Lat} and of $\cong 24\%$ in the FF_{Lat} . However, considering the differences between the systems (kind of sensor, sensor area and layout), we may consider these differences as acceptable.

Normal values

All the P_{Peak} values found in the present study fell in the reference range previously proposed for healthy subjects (Putti et al., 2007). The P_{Integral} magnitudes and the sequence of the P_{Time} along the regions were similar to those presented by Putti et al. (2007). In the present study, the highest P_{Peak} was in the FF_{Ct} (331 kPa), followed by the RF_{Med} (275 kPa) and the $RFLat$ (257 kPa), while in the aforementioned study (Putti et al., 2007) the highest P_{Peak} occurred in the G_{Toe} (280 kPa), followed by the rearfoot (which was not divided, 264 kPa) and forefoot (metatarsal heads I and II, $\cong 247$ kPa). In another study assessing a healthy population (Putti et al., 2008), the highest P_{Peak} was found in the forefoot (metatarsal heads II – 361 kPa and III – 330 kPa), followed by the G_{Toe} (321 kPa) and the rearfoot (313 kPa) using a pressure plate. The differences among these studies could be attributed to external variables such as the use of different shoes (neutral shoes vs. running shoes) or different reference systems (in-shoe pressure system vs. pressure plate).

This study presents some limitations such as i) the standardized position of the WalkinSense® sensors for the four pair of Pedar® insoles did not necessarily correspond to the point of maximal pressure for all subjects; ii) The differences between WalkinSense® and Pedar® (i.e. layout, sensor area and kind); iii) The normal plantar pressure values provided in this study can only be considered for the arrangement of the sensors proposed.

In conclusion, the plantar pressure parameters provided by the WalkinSense® were found to be repeatable and accurate. Four plantar pressure parameters in healthy

adults were analyzed and can be used as normative values for users of the device. Further investigations on gait-running analysis and on the long-term accuracy and repeatability, the between-day repeatability and the accuracy and repeatability of the spatial-temporal parameters of the WalkinSense® are needed.

6.5. References

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Chapter 7

**OVERALL DISCUSSION
&
FINAL CONCLUSIONS**

In the current PhD project three phases were carried out (Figure 1). At the beginning of the phase 1, the characteristics of the main instruments used in this project were analyzed (Appendix I). Afterwards, preliminary studies exploring the biomechanical gait features of backpackers and obese people were developed. Following the analyses of the mentioned studies we felt ourselves ready to start with the main stream of the project: description of the ground reaction forces and plantar pressures of backpackers and obese people (Chapters 2 and 3). The gait of normal-

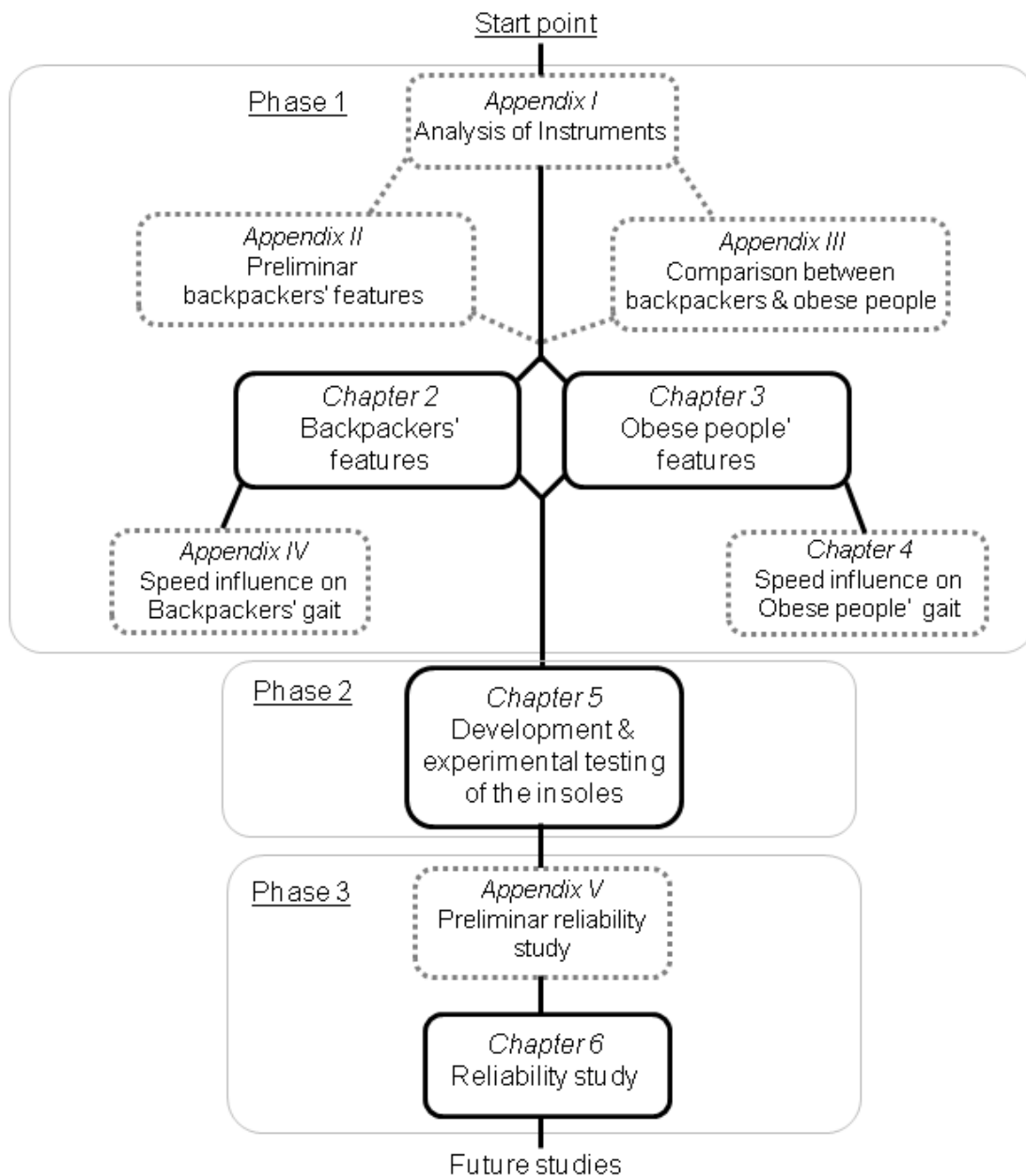


Figure 1. Development of the project. The black line boxes indicate the main stream works, and the grey detached line boxes indicate the secondary streams of the study.

weight people was also assessed in order to verify the influence of occasional and permanent load on the biomechanics of the musculoskeletal system. As complementary analysis for phase 1, we have investigated the influence of speed on the loaded gait (Chapter 4 and Appendix IV). The phase 2 was based on the analysis of phase 1 data and on the results provided by a finite element model. These analyses were the basis for developing the pressure-relieve insoles (SLS 1 and SLS 2). Both insoles were experimentally assessed while backpackers and obese people were walking. Finally, during the development of the phases 1 and 2, we felt the need for a device that allowed us to record biomechanical gait parameters in real life environments. A new device for gait analysis was identified and the accuracy and repeatability of the device was assessed (Appendix V and Chapter 6).

The phase 1 started with the analysis of the systems which could have been used in the project. During gait, forces are transferred between the human body and the ground, starting at the calcaneous and ending in the forefoot (Burnfield et al., 2004). Both force plates and plantar pressure systems are usually used with the purpose of assessing the forces that the body is receiving during any task. While the force plates are considered highly reliable for force measurements during gait (Cobb & Claremont, 1995), the reliability of the plantar pressure systems have been questioned (Nicolopoulos et al., 2000; Rosenbaum & Becker, 1997; Woodburn & Helliwell, 1996). In our first study (Appendix 1), the vertical ground reaction force (GRF) recorded by a force plate and another one recorded (reconstructed) by an in-shoe pressure system (Pedar®) were compared. This study provided two important outcomes: first, an in-shoe system with a thinner insole would be helpful in preserving the internal space of the shoe, mainly because one of the populations that we aimed to investigate was obese people; secondly, the in-shoe pressure system presents good information about the relative distribution of plantar forces; however, it underestimates its magnitude. Other studies corroborate with these findings (Nicolopoulos et al., 2000; Woodburn & Helliwell, 1996). Thus, an in-shoe pressure system with a thinner insole was selected (F-scan®), and the force plate was used to calibrate (post-test) the plantar pressure data test by test. It warranted the maintenance of the internal space of the shoes and the accuracy of the plantar pressure values.

Before starting the analysis of the occasional loaded gait, we faced some doubts in order to define the weight of the backpacks. We intended a load that was both potentially harmful for the musculoskeletal system and ecologically valid. For school children population, 10 to 15% of the body mass is considered as the limit of the

load of the backpack in order to avoid impairment (Lindstrom-Hazel, 2009). However, for the adult population a load limit which would put the human body under stress is not well established. A wide range of high loads is carried by different populations. The total load masses carried by soldiers are in average 40 kg; in some situations they could be required to carry loads of 56 to 76 kg (Reynolds et al., 1999). Korean beverage workers usually carry approximately 53 kg (ranging from 20 to 80 kg) during backpack carrying (Chung et al., 2005). People of New Zealand go tramping carrying backpacks with up to 29% of their body weight for five or more consecutive hours over distances of 11 or more kilometers per day (Lobb, 2004). In all of these populations a high incidence of injury has been described. A limit load of 30% of the body weight was proposed for female recreation hikers (Simpson et al., 2011, 2012). Even though, it is difficult to identify a potential harmful load threshold for the general adult population. The Class I Obesity (BMI > 30) is a well documented risk factor for traumatic-orthopedic injuries being considered as a possible threshold for such dysfunctions (WHO, 2000). Therefore, we have used the BMI = 30 instead of the traditional percentage of the body mass or a fixed load threshold in order to assess the musculoskeletal system over a potentially harmful condition. Some limitations exist, such as the obese people are on a permanent overload while backpackers are occasionally loaded, the mass distribution between obese and backpackers is different, and there is a wide range difference of load or percentage of the body mass inside the backpack between participants (between 19 and 30 kg). Even though, we believe that this was an adequate way for analyzing the human body dynamically and occasionally loaded over a stressful condition.

In regards to the biomechanical characterization of the loaded gait, a few studies have already analyzed the GRF of the backpackers' (Birrell et al., 2007; Chow et al., 2005; Harman et al., 2000; Simpson et al., 2012) and obese adults' gait (Browning & Kram, 2007; Messier et al., 1996). However, the influences of load carriage on the GRF gait pattern (values scaled by the body mass plus backpack mass) have not been investigated. Moreover, the studies assessing the GRF gait pattern in obese subjects are contradictory: Browning & Kram (2007) found similar horizontal components (anterior-posterior and medial-lateral) and lower vertical GRF, while Lai et al. (2008) showed higher anterior-posterior propulsive force and similar vertical GRF in obese people. Therefore, while the absolute values clearly indicate an overall overloading during backpackers and obese people's gait, the normalized ones were either not presented or suggesting some alterations on gait patterns which are not clear. Our data corroborate with this overall increase in the absolute magnitude of the

GRF during backpackers and obese individuals' walking. In terms of GRF gait pattern, both loaded populations were found to walk differently than their normal-weight counterparts. The backpackers showed a protective gait pattern in which a decreased normalized vertical GRF were found at the load acceptance and propulsive gait phases. On the other hand, the shear forces increased more than the proportion of the load, which may help to explain the higher susceptibility to blister development while load carriage. The obese participants have also presented lower normalized vertical GRF than the normal-weight participants. Thus, we can infer that the changing in the GRF pattern is an ability of the musculoskeletal system when imposed to both occasional and permanent loaded condition. However, the obese participants showed adaptations in the anterior-posterior GRF compared to the normal-weight people different than those found in the backpackers (also when compared to the normal-weight subjects). The higher period exposed to load, as in the obesity, may had promoted the ability of also decreasing the normalized shear forces during gait. Regarding the medial-lateral GRF, both loaded populations did not show any differences compared to their normal-weight peers.

Considering the in-shoe plantar pressures, to the best of our knowledge, no studies assessed these parameters either in adult backpackers or obese subjects. Rodrigues et al. (2008) and Pau et al. (2011) assessed the plantar pressure distribution in school children during quiet stance upward position. The former study did not find any influence of backpack (5, 10 and 15% of the body weight) on plantar pressure distribution, whereas the latter found higher plantar pressure peaks in midfoot and rearfoot regions (20 to 30%) while children carried their own backpacks (not a controlled load). Regarding the obese people, two studies have assessed the plantar pressures during barefoot gait (Birtane & Tuna, 2004; Hills et al., 2001). Hills et al. 2001 found the pressure peaks higher in almost all regions of the foot in the obese individuals, while Birtane & Tuna (2004) found higher values of pressure only in the midfoot region of the obese subjects. In both studies (Birtane & Tuna, 2004; Hills et al., 2001), the midfoot and rearfoot were considered as one region and only absolute data were analyzed. Our data showed that during backpackers' gait, higher absolute and similar normalized plantar pressure peaks occurred in seven out of the ten studied regions. It suggests that the plantar pressures increased quite proportionally to the weight of the backpack in the central and medial rearfoot, lateral midfoot, three parts of forefoot and great toe regions. However, the medial midfoot and little toes regions were most needed during occasional loaded gait, and the lateral rearfoot was used less. The obese participants showed differences in the plantar pressures compared to the

normal-weight participants quite different than those differences presented by the backpackers. Higher pressure peaks were observed in the central and lateral forefoot, lateral midfoot and central rearfoot regions under influence of obesity. The lateral forefoot of the obese participants was the most loaded region, while the great toe and medial rearfoot regions seemed to be protected during obese people's gait.

The midfoot region was influenced very differently as consequence of walking occasionally and permanently loaded. Previous studies have described higher pressure peaks in the midfoot in the obese compared to normal-weight people (Birtane & Tuna, 2004; Hills et al., 2001). However, a direct comparison with our data is not valid as we divided the midfoot into medial and lateral regions which appear merged in other studies. Our data reveals that the lateral region of the midfoot presents higher values in obese subjects. However, this behavior is not observed in the medial midfoot region, where similar values were found. Nyska et al. (1997) analyzed the influence of a backpack with 20 and 40 kg had on the plantar pressures of normal-weight subjects and concluded that the human foot adapts itself under loading condition by maintaining the medial longitudinal arch. These adaptations involved shifts of the load to the central and medial forefoot (Nyska et al., 1997). Our data support this maintenance of the medial longitudinal arch function with an adaptation that shifted the pressures to the lateral midfoot and lateral forefoot regions in the obese participants. However, for the backpackers our data did not agree with this theory (Nyska et al., 1997). At the medial midfoot of the backpackers, larger absolute and normalized values were found, which indicates that this region was specially needed. It may be interpreted as an adaptation in the gait pattern or as in a fail of the longitudinal arch function as a consequence of load carriage. For the lateral midfoot, the opposite behavior was observed. Similar absolute and normalized peak pressures were shown between load carriage and normal-weight walking (load free). It may indicate a protective strategy for relieving the pressure on the lateral midfoot by putting more load on the less loaded regions (medial midfoot).

At the last stage of the phase 1 (Figure 1), the influence of speed on the loaded gait was assessed. Many studies have indicated that the GRF (Chiu & Wang, 2007; Chung & Wang, 2010; Goble et al., 2003; Grabowski, 2010; Jordan, Challis, et al., 2007; Jordan, John, et al., 2007; Saha et al., 2008) and plantar pressure parameters (Pataky et al., 2008; Rosenbaum et al., 1994; Segal A, 2004; Warren et al., 2004) are influenced by speed during normal-weight subjects' gait. However, these parameters for backpackers and obese people were almost never investigated. High magnitudes of

vertical GRF may represent a continuous inability to absorb the body weight load during gait (Simpson et al., 2012) and it has been considered as a major risk factor for overuse injuries (Birrell et al., 2007). Our data suggest that the backpackers were submitted to higher total mechanical loads (vertical GRF impulse) and a lower mean vertical GRF during the slow gait compared to fast condition (Appendix IV). The backpacker had more time (larger duration of the stance phase) to dissipate force during slow gait. Considering viscoelastic properties of the musculoskeletal system, it seems to be advantageous. However, during slow gait, the backpackers showed larger magnitudes to propulsive anterior-posterior and medial-lateral impulses when compared to the fast condition. As these variables provide information that could be interpreted as the likelihood of blister development and balance disturbances (Birrell et al., 2007), respectively, at the slow gait, these negative aspects may be more pronounced. The influence of gait speed on the plantar pressures of backpackers was not investigated in this PhD thesis. Regarding the obese participants (Chapter 4), higher GRF peaks, lower vertical and medial-lateral impulses, and higher pressure peaks in the rearfoot (medial and central) and little toes were found comparing the fast to the slow gait. Comparing our obese participants to the normal-weight ones, we observed a similar alteration in the GRF peaks and plantar pressure peaks between them as a consequence of speed changing. However, the GRF impulses changed differently in the obese compared to the normal-weight participants. These data suggest that at a faster speed obese subjects may be more likely to develop plantar foot injury mainly at the load acceptance phase, while the fast gait may be a more stable condition (as indicated by the medial-lateral GRF) for this particular population.

Before starting phase 2, the data from the backpackers (Chapter 2) and obese participants (Chapter 3) were analyzed in order to establish the main features of insoles. After that, the phase 2 started (Chapter 5). With the rationale of peak pressure-relieving, a combination of different layouts and materials were selected for simulating in a finite element model, which was adapted from a previous study (Pinto et al., 2011). Two insoles (called as SLS1 and SLS 2) were considered as the most appropriate to reduce the maximum stress and pressure peaks. Thus, they were selected for manufacturing by a specialized company of shoe components: 3DCork Lda (Passos de Brandão, Portugal). These kind of devices have been shown to be effective for managing many foot problems (Bonanno et al., 2011; Colagiuri et al., 1995; Cronkwright et al., 2011; Lynch et al., 1998; Sasaki & Yasuda, 1987). They can reduce and redistribute plantar foot pressure and subsequently avoid or decrease foot pain (Burns et al., 2007). Therefore, we aimed to assess whether these manufactured

insoles would be powerful for enhancing pressure distribution during the backpackers and obese participants, or not. Analyzing the SLS1 and SLS2, several differences in GRF and plantar pressure peaks among normal-weight, backpackers and obese participants were found. In the backpackers, the SLS1 did not influence the GRF, but it positively influenced the pressure peaks. This insole showed an interesting effect in the lateral forefoot, in which it decreased the pressure peaks compared to the other conditions. In the obese people, the most consistent pressure-relieving in all forefoot and lateral midfoot regions was observed with SLS1. Therefore, this orthosis seemed to be the most appropriate one to loaded gait population (mainly for the obese). The SLS2 was able to decrease the vertical impact peak. However, it did not show positive influence on plantar pressure distribution.

Along the phases 1 and 2, we felt that it would be interesting to evaluate the gait parameters during daily life activities in order to increase the ecological validity of our study and future studies. Therefore, we looked for devices that allowed us to perform this kind of analysis. We have identified the limited time of operation and the need of a laboratorial setup as the main drawbacks for assessing biomechanical gait parameters in real life circumstances. One device that overcomes these problems in the WalkinSense® (Tomorrow Options SA, Porto, Portugal). This device which was designed for activity monitoring provides plantar pressure and spatial-temporal measurements during gait and running. Through eight removable force sensing piezoresistors, the WalkinSense® allows recording gait parameters for several days of activity. However, to our knowledge, no previous studies assessed the accuracy and repeatability of this device. Reliable measures must be ensured before starting to use a new device. To this purpose, validation studies against reliable, gold standard instruments are claimed (Bland & Altman, 1986). Therefore, we started the phase 3 performing a pilot study in which the reliability of the spatial-temporal parameters of the WalkinSense® was verified (Appendix V). The distance measurements, in particular, showed good accuracy and agreement with ground truth data for the ten meter track. But, as our main interest was assessing the plantar pressure parameters, a larger study assessing the accuracy and repeatability during static and dynamic conditions of the plantar pressure parameters was carried out. A bench experiment with ten levels of pressure selected (from 0 to 492kPa) was used to compare the Walkinsense® to the Trublu® calibration device, while a gait analysis test was carried out overlapping the WalkinSense® and the Pedar® insoles for the dynamic evaluation. The plantar pressure parameters provided by the WalkinSense® were found to be repeatable and accurate. Therefore, this device seems to be a reliable option for analyzing plantar

pressures during gait. It can be a useful tool for increasing the ecological validity of biomechanical gait analysis.

Future Work

Further investigations assessing the long-term influence of the developed insoles on the plantar pressures are recommended. The analysis of the effect of different approaches such as other therapeutic relief-insoles or shoes and strengthening of lower limb muscles would be interesting. Moreover, the influence of potential harmful conditions such as fatigue or ground inclination on the plantar pressures and GRF during backpackers and obese adults' gait could be helpful for preventing injury. Finally, the assessment of all these mentioned conditions in real life circumstances with small and portable reliable devices would be of interest.

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APPENDIX

Appendix I: Conference Proceeding I

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 PP 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.

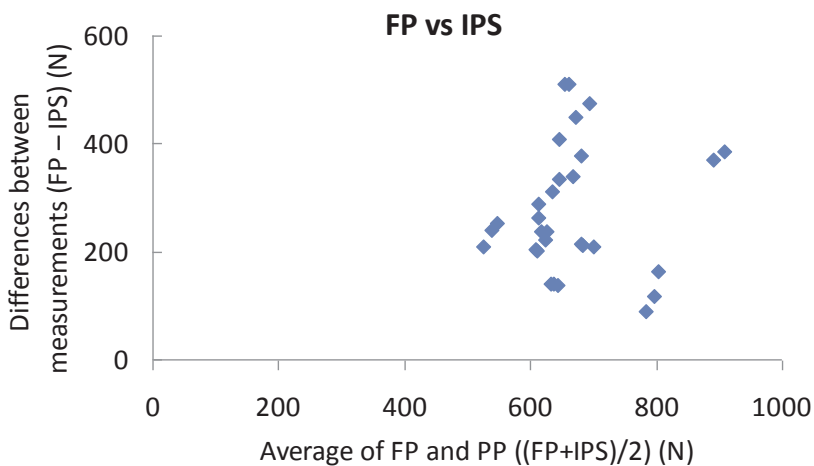


Figure 1: Differences between Force Plate (FP) and Insole Pressure System (IPS).

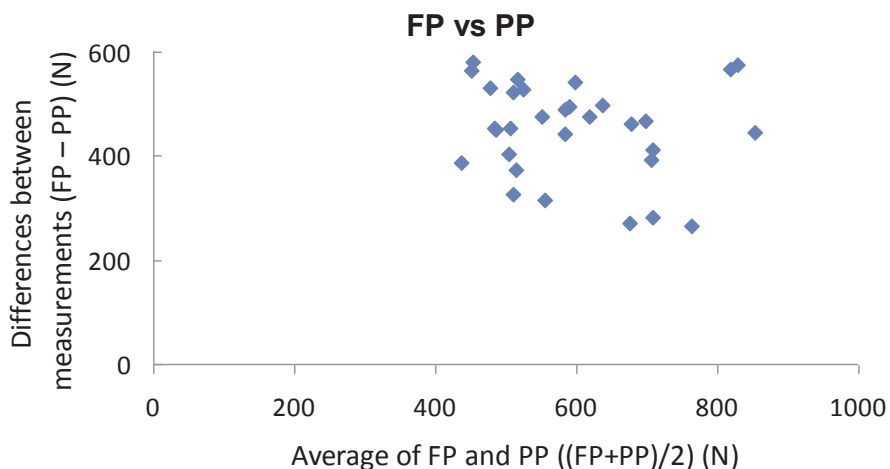


Figure 2: Differences between Force Plate (FP) and Pressure Plate (PP).

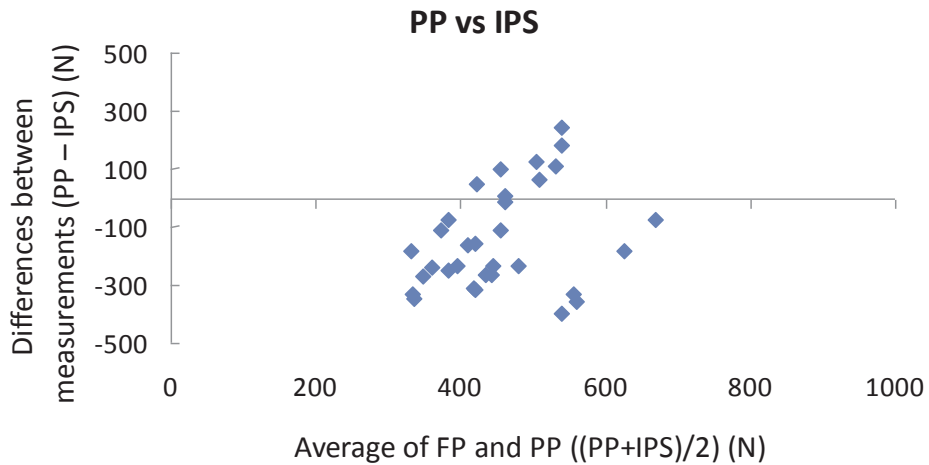


Figure 3: Differences between Pressure Plate (PP) and Insole Pressure System (IPS).

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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Appendix II: Conference Proceeding II

ANALYSIS OF THE BACKPACK LOADING EFFECTS ON THE HUMAN GAIT

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KEYWORDS: gait, load, backpack, ground reaction force.

INTRODUCTION: Gait is a simple activity of daily life and one of the main abilities of the human being. Often during leisure, labour and sports activities, loads are carried over (e.g. backpack) during gait. These circumstantial loads can generate instability and increase biomechanical stress over the human tissues and systems, especially on the locomotor, balance and postural regulation systems. According to Wearing (2006), subjects that carry a transitory or intermittent load will be able to find relatively efficient solutions to compensate its effects. These are dependent upon the walking distance and of the load characteristics - size, weight and location relatively to the body (Hsiang, 2002). Thus, these solutions should become a concerning factor (Koh, 2009) and a topic of scientific research, particularly in what concerns the inventory of its biomechanical effects and the possible strategies to be developed in order to minimize its effects.

The aim of the present study was to analyze the effects of an occasional dorso-lombar load during the gait through the use of a backpack.

METHOD: Data was collected from forty healthy subjects (twenty males: mean stature 1.75±0.07m and mass 72.01±6.75kg; twenty females: mean stature 1.63±0.06m and mass 59.45±5.71kg), students of Sport Sciences with body mass index (BMI) less than 25, aged between 18 and 45 years and without any dysfunction that affect the independent gait. The subjects were informed of the purpose of study and all signed written informed consent.

Gait characterization was accomplished through ground reaction force (GRF) analysis. To collect the GRF data, a BERTEC force plate (model: 4060-15) was used. A devoted amplifier system (BERTEC AM 6300) and a 16 bits analogical-digital conversion unit (BIOPAC) were also used. The sampling rate was established at 1000 Hz.

Each subject was assessed initially in a normal condition (without load) and then loaded (backpack condition). Data were collected regarding three valid rehearsals of each test on the force plate (right foot). Subjects carried on the backpack, fixed at the dorso-lombar region, a static load that allow the subject + backpack to reach the "total BMI" of 30. Each subject walked three times in each condition, at a self-selected velocity, along the experimental walkway (600 cm × 92 cm × 15 cm) in which the force plate was engraved.

The results for the three components of the GRF (vertical, anterior-posterior and medial-lateral) were expressed as percentages of the total weight, with and without load. The statistical analysis was conducted with SPSS 16.0 software. Data on independent variables studied were statistically analyzed by measures of central tendency (mean) and dispersion (standard deviation), and compared by paired student t-test (with vs. without load), the significance level adopted was $\alpha=0.05$.

RESULTS: The main results of the study are presented in Table 1. Results include chronometric (temporal) and the dynamometric (GRF) variables. Statistical significant differences ($p < 0.05$) are marked (*). From these we highlight an increase in the stance phase duration and a reduction in the relative magnitude of the first and second peaks of the vertical component in the load condition.

DISCUSSION: During loading, an increase in the two peak values of the vertical GRF component was observed, together with an increase of the stance phase duration. A higher value of the horizontal braking force (anterior-posterior GRF component) was also noted. On the contrary, a reduced maximal value of the latero-medial component was registered. These

results showed that even when the weight of the backpack is included in the calculations of GRF as a percentage of total weight (body + pack), the mean values obtained with and without load traduce a relevant disturbance of the dynamometric profile of the gait pattern. These results conflict with previous results from Tilbury-Davis and Hooper (1999), which evaluated the biomechanical effects of load (20 and 40 kg) in military subjects. Nevertheless, those were trained subjects in this particular task.

Table 1. Mean and standard deviation (Std.) values for the studied chronometric and dynamometric variables obtained for unloaded and loaded situations in both genders

Variables	Normal		Loaded		Confidence Interval		t	Sig.
	Mean	Std.	Mean	Std.	Lower	Upper		
Duration stance phase (s)	0.78*	0.06	0.81*	0.07	-0.055	-0.014	-3.36	0.002
First peak - Vertical Component (N/BW)	1.03*	0.04	0.99*	0.06	0.021	0.057	4.35	0.000
Time (% stance phase)	25.96	3.09	26.68	3.24	-1.674	0.200	-1.59	0.120
Minimum value between Vertical peaks (N/BW)	0.82	0.05	0.82	0.06	-0.016	0.019	0.18	0.859
Time (% stance phase)	46.10	5.95	46.02	4.32	-1.740	1.896	0.09	0.931
Second peak - Vertical Component (N/BW)	1.10*	0.05	1.07*	0.06	0.018	0.052	4.20	0.000
Time (% stance phase)	74.64*	2.46	72.64*	3.40	1.139	2.865	4.69	0.000
Braking Force - anteroposterior component (N/BW)	-0.14*	0.03	-0.15*	0.03	0.003	0.018	2.93	0.006
Time (% stance phase)	18.12	2.66	17.99	1.87	-0.655	0.956	0.38	0.708
Propulsion Force - anteroposterior component (N/BW)	0.19	0.03	0.18	0.03	-0.001	0.014	1.79	0.081
Time (% stance phase)	83.09	1.91	82.99	1.90	-0.473	0.664	0.34	0.736
Peak - mediolateral component (N/BW)	0.10*	0.02	0.09*	0.01	0.002	0.012	2.77	0.009
Time (% stance phase)	46.91	21.91	46.32	18.51	-4.898	6.081	0.22	0.829

* Statistical significance $p < 0.05$

Pierrynowski, Norman and Winter (1981) suggested that gait adjustments occurred when loads of 34kg are carried by subjects whose weight was approximately 72kg (47% body weight). The maximal load used in our research for all the studied subjects was lower than the critical absolute value of 30kg proposed by the referred authors, but sometimes higher in relative terms considering the subject's body weight (57%), which suggest the need of a revision of the boundary proposed by the referred authors. Authors also reported an attenuation of the loading and unloading rates when carrying the higher load of 40kg, suggesting a protection of the biomechanical system. This seems to be in agreement with the reduction of the relative values for the first and second peaks of the vertical GRF component for the load condition obtained in our study, combined with the increased stance duration.

CONCLUSION: The present study showed an adaptation of the subjects to the load condition, with: (i) an increase of the stance phase duration; (ii) a significant reduction of the GRF vertical component peaks, and (iii) an increase of the horizontal braking force.

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Appendix III: Conference Proceeding III

EFEITO DA SOBRECARGA PERMANENTE E OCASIONAL NOS PARÂMETROS CINÉTICOS DA MARCHA

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PALAVRAS CHAVE: Sobrecarga, Obesidade, Mochileiros, Marcha, Força de Reacção do Solo.

RESUMO: A proposta do presente estudo foi comparar o efeito da sobrecarga permanente (pessoas obesas) e ocasional (mochileiros) nos parâmetros cinéticos da marcha. Foram observadas adaptações no padrão da marcha nos indivíduos quando submetidos a sobrecarga.

1 INTRODUÇÃO

A compreensão das adaptações do sistema músculo-esquelético durante a marcha de pessoas submetidas a sobrecarga permite o estabelecimento de estratégias de prevenção e reabilitação mais seguras e eficazes para a manutenção da integridade física. Neste sentido, a proposta do presente estudo foi comparar os efeitos da sobrecarga permanente e ocasional nos parâmetros cinéticos da marcha.

2 METODOLOGIA

A amostra foi constituída por 80 participantes, dos quais 60 apresentavam índice de massa corporal (IMC)<25, idade de 22,8±3,7 anos, altura de 1,69±0,09 m e massa corporal de 65,5±9,8 kg; e 20 participantes apresentavam IMC>30, idade de 39,1±7 anos, altura de 1,69±0,1 m e massa corporal de 104,6±12,0 kg. No presente estudo realizou-se uma análise extensiométrica para a qual foi utilizada uma plataforma de força Bertec (modelo 4060-15) para recolher dados de força de reacção do solo (FRS) a uma taxa de amostragem de 1000 Hz. Cada participante caminhou a uma velocidade auto-seleccionada ao longo de um estrado de marcha, em cujo centro se encontrava a plataforma. Os participantes com IMC>30 realizaram três testes, enquanto que os

participantes com IMC<25 realizaram seis, sendo que nos três últimos testes, utilizaram uma mochila (fixada na região dorso-lombar) que continha uma massa que, quando somada à massa corporal, perfazia um IMC de 30. Tal IMC foi escolhido por ser considerado potencialmente lesivo ao aparelho locomotor [1]. Desta forma, três grupos foram estabelecidos: grupo IMC<25 sem mochila (grupo controlo), IMC<25 com mochila (grupo sobrecarga ocasional) e IMC>30 (grupo sobrecarga permanente). Foi desenvolvido um programa no software Matlab® 7.0 para cálculo das seguintes variáveis: duração da fase de apoio (FA), 1º pico (Fvt1), mínimo intermédio (Fvt2) e 2º pico (Fvt3) da componente vertical da FRS; pico de travagem (Fap_T) e propulsão (Fap_P) da componente ântero-posterior da FRS; pico da componente médio-lateral (Fml) da FRS e tempo onde ocorreram tais eventos, respectivamente: TFvt1, TFvt2, TFvt3, TFap_T, TFap_P e TFml. O impulso da componente vertical (Ivt) e da componente antero-posterior (travagem e propulsão - Iap_T e Iap_P) também foram determinados. Os dados de força foram normalizados pela massa corporal enquanto que os dados referentes às variáveis temporais foram normalizados pela duração da FA. A normalidade dos dados foi verificada por

meio do teste de Shapiro-Wilk. Para a comparação entre os grupos foi utilizado o teste t-Student e o valor de significância foi $\alpha = 0.05$. Os resultados serão apresentados como média e desvio padrão da média. Os procedimentos estatísticos foram realizados em SPSS® 17.

3 RESULTADOS

Na tabela 1 encontram-se os resultados das variáveis em estudo. Pela análise dos parâmetros temporais, observa-se que o grupo com sobrecarga permanente promove

grupos com sobrecarga quando comparados ao controlo.

4 CONCLUSÃO

Observam-se diferenças no padrão cinético da marcha dos indivíduos submetidos a sobrecarga (permanente e ocasional) quando comparados ao grupo controlo. Adaptações específicas consoante ao perfil da sobrecarga imposta também foram observadas. Assim, uma possível adaptação do aparelho locomotor parece ocorrer no sentido de minimizar as forças excessivas e potencialmente lesivas à integridade física.

Tabela 1 – Resultados de média e desvio padrão de todas as variáveis em estudo

Variáveis	Controlo		Sobrecarga Ocasional		Sobrecarga Permanente	
	Média	dp	Média	dp	Média	dp
Duração da FA (s)	0,79 ^{*&}	0,06	0,81 ^{&}	0,07	0,83 [*]	0,08
Fvt1 (N/BW)	1,02 ^{*&}	0,03	0,99 ^{&}	0,04	1,00 [*]	0,02
TFvt1 (%FA)	25,95 [†]	2,74	26,55	3,06	28,00 [*]	2,77
Fvt2 (N/BW)	0,83	0,05	0,83	0,06	0,83	0,05
TFvt2 (%FA)	46,07	5,42	45,51	3,76	46,60	4,66
Fvt3 (N/BW)	1,10 ^{*&}	0,05	1,07 ^{&#}	0,05	1,04 [#]	0,05
TFvt3 (%FA)	74,40 ^{&}	2,35	73,02 ^{&#}	2,55	75,08 [#]	1,29
Fap _T (N/BW)	-0,14 ^{&}	0,03	-0,15 ^{&#}	0,03	-0,13 [#]	0,03
TFap _T (%FA)	18,00 [†]	2,81	18,01 [#]	1,77	19,74 [#]	2,19
Fap _P (N/BW)	0,18 [†]	0,02	0,18 [#]	0,03	0,16 [#]	0,02
TFap _P (%FA)	82,88 [†]	1,96	82,81 [#]	1,90	85,20 [#]	1,45
Fml (N/BW)	0,10 ^{*&}	0,01	0,09 ^{&}	0,01	0,11 [*]	0,02
TFml (%FA)	45,31 [†]	21,35	46,27	19,18	75,29 [*]	1,91
Ivt (N/BW),s	0,590	0,04	0,600	0,04	0,598	0,03
Iap _T (N/BW),s	-0,028 ^{&}	0,00	-0,030 ^{&#}	0,00	-0,027 [#]	0,00
Iap _P (N/BW),s	0,028 ^{&}	0,00	0,030 ^{&#}	0,00	0,028 [#]	0,00

dp – desvio padrão; diferenças significativas com $p < 0,05$ entre: ^{*} grupo controlo e sobrecarga permanente; [&] grupo controlo e sobrecarga ocasional; e [#] sobrecarga ocasional e permanente.

um atraso onde ocorrem os eventos que caracterizam a marcha; enquanto que, ao analisar as variáveis relativas ao impulso, parece que as pessoas com sobrecarga permanente apresentam uma melhor adaptação ao excesso de carga, quando comparado com o grupo com sobrecarga adicional (Iap_T e Iap_P). Em relação aos picos, verifica-se uma suavização nos

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Appendix IV: Conference Proceeding IV

THE INFLUENCE OF DIFFERENT SPEEDS ON BACKPACKER'S GAIT KINETICS

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This study analyzed the influence of different speeds on ground reaction force's (GRF), impulses and mean vertical force during gait of people submitted to occasional overload (backpack). A force plate was used to record the GRF data of 60 young adult subjects walking in two different cadences: 69 steps/min (slow gait) and 120 steps/min (fast gait). During the slow gait, the impact and propulsive impulses of vertical GRF, propulsive impulse of anterior-posterior GRF, impulse of medial-lateral GRF and duration of stance phase were larger than during the fast gait; the mean vertical force was the only variable that showed larger values during fast gait. Therefore, slow gait may present a larger possibility of blister development and gait unbalance, while the fast gait, even presenting a small impulse, seems to be more harmful to the musculoskeletal system.

KEY WORDS: backpack, overload, ground reaction force, impulse.

INTRODUCTION: The backpack has been widely used by students, hikers and military as a device to transport load. As a consequence a number of studies have been conducted to identify the biomechanical and physiological impact of this occasional overload on the musculoskeletal system (Birrell et al., 2007; Browing & Kram, 2007; Knapik et al., 1996). Some of the analyzed variables were the impulse or force-time integral of the three components of the ground reaction force (GRF), and mean values of vertical force component (Jordan et al., 2007; Lewek, 2010; Vito et al., 2009). The vertical forces (impulse and mean value) provides information about impact forces, anterior-posterior impulse provides information about impact and blister development and the medial-lateral impulse may be linked to dynamic balance and stability (Birrell, et al., 2007).

Changes in walking speed seem to influence the impulse magnitudes. Previous studies found that as the walking speed increases the vertical GRF impulse decreases (Jordan, et al., 2007; Kimberlee et al., 2007; Vito, et al., 2009), while anterior-posterior GRF impulse increases (Chung & Wang, 2010; Vito, et al., 2009).

The previous studies, however, have not evaluated the effect of gait speed on GRF with additional loading from carrying a backpack. Therefore the purpose of this study was to analyze the influence of different speeds on GRF' impulses and mean vertical force during gait of people under occasional overload (backpack).

METHODS: The study was approved by the local ethical committee and all participants freely signed an informed consent term, based on Helsinki's declaration, which explained the purpose and the procedures of the study.

Participants: The sample was selected by convenience from university students of sport sciences, and was composed by 60 subjects (30 male and 30 female) with a mean age of 23.0 (± 3.7) years, mean height of 168.0 (± 9.0) cm and mean body mass of 67.8 (± 11.2) kg. All participants were physically active and did not present a body mass index (BMI) above 25, didn't have any traumatic-orthopedic dysfunction nor have difficulties on independent gait.

Instruments: A Bertec force plate (model 4060-15, Bertec Corporation, Columbus, USA), operating at 1000 Hz, was used to measure GRF and a Maelzel metronome (Wittner, Germany) to control the step frequency. Three digital non-coplanar video cameras were used for visual inspection, if necessary.

Experimental Protocol: The participants underwent three phases of testing: preparation, familiarization and testing. In the first phase the procedures to be implemented were explained to the participants and anthropometric data (height and weight) were recorded. A neutral shoe (ballet sneaker) was provided for all participants aiming to minimize the effects of different soles. For each participant the weight to raise their BMI to 30 was calculated; then a backpack was filled with sand and fixed in the central area of each subject's back; the weight placed inside the backpack ranged from 14.1 to 30.1 kg (mean weight 20.3 ± 4.4 kg). This overload was chosen because it is considered to leave the locomotor system more susceptible to injuries (Ko et al., 2010), and the additional upper body mass mimicked obesity, but with the overload in posterior rather than anterior position. In the familiarization process, the participants walked freely over a 6m walkway which had the force plate embedded in the middle; then they trained to walk with two different step frequencies: 69 steps/min (slow gait) and 120 steps/min (fast gait). Participants were asked to try to walk as naturally as possible during these controlled conditions. In this phase the researchers identified the place where the participant should begin the gait to step with his/her right foot in the center of the plate without changing the natural gait pattern. During the test the participants walked three times with a self-selected speed, three times with slow controlled gait, and three times with fast controlled gait. The present study will present data referring to slow and fast gait.

Data analysis: For the acquisition of the force plate data, Acknowledge software (BIOPAC System, California, USA) was used. These data were exported to Matlab® 7.0 (MathWorks, Massachusetts, USA) where a routine was developed to process and calculate the following variables: impact impulse of vertical GRF (Vt_i), propulsive impulse of vertical GRF (Vt_p), braking impulse of anterior-posterior GRF (AP_B), propulsive impulse of anterior-posterior GRF (AP_P), impulse of medial-lateral GRF (ML), mean vertical force (VtF) and duration of stance phase. The events used to calculate impulse variables are illustrated in Figure 1.

Statistical analysis: The mean of the three repetitions performed by each subject was computed and all the statistical procedures were performed with these mean values. The normality of the data was verified using Kolmogorov-Smirnov test and the homogeneity of the variances using Levene's test. Then seven paired t-tests were used to compare the variables between the groups. The results will be presented as mean and standard deviation and the significance level adopted was $\alpha=0.05$. All the statistical procedures were conducted using the software SPSS (v.17; SPSS Inc, Chicago, IL).

RESULTS: The data showed a normal distribution and variances homogeneity. Figure 1 shows that the impulse variables Vt_i , Vt_p , AP_P and ML for the slow gait had higher values compared to the fast gait with statistical significant differences. Despite the AP_B mean values obtained for the slow gait tended to be higher than during fast gait, differences were not statistically significant. Table 1 shows the confidence interval and level of significance of the difference between fast and slow gait obtained by statistical test for all variables. As expected, the duration of stance phase was higher at slow gait (1.091 ± 0.009 s) when compared with fast gait (0.677 ± 0.004 s). Considering the VtF, this variable presents larger magnitude during fast gait (498.9 ± 76.9 N) with statistical significant differences comparatively to slow gait (475.8 ± 71.7 N).

DISCUSSION: The purpose of this study was to compare the two different gait speeds (slow vs fast) in a severe occasional overloading situation (wearing a backpack), comparable to what a subject of BMI=30 may experience. The results suggest that, in this particular situation, the musculoskeletal system need to manage larger impulses during slow than during fast gait, while the VtF is smaller (see Fig. 1 and results). In the following studies (in non-overload conditions), as in the present study, the vertical impulse of GRF decreases with increasing speed during walking (Jordan, et al., 2007; Vito, et al., 2009), but also during running (Jordan et al., 2007). So, it seems that the influence of speed on the behavior of the vertical impulse of GRF is similar during normal and overloaded gait. However, when analyzing only the propulsive impulse (anterior-posterior GRF), Lewek et al. (2010) found,

contrasting with our results, that it increases as gait speed increases in non overloaded conditions. These findings, when compared with the results of the present study indicate a possible difference in the characteristics of a backpacker's gait when compared with normal gait (without overload).

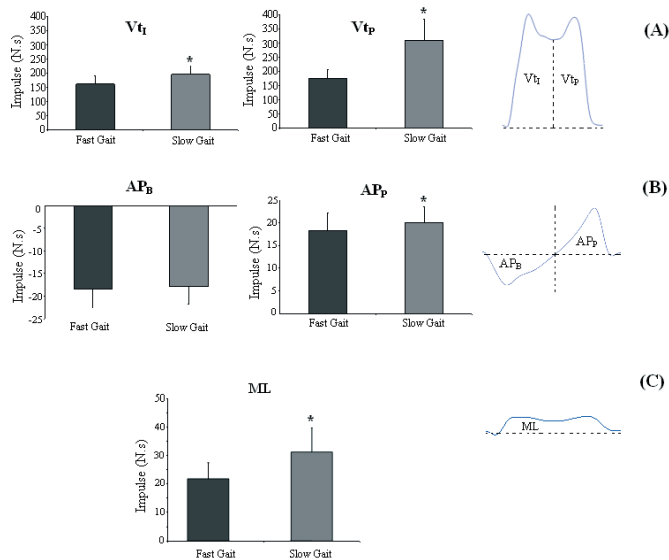


Figure 1: Comparison of impulse variables between fast and slow gait. (A) impulse of vertical GRF; (B) impulse of anterior-posterior GRF and (C) impulse of medial-lateral GRF. V_{t_i} - impact impulse of vertical GRF; V_{t_p} - propulsive impulse of vertical GRF; AP_b - braking impulse of anterior-posterior GRF; AP_p - propulsive impulse of anterior-posterior GRF; ML - impulse of medial-lateral GRF; * - statistical significant difference $p \leq 0.05$.

Table 1
Confidence interval and level of significance of the difference between fast and slow gait

Variables	95% Confidence Interval of the Difference		Level of significance (p)
	Lower	Upper	
Duration of stance phase	- 0.434	- 0.394	<0.001
Impact impulse of vertical GRF	- 43.286	- 24.899	<0.001
Propulsive impulse of vertical GRF	-149.146	- 118.050	<0.001
Braking impulse of anteroposterior GRF	- 1.638	0.559	0.329
Propulsive impulse of anteroposterior GRF	- 2.437	- 0.762	<0.001
Impulse of mediolateral GRF	- 12.564	- 9.511	<0.001
Mean vertical force	18.937	27.384	<0.001

Impulses depend on the intensity and duration of the application of force. Previous studies on unloaded subjects indicate that when the speed increases peak vertical (Browning & Kram, 2007; Caravaggi et al., 2010; Grabowski, 2010) and anterior-posterior GRF values increase. On the contrary, the duration of stance phase is reduced at higher gait speeds (Caravaggi, et al., 2010; Grabowski, 2010). Consequently, the amount of variation of these two variables will be responsible for the variation of the impulse. The analysis of the present results suggests that the duration of force application affects more the impulse outcome, being responsible for a significant increase on musculoskeletal load (total load, not peaks) during

slow gait. However, analyzing the VtF, it is possible to observe that, during fast gait, there is less time available for musculoskeletal adaptation which makes this situation potentially more aggressive than slow gait considering the viscoelastic properties of the human body tissues. Birrel et al. (2007) found an increase of the GRF' medial-lateral impulse during overloaded gait, and stated that this characteristic may be linked to a decrease in stability of gait dynamic balance. In this sense our results seem to point out that the overloaded slow gait situation may be characterized by a decreased stability when compared with fast gait. One possible limitation of the present study is the utilization of an acoustical pacer to control different gait conditions (slow and fast). However, the subjective analyzes of video images and the differences observed on the duration of stance phase seem to indicate that this methodological option didn't significantly constrain performance.

CONCLUSION: The results of the present study indicate that the backpacker, walking with a slow speed, is submitted to a higher total mechanical load (impulse) and a lesser mean vertical force when compared to fast gait. Therefore, the backpacker has more time (larger duration of stance phase) to force dissipate during slow gait, what seems to be advantageous for the musculoskeletal system, considering their viscoelastic properties. However, during slow gait the backpacker presented larger magnitudes to propulsive anterior-posterior and medial-lateral impulses when compared to fast gait; and since these variables can provide some information about blister development and balance disturbances, respectively, possibly during slow gait these negative aspects are more pronounced. Therefore, each condition (slow and fast gait) seems to have positive and negative aspects considering these kinetic variables. These gait characteristics can be useful in order to achieve adequate preparation and to promote safety during physical activities and sports performance involving load transportation.

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Appendix V: Conference Proceeding V

WALKINSENSE VALIDATION: PRELIMINARY TESTS OF MOBILITY PARAMETERS

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The purpose of this study was to perform a preliminary validation of a new electronic instrument for human movement and performance assessment in sports. Measurements of distance, walking speed, step length and frequency were acquired, for a small sample of 15 subjects in a track of 10 m length, and compared to reference data. Results show good repeatability and data agreement across several trials at three different self-selected walking speeds.

KEY WORDS: gait parameters, repeatability, measurement agreement.

INTRODUCTION: In the field of physical activity and sports monitoring, we are currently witnessing a shift in the paradigm for the assessment of human movement and performance. Empowered by the fast paced development of portable and wearable technology, research in this field can now take place in real life scenarios, under everyday and long term conditions, as opposed to short term, laboratory or otherwise controlled experiments.

This trend towards the use of wearable monitoring and recording equipment, seamlessly attached to the human body, allows effortless data capturing without disturbance or discomfort to the subject under observation (Pantelopoulos and Bourbakis, 2010). However, for these equipments to be widely accepted as research or clinical tools, they have to be validated against well known and established methods and instruments (Bland and Altman, 1986).

WalkinSense® is one such type of equipment designed for activity monitoring, combined with foot pressure evaluation and analysis of gait parameters. While many systems are available for each one of those assessments individually, the former combines all three capabilities in a single fully autonomous portable lightweight unit with accompanying analysis software. Amidst several other parameters, it provides effective measures of traveled distance, average speed, step length and frequency, together with foot pressure trends during gait cycles for extended periods of time, which can span several days of activity. Finding usefulness in the field of lower limb prophylaxis and rehabilitation, its suitability for performance assessment in sports such as athletics, football or golf, among others, is the main concern of this work.

The main purpose of this study was to perform a preliminary validation of the WalkinSense® equipment, under controlled conditions and evaluate its repeatability on several gait parameters, as well as on the accuracy of the distance measure with respect to ground truth data. Related studies evaluating similar gait parameters can be found in the literature. Al-Obaidi et al. (2003) compare the basic gait parameters of normal subjects from Kuwait with a similar group of subjects from Sweden, following a previous study reported by Öberg et al. (1993).

METHODS: This study was approved by the local ethical committee.

Participants: The sample of convenience included fifteen participants, eleven male and 4 female, all of them students enrolled at the University that hosted the study. All participants were healthy and physically active and did not have any gait impairments. The participants were on average 20.1 (± 5.5) years old, with height and body mass of 1.70 (± 0.083) m and 67.8 (± 11.2) kg, respectively.

Instruments: WalkinSense® (Tomorrow Options SA, Porto, Portugal) is a CE Mark class I electronic medical device designed to dynamically monitor human lower limbs activity. It gathers and processes quantitative and qualitative information and sends it to a computer, laptop or palmtop computer via wireless Bluetooth® connection or wired USB cable to be analyzed with the WalkinSense® software (Tomorrow Options SA, Porto, Portugal). The device contains MEMS triaxial accelerometer and gyroscope and an array of eight force sensing resistors for foot pressure measurements. The device can operate in two modes, the offline recording mode that allows data capturing for several days, and the real-time mode that can acquire data at 100 Hz and send it directly to a computer with the WalkinSense® software using the Bluetooth connection. For this validation study, we chose the real-time mode to have the detailed data, in order to perform a more complete statistical analysis.

Experimental protocol: Firstly, the procedures were explained to each participant, then anthropometrics data were obtained, followed by a familiarization period. The tests were carried out in a gym where the beginning and end of a track was marked on the floor over a ten meter distance. Each participant performed six tests: in the first two, subjects were asked to walk at a normal self-selected speed, in the third and fourth tests at a slow self-selected walking speed and the last two tests were performed with a fast self-selected speed.

The participants began the test one step before the start mark, since the device excludes the first step, and stopped over the end mark. During the test, the participants were asked to look forward and to walk as naturally as possible.

Data analysis: For data acquisition and recording, we used the WalkinSense® software. A data set with a total of 90 trials was recorded (six tests of fifteen participants). Four temporal gait parameters were analyzed: gait distance, average speed, step frequency and step length. The last three parameters were calculated considering only the three central steps at mid distance in the track.

Statistical analysis: To verify intra-individual repeatability for all variables the intra-class correlation coefficient (ICC) was calculated for all tests ($n = 90$). The results are presented as mean, standard deviation and confidence interval; these statistical procedures were conducted using SPSS (v.17; SPSS Inc, Chicago, IL, USA).

Aiming to compare the distance data provided by the WalkinSense® equipment with the track length (10 m), we used the method proposed by Bland and Altman (1986), where the difference between every test and the track distance was analyzed.

For the calculation of sample size to a future definitive validation study we used the SYSTAT (v.12; Cranes Software International, Chicago, IL, USA) software.

RESULTS: Table 1 shows the mean, standard deviation, confidence interval and ICC for all studied mobility variables and the values found by Al-Obaidi et al. (2003) for average speed, step length and frequency. It is noteworthy the high ICC values obtained for all the studied variables. Regarding the distance, we notice that a value much closer to the target value is recorded for faster walking. Figure 1 presents the comparison between gait distance data measured by WalkinSense® and the track length. Using a limit of two standard deviations, 96.3% of the measured gait distance data are in the "limits of agreement", indicating a good capability of the device to perform gait distance measurements.

Additionally, we used the standard deviation of the gait distance data of this preliminary study to calculate the sample size required for a future extended validation study. For that future purpose, the number of 50 participants was obtained in order to provide a statistical power of 95% with an alpha error level of 5%.

DISCUSSION: According to the values of ICC for all variables, it appears the data provided by the device are consistent and display good repeatability. With regard to the distance data, accuracy seems to improve for higher speeds, showing a relative error of -6.8% at the lower speed against only -1.6% for the higher speed in the 10 m track. This underestimation of the mean distance may be due to the subjects' inaccuracy to step on the beginning and ending marks of the 10 m track. These relative errors would have much less significance for longer tracks.

Table 1
Mean, standard deviation (SD), lower and upper confidence interval, intra-class coefficient correlation (ICC) of the variables and larger and lower values found by Al-Obaide et al. (2003)

Variables		Mean (SD)	Confidence Interval		ICC	Al-Obaidi et al. (2003)	
			Lower	Upper		Larger mean	Lower mean
Average Speed (m/s)	<i>Slow</i>	0.82 (0.20)	0.74	0.90	0.98	0.80 (0.16)	0.84 (0.20)
	<i>Normal</i>	1.20 (0.18)	1.13	1.26		1.08 (0.15)	1.24 (0.17)
	<i>Fast</i>	1.67 (0.16)	1.61	1.73		1.56 (0.14)	1.81 (0.22)
Step frequency (Steps/s)	<i>Slow</i>	1.40 (0.16)	1.34	1.46	0.99	1.42 (0.19)	1.59 (0.20)
	<i>Normal</i>	1.75 (0.15)	1.70	1.80		1.73 (0.15)	2.08 (0.15)
	<i>Fast</i>	2.04 (0.17)	1.97	2.10		2.18 (0.27)	2.56 (0.25)
Step length (m)	<i>Slow</i>	0.59 (0.19)	0.56	0.62	0.96	0.52 (0.07)	0.57 (0.06)
	<i>Normal</i>	0.68 (0.07)	0.66	0.70		0.59 (0.06)	0.70 (0.07)
	<i>Fast</i>	0.84 (0.07)	0.81	0.86		0.67 (0.06)	0.83 (0.07)
Gait Distance (m)	<i>Slow</i>	9.32 (1.47)	8.71	9.93	0.88	—	—
	<i>Normal</i>	9.53 (0.60)	9.31	9.75		—	—
	<i>Fast</i>	9.84 (0.58)	9.61	10.01		—	—

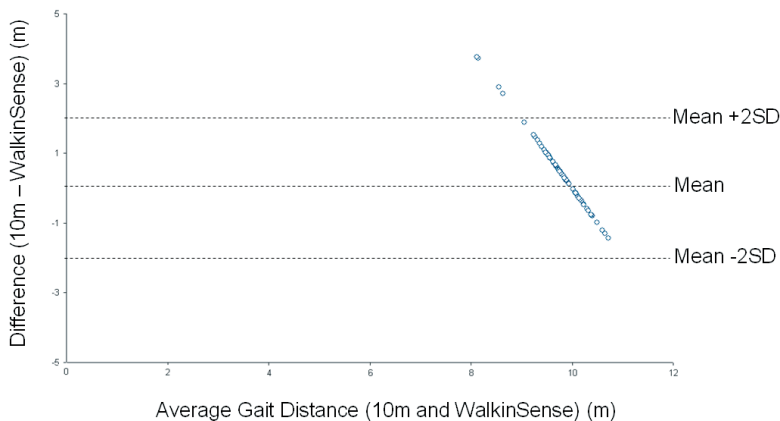


Figure 1: Comparison between gait distances measured by WalkinSense® and the track length using method proposed by Bland and Altman (1986).

For the walking speeds, step lengths and frequencies presented in Table 1 no ground truth data was available for this preliminary study. However, when comparing with reference data obtained by Al-Obaide et al. (2003) for the same gait parameters of men and women aged between 20 and 29 years of Kuwait and Scandinavia, one can observe that our results fall well within the same limits, considering data for the average walking speed, step length and frequency (see Table 1). The results of the present study could only be more similar if there was a better match between the subjects age and separated by gender.

The limitations and issues identified on this preliminary study will be properly addressed and resolved on a follow up and more extended validation work.

CONCLUSION: This preliminary study, for the validation of WalkinSense® as an instrument for human movement and performance assessment in sports, allowed us to obtain consistent data with good repeatability across all the mobility parameters analyzed. The distance measurements, in particular, showed good accuracy and agreement with ground truth data for the ten meter track. Notwithstanding, several limitations were identified, such as the sample size and the reduced number of trials, as well as the lack of ground truth for the other parameters (average walking speed, step length and frequency). This study will be followed for a more extended validation work with a larger sample of subjects. Also, it will include a comparison of WalkinSense® measurements with other gold standard or well established methods following the guidelines that were set with this preliminary validation.

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