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DEVELOPMENT OF A METHODOLOGY FOR BIOMECHANICAL GAIT EVALUATION IN OLDER ADULTS

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CANDIDATA: ANDRÉIA GOMES AIRES

Esta Dissertação de Mestrado foi julgada para obtenção do título de MESTRE EM ENGENHARIA ELÉTRICA e aprovada em sua forma final pelo Programa de Pós-Graduação em Engenharia Elétrica da Pontifícia Universidade Católica do Rio Grande do Sul.

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“Nothing in life is to be feared, it is only to be understood. Now is the time to understand more, so that we may fear less.”

Marie Curie

RESUMO

As alterações no quadro demográfico do Brasil e do mundo servem como motivação para direcionar a atenção nos indivíduos responsáveis pela alteração da distribuição populacional, os idosos. Avaliar a capacidade funcional do idoso é fundamental para determinar seu nível de independência e com isso verificar qual o potencial que estes indivíduos possuem para executar suas atividades do dia a dia. A determinação deste nível de funcionalidade pode ser realizada através do estudo das Forças de Reação do Solo (FRS) durante o ciclo da marcha, que muitas vezes é realizada de forma qualitativa, tornando os resultados altamente subjetivos e sem padronização. Neste contexto, o objetivo deste trabalho foi desenvolver uma metodologia capaz de identificar e quantificar os principais parâmetros que caracterizam as curvas vertical e ântero-posterior das FRS. Para o desenvolvimento desta metodologia, um banco de dados contendo dados força e tempo de testes de caminhada realizados por 33 participantes foi utilizado. As informações referentes aos sinais biomédicos utilizados para a realização da avaliação biomecânica da marcha de idosos foram capturados por 8 plataformas de força, compostas por 12 sensores de força cada. A análise e o tratamento dos dados brutos de força foram realizados com o auxílio de um algoritmo desenvolvido para o processamento do sinal de FRS. Como resultado foram identificados e calculados os valores do primeiro pico de força vertical, da menor força entre os picos de força vertical, do segundo pico de força vertical, pico de frenagem da força ântero-posterior, pico de propulsão da força ântero-posterior, bem como os tempos para estes eventos da marcha. Os impulsos das forças verticais e ântero-posteriores também foram calculados através da área abaixo da curva de força pelo tempo. Adicionalmente, os valores encontrados no processamento dos dados provenientes do banco de dados de idosos, utilizando a metodologia proposta, foram comparados com outros estudos que também avaliaram as forças de reação do solo na população idosa. O desenvolvimento desta metodologia mostrou-se eficiente na avaliação de parâmetros cinéticos e temporais relacionados a marcha de idosos. Além disso, servirá como importante ferramenta para análise biomecânica da marcha de diferentes populações proporcionando novas investigações e avaliações.

Palavras-Chaves: Análise de marcha. Biomecânica. Forças de reação do solo. Idosos.

ABSTRACT

Changes in the demographic profile of Brazil and the world serve as a motivation to direct attention to the population that has been changing the shape of the age pyramid, the elderly. Assessing the functional capacity of older adults is crucial to determine their level of independence and verify the potential that these individuals must perform their activities of daily living. Since functional capacity means the ability of an individual to maintain the physical and mental functions required to preserve their independence, assessing how older adults walk is essential to ensure that this activity of daily living is carried out safely. The determination of this level of functionality can be performed through the study of ground reaction forces (GRF) during the gait cycle, which is often performed in a qualitative way, making the results highly subjective and without standardization. In this context, the objective of this work was to develop a methodology for identifying and quantifying the main parameters that characterize the vertical and anterior-posterior curves of the GRF. This methodology was created using a database containing raw data from walking tests performed by 33 participants. The force signals used to perform the biomechanical assessment of gait in the elderly were captured by 8 force platforms, composed of 12 strain gauges each. An algorithm was developed for processing the GRF signal, transforming the raw force data into discrete gait parameters. As a result, the methodology delivers the values of the first vertical peak force, the lowest force between the two maximums vertical peak force, the second vertical peak force, braking peak, propulsion peak, as well as the times for these gait events. The impulses of the vertical and anterior-posterior forces were also calculated through the area under the force-time curve. Additionally, the values found in the processing of data from the elderly database, using the proposed methodology, were compared with other studies that also evaluated ground reaction forces in the elderly population. The development of this methodology proved to be efficient in the evaluation of kinetic and temporal parameters related to the gait of older adults. In addition, it will serve as an important tool for biomechanical analysis of the gait of different populations, providing new investigations and evaluations.

Keywords: Gait analysis. Biomechanics. Ground reaction forces. Older adults.

LIST OF FIGURES

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ABBREVIATION AND ACRONYMS

GRF	Ground reaction force
FFT	Fast Fourier Transform
F	Force
t	Time
I	Impulse
Fx	Mediolateral component of ground reaction force
Fy	Anterior-posterior component of ground reaction force
Fz	Vertical component of ground reaction force
FP	Force platform
ITD	Instrumented treadmill
fs	Sampling frequency
CV	Coefficient of variation

CONTENTS

1 INTRODUCTION

The study of human movement, specially locomotion, is as ancient as walking became the most performed activity by humans, not only as a physical activity, but also as means of locomotion. Gait analysis goes back centuries motivated by the necessity of understanding how humans walk and the importance of the normal and pathological gait study (ANDRIACCHI; ALEXANDER, 2000; BAKER, 2007; MÜNDERMANN; CORAZZA; ANDRIACCHI, 2006).

Some diseases and conditions caused by age may affect the normal walking (CALDAS et al., 2017). Thus, the analysis of this activity is relevant because it provides information on the functional fitness of individuals and can be used as an evaluation tool for the effectiveness of health treatment or physical training (MURO-DE-LA-HERRAN; GARCÍA-ZAPIRAIN; MÉNDEZ-ZORRILLA, 2014), which are indispensable requirements related to the treatment or monitoring of the health and welfare of the elderly.

The worth of studying the aspects related to this population and how their lives may be affected by the improvements achieved with advances in science are well established. Based on this fact a document pointed evidence-based actions to cope with the social and economic changes in ageing (JUDGE, 2020).

Some important facts about aging should be noted and included in the plans of governments, professionals, society and the private sector to respect the lives of older people and their families. In developing countries, the number of persons aged over 60 years is expected to grow faster than in more developed regions. Projections indicate that by 2050 there will be more people aged 60 years or over than persons aged 10-24 years (JUDGE, 2020).

Considering the growing number of older people, and increasing life expectancy, investigating how ageing affects the walking pattern becomes an important role in the study of human movement. Gait assessment is performed for various purposes, in clinical situations, gait observation helps in diagnoses and monitoring the effects of interventions. However, this observational situation is subject to an evaluator error.

Gait Analysis is an evaluation method that can be done through the application of biomechanical techniques for movement analysis. Biomechanics is a study of the mechanical aspects related to living organisms, like the internal forces produced by muscles and the external forces acting on the human body (MURO-DE-LA-HERRAN; GARCÍA-ZAPIRAIN; MÉNDEZ-ZORRILLA, 2014; WINTER, 2009).

The biomechanical assessment of human movement describes and investigates phenomena and modifications that occur in the human body and consequently affect the mechanisms of movement (HALL, 2014). Advances in the development of new equipment, as well as in the development of new methodologies and analysis techniques allow the evaluation of different gait parameters, generating more efficient and reliable results and diagnoses about the ambulation of the evaluated ones. The most advanced techniques in gait analysis involve methods based on kinematic, kinetic and electromyographic analysis, as well as the use of accelerometers, gyroscopes, among others (VAVERKA et al., 2015).

Kinematic and electromyographic analysis require very elaborate and time demanding processes, such as fixation of reflective markers and electrodes on the patient's body. In addition, the data acquisition system requires specific training and a wide knowledge of anatomy for proper placement of sensors and markers. On the other hand, the evaluation of ground reaction force (GRF) curves acquired with force platforms has a considerable advantage when compared to these two other techniques due to the possibility of immediate visual inspection of a test performed (VAN KOOTEN et al., 2018).

The evaluation of gait parameters during walking can be used to control ageing and pathological processes by detecting instabilities and deviations of the normal gait profile (VAN KOOTEN et al., 2018). Assessing the functional capacity of older adults and understand how mobility is altered during aging is extremely important to guarantee the quality of life of this population. The evaluation of the daily activities performed by older adults is necessary to ensure their independence (OH-PARK; WANG; VERGHESE, 2011).

One of the greatest roles of physicians, whether in diseases diagnosis or treatment evaluation is to identify the outcomes of the interventions. This procedure is often performed subjectively, or by different evaluators throughout the treatment, which may impact the results of the evaluation. The question of how the trials should be performed in the elderly and which parameters should be considered, highlights the need for the development of standards and reference guides. Thus, studies on gait patterns greatly improve data quality and the ability to quantify and interpret results to help healthcare professionals analyse the phenomenon of interest.

Kinetic gait analysis requires the use of force platforms associated with some specific software for collecting and processing biological signals. These tools can be purchased from the hardware supplier or developed independently. The commercial software available in our laboratory does not present all the variables of interest in our studies. In addition, it is not possible to change the evaluated parameters.

Therefore, creating solutions that reduce this subjectivity through the development of algorithms and mathematical models that can understand the human body and allow restoring, simulating or analysing a variety of movements turns possible to perform evaluations accurately and with greater repeatability.

2 PURPOSES

Based on the above, the general purpose of this study is to develop a methodology for biomechanical gait evaluation in older adults, assessing discrete parameters of the Ground Reaction Forces (GRFs) during walking.

To achieve the purpose of this dissertation, the following specific goals were addressed:

- To provide a comprehensive review of biomechanical gait analysis to identify the main GRF parameters;
- To develop a protocol for collecting GRF data during elderly walking;
- To create an algorithm to analyse and process the GRF signal, based on the parameters identified;
- To evaluate different aspects of gait variability;
- To compare selected gait GRF parameters studied here with the findings by other studies.

3 LITERATURE REVIEW AND STATE-OF-THE-ART

3.1 Ageing

The decrease in fertility rates, ally with an increase in life expectancy, causes the ageing of the world's populations. The socioeconomic level of each country determines the age at which an individual begins to be considered elderly. The World Health Organization (WHO) considers that in developing countries, such as Brazil, people over 60 are considered elderly, while in developed countries at 65 years of age, the individual is considered elderly.

The ageing process consists of a complex interaction among biological and cognitive functions, where nervous, muscular, and skeletal systems change irreversibly and progressively. The Pan American Health Organization (PAHO) considers ageing as “a sequential, individual, accumulative, irreversible, universal, non-pathological process, of deterioration of a mature organism, proper to all members of a species, so that the time makes you less able to cope with environmental stress and, therefore, increase your chance of death ”.

The monitoring of the ageing process is carried out in an interdisciplinary manner by health professionals. The clinical and biomechanical evaluation of gait in the elderly should be implemented to control alterations in daily life activities and the appearance of pathological processes in this population (HAUSDORFF, 2005).

3.2 Gait

Gait was defined as “the manner or style of walking”, while walking is defined as “method of locomotion involving the use of the two legs alternately to provide both, support and propulsion, with at least one foot being in contact with the ground at all times" (WHITTLE, 2007).

In order to identify which changes affect the locomotion of the elderly, it is necessary to understand what are the characteristics of a healthy gait, even if there are no determinant patterns of a “normal” gait. Therefore, an alternative is to compare the gait cycle of healthy young individuals, with elderly people of different age groups (WATELAIN et al., 2000).

The gait cycle can be defined as the interval of time and space between two gait events, normally the beginning of the cycle is considered with the first contact of the foot with the ground until the moment the same foot touches the ground again, but other events can be considered (WHITTLE, 2007). The gait cycle is divided into two main phases, the stance phase and the swing phase, related to the side of the evaluated limb, right or left. The stance phase or support phase, occurs when the foot is in contact with the ground, and the swing phase, when there is no foot contact with the ground (MENZ; LORD; FITZPATRICK, 2003).

These two phases are subdivided into others, according to the positioning of the lower limbs and their functionality, normally divided into another 8 events, as illustrated in Figure 1. In each gait cycle considered normal and symmetrical, the stance phase lasts approximately 60% of the cycle and the swing phase approximately 40% (PERRY, 1992).

Walking is characterized by presenting a phase where both feet are in contact with the ground, the double support phase, and in a complete gait cycle there are two periods of double support, lasting 10% of the cycle. When only one member is in contact with the ground, we call it single support or single limb stance and there are two periods of the single support (LIPPERT, 2011; VAUGHAN; DAVIS; O'CONNOR, 1999; WHITTLE, 2007).

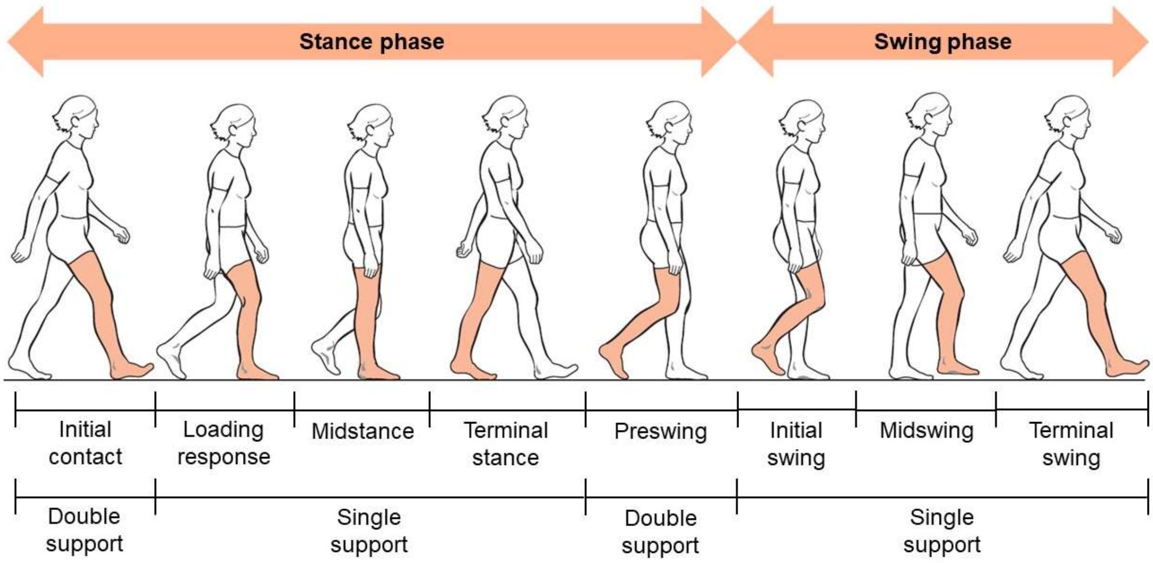
The initial contact also called a 'heel strike', is the first event of the stance phase and begins when the foot touches the ground. As soon as the ground reaction forces start to act in the movement, between 0% and 10% of the cycle, the loading response event starts, during the double support. At this moment the foot approaches the ground by plantarflexion of the ankle (LIPPERT, 2011; VAUGHAN; DAVIS; O'CONNOR, 1999; WHITTLE, 2007).

As the movement advances (10% to 30% of the cycle) the midstance phase occurs, the forefoot lowered to the ground is called 'foot flat'. Simultaneously, the contralateral limb starts the swing phase, which means the double support phase ends and the first single support phase begins. When the contralateral limb passes the limb in midstance phase, the body's centre of gravity is at the highest position (LIPPERT, 2011; VAUGHAN; DAVIS; O'CONNOR, 1999; WHITTLE, 2007).

The terminal stance event (30% to 50% of the cycle) when the heel raises off the floor. The ankle starts to plantar flexion and progresses to push-off, starting the last division of the stance phase, the pre-swing phase (50% to 60% of the cycle). At this moment, the ankle plantar flexors are actively pushing the body forward, because of this, sometimes this event is called the propulsion phase (LIPPERT, 2011; VAUGHAN; DAVIS; O'CONNOR, 1999; WHITTLE, 2007).

The swing phase begins with the release of the foot from the ground, in this phase there is the activation of the hip flexor muscles so that the acceleration of the front limb occurs. The medium swing phase coincides with the medium support phase of the opposite leg, it is the moment that the foot passes under the body, this phase is followed by the terminal swing that is characterized by the deceleration of the limb being prepared for the next initial contact , and thus the gait cycle is completed (LIPPERT, 2011; VAUGHAN; DAVIS; O'CONNOR, 1999; WHITTLE, 2007).

Figure 1 - Divisions of Gait Cycle



Source: adapted from Lippert (2011).

3.3 Physiological variables

The physiological adaptations associated with ageing, such as the decrease in muscle strength, caused by the loss of neurons, alteration of muscle fibres and aerobic capacity, affect the movement of older adults. Another impairment in the musculoskeletal system of the elderly is the decreased range of motion, caused by changes in connective tissues, ligaments, joint capsules, aponeuroses, tendons and skin (HAUSDORFF, 2005).

The musculoskeletal system performs an important role in locomotion and balance in humans. With advancing age, there is a significant decrease in the number of motor units, as well as changes in muscle tissue, which results in changes in the number and morphology of muscle fibres. In addition to these factors inherent to ageing, inadequate nutrition and hormonal changes lead to the phenomenon called sarcopenia, which influences muscle weakness, osteoporosis and metabolic syndromes, which can even cause death (GOODPASTER et al., 2006; WILKINSON; PIASECKI; ATHERTON, 2018).

In this sense, muscle weakness is indicated as a risk factor for high mortality in the elderly, as it affects muscle functionality, increases the risk of falls, which can lead to disability and loss of independence (GOODPASTER et al., 2006).

Ageing causes changes in muscle architecture that alter the arrangement and orientation of the fibres in relation to the line of force generation produced by the muscle. There is a reduction in the length of muscle fibres, related to the loss of sarcomeres in series, and a decrease in the angle of penetration of muscle fibres, consequently decreasing the anatomical and physiological cross-sectional area, related to the loss of sarcomeres in parallel, reducing the muscle volume, which may explain the decrease in strength production capacity of the elderly (BAPTISTA; VAZ, 2009; LIEBER; FRIDÉN, 2001).

The changes observed in cell morphology indicate the direct relationship between ageing and muscle architecture, as they alter the mechanical properties of the muscle, influencing the capacity to produce force in the different lengths of the muscle-tendon unit and gestures speeds, causing the activities of daily living are affected by the different architectural characteristics of the muscles of the human body (BAPTISTA; VAZ, 2009).

3.4 Biomechanical variables

Postural changes and changes in biomechanical variables caused by ageing cause an increase in energy costs during displacement. The reasons for these changes are the reduction in the range of motion of the hip extension, as well as the change in step width and walking cadence (WERT et al., 2010). When the gait of an elderly person is compared to that of an adult, there is a decrease in speed, shorter stride length and less plantar flexion in the propulsive phase (WINTER; PATLA; FRANK, 1990).

These biomechanical changes combined with the physiological changes mentioned above, cause the execution of a simple and everyday task to be affected, altering the natural patterns of walking and making other motor tasks more difficult, as they require greater effort and energy expenditure. In view of this, clinical and biomechanical gait assessment should be carried out for the safety and health promotion of the elderly.

3.4.1 Temporal-spatial variables

Several parameters can be used to describe and assess the gait of the elderly. For example, it is possible to use the distances and times of events related to the cycle or characterization of the gait. The changes related to ageing reflect the adoption of a more cautious strategy of movements during displacement, in order to increase stability and prevent falls. These preventive measures affect the parameters related to the distance (positioning) between the limbs during walking, as well as in the times that characterize the subphases of the gait cycle. What is most commonly observed is the reduction of stride size, as well as the increase or permanence of the stride width (MENZ; LORD; FITZPATRICK, 2003; WINTER; PATLA; FRANK, 1990).

When the temporal parameters are evaluated, comparing young adults with the elderly, in general, the time of contact with the foot on the ground and the time to complete a stride cycle increase, proportionally to the increase in age. Possible causes for this event are an increase in joint stiffness and changes in the movement of the foot during walking, in order to increase safety and balance (HAUSDORFF, 2005; WINTER; PATLA; FRANK, 1990). Moreover, self-selected speed and stride length in older adults are shorter when compared to younger (LARISH; MARTIN; MUNGIOLE, 1988).

3.4.2 Kinematic variables

The movements of the body segments during gait can be evaluated through kinematic analysis of the movement that describes the position, speed and acceleration, without considering the forces acting on the body. Changes in these variables involve several causes, such as joint stiffness, sarcopenia, altered balance, among others (KIRTLEY, 2006).

One of the most studied kinematic variables for assessing gait in the elderly is walking speed, which can be calculated by dividing the distance covered by the time used to cover the route. All people have a natural speed (self-selected), but this speed can be adjusted according to specific conditions (KIRTLEY, 2006).

Several studies show that walking speed decreases with advancing age (HAUSDORFF, 2005; MENZ; LORD; FITZPATRICK, 2003; PRINCE et al., 1997). This occurs in order to maintain a conservative and safe gait pattern, by controlling other space-time variables, such as stride size and frequency, causing a decrease in speed and an increase in stride time variability (MENZ; LORD; FITZPATRICK, 2003).

In fact, gait speed in the elderly is the most observed parameter. As the speed is a result of the association between the stride size and the cadence (frequency), it is estimated that this decrease is mainly influenced by the decrease in the step length. Cadence is related to the number of steps taken per minute, and does not appear to change during ageing (KIRTLEY, 2006).

Another explanation for the decrease in speed lies in the increase in the moment of double support in the gait cycle, as a way of compensating and protecting the instability caused by the loss of balance during the walking motion (MENZ; LORD; FITZPATRICK, 2003).

In addition to gait speed, variables such as horizontal head acceleration associated with decreased control of this acceleration, make it difficult to stabilize the body segment, influencing the visual pattern of the elderly.

3.4.3 Kinetic variables

Knowing the behaviour of forces and their results is essential for understanding the causes of any movement. The analysis of these forces that cause displacement is called the kinetic analysis of movement. During ageing, the distribution of forces, torques and joint moments is altered, reinforcing the importance of studying gait kinetic (KNUDSON, 2007).

One of the changes observed in the kinetic variables is the decrease in the propulsion force. This situation can be explained by the reduction of the muscular strength of the plantar flexors and the modification of the gait pattern in order to guarantee a safe displacement, since during the impulse the elderly person needs to make an upward and forward movement, which can cause destabilization during the gait (HAUSDORFF, 2005; WINTER; PATLA; FRANK, 1990).

In addition, energy absorption by the knee is greater in the elderly when compared to young adults, especially during the propulsion phase, where the elderly absorb almost 50% of the energy generated by the movement, and the young absorb 16%(HAUSDORFF, 2005; WATELAIN et al., 2000).

Muscle powers are also affected throughout the gait cycle. In the terminal support phase, for example, the second peak of hip power is higher in the elderly when compared to young adults, because to keep the base stable, the elderly spend more energy in controlling the hip (WATELAIN et al., 2000).

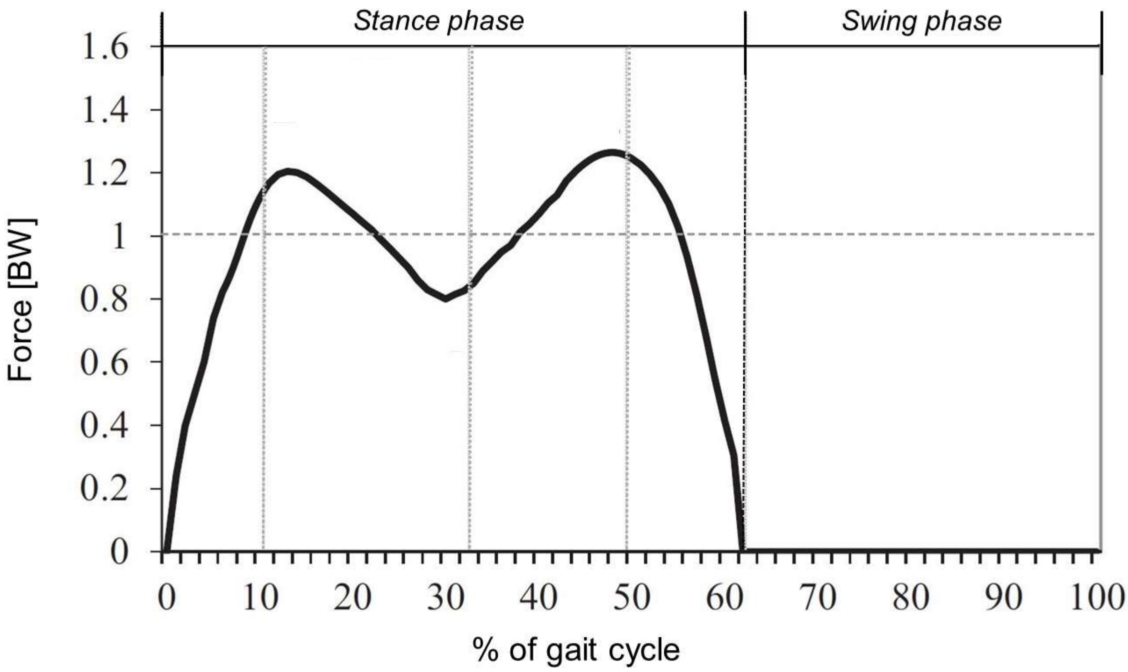
3.5 Characterization of GRF gait pattern

In the biomechanical analysis of movement, the physical magnitude called “force” represents the action that a body exerts on another, whether on rigid bodies or deformable surfaces. The effects of forces applied to these bodies can be assessed using Newton's Laws (ROBERTSON DGE, CALDWELL GE, HAMIL J, KAMEN G, 2013).

Newton's third law, known as the Law of Action and Reaction, contributes to the evaluation of human movement, as it explains how the forces applied to an object can cause movement in the opposite direction of application of these forces if the object's inertia or force is greater than the force acting on it (KNUDSON, 2007).

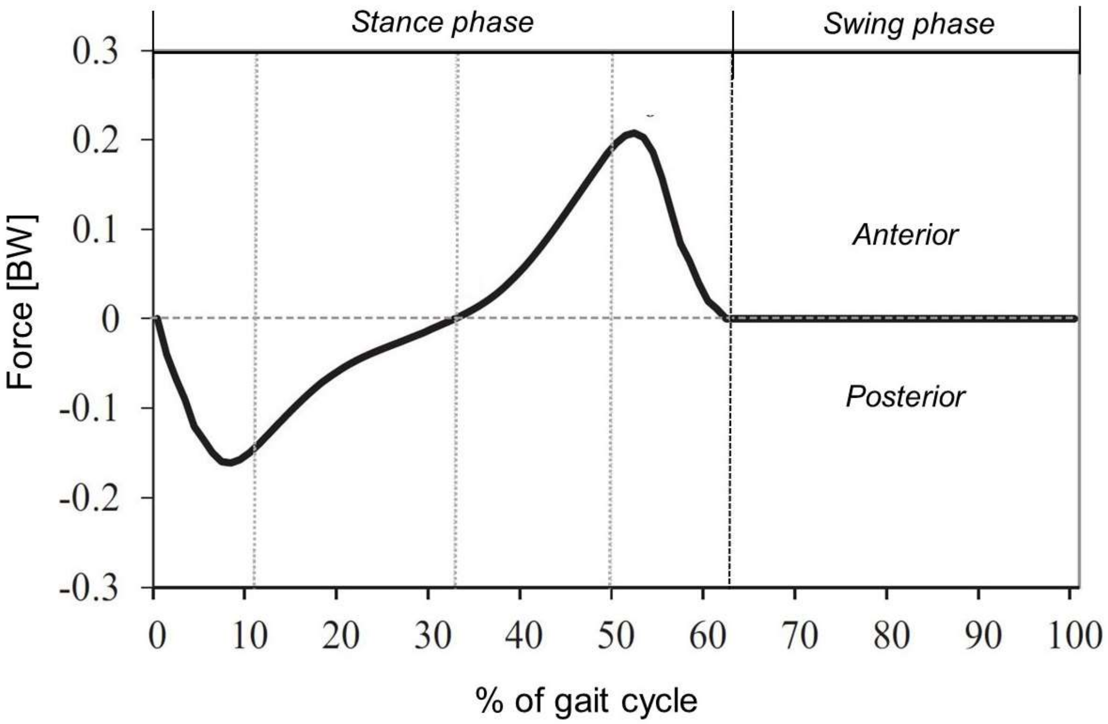
During gait, when the individual's foot touches the ground, ground reaction force (GRF), which is equal in magnitude and opposite in direction from the force that the foot makes on the ground, is applied on the subject. The resulting GRF vector has three components, vertical (Figure 2), anterior-posterior (Figure 3) and mediolateral (Figure 4). The direction of this resulting vector depends on the position of the lower limbs during the movement, and how much force the person applies in contact with the ground, which also determines the magnitude of the 3 components of the GRF.

Figure 2 - Typical vertical GRF curve of a normal gait



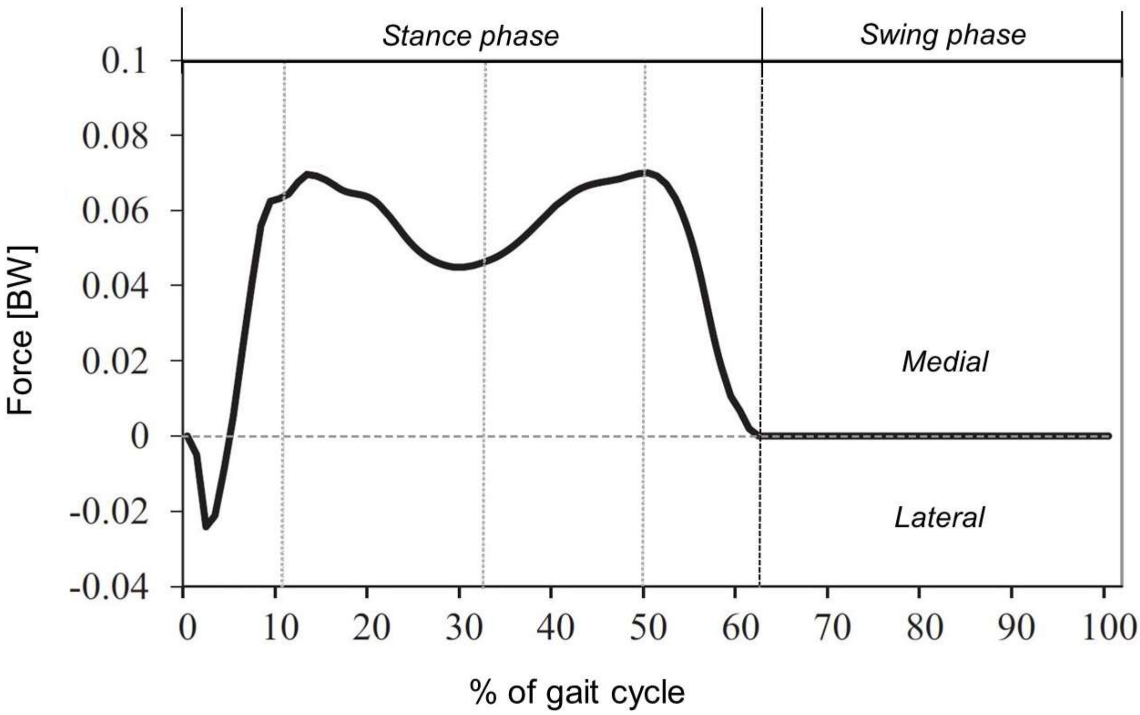
Source: adapted from Chockalingam, Healy, Needham (2016).

Figure 3 - Typical anterior-posterior GRF curve of a normal gait



Source: adapted from Chockalingam, Healy, Needham (2016).

Figure 4 - Typical mediolateral GRF curve of a normal gait



Source: adapted from Chockalingam, Healy, Needham (2016).

3.6 State-of-the-art

A systematic literature review was conducted in PubMed and Scopus to identify references related to kinetic gait analysis in older adults. A search protocol was designed to promote a consistent review of studies related to the area of interest, to ensure transparency, accountability and integrity of the research. The protocol was based on the Preferred Reporting Items for Systematic Reviews and Meta-Analyses (PRISMA) statement.

The following key search terms were used: (gait) AND ("gait analysis") AND (elderly) OR ("older adults") AND (walk*) AND (kinetic*) AND ("ground reaction force").

The research retrieved a total of 197 publications across the two selected databases. The results were then imported to Mendeley to remove duplicate records, which were only 3 articles. The remaining 194 articles were screened for their relevance based on the title and abstract, and selected according to the following criteria: 1) gait analysis using force plates or instrumented treadmill; 2) kinetic analysis through GRF; 3) adults aged > 60 years; 4) walking on a level surface; 5) walking without assistance.

The screening resulted in a total of 38 studies that were evaluated for eligibility and 29 full-text articles meet the criteria to be included in the review. The PRISMA flow diagram in Figure 5 describes the review process. There were no articles included in quantitative synthesis step in the flow, as we do not perform a meta-analysis. Relevant data (e.g. purpose of the study, participants' characteristics, kinetic methods used, variables of interest, and results) were reported in Table 1, when available.

Figure 5 - Prisma flow diagram of the search strategy and its results

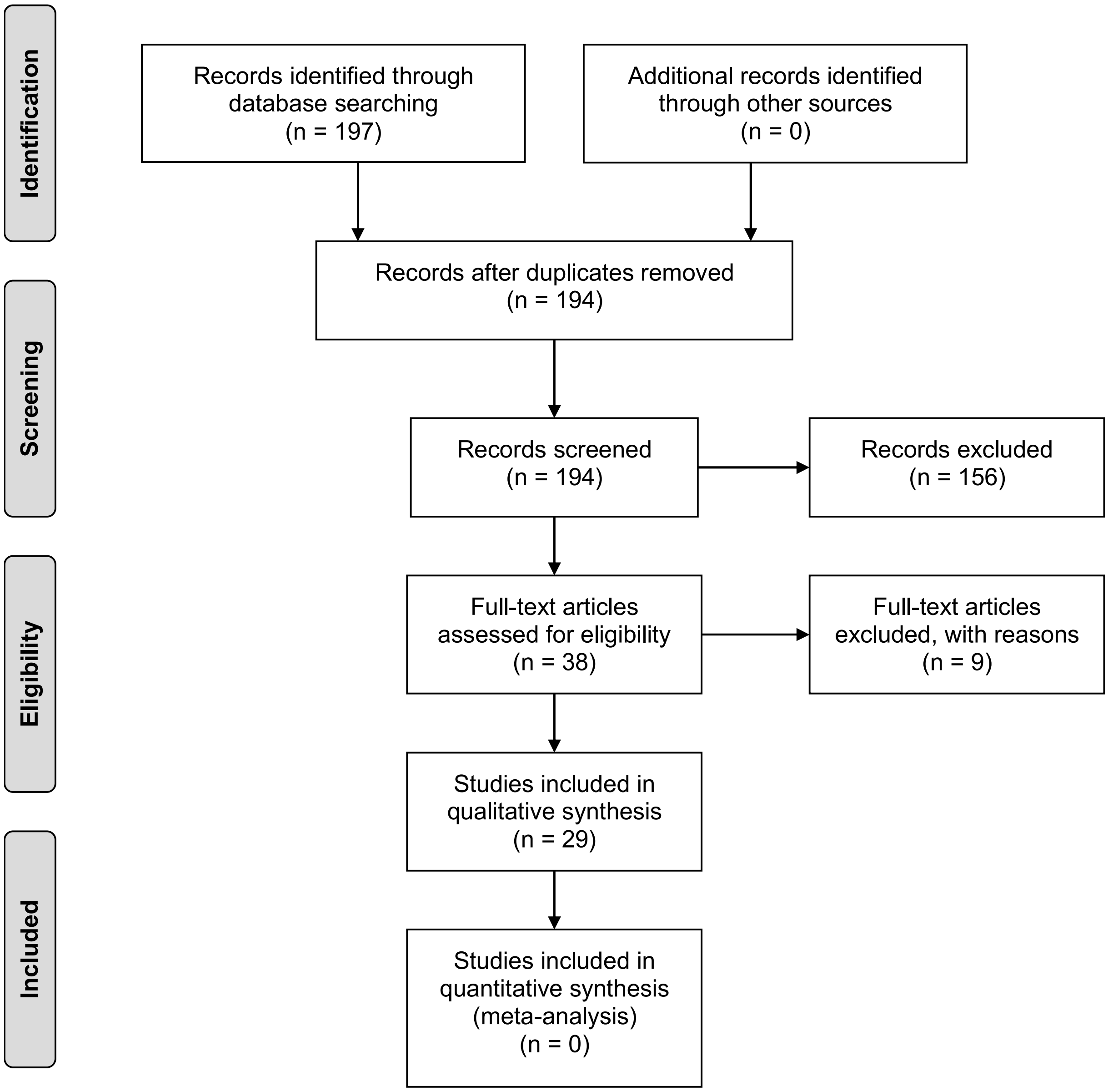


Table 1 - Investigation of studies on kinetic gait analysis in older adults

Study	Objective	Participants	Methods	Parameters	Results/Observation
(ALAM et al., 2017)	Differentiate PD patients from healthy controls	n = 47; 29 PD patients (71.1 ± 8.05 years), 18 older adults (71.6 ± 6.6 years)	8 foot sensors	Vertical GRF	Vertical GRF data obtained non-invasively from wearable devices, in combination with a Support Vector Machine (SVM) classifier trained can differentiate healthy and pathological gaits
(ANDERSON; MADIGAN, 2013)	Examine the relative contributions of age-related differences in femoral loading during walking and age-related differences in femoral bone mineral density to age-related differences in strains in the proximal femur	n = 10; 5 elderly (79.4 ± 4.6 years); 40 youger (25.0 ± 4.3 years)	A six-degree-of-freedom FP, f_s=1000 Hz	GRF (peak values), hip joint contact forces, and hip flexor forces	GRF peaks were lower in older adults compared to young adults
(ARENA et al., 2017)	Investigate obesity and age-related differences in the required friction while walking	n = 65; 20 young non-obese (24.5 ± 3.5 years), 20 young obese (23.5 ± 3.2 years); 14 older adults non-obese (66.6 ± 4.9 years); 11 older adults obese (70.5 ± 7.4 years)	1 FP (Bertec)	Peak required coefficient of friction (RCOF), vertical GRF at peak RCOF, shear GRF at peak RCOF	No differences in required friction were found between non-obese and obese younger adults

(BEGG, 2006)	Test whether an artificial neural network can be applied to model relationships between the GRF–time parameters	n = 27; 14 young (21.2 ± 1.3 years) and 13 older adults (67.6 ± 4.8 years)	1 FP (AMTI), f_s= 200 Hz	Vertical, anterior-posterior and mediolateral GRFs	The method was able to accurately estimate stance time and push-off force/time data
(BEGG; KAMRUZZAMAN, 2005)	Automated recognition of young/old gait through space temporal, kinetic and kinematics parameters	n=24; 12 elderly (68.8±7.46 years) and 12 youngers (28.1 ± 7.56 years)	2 FPs	Vertical GRF (1st and 2nd peak)	Discrimination between young and elderly walking and mapping the underlying data structure relating to young and ageing populations
(BEGG; KAMRUZZAMAN, 2006)	Test whether an artificial neural network could be applied to detect gait changes due to ageing using standard gait features that are recorded during gait analysis	n = 24; 12 young (28.1 ± 5.6 years) and 12 older adults (68.8 ± 4.6 years)	1 FP (AMIT) and a 2D Mot.Analysis system (Peak Performance Inc, USA)	Vertical and anterior-posterior GRF (peaks)	An artificial neural network were able to differentiate young/old gait with an accuracy of 83.3% across all participants
(BIGGS et al., 2019)	Identify which biomechanical features of OA significantly change following surgery, applying Principal Component Analysis (PCA) combined with a classification method based on a Dempster-Shafer Theory (DST)	n = 60; 30 participants with knee OA (70.7 ± 8.3 years) and 30 healthy participants (39.8 ± 17.6 years)	2 FPs (Bertec), f_s=1080 Hz	Vertical, anterior-posterior and mediolateral GRFs	The classifier successfully discriminated between OA and healthy gait in all 60 cases
(FAVARO et al., 2019)	Evaluate the GRF in young and old individuals under the influence of simulated overweight in different gait speeds	n = 30; 15 young (22 ± 3.7 years) and 15 older adults (69.8 ± 6.4 years)	1 FP	Vertical and anterior-posterior GRF peaks, Impulses	Differences were found only in the anterior-posterior GRF and for deceleration impulse

(FUKUCHI; FUKUCHI; DUARTE, 2018)	Create a dataset of kinematics and kinetics data on healthy young and older adults in both the treadmill and over ground environments	n=42; 24 young adults (27.6 ± 4.4 years) and 18 older adults (age 62.7 ± 8.0 years)	5 FPs and a dual-belt instrumented treadmill	Vertical, anterior-posterior and mediolateral GRFs, ankle force, ankle moments and ankle power	Creation of a public dataset containing raw and processed kinematics and kinetics data on both over ground and treadmill walking trials at a range of gait speeds in both young and older healthy adults
(GINNERUP-NIELSEN et al., 2015)	Evaluate the efficacy of a specialized rosehip powder on the biomechanical knee joint function during walking in individuals with knee-related walking limitations compared with placebo	n = 100; 50 older adults - Rosehip (67.5 ± 9.0 years) and 50 older adults - Placebo (66.12 ± 9.69 years)	2 FPs (OR-6-5-1000, AMTI), f_s=1500 Hz.	Vertical GRF (1st and 2nd peak)	There were no statistically significant group differences in the vertical GRF
(HITZ et al., 2018)	Analyse the impact of the moving fluoroscope on the gait characteristics, specifically the time distance parameters, whole body kinematics, and GRF of young and elderly participants	n = 19; 10 young (24.5 ± 3.0 years) and 9 older adults (61.6 ± 5.3 years)	5 FPs (Kistler), f_s=2000 Hz	Vertical GRF (1st peak, valley and 2nd peak), loading rate (bn) and unloading rate (en)	There were no statistically significant differences in parameters analysed between the conditions
(HSU et al., 2015)	Investigate the immediate and long-term effects of laterally-wedged (LW) insoles on the lower limb joint biomechanics and knee abductor moment (KAM)	n = 10 patients with bilateral medial knee OA (66 ± 5.3 years)	2 FPs (OR-6-7-1000, AMTI), f_s=1080 Hz	GRFs, moments and COP	The (KAM) with LW insoles at baseline (initially) was significantly reduced when compared to the barefoot condition ($p < 0.05$), suggesting that the LW insoles were effective in reducing unfavourable loading at the knee immediately upon wearing the insoles.

(JONES et al., 2016)	Compare the gait of unicompartmental knee arthroplasty (UKA) and total knee Arthroplasty (TKA) patients with healthy controls, using a machine-learning approach	n = 145; 12 elderly (UKA - mean age 65 years), 12 elderly (TKA - mean age 68) and 121 healthy controls (mean age 32 years)	ITD, f_s=100 Hz	Vertical GRF	There was no significant difference between the TKA and UKA groups for age
(JONES et al., 2016)	Test whether support moment and individual joint contributions differed between post-stroke participants with different gait function	n = 37; 14 non-hemiplegic elderly (NE) (62.86 ± 5.48 years); 12 post-stroke hemiplegic elderly (60.58 ± 7.05 years) and 11 post-stroke hemiplegic elderly walking with a cane (58.91 ± 8.31 years)	2 FPs (OR6-5, AMTI), f_s=60 Hz	Vertical GRF (1st and 2nd peak) and temporal occurrence of the peaks	All participants in the NE group showed a double peak and trough (bimodal-shape) in the vertical GRF curve
(LAROCHE et al., 2016)	Examine how manipulation of strength-to-weight ratio (S:W) affects self- selected walking speed and if acutely increasing S:W affects walking performance differently for normal weight, overweight, and obese older adults who possess different levels of S:W	n = 27; 9 normal weight (70,9 ± 6,2 years), 9 overweight (70,2 ± 4,5 years) and 9 obese participants (70,3 ± 6,1 years)	ITD	Vertical GRF (1st and 2nd peak)	Low S:W was associated with slower preferred walking speed

(LAROCHE; COOK; MACKALA, 2012a)	Determine how knee extensor strength asymmetry influences gait asymmetry and variability	n = 24 women; 13 symmetrical strength-SS (70.9 ± 4.7 years) and 11 asymmetrical strength -AS (72.5 ± 4.6 years)	ITD (Gaitway II;Kistler), f_s=100 Hz	Vertical GRF peaks	Gait variability and asymmetry wee greater in older women with strength asymmetry
(MARTÍNEZ-RAMÍREZ et al., 2013)	Investigate how mobility characteristics during walking, relate to gait velocity and questionnaire outcomes of patients with hip osteoarthritis in an outpatient setting	n = 22 elderly with hip osteoarthritis (63 ± 10 years)	Instrumented Force Shoes (IFS) (Xsens Technologies)	Vertical GRF and time parameters	Gait parameters correlated significantly with velocity, although symmetry index parameters were not
(PARVATANENI et al., 2009)	Examine gait kinetics and kinematics and the metabolic demands associated with over ground and treadmill walking in healthy adults over the age of 50 years	n = 10 (5 women and 5 men, mean age 60.6 ± 7.4 years)	2 FPs (AMIT) and 1 ITD (Kistler)	Vertical GRF	Significant kinetic and metabolic differences were found between the two modes of walking, treadmill and over ground
(PATERSON et al., 2017)	Examine sex- and obesity-related differences in knee biomechanics relevant to knee arthroplasty in a group of people with severe knee AO	n = 34 patients with severe knee OA (70.0 ± 7.2 years)	2 FPs (OR6-6-2000 AMTI), f_s=1200 Hz	Knee adduction moment (KAM), KAM impulse, knee flexion moment, vertical GRF peak	Men had a higher absolute peak KAM, KAM impulse and peak GRF compared to women
(PENN et al., 2019)	Analyse the gait patterns of patients with Parkinson's disease (PD) while walking on a non-motorized treadmill (NMT)	n = 25; 12 patients with PD (77.6 ± 7.1 years), 13 older adults - control group (61.0 ± 10.8 years)	ITD	Vertical GRF	A significantly lower VGRF/BW ratio was noted during both comfortable and maximal walking speeds in the PD group compared with the controls

(PÖTZELSBERGER et al., 2015)	The aim of the study was to examine the effects of a 12-week recreational skiing intervention in people with total knee arthroplasty (TKA) on functional performance	n=23 older adults (71 ± 5 years). 10 participants with a unilateral TKA and 13 participants in the control group	Pedar mobile system	GRF curve and asymmetry index from force variables	Alpine skiing as a leisure-time activity has a beneficial effect on gait performance and leads to a more balanced load distribution between the legs during daily activities
(ROUHANI et al., 2011)	Design and validate the measurement of ankle kinetics (force, moment, and power) during consecutive gait cycles and in the field using an ambulatory system	n = 22; 12 participants with ankle osteoarthritis (58 ±13 years) and 10 healthy participants (61 ±13 years)	Custom-made shoes embedding a pressure insole (Pedar, Novel, DE) and a FP	Vertical, anterior-posterior and mediolateral GRFs, ankle force, ankle moments and ankle power	GRF assessed by ambulatory system compared to stationary system showed median NRMSE<3% and R>0.94 for both patients and healthy individuals
(RUTHERFORD et al., 2017)	Compare a group of individuals with moderate medial compartment knee osteoarthritis (MOA) to both an age-matched asymptomatic group of older adults and an asymptomatic group of young adults	n = 60; 20 young adults (25 ± 2 years), 20 older adults (61 ± 7 years), 40 older adults with OA (61 ± 7 years)	GaitRITE instrumented walkway (CIR Systems, USA), f_s=2000 Hz	GRF and moments	Individuals with knee OA have distinct biomechanics and muscle activation patterns when compared to age-matched asymptomatic adults and younger adults
(SALEH et al., 2018)	Investigate the effects of ageing on different gait parameters such as the spatiotemporal, kinetic and kinematic	n = 12; 6 elderly (62±3.9 years) and 6 young (23±1.5 years)	Moticon Insoles	Force on heel to body weight ratio (HBW), force on toe to body weight ratio (TBW) and the COP	No significant differences were found in HBW and TBW between the groups

(SHARMA; MCMORLAND; STINEAR, 2015)	Characterize GRFs acting on the limbs during gait initiation (GI) after stroke	n = 46; 18 chronic stroke patients (mean age 67.6 years), 28 healthy older adults (mean age 67.6 years)	2 FPs (AMTI), f_s=50 Hz	1st anterior-posterior GRF peak, positive and negative medial-lateral GRF peaks	An effect of side-of-lesion was revealed in average peak lateral ground reaction force data
(TAKAHASHI et al., 2004)	Evaluate the relationship between knee pain and various indicators of the combined performance of the lower extremity and to determine whether the classification of vertical GRF correlates with gait parameters and functional performance	n = 130 (mean age 80 years with a range of 65–94 years)	Gait Scan 8000 - thin-film sensor walkway (Nitta Co. Ltd, Osaka, Japan)	Vertical GRF	No significant association was found between knee pain and timed up and go, functional reach test, or gait parameters in elderly female participants
(TODA; NAGANO; LUO, 2013)	Examine the relation between the antero-posterior ground reaction forces and hip, knee, and ankle joint moments during walking in elderly and the younger individuals	n = 80; 40 elderly (mean age 70,1 years); 40 younger (mean age 23,2 years)	8 FPs	Anterior-posterior GRF (peak values)	Anterior and posterior components of GRF were lower in the elderly participants
(WANG et al., 2018)	Develop an automatic feature extraction method to analyse patterns from high-dimensional autocorrelated gait waveforms	n = 74; 43 patients with total knee arthroplasty -TKA (69.9 ±8.5 years) and 31 healthy individuals (69.9 ±8.0 years).	2 FPs (Kistler and AMTI), f_s=1000 Hz	GRF waveforms in anterior-posterior and vertical directions.	The proposed method could capture virtually all significant differences between TKA patients and the controls

(YEO et al., 2019)	Compare clinical and radiologic outcomes and perform gait analysis using three-dimensional (3D) spatiotemporal, kinematic, and kinetic parameters during walking between two alignment methods in robotic-assisted TKA	n = 60 elderly; 30 mechanical alignment- MA (age 74 ± 5.16 years), 30 kinematic alignment group - KA (72 ± 5.52 years)	2 FPs (Kistler)	Vertical, anterior-posterior and mediolateral GRFs	Vertical and anterior-posterior GRFs showed no significant difference between the two groups

FP = force platform; ITD = instrumented treadmill; f_s = sampling frequency.

The current literature in the field of biomechanical gait analysis vary widely in terms of methodology and procedure, equipment, interpretation of data and presentation of results (BOYER et al., 2017). Some of the studies analysed do not present in a discriminatory way the methods used in the gait analysis, for this reason, some information was not reported in Table 1, such as the sampling rate of the collections, the parameters related to the GRF curve, the statistical treatment performed, among others.

After evaluating the selected studies, it is clear that the use of force or similar platforms (instrumented treadmills, instrumented insoles) is performed for different purposes, comparison of different populations or interventions, prediction of falls in the elderly, development of gait analysis models and assess what are the causes of human movement in different situations. The use of commercial software facilitates the evaluation, however, it often limits the reproducibility of the study, since the laboratories are equipped with different equipment from different brands in the field of biomechanics.

The main intention of the state of art was to review the current studies in older adults' gait analysis, investigate the intervention or disease that most affect older adults and how gait is assessed through different types of equipment. It helped to develop our methodology and choose the most appropriate gait parameters for evaluation.

Analysing the methodology proposed by different researches, the studies showed a sampling rate ranging from 50 Hz to 2000 Hz, but there is no justification for the chosen frequency and in some cases the frequency is not even mentioned. Also, one of the most diversified items when the studies are evaluated is the investigated kinetic parameters and how the results are presented. Not all studies showed or explained how the variables were obtained and what were the variables used for kinetic evaluation.

Some studies discuss GRF analysis, but do not specify which parameters were analysed for the purpose. The researchers present the results in different ways, but the quantitative and absolute values of the measured variables are not always presented. Moreover, some articles show the force curves, but do not show the values (FUKUCHI; FUKUCHI; DUARTE, 2018; GINNERUP-NIELSEN et al., 2015).

This lack of standardization difficult the literature review. For example, some studies seek to compare different populations, between young and elderly, or between healthy elderly and elderly people with some movement limitation. A point to be observed, and that has already been pointed out by another study (ROBERTS; MONGEON; PRINCE, 2017), is the number of individuals evaluated. The study with the largest number of participants showed n = 145 but only 24 individuals were elderly, the other 121 participants were part of the control group (JONES et al., 2016). TAKASHI showed a sample of 130 individuals (TAKAHASHI et al., 2004), and GINNERUP-NIELSEN a sample of 100 volunteers (GINNERUP-NIELSEN et al., 2015), but in these cases, all individuals were elderly.

The results of this review support the idea of a huge variability of applied methods and investigated parameters, without a clear and standardized way of communicating the results and, therefore, not being easily reproducible. It is clear that the use of force platforms or similar (instrumented treadmills, instrumented insoles) is performed for different purposes, comparison of different populations, prediction of falls in the elderly, and investigating physical and mental illnesses.

For this reason, elucidating the biomechanical gait assessment techniques in older adults, as well as the parameters involved in the investigation, need to be better standardized, so that comparisons can be made, and the studies can be replicated in other environments.

4 METHODS

This study proposes a method to analyse older adults´ gait GRF signals. This approach can be used to directly calculate gait parameters provided by digital signal processing and use these parameters to analyse and investigate the pattern of multiple gait cycles, facilitating the clinical interpretation of gait in this population.

4.1 Older adult's dataset

Participants were recruited from a physical activity program performed in our University. Thirty-three women able to walk without any assistance were randomly sampled. All volunteers were briefed in advance about the purpose and experimental procedure of the study, all volunteers signed a written consent form before beginning the tests. The Participants' age, height, and weight were (mean ± standard deviation [SD]) 70.45 ± 6.92 years, 154.76 ± 7.28 cm, and 72.65 ± 14.00 kg, respectively. Table 2 details the demographic data for the participants.

Table 2 - Demographic data for the volunteers

Volunteer nº	Age (years)	Height (cm)	Weight (kg)
1	80.00	159.00	74.72
2	60.00	158.00	78.54
3	71.00	147.00	67.91
4	77.00	150.00	66.59
5	71.00	181.00	89.51
6	60.00	151.00	85.26
7	78.00	145.00	62.46
8	79.00	148.00	62.77
9	66.00	159.00	92.76
10	80.00	161.00	81.90
11	70.00	149.00	79.73
12	82.00	158.00	64.93
13	66.00	148.00	60.87
14	65.00	158.00	50.60

15	69.00	163.00	94.54
16	61.00	156.00	73.83
17	68.00	150.00	57.52
18	63.00	156.00	93.67
19	66.00	157.00	67.85
20	71.00	159.00	75.51
21	66.00	151.00	52.55
22	68.00	166.00	104.70
23	82.00	157.00	78.97
24	74.00	143.00	55.93
25	82.00	146.00	58.21
26	70.00	154.00	80.18
27	62.00	150.00	52.08
28	72.00	156.00	91.10
29	81.00	151.00	64.41
30	69.00	151.00	59.11
31	68.00	160.00	78.60
32	64.00	158.00	74.77
33	64.00	151.00	65.29

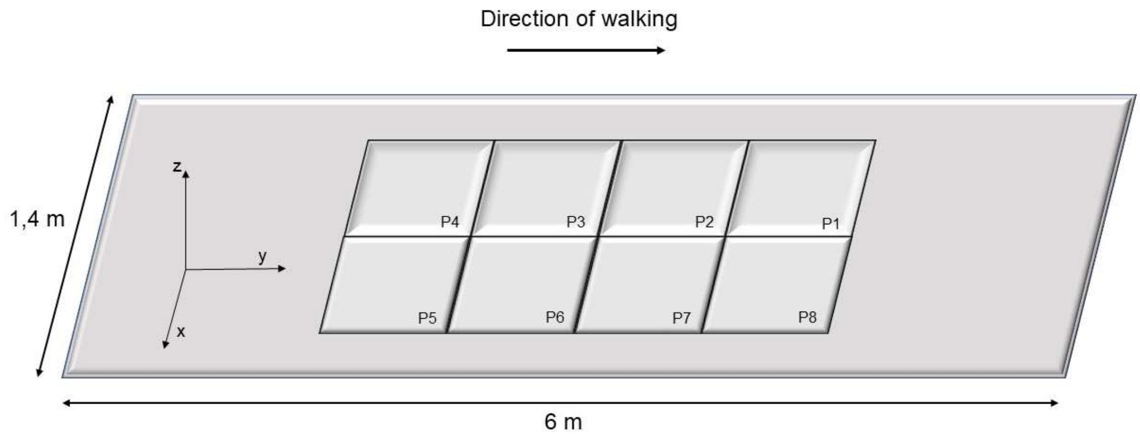
4.2 Experimental protocol

Participants were allowed to wear their usual clothes and shoes and walk with their preferred (normal) speed along a 6 m long by 1.4 m wide walkway, containing 8 force platforms positioned in pairs at the centre of the walkway. The scheme of the gait measurement is presented in Figure 6.

To allow familiarization with walking on the platforms and the environment, the individuals performed two tests before recording data. The participants performed at least five gait trials and a rest period of two minutes was given between the tests.

The data acquisition protocol was previously approved by the University Research Ethics Committee (CAEE 55674116.3.0000.5336).

Figure 6 - Walkway and force platforms



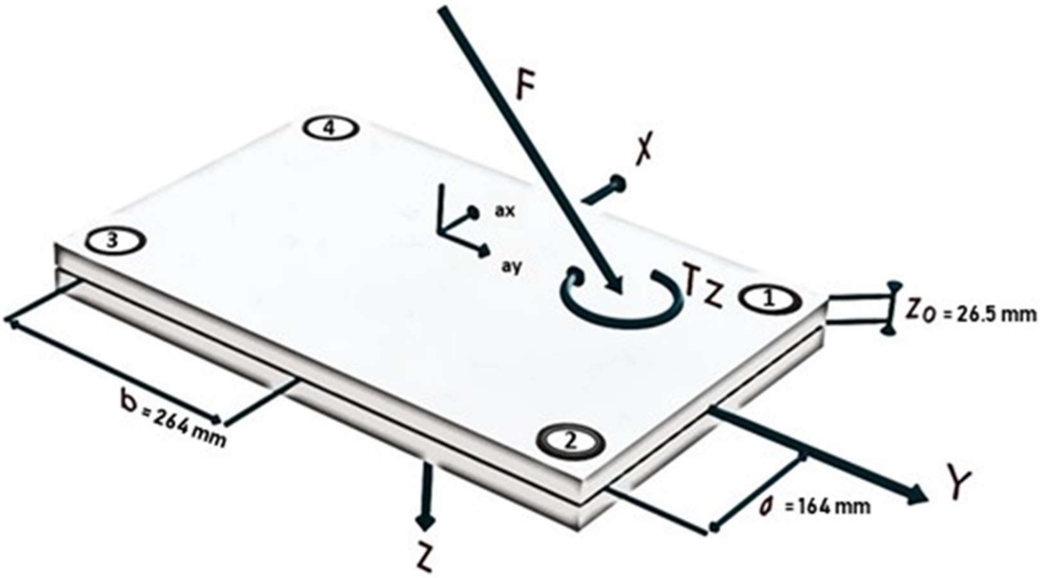
P1 – P8: force platforms.
Direction of the movement to determine the orientation of the forces.

4.3 Instrumentation

The force plates (BTS P6000, BTS Bioengineering, Italy) used in this study are in agreement with the minimum requirements for GRF capture during human gait. According to the user manual the equipment is classified as a medical device and complies the European Directive 93/42 /CEE (and its amendments) and has the INMETRO - National Institute of Metrology, Standardization and Industrial Quality - warranty seal.

All force platforms are fully digital and equipped with 12 strain gauges each (Figure 7), there are three sensors placed at each platform corner. Which make the equipment suitable for dynamic force measurement and are large enough (40 mm x 60 mm) to hold an entire foot. The system coordinate is used for X, Y and Z axes, mediolateral axis, anterior-posterior axis (direction of motion) and vertical axis (proximal-distal) of coordinates, respectively. Each sensor has a detection capacity of up to ± 2000 N at each of the coordinates (x, y and z), through accurate high-frequency analysis.

Figure 7 - Geometric configuration of a force platform

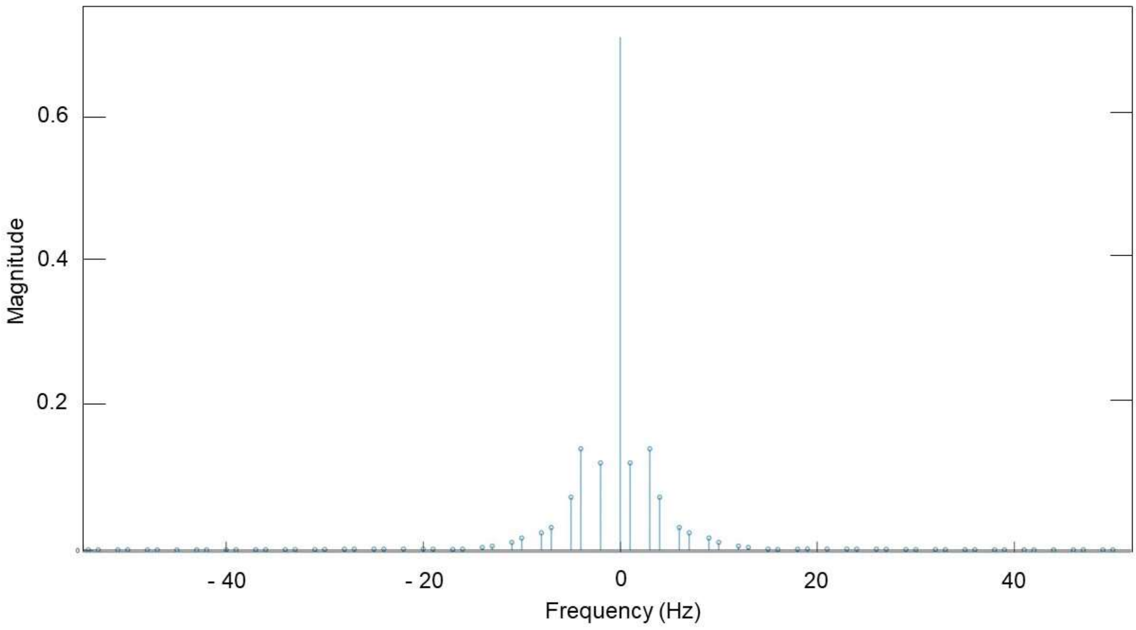


Force platform dimensions and location of the 12 strain gauges.
F: resultant force vector obtained by the association of the three force directions (x, y, z);
Tz: vertical torque.

The data set used for this methodology was composed of samples with 100 Hz to 1000 Hz of sampling rate, which were collected on different days and in varied situations. The fundamental most abrupt components of human gait frequency are less than 10 Hz (ROBERTSON DGE, CALDWELL GE, HAMIL J, KAMEN G, 2013). Thus, the 100 Hz to 1000 Hz band of frequency is safe to prevent aliasing of the signal and respect the Nyquist sampling theorem, since the sampling rate of the signal must be at least twice the highest frequency in the signal.

To reinforce the chosen frequency, a Fast Fourier Transform (FFT) was applied to a vertical force signal. The signal spectrum confirmed that the one-step signal in the frequency domain and their respective display frequencies are predominantly below 20 Hz (Figure 8).

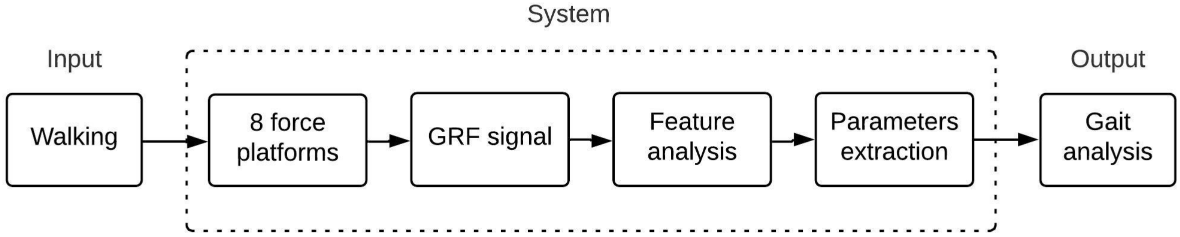
Figure 8 - FFT of vertical GRF signal



4.4 Data collection

Figure 9 shows the block diagram of the system for assessing and analysing GRF in the elderly. The set of input variables is obtained from walking tests performed on force platforms, the system consists of signal processing to determine specific parameters for gait evaluation in older adults, making it possible for gait analysis to be performed on system output.

Figure 9 - System block diagram



Ground reaction forces were measured as a function of time by 8 force platforms during the stance phase of the gait cycle. Force and time data recorded are exported to a text file containing an 8-column matrix, which contains the force values (in

Newtons) in the 3 coordinates (x-mediolateral; y-anterior-posterior; z-vertical) for each side of the body, right and left, associated with the acquisition time, for subsequent upload to the developed analysis routine, in order to process the signal and define the parameters of interest.

The data recorded is called as raw data and it must be treated before performing the gait analysis. For that purpose, a custom algorithm written in Octave (GNU v.4.4.1) was created to process and analyse these raw data recorded with the help of a BTS Software interface (Data acquisition system), developed by the company of the force platforms.

Applying a filter to the captured signal should only be used if the noise is a clear contaminant of the signal and if it interferes with the analysis. Filter selection is an empirical procedure since the noise is never completely known, so the chosen filter must be carefully selected, when necessary.

The force platforms already have built-in filters and are not susceptible to noise, so the data was not filtered after capture to avoid losing any important information. The GRF components captured by the platforms are calculated by the sum of the forces measured by the sensors allocated at the corners of the plate, as set out below.

Mediolateral force (Fx) is calculated from the sum of the force components measured on the x-axis, identified by sensors located at corner 1 and corner 2 summed with the forces identified by sensors placed at corner 3 and corner 4.

$$Fx = fx12 + fx34$$

Where,

$$fx12 = force\ on\ x - axis\ measured\ by\ sensor\ 1 + sensor\ 2$$

$$fx34 = force\ on\ x - axis\ measured\ by\ sensor\ 3 + sensor\ 4$$

Anterior-posterior force (Fy) is calculated from the sum of the force components measured on the y-axis, identified by sensors 1 and 4, added to the forces identified by sensors 2 and 3.

$$Fy = fy14 + fy23$$

Where,

$$fy14 = force\ on\ y - axis\ measured\ by\ sensor\ 1 + sensor\ 4$$

$$fy23 = force\ on\ y - axis\ measured\ by\ sensor\ 2 + sensor\ 3$$

Vertical force (Fz) is calculated from the sum of the force components measured on the z-axis, identified by sensors 1, 2, 3 and 4.

$$Fz = fz1 + fz2 + fz3 + fz4$$

Where,

$$fz1 = force\ on\ z - axis\ measured\ by\ sensor\ 1$$

$$fz2 = force\ on\ z - axis\ measured\ by\ sensor\ 2$$

$$fz3 = force\ on\ z - axis\ measured\ by\ sensor\ 3$$

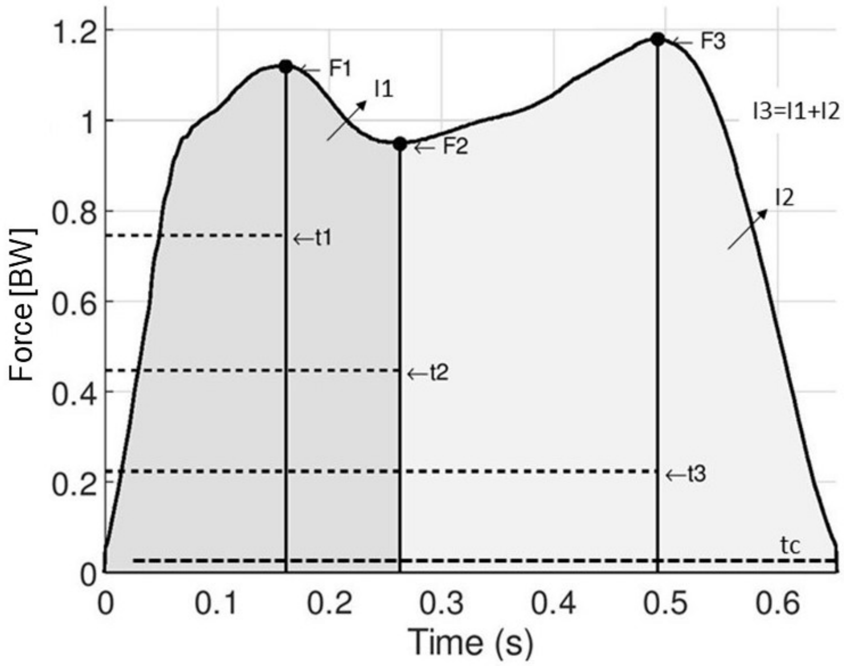
$$fz4 = force\ on\ z - axis\ measured\ by\ sensor\ 4$$

4.5 Data processing

The raw data containing the three components of GRF recorded by the force plates are then submitted to data processing. The parameters detection were made by an algorithm developed in this study, addressing the main points of interest for gait evaluation.

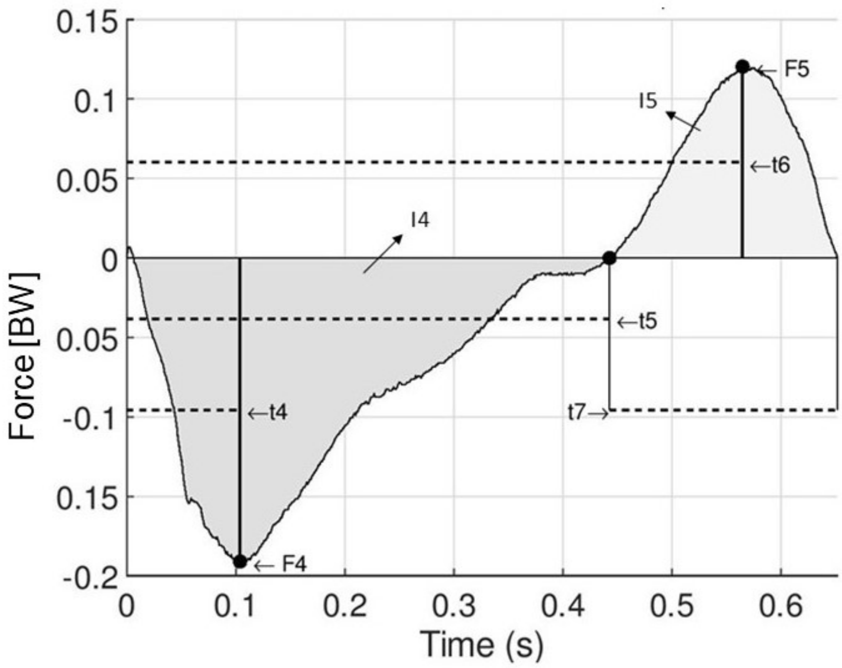
The study was carried out using the vertical component (Fz), figure 10, and anterior-posterior component (Fy), figure 11. The mediolateral component (Fx) was not included in this study due to the high variability (MASANI; KOUZAKI; FUKUNAGA, 2002; VAVERKA et al., 2015). All the parameters identified are related to these two curves and were identified through mathematical strategies implemented in the signal processing routine.

Figure 10 - Vertical GRF curve from the developed methodology



F1: first vertical peak; F2: vertical valley, F3: second vertical peak, t1: time to F1, t2: time to F2, t3: time to F3, tc: time of stance phase I1: impulse of load response and midstance (light gray), I2: impulse of terminal stance and preswing (dark gray), I3: total impulse of vertical GRF.

Figure 11 - Vertical GRF curve from the developed methodology



F4– braking peak, F5: propulsion peak, t4: time to F4, t5: time of braking phase (zero crossing), t6: time for F5, t7: time of propulsive phase, I4: braking impulse, I5: propulsive impulse, I6 (I4+I5): total impulse of anterior-posterior GRF.

The algorithm does the step by step extraction of gait features from Fz and Fy. The gait parameters, computed for the left and the right side, were: tc - time of ground foot contact (stance phase) - computed from initial contact of foot and ends at toe-off of the same limb, F1 - first vertical peak, F2 – the minimal force between the first and second peaks (vertical valley), F3 - second vertical peak, F4 – first anterior-posterior peak (braking peak), F5 – second anterior-posterior peak (propulsive peak), t1 – time to F1, t2 – time to F2, t3 – time to F3, t4 – time to F4, t5 – time of braking phase (zero crossing), t6 – time for F5, t7 – time of propulsive phase, I1 - impulse of load response and midstance, I2 - impulse of terminal stance and preswing, I3 - total impulse of vertical GRF, I4 – braking impulse, I5 – propulsive impulse, I6 – total impulse of anterior-posterior GRF.

When foot touches the ground the force sensors start to capture the reaction signal of the lower limbs. Initial contact and toe-off events were defined when the vertical GFR exceeded 1% of the maximum absolute force and dropped below 1% of the maximum absolute force, respectively. At each step, the algorithm determines whether the input signal is valid based on the characteristics of the GRF curve over time, which is expected to follow the normal "M" curve, due to this shape resembles the letter "M".

The characteristic vertical GRF force-time curve shows two peaks of force and a valley (Fig. 10). The reaction force signal increases to a maximum value, related to the first event of the gait cycle, initial contact. So, we have our first parameter of interest, F1. The second peak of force, F3, occurs at the end of the support phase and relates to the moment of propulsion of the limb to start the swing phase. The valley, F2, found between the first and second peak corresponds to the moment when the foot is in the medium support position, and the contralateral limb is in the swing phase.

The important values to be detected are the force values related to the gait cycle, the times of occurrence of these events, as well as impulses, characterized by the areas below the curves. These values are the points of interest of the study, as they characterize the gait of the elderly and enable the interpretation and intervention of professionals in the treatment of this population.

After the step detection, the amplitude values of the force, Fz (vertical) and Fy (anterior-posterior) were normalized by dividing the force by the product of body mass times acceleration due to gravity (body weight-BW), in order to allow comparison between the participants, as in (1):

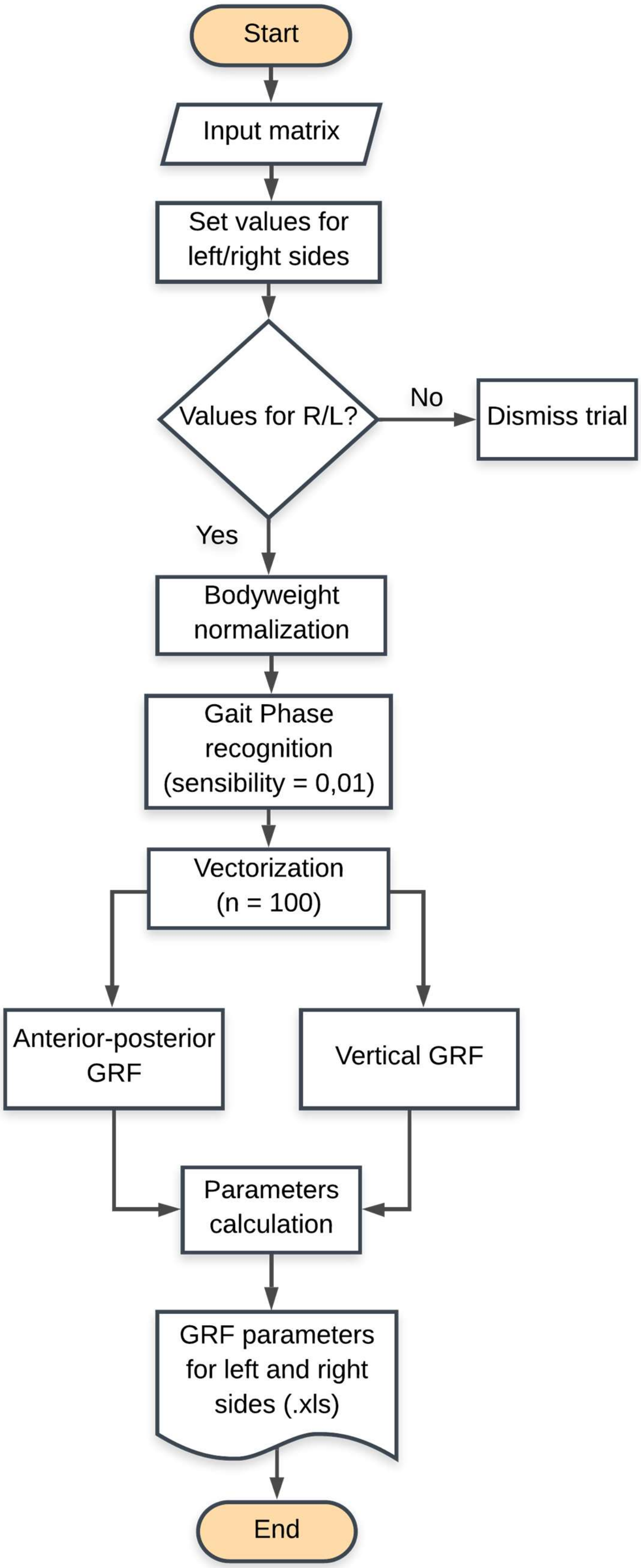
$$\text{Normalized GRF} = \text{GRF}/\text{BW} \tag{1}$$

This normalized signal is then segmented into a vector of size n = 100 observations, which corresponds to a duration of stance phase. This vectorization process was important to allow the evaluation of gait analysis collected with different sampling frequencies.

Identification of the GRFs parameters is the next step of processing. After the successful recognition of the forces, the system inputs are generated, based on the force values related to the acquisition time of the signal. Data was collected for the right and left legs of the individuals and the values for each trial were calculated from the mean values of the steps performed by each volunteer.

The overall data processing is shown in the flow diagram (Figure12).

Figure 12 - Algorithm flowchart



4.6 Statistical Analysis

All collected data were statistically analysed with IBM SPSS statistical software, version 21, statistical significance was set at $p < 0,05$. Descriptive statistics were calculated for all gait parameters and data was expressed as mean ± standard deviation. Variables were assessed for normality using the Shapiro-Wilk test. When normality assumption was not met, nonparametric statistics were used.

4.6.1 Variability measures

Within-subject variability was calculated from the performed trials in each session, and between-subject variability was calculated to ensure normal distribution of all parameters.

The investigation of gait variability, the magnitude of parameters fluctuations, is an additional measure of human locomotion, which has widely been used as a way to quantify spatiotemporal inconsistency of strides (HAUSDORFF, 2005).

This fluctuation also can be observed in the kinematic, electromyographic and kinetic measurements, for within-subjects, in repeated measurements over time or different interventions, or between subjects (CHAU; YOUNG; REDEKOP, 2005), these two types of variability represent the total variation (HAZARD MUNRO et al., 2005). In this study, intra-subject and inter-subject variability were investigated for each gait variable calculated by the algorithm.

In biomechanical gait analysis, variability can be measured through the coefficient of variation (CV). The CV is the ratio between the standard deviation (SD) to sample mean ($\bar{X}$), typically given on a percentage basis (WINTER, 2009):

$$CV = \frac{SD}{\bar{X}} \times 100 \qquad (2)$$

These measures of gait parameters variability can help identify instabilities during walking and as consequence predict falls (HAUSDORFF et al., 2003), and can be used to compare normally distributed data with respect to their variability (OSPINA; MARMOLEJO-RAMOS, 2019; WINTER, 2009). As suggested by some authors

(WHITE et al., 1999) data variations (CV) below 12.5% can be accepted as a normal level of variability for values related to gait.

To determine the trial-to-trial variability of the parameters, the volunteers underwent at least 5 valid gait trials, so that measures of left and right limbs were used to calculate the mean for each trial of each subject.

The variability between subjects was measured for the same parameters as for the within-subject variability, however, the data for the analysis were from all thirty-three older adults, mean was calculated for all right and left steps and for all valid gait trials.

4.7 Comparison with literature

Parametric statistics were used when the variables to be compared (GRF) followed a normal distribution ($p > 0.05$, Shapiro-Wilk) and the variations were homogeneous (Levene's test, $p > 0.05$). The parametric t tests compared the means of the forces calculated by our methodology with other studies, carried out in a similar population.

5 RESULTS AND DISCUSSION

5.1 General results

The methodology developed enabled the collection of several parameters regarding gait biomechanics such as force-time characteristics, peaks of vertical and anterior-posterior ground reaction force components, spatiotemporal parameters and impulses. It can be used by researchers, physiotherapists, clinicians or others interested in human movement as a tool to assess interventions, investigate the patient's walking pattern, diagnose gait disorders, compare different populations, among other applications.

The protocol developed consists in 5 gait trials in a preferred speed, where the individual is instructed to walk naturally on the force platforms. The choice of this number of trials is highly related to the number of platforms used in the methodology. When other studies are analysed, the number of tests performed varies considerably, even if a similar population is assessed (VAN KOOTEN et al., 2018).

Having 8 force platforms to capture the signal of ground reaction force, allows that during a test the volunteer performs one complete gait cycle. In order that the captured signal can be processed, the evaluator needs to ensure that the volunteer does not step with both feet on the same platform, otherwise the trial must be repeated. One of the most important instructions and precaution that must be reinforced is to make sure the subjects walk without attempting to place the foot on a specific place, avoiding stepping at the platforms junctions, as it is common to note that some participants are concerned with the performance of the test.

Another point that needs to be considered is the distractions of the research environment. The tests carried out in this protocol were for the assessment of gait during walking, without a dual-task, so it was important to ensure that the elderly would not talk or perform any other cognitive task while walking on the platforms.

However, gait assessment could, at another time, be performed during the double task using the same methodology, with attention to the intrinsic characteristics of the double task situation during the walk.

When correct measures are taken, the force signal captured by the platforms can be processed using the algorithm developed to quantify the variables of interest. The start of the signal processing happens by choosing the files containing the raw data provided by the software used for data capture. The algorithm developed for this methodology is capable of using folders, containing the raw data, for one or more tests necessary for the evaluation. The routine developed to identify the parameters of the force curves begins with the identification of the participant. The evaluator manually informs the participant's name and mass.

The results of the data processing are two useful documents for gait analysis. One file (.xls) containing the values of all parameters previously defined (Figure 13), divided into left and right sides, for each test performed by the volunteer, as well as the mean and the standard deviation for all steps. The other output of the system is the force-time curves of all steps performed, which allows verification of the shape of the curves (Figure 14 and 15).

Figure 13 - GRF parameters for a participant from all trials performed

	A	B	C	D	E	F	G	H	I	J	K	L	M	N	O	P	Q	R	S	T
1	tc	F1	t1	F3	t3	F2	t2	F4	t4	t6	F5	t5	t7	I1	I2	I3	I4	I5	I6	
2	0,752	1,425	0,177	1,557	0,536	1,258	0,291	0,251	0,613	0,208	0,108	0,108	0,644	0,33	0,538	0,868	0,011	0,014	0,026	
3	0,723	1,452	0,229	1,551	0,53	1,353	0,361	0,111	0,618	0,111	0,111	0,383	0,34	0,417	0,426	0,843	0,023	0,019	0,042	
4	0,683	1,437	0,167	1,559	0,515	1,311	0,39	0,237	0,591	0,204	0,091	0,397	0,286	0,481	0,34	0,82	0,045	0,035	0,08	
5	0,707	1,48	0,202	1,596	0,526	1,279	0,303	0,211	0,619	0,205	0,123	0,447	0,26	0,339	0,492	0,832	0,042	0,03	0,072	
6	0,691	1,463	0,211	1,515	0,493	1,304	0,359	0,238	0,605	0,16	0,12	0,415	0,276	0,403	0,394	0,797	0,034	0,035	0,07	
7	0,715	1,477	0,241	1,515	0,546	1,29	0,372	0,238	0,633	0,135	0,146	0,452	0,263	0,4	0,408	0,808	0,036	0,034	0,07	
8	0,709	1,512	0,224	1,561	0,556	1,238	0,383	0,272	0,621	0,185	0,123	0,404	0,305	0,427	0,387	0,814	0,037	0,042	0,079	
9	0,759	1,512	0,232	1,508	0,58	1,272	0,371	0,229	0,665	0,247	0,124	0,441	0,318	0,422	0,452	0,873	0,055	0,037	0,092	
10	0,692	1,489	0,219	1,461	0,536	1,294	0,367	0,27	0,62	0,183	0,12	0,409	0,283	0,416	0,378	0,793	0,038	0,039	0,077	
11	0,757	1,449	0,247	1,515	0,571	1,313	0,355	0,231	0,671	0,202	0,139	0,463	0,294	0,369	0,485	0,853	0,042	0,036	0,078	
12																				
13	0,7188	1,4696	0,2149	1,5338	0,5389	1,2912	0,3552	0,2288	0,6256	0,184	0,1205	0,3919	0,3269	0,4004	0,43	0,8301	0,0363	0,0321	0,0686	
14	0,02832	0,02975	0,02628	0,03816	0,02583	0,03216	0,03257	0,04533	0,02495	0,03947	0,0154	0,10317	0,11407	0,04461	0,06098	0,02847	0,01209	0,00888	0,01967	
15																				
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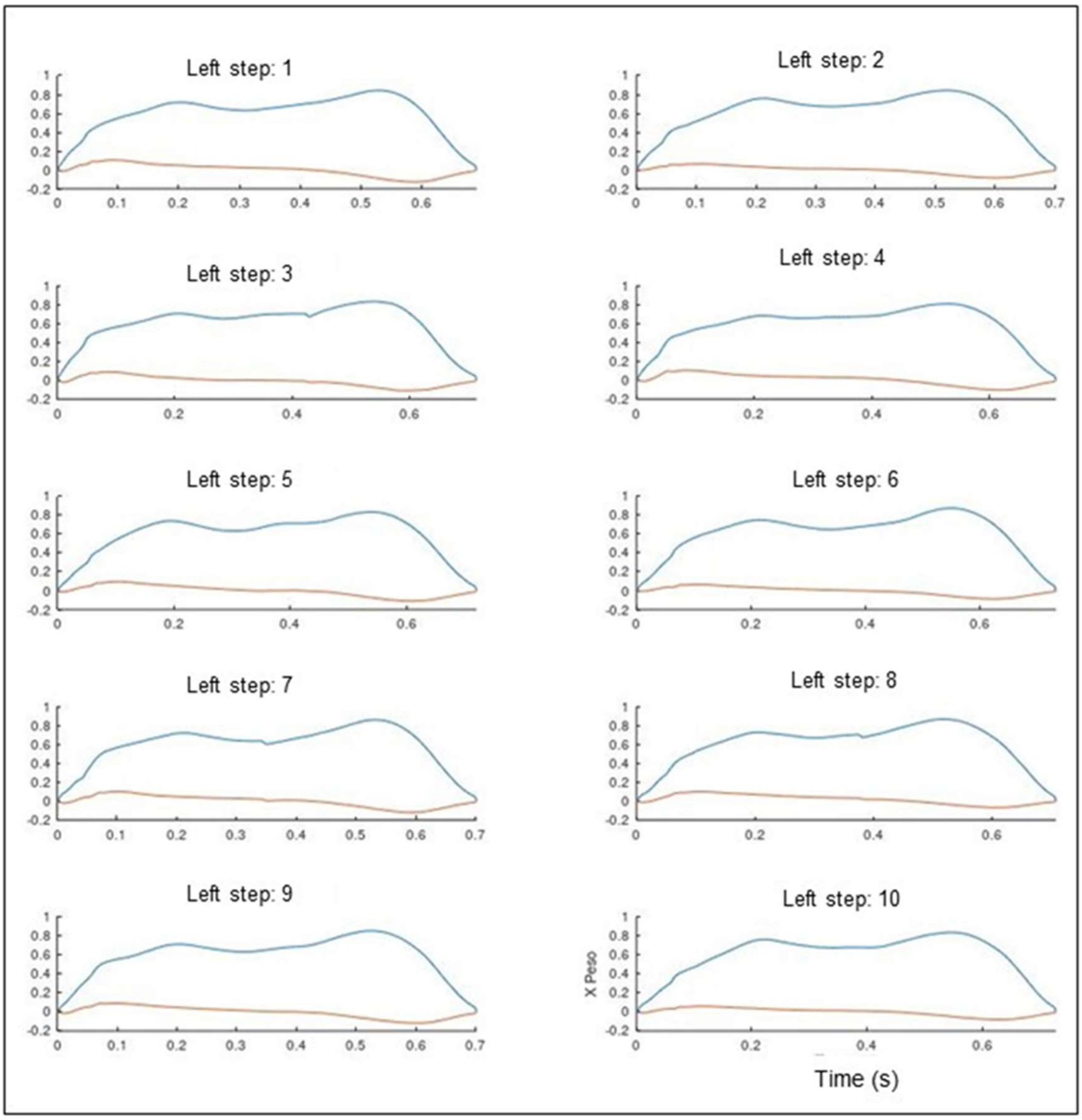
left right

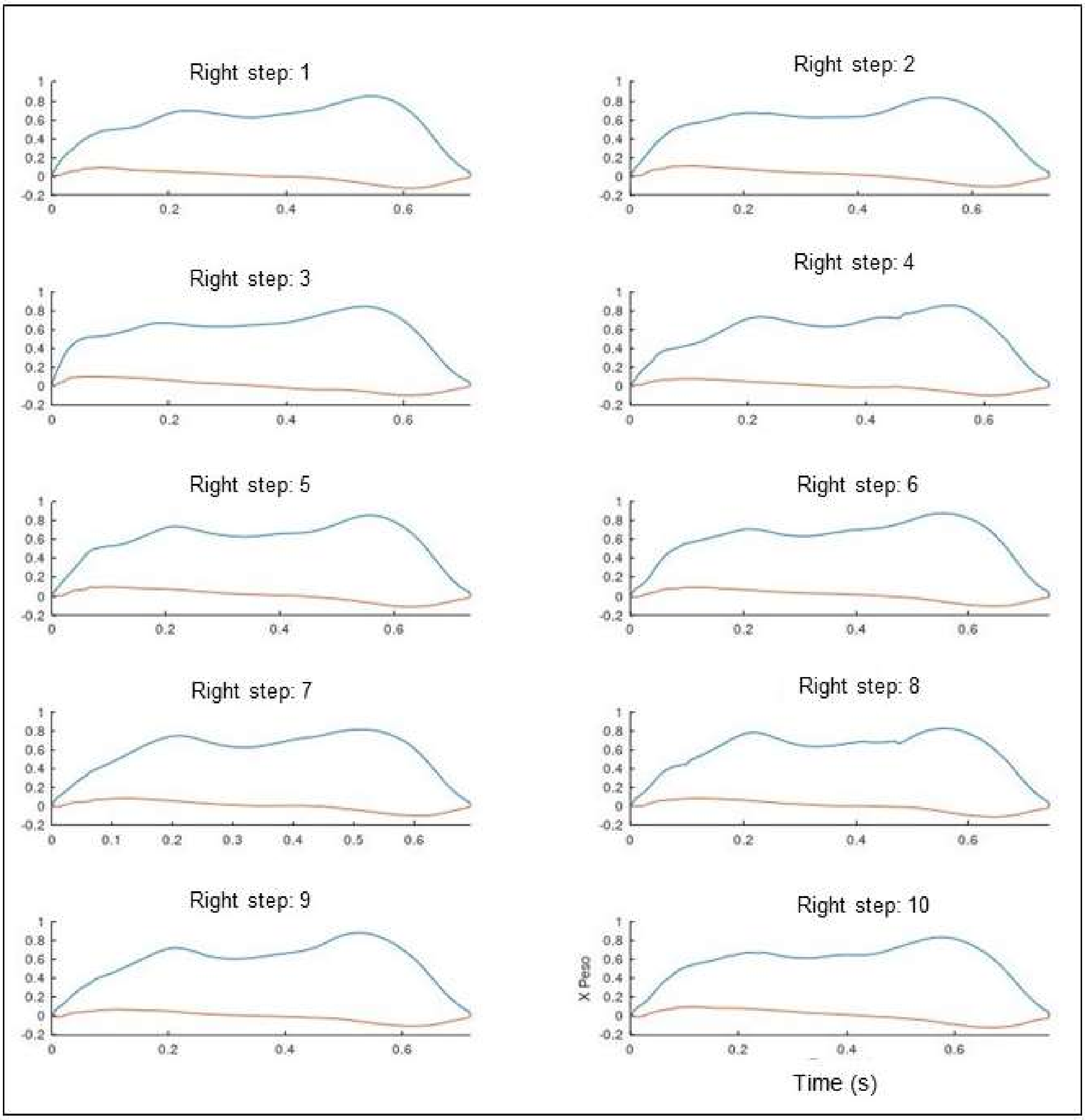
A print screen of the file provided by the algorithm with all the parameter values measured for each subject.
Note: the document presents the parameters for left and right limbs.

Figure 14 - Representation of curves for the left foot from all trials processed by the algorithm



Note: the blue line represents the vertical GRF and the orange line represents the anterior-posterior GRF. The vertical axis (Force) is normalized by body weight (N/kg) and horizontal axis (time) is in absolute value (s).

Figure 15 - Representation of curves from the right foot from all trials processed by the algorithm



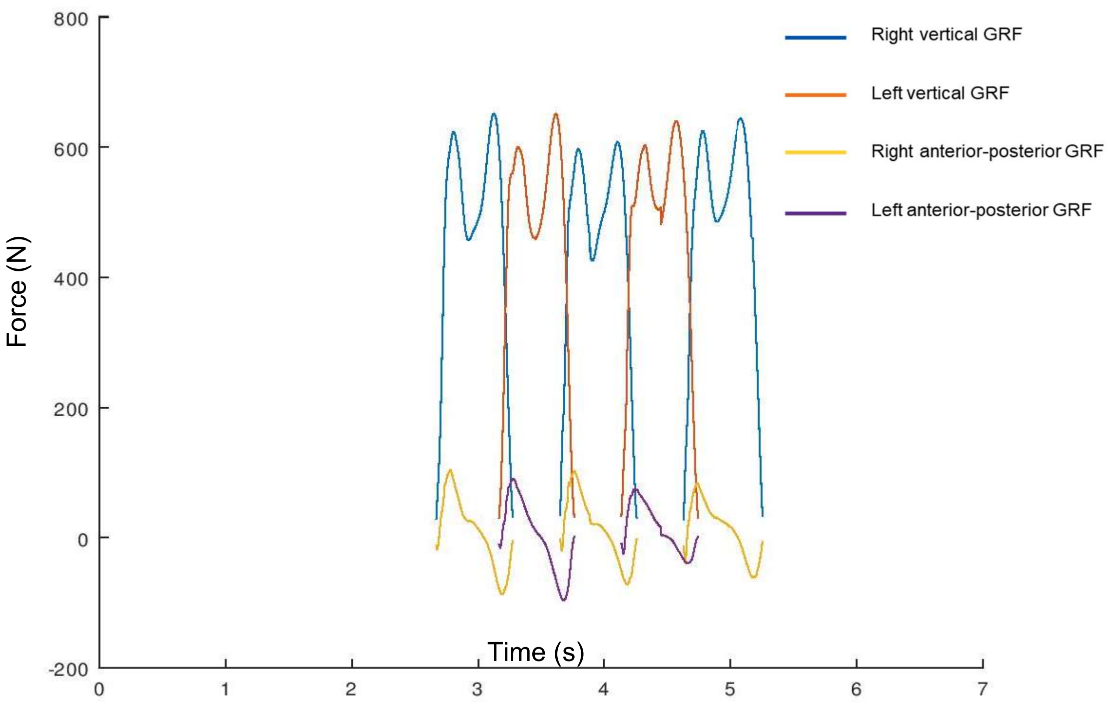
Note: the blue line represents the vertical GRF and the orange line represents the anterior-posterior GRF. The vertical axis (Force) is normalized by body weight (N/kg) and horizontal axis (time) is in absolute value (s).

Our results show that there is a wide variety of strategies for analysing gait in different situations. The choice of the appropriate approach, as well as the usefulness of the parameters addressed in the investigation of the research problem are crucial factors for a reliable assessment. For this reason, we investigated what was the importance of each parameter related to the GRF in the analysis of older adults' gait.

5.2 GRF curves

Typical GRF signal is presented in Figure 16 plotted against time for a individual walking at a self-selected speed, where vertical and anterior-posterior components from left and right limbs are illustrated during a trial. The vertical component of GRF (Fz) is the largest component of resultant force measured by the force plate. During this test, the participant performed 3 steps with the right foot, and 2 steps with the left foot.

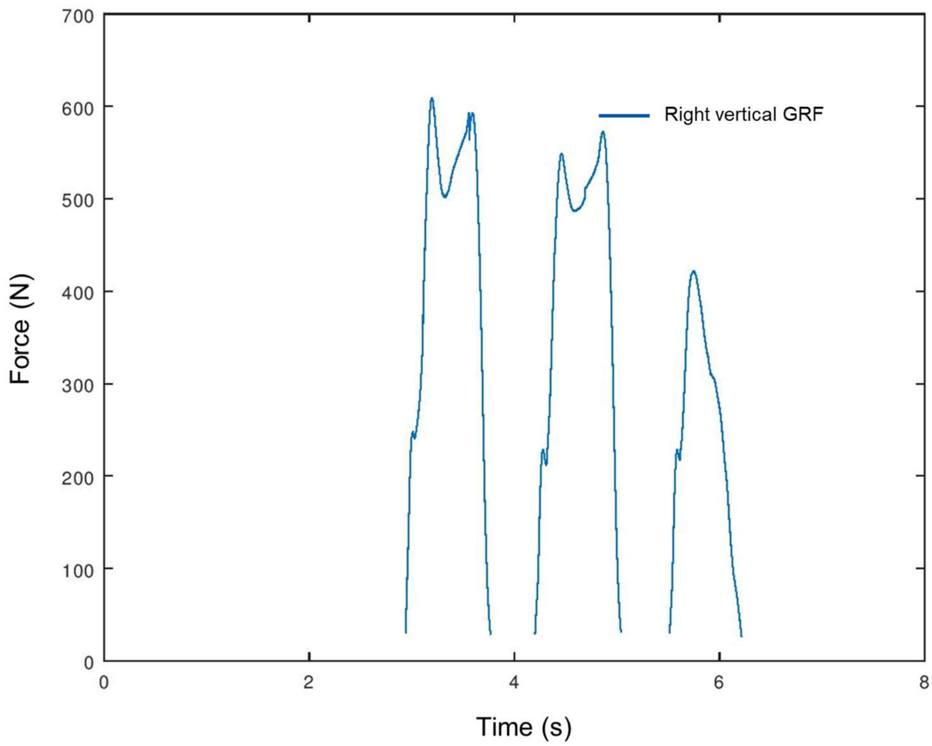
Figure 16 – Ground reaction forces walking curves



Qualitative analysis of the force curve may be a useful and simple tool and can be used for complementing biomechanical gait analysis. The classification of the vertical GRF was assessed for investigating if the changes in the “M” shape can be related to functional performance in older females (TAKAHASHI et al., 2004). The authors found that the shape of the vertical GRF was significantly correlated with gait parameters and functional performance, reinforcing the importance of interpreting force curves.

There are different factors that can change the pattern of the force curve, some related to diseases, others related to the way the participant walks (FUKUCHI; FUKUCHI; DUARTE, 2019), and the way foot touches the force platform (BEGG; RAHMAN, 2000). Analysing the vertical curve for the right limb of a volunteer in Figure 17, it is possible to notice that in the last step, the individual probably did not place the entire foot on the platform, because the "M" shape is not complete. To obtain accurate data, it is essential that the foot completely contacts the plate during the signal recording.

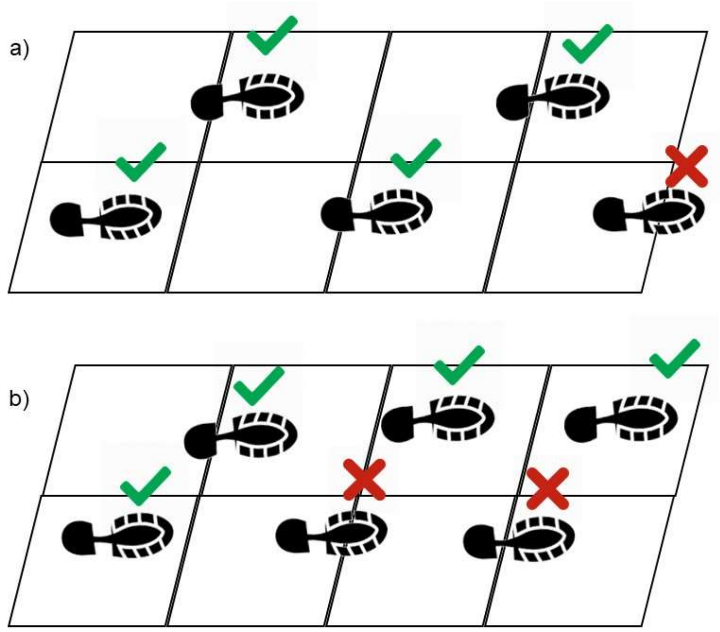
Figure 17 - Vertical force signal



Note: Ground Reaction Forces (GRF) curves obtained during a walk test of 2 correct steps and 1 incomplete step.

During the gait analysis, the situations illustrated in figure 18 are quite common, and occurs when the participant does not place his foot entirely on the platform (a) or when participants perform simultaneous contact with both feet in the same platform (b). However, when the quantitative analysis of this signal is performed, and the points of interest need to be identified, the incomplete step cannot be incorporated into the analysis.

Figure 18 - Correct and wrong footsteps during a trial



a) last step was not completely taken on the force platform, b) the last two right steps were taken on the same platform.

To solve this problem, the algorithm does not use the signal from incomplete steps. This happens through the classification of the curves, where during the processing of the vertical force signal the two force peaks and the valley between them are not identified, thus this step is discarded from the analysis. It is important to note that these failures in data capture do not drastically affect the collection performed since in this study we have 8 force platforms, which guarantee the collection of the necessary number of steps to have at least one essay. Moreover, even with fewer platforms, it is possible to carry out mathematical procedures to deal with this situation (BEGG; RAHMAN, 2000).

5.3 Gait analysis parameters

All parameters were averaged over all trials performed for each foot for each volunteer. The data from the GRF digital signal processed of the older adults' dataset is shown below (Table 3) as mean ± standard deviation (SD).

Table 3 - Gait parameters of the vertical GRF of the current data set

Parameter	L	R	Mean
F1 (N/kg) – f*irst vertical peak*	0,99 ± 0.11	0.98 ± 0.10	0.98 ± 0.09
F2 (N/kg) – v*ertical valley*	0.82± 0.11	0.81 ± 0.13	0.81 ± 0.10
F3 (N/kg) – s*econd vertical peak*	1.02 ± 0.09	1.03 ± 0.09	1.03 ± 0.08
tc (s) – *time of foot contact*	0.73 ± 0.09	0.73 ±0.08	0.73 ± 0.08
t1 (s) – *time to F1*	0.21 ± 0.05	0.21 ± 0.05	0.21 ± 0.05
t2 (s) – *time to F2*	0.36 ± 0.06	0.36 ± 0.07	0.36 ± 0.07
t3 (s) – *time to F3*	0.54 ± 0.05	0.55 ± 0.06	0.55 ± 0.06
I1 (N.s/kg) – i*mpulse of load response and midstance*	0.26 ± 0.04	0.26 ± 0.04	0.26 ± 0.03
I2 (N.s/kg) –i*mpulse of terminal stance and preswing*	0.29 ± 0.04	0.29 ± 0.04	0.29 ± 0.04
I3 (N.s/kg) – t*otal impulse of vertical GRF*	0.55 ± 0.07	0.55 ± 0.06	0.55 ± 0.06

Note: data presented as mean ± standard deviation. Forces and impulses were normalized by body weight; L: values for left limb; R: values for right limb; Mean: between right and left limbs.

Table 4 - Gait parameters of the anterior-posterior GRF of the current data set

Parameter	L	R	Mean (L+R)
F4 (N/kg) – *braking peak*	0.11 ± 0.03	0.13 ± 0.04	0.12 ± 0.03
F5 (N/kg) – propulsive *peak*	0.13 ± 0.03	0.13 ± 0.03	0.13 ± 0.03
t4 (s) – *time to F4*	0.12 ± 0.02	0.13 ± 0.03	0.12 ± 0.03
t5 (s) – *duration of braking phase*	0.62 ± 0.06	0.63 ± 0.07	0.62 ± 0.07
t6 (s) – *time to F5*	0.38 ±0.05	0.41 ± 0.06	0.39 ± 0.05
t7 (s) – *duration of propulsive phase*	0.35 ± 0.06	0.33 ± 0.05	0.34 ± 0.05
I4 (N.s/kg) – *braking impulse*	0.02 ± 0.00	0.02 ± 0.01	0.02 ± 0.00
I5 (N.s/kg) – *propulsive impulse*	0.02 ± 0.01	0.02 ± 0.01	0.02 ± 0.00
I6 (N.s/kg) – *Total impulse of anterior-posterior GRF*	0.04 ± 0.01	0.04 ± 0.01	0.04 ± 0.01

Note: data presented as mean ± standard deviation. Forces and impulses were normalized by body weight; L: values for left limb; R: values for right limb; L+R: mean between right and left limbs.

5.4 Statistical Analysis

Normality test showed that GRF peaks for vertical and anterior-posterior curves were normally distributed. Temporal variables showed non-normal distribution for t2 and t4 and normal distribution for tc, t1, t3, t5, t6 and t7.

Table 5 - Normality test for vertical and anterior-posterior GRF parameters

	Shapiro-Wilk		
	Statistics	**df**	**Sig.**
F1 (N/kg) – f*irst vertical peak*	0.957	33	0.212
F2 (N/kg) – v*ertical valley*	0.955	33	0.192
F3 (N/kg) – s*econd vertical peak*	0.969	33	0.466
F4 (N/kg) – *braking peak*	0.967	33	0.406
F5 (N/kg) – propulsive *peak*	0.984	33	0.884
tc (s) – *time of foot contact*	0.969	33	0.453
t1 (s) – *time to F1*	0.980	33	0.777
t2 (s) – *time to F2*	0.857	33	0.000*
t3 (s) – *time to F3*	0.960	33	0.265
t4 (s) – *time to F4*	0.932	33	0.041*
t5 (s) – *duration of braking phase*	0.966	33	0.374
t6 (s) – *time to F5*	0.979	33	0.766
t7 (s) – *duration of propulsive phase*	0.939	33	0.065

df: degrees of freedom; Sig.: significance (p value).
* $p < 0.05$

5.4.1 Variability Results

The mean and SD values for the following measured parameters were used to calculate variability: force in Newton (N/kg - expressed relative to body mass), time in seconds (s) and impulse in N.s/kg (expressed relative to body weight), the CV was expressed as a percentage.

5.4.1.1 Variability of force

Since the coefficients of variation for force present a non-normal distribution (Table 6), non-parametric test (Friedman test) were applied to compare parameters of variability of forces.

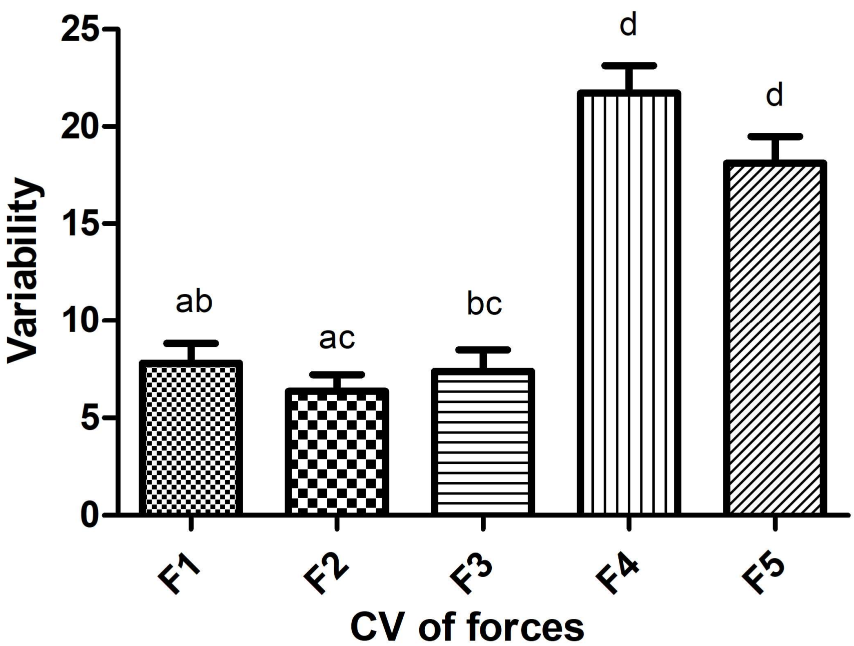
Table 6 - Normality test for variability (CV%) of forces

Coefficient of variation (CV)			
	Shapiro-Wilk		
	Statistics	df	Sig.
CV_{F1}	0.801	33	0.000*
CV_{F2}	0.849	33	0.000*
CV_{F3}	0.747	33	0.000*
CV_{F4}	0.944	33	0.091
CV_{F5}	0.910	33	0.010*

df: degrees of freedom; Sig.: significance (p value).
* *p < 0.05*

Intra-subjects' variability of the force parameters is shown in Figure 19, for vertical GRF parameters (F1, F2, F3) and for anterior-posterior GRF parameters (F4, F5). The CV was calculated using mean and SD from the trials performed by each volunteer, as described before (Equation 2).

Figure 19 - Intra-subjects comparison of CVs of the force peaks



Note: Columns with the same letters indicate no statistical difference between the CVs ($p < 0.05$).

Variability for the first vertical peak (F1) ranges from 1.75% to 28.60%, for vertical valley (F2) from 0.98% to 19.31%, for the second vertical peak (F3) from 1.46% to 29.42%, for the first anterior-posterior peak (F4) from 2.30% to 44.21% and for the second anterior-posterior peak (F5) from 1.73% to 44.35%.

Vertical and anterior-posterior forces are related to the mass-acceleration product and represent the gravitational forces acting on body and this range of values in force data can be explained by different sources, related to external environments such as instrumentation, methodologies and evaluators. In the same way as factors related to internal environments, such as temporal dynamics of neuromotor control, musculoskeletal pathologies, ageing, also could be associated with variability in gait data (CHAU; YOUNG; REDEKOP, 2005).

The intersubject variability shows how the parameters oscillate among the participants. The results can be used to assess a specific population. The mean values, SD and CV (%) of vertical GRF (F1, F2, F3) and anterior-posterior GRF (F4, F5) are summarized in a Table 7.

Table 7 - Intersubjects variability of GRF

	Mean	SD	CV %
F1 (N/kg) – f*irst vertical peak*	0.98	0.09	9.27
F2 (N/kg) – v*ertical valley*	0.81	0.10	12.49
F3 (N/kg) – s*econd vertical peak*	1.03	0.08	7.72
F4 (N/kg) – *braking peak*	0.12	0.03	29.01
F5 (N/kg) – propulsive *peak*	0.13	0.03	24.93

Mean values of the current data set of right and left limbs. Forces were normalized by body weight.

Like intra-variability results, table 7 shows that the variations were greater in anterior-posterior force than in vertical forces. This result is similar to that of previous studies (MASANI; KOUZAKI; FUKUNAGA, 2002; WINTER, 1984), which revealed differences between force directions. Mathematically this was expected, since the magnitude of the antero-posterior force is less than the vertical, generating a lower average of values, consequently increasing CV.

Winter (1984), compared the gait variability of 16 participants walking at different speeds, included their natural speed, calculating the CV for forces and moments. The author found that the variability of anterior-posterior forces was greater (20%) than the variability presented by vertical forces (7%), for natural velocity.

Other study (MASANI; KOUZAKI; FUKUNAGA, 2002) investigated whether the variation in walking speed influenced CVs for 10 male individuals and reported that gait variability has different values for certain speeds. Overall, the results showed similarity with the current study, lower CVs for vertical parameters, in comparison to anterior-posterior parameters, of all gait speeds tested.

5.4.1.2 Variability of temporal parameters

When variability of temporal parameters is analysed, the differences between parameters related to vertical forces (CV_{t1}, CV_{t2}, CV_{t3}) and anterior-posterior forces (CV_{t4}, CV_{t5}, CV_{t6}, CV_{t7}) are not easily detectable. The CVs for temporal parameters present a non-normal distribution (Table 8) for some variables, because of this non-parametric test was applied to compare the variability of temporal parameters.

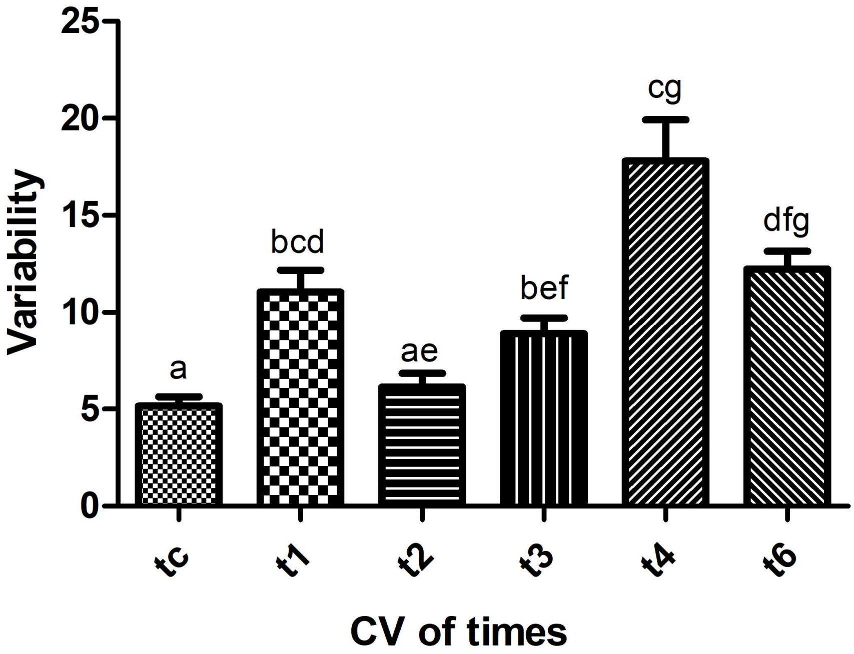
Table 8 - Normality test for variability (CV%) of time

Coefficient of variation (CV)			
	Shapiro-Wilk		
	Statistics	df	Sig.
CV_{tc}	0.918	33	0.016*
CV_{t1}	0.890	33	0.003*
CV_{t2}	0.853	33	0.000*
CV_{t3}	0.949	33	0.126
CV_{t4}	0.899	33	0.005*
CV_{t5}	0.848	33	0.000*
CV_{t6}	0.939	33	0.062
CV_{t7}	0.949	33	0.124

df: degrees of freedom; Sig.: significance (p value).
* *$p < 0.05$*

The intra-subject variability of the temporal parameters can be observed in Figure 20, which is related to time parameters behaviour for the same individual for all gait trials performed.

Figure 20 - Intra-subjects comparison of CVs of the temporal parameters



Note: Columns with the same letters indicate no statistical difference between the CVs ($p < 0.05$).

Table 9 shows how the variability of time parameters behave for all trials of all participants.

Table 9 - Intersubjects variability of temporal parameters

Temporal parameters (s)	Mean (n=33)	SD	CV %
tc (s) – *time of foot contact*	0.73	0.08	10.91
t1 (s) – *time to F1*	0.21	0.05	22.22
t2 (s) – *time to F2*	0.36	0.07	18.26
t3 (s) – *time to F3*	0.54	0.05	9.67
t4 (s) – *time to F4*	0.12	0.03	20.81
t5 (s) – *duration of braking phase*	0.63	0.07	10.49
t6 (s) – *time to F5*	0.39	0.05	12.25
t7 (s) – *duration of propulsive phase*	0.34	0.05	14.59

Mean values of the current data set of right and left limbs.

Temporal parameters of gait variability were investigated by other researches suggesting that the coefficient of variation (CV) for time and width stride could be highly related to the falls in older adults (HAUSDORFF, 2005; HAUSDORFF; RIOS; EDELBERG, 2001; LAROCHE; COOK; MACKALA, 2012b; OSPINA; MARMOLEJO-RAMOS, 2019; WINTER, 1984).

Time variability can also be related to the energy cost of walking since the alteration of stepping can change muscle activity and influence muscle adaptation. The compensation for alterations to the normal gait requires an increase in VO_2 to meet the additional energy for managing the muscle requirements, altering variability during the phases of gait (WERT et al., 2010).

5.4.1.3 Variability of impulse

As seen in Figure 21 and Table 10, the inter-subject and intra-subject impulse variability follow the same pattern of force variability, where CV's are higher for anterior-posterior parameters (I4, I5, I6) than for vertical parameters (I1, I2, I3).

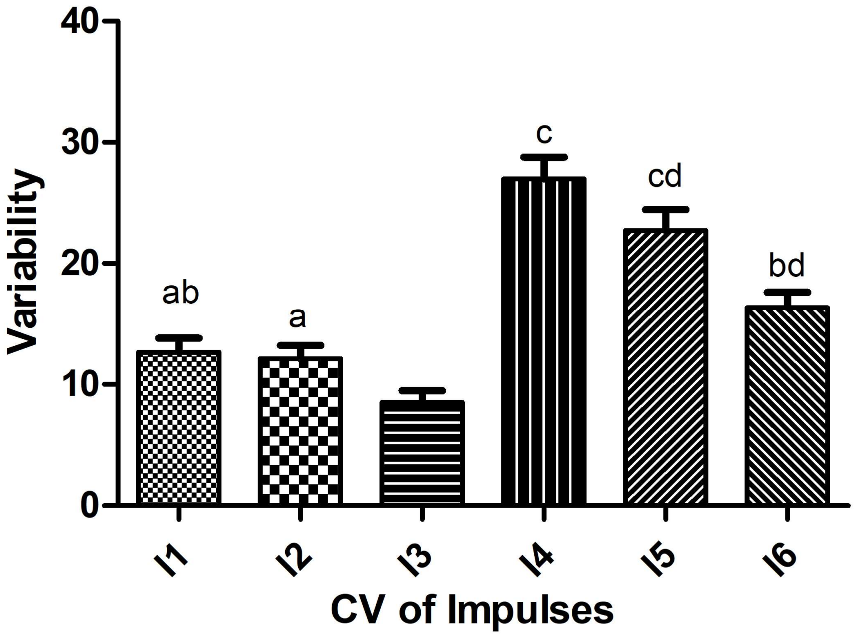
Table 10 - Normality test for variability (CV%) of impulse

Coefficient of variation (CV)

	Shapiro-Wilk		
	Statistics	df	Sig.
CV_{I1}	0,918	33	0.016*
CV_{I2}	0.916	33	0.014*
CV_{I3}	0.846	33	0.000*
CV_{I4}	0.961	33	0.282
CV_{I5}	0.950	33	0.137
CV_{I6}	0.975	33	0.635

df: degrees of freedom; Sig.: significance (p value).
* *p < 0.05*

Figure 21 - Intra-subjects comparison of CVs of the impulses



Note: Columns with the same letters indicate no statistical difference between the CVs ($p < 0.05$).

Table 11 - Intersubjects variability of impulses

Impulse	Mean (n=33)	SD	CV %
I1 (N.s/kg) – i*mpulse of load response and midstance*	0.26	0.03	13.16
I2 (N.s/kg) –i*mpulse of terminal stance and preswing*	0.29	0.04	13.23
I3 (N.s/kg) – t*otal impulse of vertical GRF*	0.55	0.06	10.77
I4 (N.s/kg) – *braking impulse*	0.02	0.00	22.53
I5 (N.s/kg) – *propulsive impulse*	0.02	0.00	22.67
I6 (N.s/kg) – *Total impulse of anterior-posterior GRF*	0.04	0.01	20.81

Note: Mean values of the current data set of right and left limbs.

The CV can be used for many purposes, in different situations and populations. Some researchers have studied the variability of the GRF for transtibial amputees (SVOBODA et al., 2012).

Studies relate gait variability to instability and risk of fall based on the decreased ability of the central neuromuscular system to regulate and support gait patterns (HAUSDORFF, 2005; HAUSDORFF; RIOS; EDELBERG, 2001; WHITE et al., 1999).

Variability has a widely application on gait analysis, for varies purposes on different populations. It can be used to check repeatability or reproducibility, to predict falls in older adults, patients with Parkinson's Disease, children with Cerebral Palsy and gait amputees (SVOBODA et al., 2012), for validate gait analysis devices (LIU; INOUE; SHIBATA, 2010).

5.5 Literature comparison

Some of the gait parameters studied in our methodology were compared with other works. The purpose of the comparison was more than the verification of the force values, but also to analyse how other authors perform the quantification of the parameters and whether these values can be compared to our work.

Table 12 - Vertical force peak comparison

	Author	n	Mean	SD	p
F1	Current Study	33	0.98	0.09	-
	Larish et al. (1988)	13	1.00	0.03	0.28
	Anderson and Madigan (2013)	5	1.05	0.04	0*
	Kim and Kim (2017)	14	1.01	0.06	0.02
	Hitz et al. (2018)	9	1.14	0.06	0*
F2	Current Study	33	0.81	0.10	-
	Larish et al. (1988)	13	0.92	0.02	0*
	Hitz et al. (2018)	9	0.77	0.08	0*
F3	Current Study	33	1.03	0.07	-
	Larish et al. (1988)	13	1.02	0.02	0.56
	Anderson and Madigan (2013)	5	1.06	0.04	0.02
	Kim and Kim (2017)	14	1.03	0.07	0.89
	Hitz et al. (2018)	9	1.12	0.07	0*

*$p < 0.05$ indicates difference between our methodology and the literature.

F1: first vertical peak; F2: vertical valley; F3: second vertical peak

The force peaks values found in the literature were compared to the values calculated from our methodology in order to verify the similarity between the studies. The vertical forces comparison are shown in table 9. Regarding authors comparison, the values for the first vertical peak (F1) presented statistical difference between the results found by the studies investigated (ANDERSON; MADIGAN, 2013), (KIM; KIM, 2017) and (HITZ et al., 2018), and did not present difference when compared to (LARISH; MARTIN; MUNGIOLE, 1988).

These differences may arise due to the application of different methods of data collection, slower or faster gait, since the speed influences the GRF parameters, other laboratory environments, etc. This must be considered so that comparisons made by different studies can be considered reliable.

This might be explained by the fact that in some studies (HITZ et al., 2018) the number of the volunteers evaluated were very low (n=9) and their majority were males (n=6) against females (n=3), while participantss in the present study were all women (n=33). Another observation that may explain the statistical difference is the lower average age of the population studied by Hitz et al. (2018), (61.6 ± 5.3 years), compared to our study (70.45 ± 6.92 years).

The reduced number of participants is also a point to consider, since some studies (ANDERSON; MADIGAN, 2013), evaluated only five older adults (2 men and 3 women). The velocity adopted by Anderson and Madigan (2013), was a controlled speed of 1.1 m/s, which is also a different condition from our self-selected speed, 0.92 m/s.

6 CONCLUSION

The main purpose of this study was to elucidate and assist in the investigation of the main characteristics related to gait in the elderly, confirming that there is a high applicability of kinetic analysis in gait evaluation. This was possible through the development of a methodology for biomechanical evaluation of walking.

In this work, a new methodology for gait biomechanical assessment in the elderly was developed, based on GRF curves and their parameters of interest. The method was able to obtain discrete values for the variables that characterize the biomechanical pattern of walking, as well as showing the characteristic curves of the elderly population.

Specific computational tools assist in the analysis and interpretation of biomechanical data. The developed algorithm facilitates the processing of data by the evaluator, allowing users interested in clinical and scientific applications to have greater autonomy to perform diagnostics and evaluations, without the need for deep knowledge of computing or programming.

When gait analysis is studied, several factors must be considered. It is not just the age of the individual that influences the ground reaction forces patterns, but a network of decisions and situations, leading to the most appropriated methodology.

The proposed methodology is helping our laboratory to conduct several studies and is considered by our group a biomedical engineering solution, since combines engineering tools to solve health sciences problems.

REFERENCES

ALAM, M. N. et al. Vertical ground reaction force marker for Parkinson’s disease. **PLoS ONE**, v. 12, n. 5, 2017.

ANDERSON, D. E.; MADIGAN, M. L. Effects of age-related differences in femoral loading and bone mineral density on strains in the proximal femur during controlled walking. **Journal of Applied Biomechanics**, v. 29, n. 5, p. 505–516, 2013.

ANDRIACCHI, T. P.; ALEXANDER, E. J. Studies of human locomotion: Past, present and future. **Journal of Biomechanics**, v. 33, n. 10, p. 1217–1224, 2000.

ARENA, S. L. et al. Required friction during overground walking is lower among obese compared to non-obese older men, but does not differ with obesity among women. **Applied Ergonomics**, v. 62, p. 77–82, 2017.

BAKER, R. The history of gait analysis before the advent of modern computers. **Gait and Posture**, v. 26, n. 3, p. 331–342, 2007.

BAPTISTA, R. R.; VAZ, M. A. Arquitetura muscular e envelhecimento: adaptação funcional e aspectos clínicos; revisão da literatura. **Fisioterapia e Pesquisa**, v. 16, n. 4, p. 368–373, dez. 2009.

BEGG, R. K. Neural network-based prediction of missing key features in vertical GRF-time recordings. **Journal of Medical Engineering and Technology**, v. 30, n. 5, p. 315–322, 1 set. 2006.

BEGG, R. K.; RAHMAN, S. M. A method for the reconstruction of ground reaction force-time characteristics during gait from force platform recordings of simultaneous foot falls. **IEEE Transactions on Biomedical Engineering**, v. 47, n. 4, p. 547–551, 2000.

BEGG, R.; KAMRUZZAMAN, J. A machine learning approach for automated recognition of movement patterns using basic, kinetic and kinematic gait data. **Journal of Biomechanics**, v. 38, n. 3, p. 401–408, 2005.

BEGG, R.; KAMRUZZAMAN, J. Neural networks for detection and classification of walking pattern changes due to ageing. **Australasian Physical and Engineering Sciences in Medicine**, v. 29, n. 2, p. 188–195, 2006.

BIGGS, P. R. et al. Which osteoarthritic gait features recover following total knee replacement surgery? **PLoS ONE**, v. 14, n. 1, p. 1–14, 2019.

BOYER, K. A. et al. Systematic review and meta-analysis of gait mechanics in young and older adults. **Experimental Gerontology**, v. 95, p. 63–70, 2017.

CALDAS, R. et al. A systematic review of gait analysis methods based on inertial sensors and adaptive algorithms. **Gait & Posture**, v. 57, p. 204–210, set. 2017.

CHAU, T.; YOUNG, S.; REDEKOP, S. Managing variability in the summary and comparison of gait data. **Journal of NeuroEngineering and Rehabilitation**, v. 2, n. 1, 2005.

CHOCKALINGAM, N.; HEALY, A.; NEEDHAM, R. Interpreting Ground Reaction Forces in Gait. In: **Handbook of Human Motion**. [s.l.] Springer International Publishing, 2016. p. 609–620.

FAVARO, S. O. V. B. et al. Ground reaction force analysis of the young and elderly gait on the influence of the overweight simulation. **Journal of Physical Education (Maringa)**, v. 30, n. 1, p. 1–10, 2019.

FUKUCHI, C. A.; FUKUCHI, R. K.; DUARTE, M. A public dataset of overground and treadmill walking kinematics and kinetics in healthy individuals. **PeerJ**, v. 2018, n. 4, 2018.

FUKUCHI, C. A.; FUKUCHI, R. K.; DUARTE, M. Effects of walking speed on gait biomechanics in healthy participants: A systematic review and meta-analysis. **Systematic Reviews**, v. 8, n. 1, 2019.

GINNERUP-NIELSEN, E. et al. Improved gait in persons with knee related mobility limitations by a rosehip food supplement: A randomized, double-blind, placebo-controlled trial. **Gait & Posture**, v. 42, n. 3, p. 340–347, 2015.

GOODPASTER, B. H. et al. The loss of skeletal muscle strength, mass, and quality in older adults: The Health, Aging and Body Composition Study. **Journals of Gerontology - Series A Biological Sciences and Medical Sciences**, v. 61, n. 10, p. 1059–1064, 2006.

HALL, S. Basic Biomechanics. **McGraw-Hill Higher Education.**, v. 53, n. 9, p. 630, 2014.

HAUSDORFF, J. M. et al. Impaired regulation of stride variability in Parkinson's disease subjects with freezing of gait. **Experimental Brain Research**, v. 149, n. 2, p. 187–194, 2003.

HAUSDORFF, J. M. Gait variability : methods , modeling and meaning Example of Increased Stride Time Variability in Elderly Fallers Quantification of Stride-to-Stride Fluctuations. **Journal of neuroengineering and rehabilitation**, v. 2, n. 1, p. 19, 2005.

HAUSDORFF, J. M.; RIOS, D. A.; EDELBERG, H. K. Gait variability and fall risk in community-living older adults: A 1-year prospective study. **Archives of**

Physical Medicine and Rehabilitation, v. 82, n. 8, p. 1050–1056, 2001.

HAZARD MUNRO, B. et al. **Statistical Methods for Health Care Research FIFTH EDITION • A Wollen Kluwer Company**. [s.l: s.n.].

HITZ, M. et al. Influence of the moving fluoroscope on gait patterns. **PLoS ONE**, v. 13, n. 7, p. 1–16, 2018.

HSU, W.-C. et al. Immediate and long-term efficacy of laterally-wedged insoles on persons with bilateral medial knee osteoarthritis during walking. **BioMedical Engineering Online**, v. 14, n. 1, 2015.

JONES, G. G. et al. Gait comparison of unicompartmental and total knee arthroplasties with healthy controls. **Bone and Joint Journal**, v. 98-B, n. 10, p. 16–21, 2016.

JUDGE, J. Decade of Healthy Ageing 2020-2030. In: [s.l: s.n.].

KIM, W. S.; KIM, M. J. Individual joint contribution to body weight support in the affected lower limb during walking in post-stroke hemiplegia. **Topics in Stroke Rehabilitation**, v. 24, n. 3, p. 170–176, 2017.

KIRTLEY, C. **Clinical Gait Analysis**. [s.l.] Elsevier Ltd, 2006.

KNUDSON, D. V. **Fundamentals of biomechanics**. 2nd. ed. New York: Springer, 2007.

LARISH, D. D.; MARTIN, P. E.; MUNGIOLE, M. Characteristic Patterns of Gait in the Healthy Old. **Annals of the New York Academy of Sciences**, v. 515, n. 1, p. 18–32, 1988.

LAROCHE, D. P. et al. Augmenting strength-to-weight ratio by body weight unloading affects walking performance equally in obese and nonobese older adults. **Age**, v. 38, n. 1, p. 1–14, 2016.

LAROCHE, D. P.; COOK, S. B.; MACKALA, K. Strength asymmetry increases gait asymmetry and variability in older women. **Medicine and Science in Sports and Exercise**, v. 44, n. 11, p. 2172–2181, 2012a.

LAROCHE, D. P.; COOK, S. B.; MACKALA, K. Strength Asymmetry Increases Gait Asymmetry and Variability in Older Women. **Med Sci Sports Exerc**, v. 44, n. 11, p. 2172–2181, 2012b.

LIEBER, R. L.; FRIDÉN, J. **Clinical significance of skeletal muscle architecture**. Clinical Orthopaedics and Related Research. **Anais**...Lippincott Williams and Wilkins, 2001

LIPPERT, L. S. **Clinical Kinesiology and Anatomy**. 5th. ed. [s.l.] F. A. Davis Company, 2011.

LIU, T.; INOUE, Y.; SHIBATA, K. A Wearable Ground Reaction Force Sensor System and Its Application to the Measurement of Extrinsic Gait Variability. **Sensors**, v. 10, n. 11, p. 10240–10255, 16 nov. 2010.

MARTÍNEZ-RAMÍREZ, A. et al. Pre-operative ambulatory measurement of asymmetric lower limb loading during walking in total hip arthroplasty patients. **Journal of NeuroEngineering and Rehabilitation**, v. 10, n. 1, 2013.

MASANI, K.; KOUZAKI, M.; FUKUNAGA, T. Variability of ground reaction forces during treadmill walking. **Journal of Applied Physiology**, v. 92, n. 5, p. 1885–1890, 2002.

MENZ, H. B.; LORD, S. R.; FITZPATRICK, R. C. Age-related differences in walking stability. **Age and Ageing**, v. 32, n. 2, p. 137–142, mar. 2003.

MÜNDERMANN, L.; CORAZZA, S.; ANDRIACCHI, T. P. The evolution of methods for the capture of human movement leading to markerless motion capture for biomechanical applications. **Journal of NeuroEngineering and Rehabilitation**, v. 3, p. 1–11, 2006.

MURO-DE-LA-HERRAN, A.; GARCÍA-ZAPIRAIN, B.; MÉNDEZ-ZORRILLA, A. Gait analysis methods: An overview of wearable and non-wearable systems, highlighting clinical applications. **Sensors (Switzerland)**, v. 14, n. 2, p. 3362–3394, 2014.

OH-PARK, M.; WANG, C.; VERGHESE, J. Stair negotiation time in community-dwelling older adults: normative values and association with functional decline. **Archives of physical medicine and rehabilitation**, v. 92, n. 12, p. 2006–11, dez. 2011.

OSPINA, R.; MARMOLEJO-RAMOS, F. Performance of Some Estimators of Relative Variability. **Frontiers in Applied Mathematics and Statistics**, v. 5, n. August, 2019.

PARVATANENI, K. et al. Kinematic, kinetic and metabolic parameters of treadmill versus overground walking in healthy older adults. **Clinical Biomechanics**, v. 24, n. 1, p. 95–100, 2009.

PATERSON, K. L. et al. The influence of sex and obesity on gait biomechanics in people with severe knee osteoarthritis scheduled for arthroplasty. **Clinical Biomechanics**, v. 49, n. July 2016, p. 72–77, 2017.

PENN, I. W. et al. Speed and temporal adaptations during nonmotorized treadmill walking in Parkinson disease and nondisabled individuals.

International journal of rehabilitation research, v. 42, n. 2, p. 126–132, 2019.

PERRY, J. **Gait Analysis: Normal and Pathological Function**. 1st. ed. Thorofare: SLACK Incorporated, 1992. v. 159

PÖTZELSBERGER, B. et al. Alpine Skiing With total knee ArthroPlasty (ASWAP): Effects on gait asymmetries. **Scandinavian Journal of Medicine and Science in Sports**, v. 25, n. S2, p. 49–59, 2015.

PRINCE, F. et al. Gait in the elderly. **Gait and Posture**, v. 5, n. 2, p. 128–135, 1997.

ROBERTS, M.; MONGEON, D.; PRINCE, F. Biomechanical parameters for gait analysis: a systematic review of healthy human gait. 2017.

ROBERTSON DGE, CALDWELL GE, HAMIL J, KAMEN G, W. S. **Research Methods in Biomechanics**. [s.l: s.n.].

ROUHANI, H. et al. Ambulatory measurement of ankle kinetics for clinical applications. **Journal of Biomechanics**, v. 44, n. 15, p. 2712–2718, 2011.

RUTHERFORD, D. et al. The effect of age and knee osteoarthritis on muscle activation patterns and knee joint biomechanics during dual belt treadmill gait. **Journal of Electromyography and Kinesiology**, v. 34, p. 58–64, 2017.

SALEH, M. M. et al. Effects of Aging on Human Gait Stability. **2017 IEEE International Conference on Computational Intelligence and Computing Research, ICCIC 2017**, p. 1–5, 2018.

SHARMA, S.; MCMORLAND, A. J. C.; STINEAR, J. W. Stance limb ground reaction forces in high functioning stroke and healthy subjects during gait initiation. **Clinical Biomechanics**, v. 30, n. 7, p. 689–695, 2015.

SVOBODA, Z. et al. Variability of kinetic variables during gait in unilateral transtibial amputees. **Prosthetics and Orthotics International**, v. 36, n. 2, p. 225–230, 2012.

TAKAHASHI, T. et al. Vertical ground reaction force shape is associated with gait parameters, timed up and go, and functional reach in elderly females. **Journal of Rehabilitation Medicine**, v. 36, n. 1, p. 42–45, 2004.

TODA, H.; NAGANO, A.; LUO, Z. **Contribution of lower extremity joint moment on ground reaction force during walking in elderly and younger people**. 2013 IEEE/SICE International Symposium on System Integration, SII 2013. **Anais**...2013

VAN KOOTEN, D. et al. Are there associations with age and sex in walking stability in healthy older adults? **Gait and Posture**, v. 60, n. September 2016, p. 65–70, 2018.

VAUGHAN, C.; DAVIS, B.; O'CONNOR, J. C. **Dynamics of Human Gait : Human Kinetics**. 2nd. ed. Cape Town: Kiboho Publishers, 1999.

VAVERKA, F. et al. System of gait analysis based on ground reaction force assessment. **Acta Gymnica**, v. 45, n. 4, p. 187–193, 31 dez. 2015.

WANG, W. et al. Assessment of Gait Characteristics in Total Knee Arthroplasty Patients Using a Hierarchical Partial Least Squares Method. **IEEE Journal of Biomedical and Health Informatics**, v. 22, n. 1, p. 205–214, 2018.

WATELAIN, E. et al. Gait pattern classification of healthy elderly men based on biomechanical data. **Archives of Physical Medicine and Rehabilitation**, v. 81, n. 5, p. 579–586, maio 2000.

WERT, D. M. et al. Gait Biomechanics, Spatial and Temporal Characteristics, and the Energy Cost of Walking in Older Adults With Impaired Mobility. **Physical Therapy**, v. 90, n. 7, p. 977–985, 1 jul. 2010.

WHITE, R. et al. The variability of force platform data in normal and cerebral palsy gait. **Clinical Biomechanics**, v. 14, n. 3, p. 185–192, 1999.

WHITTLE, M. W. **Gait Analysis: an introduction**. 4th. ed. [s.l.] Butterworth-Heinemann, 2007.

WILKINSON, D. J.; PIASECKI, M.; ATHERTON, P. J. The age-related loss of skeletal muscle mass and function: Measurement and physiology of muscle fibre atrophy and muscle fibre loss in humans. **Ageing Research Reviews**, v. 47, p. 123–132, 1 nov. 2018.

WINTER, D. A. Kinematic and kinetic patterns in human gait: Variability and compensating effects. **Human Movement Science**, v. 3, n. 1–2, p. 51–76, 1984.

WINTER, D. A. **Biomechanics and Motor Control of Human Movement: Fourth Edition**. [s.l: s.n.].

WINTER, D. A.; PATLA, A. E.; FRANK, J. S. **Assessment of balance control in humansMedical Progress through Technology**, maio 1990.

YEO, J. H. et al. No difference in outcomes and gait analysis between mechanical and kinematic knee alignment methods using robotic total knee arthroplasty. **Knee Surgery, Sports Traumatology, Arthroscopy**, v. 27, n. 4, p. 1142–1147, 2019.

APPENDIX A - ALGORITHM

```
#GaitMaster
clear all
close all

pasta=uigetdir();
arquivos=dir(fullfile(pasta,'*.emt'));
letras=['a':'z'];
ensaioqtd=length(arquivos);

disp('Insira dados do paciente');
disp(' ');
arquivo=input('Nome: ','-ascii');
massakg=input('Peso(kg): ')*9.802;
disp(' ');
ctzero=time;
disp('Inicializando Ensaios');

for i=1:ensaioqtd
  predata=importdata(fullfile(pasta,arquivos(i).name),'\t',11);
  predata=predata.data;
  predata(end-1:end,:)=[];
  alldata.(letras(i))=predata;
endfor

disp(' ');
disp('Ensaios prontos');


lsteps=0;
rsteps=0;

for i=1:ensaioqtd
   [analise(i)]=GaitAnalizer(alldata.(letras(i)),massakg);
   [a,b]=size(analise(i).lvt);
   lsteps(i)=b;
   [a,b]=size(analise(i).rvt);
   rsteps(i)=b;
   disp(strcat('Ensaio: ',num2str(i),' Carregado'));
endfor

disp(' ');
disp(strcat(num2str(ensaioqtd),' ensaios carregados'));
disp(' ');
%%Plotando  gráficos=-=-=-=-=-=-=-=-=-=-=-=-=-=-=-=-=-=-=-=-=-=-=-=-=-=-=-=-=-=-
=-=-=

plotados=0; %% variável auxiliar
```

```
numdata=19; %% quantidade de dados plotados (tc, F1, t1..etc)

figure('units','Normalized','position',[0.1,0.08,0.3*ceil(sum(lsteps)/5),0.8]);
for i=1:ensaioqtd
  if isempty(analise(i).esqpts)==0;
    for k=1:lsteps(i)
        plotados=plotados+1;
        subplot(5,ceil(sum(lsteps)/5),plotados);
        hold on
        plot(analise(i).ltempo(:,k),analise(i).lvt(:,k));
        plot(analise(i).ltempo(:,k),analise(i).lap(:,k));
        grid on
        xlim([0 max(analise(i).ltempo(:,k))]);
        title(strcat('Passo Esquerdo:',num2str(plotados)));

        %%Separado pontos de interesse
        for j=1:numdata
          esqpts(plotados,j)=analise(i).esqpts(k,j);
        endfor
    endfor
   endif
endfor
xlabel('Tempo(s)');
ylabel('X Peso');
hold off

plotados=0;
figure('units','Normalized','position',[0.1,0.08,0.3*ceil(sum(rsteps)/5),0.8]);
for i=1:ensaioqtd
  if isempty(analise(i).dirpts)==0;
    for k=1:rsteps(i)
        %%plotando dados;
        plotados=plotados+1;
        subplot(5,ceil(sum(rsteps)/5),plotados);
        hold on
        plot(analise(i).rtempo(:,k),analise(i).rvt(:,k));
        plot(analise(i).rtempo(:,k),analise(i).rap(:,k));
        grid on
        xlim([0 max(analise(i).rtempo(:,k))]);
        title(strcat('Passo Direito:',num2str(plotados)));

        %%Separado pontos de interesse
        for j=1:numdata
          dirpts(plotados,j)=analise(i).dirpts(k,j);
        endfor
    endfor
   endif
endfor
xlabel('Tempo(s)');
ylabel('X Peso');
```

```
hold off

for i=1:numdata
  dirstd(i)=std(dirpts(:,i));
  dirmedia(i)=sum(dirpts(:,i))/length(dirpts(:,i));
  esqstd(i)=std(esqpts(:,i));
  esqmedia(i)=sum(esqpts(:,i))/length(esqpts(:,i));
endfor

disp(strcat('tempo computacional:',num2str(time-ctzero),' segundos'));
disp('-----------------');


%%Salvando EXCELL
cabeca=transpose(cellstr(['tc';'F1';'t1';'F3';'t3';'F2';'t2';'F4';'t4';'t6';'F5';'t5';'t7';'I1';'I2
';'I3';'I4';'I5';'I6']));
xlswrite(arquivo,cabeca,'Esquerdos','A1');
xlswrite(arquivo,cabeca,'Direitos','A1');
xlswrite(arquivo,esqpts,'Esquerdos','A2');
xlswrite(arquivo,dirpts,'Direitos','A2');

xlswrite(arquivo,[esqmedia;esqstd],'Esquerdos',strcat('A',num2str(sum(lsteps)+3
)));
xlswrite(arquivo,[dirmedia;dirstd],'Direitos',strcat('A',num2str(sum(rsteps)+3)));

disp('Programa Finalizado!');

#GaitAnalizer

function [outdata] = GaitAnalizer (matrix,massakg)
    [linha, coluna]=size(matrix);

    %%remove valores não-numéricos da matrix original
    for x=1:linha
       for y=1:coluna
          if(isnan(matrix(x,y)))
            matrix(x,y)=0;
          end
      end
    end

    %%Indice e tempo de registro
    frame=matrix(:,1);
    tempo=matrix(:,2);
    tempo=tempo(1:max(find(tempo)));

    %%medidas do pé direito
    rap=matrix(1:length(tempo),3)/massakg; %%anteroposterior
    rvt=matrix(1:length(tempo),4)/massakg; %%vertical
```

```
   %%medidas do pé esquerdo
   lap=matrix(1:length(tempo),6)/massakg; %%anteroposterior
   lvt=matrix(1:length(tempo),7)/massakg; %%vertical

   %%Controle de sensibilidade dos detectores de borda e eixos gráficos
   patamar=0.01;
   sensi=0.01;

   %%Tamanho do vetor padrão para análise de dados
   stdvec=100;

   %%Cria o passo padrão fazendo a média entre cada passo.=-=-=-=-=-=-=-=-
=-=-=-=-=-=-=-=-=-=-=-=-=-=-=-=-=
   rodando=1;
   lstepqtd=0;
   lvtstep=0;
   lapstep=0;

   %%Isola cada passo e cria um vetor padrão com as médias de cada um.
   %%Pé Esquerdo
   while rodando==1;
     if sum(lvt)!=0
       lstepqtd=lstepqtd+1;

ltempo(lstepqtd,:)=sizeadj(tempo(bordasub(lvt,patamar,sensi):bordadesc(lvt,pat
amar,sensi)),stdvec);
       ltempo(lstepqtd,:)=ltempo(lstepqtd,:)-ltempo(lstepqtd,1);

       %Antero-Posterior

lapmat(lstepqtd,:)=sizeadj(lap(bordasub(lvt,patamar,sensi):bordadesc(lvt,patam
ar,sensi)),stdvec);
       lap=lap(bordadesc(lvt,patamar,sensi):end);

       %Força Vertical

lvtmat(lstepqtd,:)=sizeadj(lvt(bordasub(lvt,patamar,sensi):bordadesc(lvt,patamar
,sensi)),stdvec);
       lvt=lvt(bordadesc(lvt,patamar,sensi):end);
     else
       rodando=0;
     endif
   endwhile

   %%Pé  Direito-=-=-=-=-=-=-=-=-=-=-=-=-=-=-=-=-=-=-=-=-=-=-=-=-=-=-=-=-=-=-=-=-=-
=-=-=-=-=-=-=-=-=-=-=-=-=-=-=-
   rodando=1;
   rstepqtd=0;
   rvtstep=0;
   rapstep=0;
```

```
 %%Reset do vetor de tempo.
  tempo=matrix(:,2);
  tempo=tempo(1:max(find(tempo)));

  while rodando==1;
    if sum(rvt)!=0
      rstepqtd=rstepqtd+1;

rtempo(rstepqtd,:)=sizeadj(tempo(bordasub(rvt,patamar,sensi):bordadesc(rvt,pa
tamar,sensi)),stdvec);
      rtempo(rstepqtd,:)=rtempo(rstepqtd,:)-rtempo(rstepqtd,1);

      %Antero-Posterior

rapmat(rstepqtd,:)=sizeadj(rap(bordasub(rvt,patamar,sensi):bordadesc(rvt,pata
mar,sensi)),stdvec);
      rap=rap(bordadesc(rvt,patamar,sensi):end);

      %Força Vertical

rvtmat(rstepqtd,:)=sizeadj(rvt(bordasub(rvt,patamar,sensi):bordadesc(rvt,patam
ar,sensi)),stdvec);
      rvt=rvt(bordadesc(rvt,patamar,sensi):end);
    else
      rodando=0;
    endif
  endwhile

  %%Análise Individual de cada passo=-=-=-=-=-=-=-=-=-=-=-=-=-=-=-=-=-=-=-=-=-
=-=-=-=-
  tmedio=(sum(ltempo(:,end))+sum(rtempo(:,end)))/(lstepqtd+rstepqtd);

  %Pé Esquerdo
    lvtmat=transpose(lvtmat);
  lapmat=transpose(lapmat);
  ltempo=transpose(ltempo);

  %Limpando curvas indesejadas....
  for ck=lstepqtd:-1:1
    %%Busca pontos de inflexão na curva para determinar usabilidade
    lvtpontos=findpeaks(lvtmat(:,ck),20);
    ptqtd=size(lvtpontos);

    if ptqtd(1)!=3 || ltempo(end,ck)>tmedio*1.2
      lvtmat(:,ck)=[];
      lapmat(:,ck)=[];
      ltempo(:,ck)=[];
      lstepqtd=lstepqtd-1;
      ck=ck-1;
```

```
    endif
  endfor

  %% Pé direito - Análise individual passo-a-passo -=-=-=-=-=-=-=-=-=-=-=-=-=-
=-=-

  rvtmat=transpose(rvtmat);
  rapmat=transpose(rapmat);
  rtempo=transpose(rtempo);

  for kk=rstepqtd:-1:1
    rvtpontos=findpeaks(rvtmat(:,kk),20);
    ptqtd=size(rvtpontos);

    if ptqtd(1)!=3 || rtempo(end,kk)>tmedio*1.2
      rvtmat(:,kk)=[];
      rapmat(:,kk)=[];
      rtempo(:,kk)=[];
      %%e finge que nada aconteceu...
      rstepqtd=rstepqtd-1;
    endif
  endfor

  %%Plotando análise individual dos passos...=-=-=-=-=-=-=-=-=-=-=-=-=-=-=-=-
=-=-=
  stepqtd=max(lstepqtd,rstepqtd);

  loutvec=[1,2,3,4,5,6,7,8,9,10,11,12,13,14,15,16,17,18,19];
  routvec=loutvec;

  for kk=1:rstepqtd
    %%Acha os pontos de inflexão Vertical
    rvtspots=findpeaks(rvtmat(:,kk),20);
    tp1=rvtspots(1,2:3);
    tv=rvtspots(2,2:3);
    tp2=rvtspots(3,2:3);

    %%Acha os pontos de interesse antero-posterior
    [rapmaxval,rapmaxpt]=max(rapmat(:,kk));
    [rapminval,rapminpt]=min(rapmat(rapmaxpt:end,kk));
    rapminpt=rapminpt+rapmaxpt-1;
    rapmid=rapmaxpt+fntransit(rapmat([rapmaxpt:rapminpt],kk));


    tc=rtempo(end,kk);
    F1=rvtmat(tp1(2),kk);
    t1=rtempo(tp1(2),kk);
    F2=rvtmat(tp2(2),kk);
    t2=rtempo(tp2(2),kk);
    F3=rvtmat(tv(2),kk);
```

```
  t3=rtempo(tv(2),kk);
  F4=abs(rapminval);
  t4=rtempo(rapminpt,kk);
  F5=rapmaxval;
  t5=rtempo(rapmaxpt,kk);
  t6=rtempo(rapmid,kk);
  t7=rtempo(end,kk)-rtempo(rapmid,kk);
  it1=integral(rtempo(1:tv(2),kk),rvtmat(1:tv(2),kk));
  it2=integral(rtempo(tv(2):end,kk),rvtmat(tv(2):end,kk));
  it3=integral(rtempo(:,kk),rvtmat(:,kk));
  it4=integral(rtempo(1:rapmid,kk),rapmat(1:rapmid,kk));
  it5=integral(rtempo(rapmid:end,kk),rapmat(rapmid:end,kk));
  it6=it4+it5;

  routvec=[routvec;tc,F1,t1,F2,t2,F3,t3,F4,t4,F5,t5,t6,t7,it1,it2,it3,it4,it5,it6];
endfor

for kk=1:lstepqtd
  %%Achando ponto de interesse da curva vertical
  lvtspots=findpeaks(lvtmat(:,kk),20);
  tp1=lvtspots(1,2:3);
  tv=lvtspots(2,2:3);
  tp2=lvtspots(3,2:3);

  %%Achando ponto de interesse da curva Anteroposterior
  [lapmaxval,lapmaxpt]=max(lapmat(:,kk));
  [lapminval,lapminpt]=min(lapmat(lapmaxpt:end,kk));
  lapminpt=lapminpt+lapmaxpt-1;
  lapmid=lapmaxpt+fntransit(lapmat([lapmaxpt:lapminpt],kk));

  %Dados Gerais
    ##tc = tempo total de contato com o solo]
    ##F1, t1 = força e tempo para o primeiro pico da vertical
    ##F2, t2 = força e tempo para o vale da vertical
    ##F3, t3 = força e tempo para o segundo pico da vertical
    ##F4, t4 = força e tempo para o mínimo da curva anteroposterior
    ##F5, t5 = força e tempo para o máximo da curva anteroposterior
    ##t6 = tempo para o ponto de transição da curva anteroposterior
    ##t7 = tempo de transição até o final do contato da curva anteroposterior
    ##i1 = integral sob a área entre o t=0 e o vale da vertical
    ##i2 = integral sob a area entre o vale da vertical e t=max
    ##i3 = integral total da curva vertical
    ##i4 = integral entre t=0 e o ponto de transição da curva anteroposterior
    ##i5 = integral entre o ponto de transição e o final da curva anteroposterior
    ##i6 = integral total da curva anteroposterior

  tc=ltempo(end,kk);
  F1=lvtmat(tp1(2),kk);
  t1=ltempo(tp1(2),kk);
  F2=lvtmat(tp2(2),kk);
```

```
      t2=ltempo(tp2(2),kk);
      F3=lvtmat(tv(2),kk);
      t3=ltempo(tv(2),kk);
      F4=abs(lapminval);
      t4=ltempo(lapminpt,kk);
      F5=lapmaxval;
      t5=ltempo(lapmaxpt,kk);
      t6=ltempo(lapmid,kk);
      t7=ltempo(end,kk)-ltempo(lapmid,kk);
      it1=integral(ltempo(1:tv(2),kk),lvtmat(1:tv(2),kk));
      it2=integral(ltempo(tv(2):end,kk),lvtmat(tv(2):end,kk));
      it3=integral(ltempo(:,kk),lvtmat(:,kk));
      it4=integral(ltempo(1:lapmid,kk),lapmat(1:lapmid,kk));
      it5=abs(integral(ltempo(lapmid:end,kk),lapmat(lapmid:end,kk)));
      it6=it4+it5;

      loutvec=[loutvec;tc,F1,t1,F2,t2,F3,t3,F4,t4,F5,t5,t6,t7,it1,it2,it3,it4,it5,it6];
    endfor

    loutvec(1,:)=[];
    routvec(1,:)=[];
    %%Controle de resolução do arquivo de saída
    resol=1000;
    outdata.esqpts=(round(loutvec*resol))/resol;
    outdata.dirpts=(round(routvec*resol))/resol;
    %Tempos
    outdata.ltempo=ltempo;
    outdata.rtempo=rtempo;
    %Vetores do pé Esquerdo
    outdata.lvt=lvtmat;
    outdata.lap=lapmat;
    %Vetores do Pé Direito
    outdata.rvt=rvtmat;
    outdata.rap=rapmat;

endfunction
#DataGen
clear all

disp('Digite 1 para criar banco de dados inicial');
op=input('  2 para agregar a um já existente: ');

if op==1
  disp('Banco de dados inicial selecionado.')

  [arquivo, pasta]=uigetfile();
  disp(strcat('Carregando: ',arquivo));
  arquivo=strcat(pasta,arquivo);
  dados=importdata(arquivo);
  matrix=dados.data;
```

```
if dados.textdata{1,1}(1,1)=='@'
  text=dados.textdata{1,1};
  massakg=str2num(text(2:end))*9.802;
else
  massakg=input('Sem peso do paciente fornecido. Favor inserir valor:');
endif


[linha, coluna]=size(matrix);

%%remove valores não-numéricos da matrix original
for x=1:linha
   for y=1:coluna
      if(isnan(matrix(x,y)))
        matrix(x,y)=0;
      end
  end
end

%%Indice e tempo de registro
frame=matrix(:,1);
tempo=matrix(:,2);
tempo=tempo(1:max(find(tempo)));

%%medidas do pé direito
rap=matrix(1:length(tempo),3)/massakg; %%anteroposterior
rvt=matrix(1:length(tempo),4)/massakg; %%vertical
rmd=matrix(1:length(tempo),5)/massakg; %%mediolateral

%%medidas do pé esquerdo
lap=matrix(1:length(tempo),6)/massakg; %%anteroposterior
lvt=matrix(1:length(tempo),7)/massakg; %%vertical
lmd=matrix(1:length(tempo),8)/massakg; %%mediolateral
disp('--------------------------------');
disp('Dados Carregados!');

%%Controle de sensibilidade dos detectores de borda.
patamar=0.01;
sensi=0.01;

%%Limites do eixo Y para os gráficos
miny=1.3*min([min(lap),min(rap),min(lmd),min(rmd)]);
maxy=1.3*max(max(lvt),max(rvt));


%%Tamanho do vetor padrão para análise de dados
stdvec=100;
```

```
  %%Cria o passo padrão fazendo a média entre cada passo.=-=-=-=-=-=-=-=-=-
=-=-=-=
  rodando=1;
  lstepqtd=0;
  lvtstep=0;
  lapstep=0;
  lmdstep=0;


  %%Isola cada passo e cria um vetor padrão com as médias de cada um.
  %%Pé Esquerdo
  while rodando==1;
    if sum(lvt)!=0
      lstepqtd=lstepqtd+1;

ltempo(lstepqtd,:)=sizeadj(tempo(bordasub(lvt,patamar,sensi):bordadesc(lvt,pat
amar,sensi)),stdvec);
      tempo=tempo(bordadesc(lvt,patamar,sensi):end);

      %Médio-Lateral

lmdmat(lstepqtd,:)=sizeadj(lmd(bordasub(lvt,patamar,sensi):bordadesc(lvt,pata
mar,sensi)),stdvec);
      lmd=lmd(bordadesc(lvt,patamar,sensi):end);

      %Antero-Posterior

lapmat(lstepqtd,:)=sizeadj(lap(bordasub(lvt,patamar,sensi):bordadesc(lvt,patam
ar,sensi)),stdvec);
      lap=lap(bordadesc(lvt,patamar,sensi):end);

      %Força Vertical

lvtmat(lstepqtd,:)=sizeadj(lvt(bordasub(lvt,patamar,sensi):bordadesc(lvt,patamar
,sensi)),stdvec);
      lvt=lvt(bordadesc(lvt,patamar,sensi):end);
    else
      rodando=0;
        if lstepqtd>1
          lmdstep=transpose(sum(lmdmat)/lstepqtd);
          lapstep=transpose(sum(lapmat)/lstepqtd);
          lvtstep=transpose(sum(lvtmat)/lstepqtd);
        else
          lmdstep=trasnpose(lmdmat);
          lapstep=transpose(lapmat);
          lvtstep=transpose(lvtstep);
        endif
    endif
  endwhile
```

```
%%Pé Direito
rodando=1;
rstepqtd=0;
rvtstep=0;
rapstep=0;
rmdstep=0;

%%Reset do vetor de tempo.
tempo=matrix(:,2);
tempo=tempo(1:max(find(tempo)));

while rodando==1;
  if sum(rvt)!=0
    rstepqtd=rstepqtd+1;

rtempo(rstepqtd,:)=sizeadj(tempo(bordasub(rvt,patamar,sensi):bordadesc(rvt,pa
tamar,sensi)),stdvec);
    tempo=tempo(bordadesc(rvt,patamar,sensi):end);

    %Médio-Lateral

rmdmat(rstepqtd,:)=sizeadj(rmd(bordasub(rvt,patamar,sensi):bordadesc(rvt,pata
mar,sensi)),stdvec);
    rmd=rmd(bordadesc(rvt,patamar,sensi):end);

    %Antero-Posterior

rapmat(rstepqtd,:)=sizeadj(rap(bordasub(rvt,patamar,sensi):bordadesc(rvt,pata
mar,sensi)),stdvec);
    rap=rap(bordadesc(rvt,patamar,sensi):end);

    %Força Vertical

rvtmat(rstepqtd,:)=sizeadj(rvt(bordasub(rvt,patamar,sensi):bordadesc(rvt,patam
ar,sensi)),stdvec);
    rvt=rvt(bordadesc(rvt,patamar,sensi):end);
  else
    rodando=0;
      if rstepqtd>1
        rmdstep=transpose(sum(rmdmat)/rstepqtd);
        rapstep=transpose(sum(rapmat)/rstepqtd);
        rvtstep=transpose(sum(rvtmat)/rstepqtd);
      else
        rvtstep=transpose(rvtmat);
        rmdstep=transpose(rmdmat);
        rapstep=transpose(rapmat);
      endif
  endif
endwhile
```

```
%%Criando tabela de dados


  for 1:stdvec
     minc(c)=min([mincurve(c), mincurve2(c)]);
     maxc(c)=max([maxcurve(c), maxcurve2(c)]);
     midc(c)=
  endfor


  disp('----------------------------------');
  disp('Dados Processados!');
  disp('Banco de dados inicial criado!');
endif
```

MARISTA

PUCRS

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