Letícia Munhoz de Avellar

Development and Evaluation of Camera-based System for Analysis of Dual-task in Older Adults: Gait combined with Prehension

Vitoria - ES

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Dissertation submitted to the Postgraduate Program in Electrical Engineering from the Technological Center of the Federal University of Espirito Santo, as a partial requirement for obtaining a Master's Degree in Electrical Engineering focused on Robotics.

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Supervisor: Prof. Dr. Anselmo Frizera Neto

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Development and Evaluation of Camera-based System for Analysis of Dual-task in Older Adults: Gait combined with Prehension

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"Be the change you want to see in the world."

(Mahatma Gandhi)

Abstract

As the age advances, it compromises the performance of motor abilities leading to an increased risk of falls. Recent studies suggest that there is a relationship between cognitive impairment and gait abnormalities and the gait performance has been used as predictor for cognitive decline and fall status, mainly in older adults with history of falls, also called faller older adults. The *dual-task paradigm* is a reference method that assess cognitive impairments, through the performance of the gait with another task simultaneously, such the combination of gait and prehension task, that is widely performed during activities of daily life. The advance in new technologies has facilitated the development of an objective evaluation of different movement parameters, such as accelerometers, force platforms and cameras. The *Kinect* sensor (Microsoft, USA) has been used for clinical motion analysis due to the low cost when compared with the expensive gold standard motion capturing systems. In addition, the *Leap Motion Controller* (Leap Motion, Inc., USA), also based on camera, has been used for analysis of hand movement. This work presents the development and evaluation of an accessible camera-based system using a sensor network composed by *Kinect* and *Leap Motion Controller* sensors to assess the gait and prehension parameters of fallers and non-fallers older adults (FOA and OA) under *dual-task* condition, gait combined with prehension. Firstly, experimental validations were performed in order to evaluate the feasibility and reliability of the selected sensors, and later, a clinical validation was applied on FOA and OA. The clinical validation protocol was divided in two conditions (*walking through* and *dual-task*) and was applied on twenty older adults (n=10 for each group). Results of the experimental validations showed low variations between *Kinect v2 system* and a commercial system, and low errors of *Leap Motion Controller system* in static scenario. In dynamic scenarios, an approach was developed to decrease the errors. Results of the clinical validation showed smaller step and stride lengths mean, and center of mass (CoM) mean velocity for FOA. In addition, the both groups decreased the CoM velocity under dual-task condition, however, FOA significantly decreased the step and stride lengths, with higher variability, in this condition. The FOA required longer movement time to perform the prehension task while walking, besides slowing down the CoM velocity in the prehension moment, showing a performance more conservative. Results showed similarity with previous studies and the system developed was capable to acquire the required parameters and evaluate the *dual-task paradigm* in the older adults. Future works involve the improvement of materials and techniques used in this work, and analysis of more parameters.

Keywords: *Dual-Task Paradigm*, Gait, Prehension, Older Adults, Faller Older Adults, Camera-Based Systems.

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1 Introduction

This Chapter presents the motivation to carry out this research, showing the main approached problem. The objectives and the justification also will be presented, showing the importance of the development of this work, contextualizing works of literature. The Chapter ends with the dissertation structure.

1.1 Motivation

The porcentage of older adults is increasing and the rapidly aging population is a global problem (MUSTUFA et al., 2015). The number of older persons is expected to exceed the number of children under 15 for the first time in 2050 (PIERLEONI et al., 2015). This increase of life expectancy is due to improvements in the quality of life and advances in medicine, besides the birth control, and this population ageing has significant social and economic implications at the individual, family, and societal levels (UNFPA; INTERNATIONAL, 2012).

As the age advances, changes occur that compromise the performance of motor abilities, such as mechanisms of postural control, gait and balance, decrease of functional capacity and difficulty adapting to the environment, leading to an increased risk of falls (GOMES et al., 2016). The gait disturbance and cognitive impairment in older adults are two major issues that increase the chances of a disability life (AUVINET et al., 2017).

Falls are the leading cause of injury-related hospitalization among people 65 years and older in society, injury is the fifth leading cause of death in elderly people and most of these fatal injuries are related to falls (KANNUS et al., 2005). According to Mustufa et al. (2015), falls due to locomotive impairment are a major cause of injury and stress in older adults resulting in incapacity, hospitalization, and subsequent social isolation. Reports show that one-third of older adults living at home fall at least one time every year.

Executive function refers to a variety of higher cognitive processes that modulate and use information from the posterior cortical sensory systems to produce behavior (SHERIDAN et al., 2003). These include the ability to plan, iniate, and monitor goal-directed behavior, with the flexibility to update goals when presented with new information (MCKINLAY et al., 2010), which are at the basis of the ability to manage independent activities of daily living (YOGEV-SELIGMANN; HAUSDORFF; GILADI, 2008).

Growing evidences suggest that executive functions play an important role in the ability to perform a motor and cognitive task simultaneously in older adults (LAMOTH et al., 2011). According to Sheridan et al. (2003), the attention is a dynamic executive function driven by sensory perception and the need to select a preferred stimulus for a particular action while ignoring the irrelevant. There are three types of attention: selective, sustained and divided. The selective attention is the ability to focus on a single relevant stimulus while ignoring irrelevant stimuli and the sustained attention is the maintenance of focused attention during an extended period. Lastly, the divided attention is the ability to focus on several relevant stimuli simultaneously (SHERIDAN et al., 2003).

Traditionally, dynamic stability during gait was considered an automatic or reflex-controlled task responses, requiring minimal cognitive resources (HOLLMAN et al., 2007). However, recent evidence suggests that maintaining stability requires sensorimotor and cognitive processes, particularly executive function and attention, with a relationship between cognitive impairment and gait abnormalities (HOWCROFT et al., 2014). Therefore, the gait performance is widely used as a predictor for cognitive decline (VERGHESE et al., 2007), falls status (BEAUCHET et al., 2009), quality of life (HIRVENSALO; RANTANEN; HEIKKINEN, 2000) and longevity (STUDENSKI, 2011).

Older adults with a history of falls (FOA) present some changes in the gait pattern, such as a decrease in stride length and velocity, and an increase in gait variability and double support time, and these changes are even more evident when two motor tasks are combined (RINALDI; MORAES, 2016). Due to increase gait stability and decrease fall risk, older individuals adopt a conservative gait pattern, that may require more cognitive control and result in gait deterioration under attention-demanding, dual-task (DT) conditions (LAMOTH et al., 2011; HOWCROFT et al., 2014).

The *dual-task training* is an increasingly rehabilition strategy, which aims to facilitate, through simultaneous functional activities, the allocation of attention resources, thus reducing interference in the DT (MENDEL; BARBOSA; SASAKI, 2015). During the DT test the subject performs an attention-demanding task, while performs a secundary task simultaneously. Also called *dual-task paradigm*, this become the reference method to assess cognitive impairments (AUVINET et al., 2017).

According to the performed activity, the dual-task can be motor or cognitive-motor (MENDEL; BARBOSA; SASAKI, 2015). There are many types of dual-task, such Howcroft et al. (2014) showed in his work, in which gait tests were performed combined with cognitive task of verbal fluence and the participants should walk and speak words beginning with the letters A, F or S. In Auvinet et al. (2017), the participants should walk along a straight 30 meters corridor counting aloud backwards from 50 subtracting serial 1 second (one by one).

Other dual-task type is the combination of gait and prehension task (motor dual-task), which is widely performed during activities of daily life (RINALDI; MORAES, 2016). Diermayr, Mcisaac e Gordon (2011) investigated the aging effects on grasp control when walking and transporting an object. The changes that occur in gait pattern can be used as predictors to reduce the frequency of falls, to identify diagnostic measures and to develop prevention of such falls (MUSTUFA et al., 2015).

The aim of this work is to assess gait and balance using the *dual-task paradigm* in fallers and non-fallers older adults, analyzing gait and prehension parameters, identifying changes in pattern, in order to predict falls.

1.2 Objective

1.2.1 General objective

The main goal of this work is to present the development of a camera-based system for acquisition of gait and prehension parameters to analyze dual-task in fallers and non-fallers older people.

1.2.2 Specific objectives

1. To develop algorithms to estimate the spatio-temporal gait parameters;
2. To develop a camera-based system for acquisition of spatio-temporal prehension parameters;
3. To integrate the camera-based acquisition systems of prehension and gait parameters;
4. To perform experimental validations of the both systems;
5. To perform a clinical validation on fallers and non-fallers older adults.
6. To analyze and to compare the obtained parameters with others systems in literature.

1.3 Justification

The biomechanics of human movement can be defined as the interdiscipline that describes, analyzes, and assesses human movement. A wide variety of physical movements are involved, since from the gait of the physically handicapped to the performance of a athlete. The physical and biological principles that apply are the same in all cases, excepting the specific movement tasks and the level of detail that is being assessed the performance of each movement for each application (WINTER, 2009). The human movement analysis mainly includes gait analysis, posture and trunk movement analysis, and upper limb movement analysis (WONG et al., 2015).

Gait analysis is the systematic study of human walking, performed by collecting kinematic and kinect data. This analysis is applied in different fields. Clinically, gait analysis is used for the assessment of gait pathologies, the prevention of pressure ulcers in diabetes or the assessment of the course of an orthopaedic disease (CREA et al., 2014).

The gait of the elderly is subject to two influences: the effects of age itself and the effects of pathological conditions, which become more common with advancing age, such as osteoarthritis and parkinsonism. Typically, the age-related changes in gait begin around 60 to 70 years of age, with a decrease in stride length, an increase in cycle time (decreased cadence) and an increase in the walking base. Others changes can be observed, such as an increase in the duration of the stance phase related with the gait cycle (WHITTLE, 2007).

The overwhelming advance in new technologies has facilitated the development of accurate and reliable devices and techniques that permit an objective evaluation of different gait parameters, providing a large amount of information related to a subject's gait. Currently, the analysis includes joint angles, angular velocities, angular accelerations (kinematic analysis); ground reaction forces, joint forces, moments, and powers (kinetic analysis); and dynamic electromyographic activity (EMG analysis) (WEBSTER, 2015).

The systems used to study the human movement can be classified according to two different approaches: those based on non-wearable sensors or on wearable sensors. The non-wearable sensors systems can be based on imaging processing, such as cameras, or based on floor sensor, such as force platforms. The wearable sensors systems use sensors located on several parts of the body, such as accelerometers, gyroscopic sensors and magnetometers (HERRAN; GARCÍA-ZAPIRAIN; MÉNDEZ-ZORRILLA, 2014).

Sensors such as accelerometers are low cost compared to more commonly used equipment for movement analysis, and are not restricted to a laboratory environment. However, these sensor are coupled to the body and can change the natural movement, besides being an inconvenience to the participant (KAVANAGH; MENZ, 2008). While the force platform are non-wearable sensor and reliable and accurate, however the analysis is limited since the surface area is small (WHITTLE, 2007).

The camera-based systems have progressed towards more accurate, automated systems (COLYER et al., 2018). These systems are divided in two types: marker-based and markerless systems. The marker-based systems are known as gold standard in motion capturing systems. However, the use of markers can affect and change the natural movement (WEBSTER, 2015). While the markerless approach is towards a fully automatic and non-invasive system, providing a major breakthrough for research and practice within rehabilitation (COLYER et al., 2018).

Recently, the *Kinect* sensor has been used for capturing of human movement, especially for clinical and scientific motion analysis of gait and the detection of falls. Due to the low cost and the markerless approach, the use of the *Kinect* sensor is a cost-efficient alternative to expensive gold standard motion capturing systems (MÜLLER et al., 2017) in clinics and rehabilitation centers of developing countries where the resources are more scarce. In addition, the *Kinect* is a non-wearable sensor, leaving the movement more free and natural.

In addition of systems for human gait analysis, camera-based systems such as *Leap Motion Controller* has been used for analysis of hand movement. These sensors allow the acquisition of relevant parameters in the study of prehension (WEICHERT et al., 2013).

This work proposes the development of an accessible camera-based system capable to investigate the changes in the gait when combined with prehension task, using a sensor network composed by *Kinect v2* and *Leap Motion Controller* sensors.

1.4 Dissertation Structure

This dissertation is divided into six chapters. Chapter 1 exposes a brief contextualization, with motivation and justification of this work, along with the research objectives.

Chapter 2 presents the theoretical background, showing the concepts related to biomechanics of the gait and the prehension in the clinical environment, besides the systems used for motion analysis.

Chapter 3 presents the materials used to development of the system, the techniques used to data processing and parameters estimation, and the selected parameters to be analyzed.

Chapter 4 exposes the experimental protocols performed in order to validate each sensor, and lastly, the clinical validation in order to assess faller and non-faller older adults.

Chapter 5 contains the results related to the experiments that were performed, described in Chapter 4. Furthermore, the obtained results are discussed and compared with previous studies using others systems.

Chapter 6 presents the acquired conclusions of work, as well as the future works to improvements and continuation of the study.

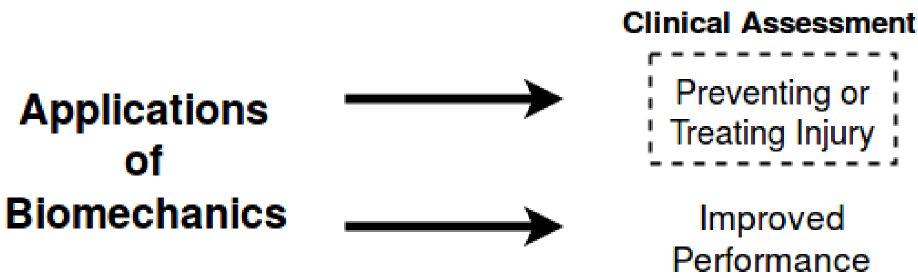
2 Theoretical Background

In this chapter, concepts and some related works about motion analysis will be presented. It will be approached about the biomechanics of the human gait and prehension and their applications on clinical assessment. Furthermore, will be presented the current systems and their techniques for motion analysis.

2.1 Motion Analysis for Clinical Assessment

Biomechanics has been defined as the study of the movement of living things using the science of mechanics (HATZE, 1974). The biomechanics of human movement consists of the study that describes, analyzes and assesses human movement (WINTER, 2009). It provides conceptual and mathematical tools that are necessary for understanding how living things move and how kinesiology professionals might improve movement or make movement safer (KNUDSON, 2007), as shown in Figure 1.

Figure 1 – Applications of Biomechanics.



Fonte: Adapted from (KNUDSON, 2007)

The science of human motion analysis has advanced rapidly over the last decade. Study of human gait characteristics may be useful for clinical applications and may benefit the various groups suffering from gait-related disorders (HERRAN; GARCÍA-ZAPIRAIN; MÉNDEZ-ZORRILLA, 2014).

2.1.1 Biomechanic of the Human Gait

The human gait can be defined as a complex and cyclical process, characterized by periods of loading and unloading of the limbs (KIRTLEY, 2006). It requires the synergy of muscles, bones, and nervous system, mainly aimed at supporting the upright position and maintaining balance during static and dynamic conditions (TABORRI et al., 2016).

The gait cycle is defined as the time interval between two successive occurrences of the same event of walking (WHITTLE, 2007). Such gait events have different classifications in the literature, however for all classifications the cycle is divided in two phases: *stance* and *swing*.

The *stance* phase is a term used to designated the period when the foot is on the ground (WHITTLE, 2007). This phase begins with the first contact of one foot and ends with the next contact of the same (ipsilateral) foot, which will be the initial contact of the next cycle (KIRTLEY, 2006). The *stance* phase can be divided in five subphases, according to the classification of Vaughan, Davis e O'Connor (1999):

1. *Heel strike*, initiates the gait cycle and represents the point at which the body's center of gravity is at its lowest position.
2. *Foot-flat*, is the time when the plantar surface of the foot touches the ground.
3. *Midstance*, occurs when the swinging (contralateral) foot passes the stance foot and the body's center of gravity is at its highest position.
4. *Heel off*, occurs as the heel loses contact with the ground and pushoff is initiated.
5. *Toe off*, terminates the *stance* phase as the foot leaves the ground.

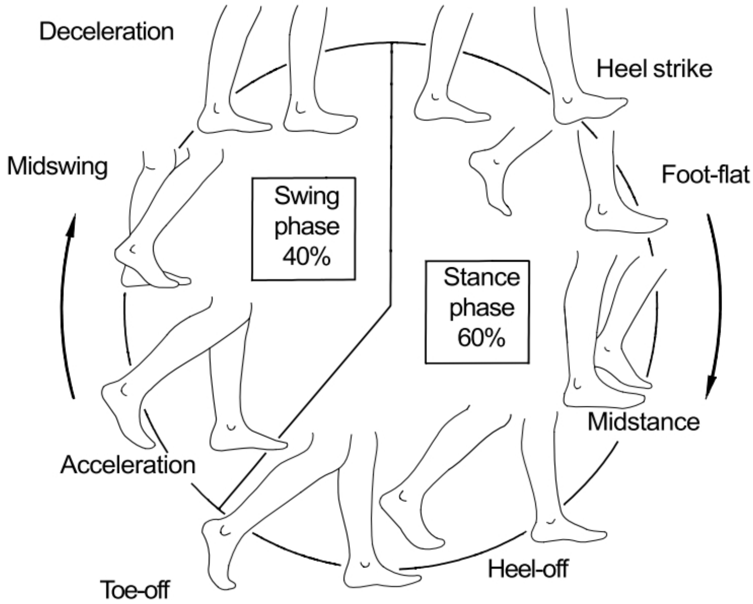
The *swing* phase corresponds to the moment when the foot is oscillation, without contact with the ground. This phase begins when the foot leaves the ground. The *swing* phase can be divided in three subphases, according to the classification of Vaughan, Davis e O'Connor (1999):

1. *Acceleration*, begins as soon as the foot leaves the ground and the subject activates the hip flexor muscles to accelerate the leg forward.
2. *Midswing*, occurs when the foot passes directly beneath the body, coincidental with midstance for the other foot.
3. *Deceleration*, describes the action of the muscles as they slow the leg and stabilize the foot in preparation for the next *heel strike*.

Figure 2 shows the gait cycle divided in phases and subphases, according to the traditional nomenclature.

This traditional classification best describes the gait of normal subjects. However, for patients with pathologies which affect the gait, this approach is not well described (VAUGHAN; DAVIS; O'CONNOR, 1999). Thus, an alternative classification was developed and is widely used in the literature, as described by Perry (1992). The subphases of this classification are listed below.

Figure 2 – The traditional nomenclature for describing events of the normal human gait.



Fonte: (VAUGHAN; DAVIS; O'CONNOR, 1999)

1. *Initial contact*, includes the moment when the foot just touches the floor.

2. *Loading response*, initial double stance period, this phase begins with initial floor contact and continues until the other foot is lifted for swing.

3. *Midstance*, the first half ofthe single limb support interval, this phase begins as the other foot is lifted and continues until body weight is aligned over the forefoot.

4. *Terminal stance*, This phase completes single limb support, and begins with heel rise and continues until the other foot strikes the ground. Throughout this phase body weight moves ahead of the forefoot.

5. *Preswing*, final phase of stance is the second (terminal) double stance interval in the gait cyc1e. It begins with initial contact of the opposite limb and ends with ipsilateral toe-off.

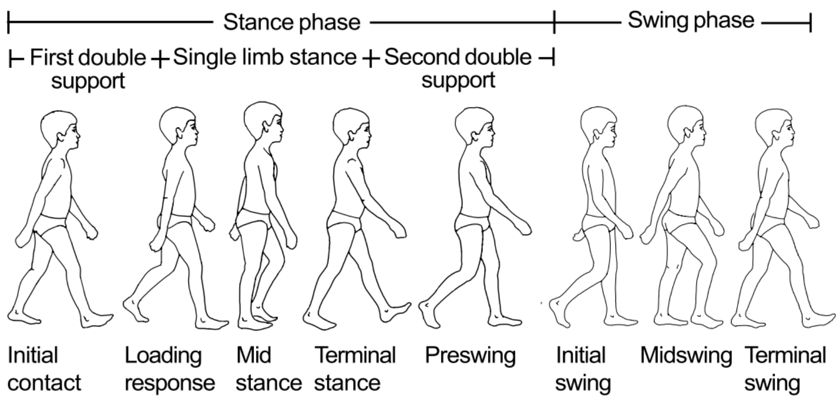
6. *Initial swing*, first phase is approximateIy one-third of the swing period. It begins with lift of the foot from the floor and ends when the swinging foot is opposite the stance foot.

7. *Midswing*, second phase of the swing period begins as the swinging limb is opposite the stance limb. The phase ends when the swinging limb is forward and the tibia is vertical.

8. *Terminal swing*, final phase ofswing begins with a vertical tibia and ends when the foot strikes the floor.

Figure 3 shows the gait cycle divided in phases and subphases, according to this alternative classification.

Figure 3 – An alternative nomenclature for describing events of the general human gait, to be applied to any type of gait.



Fonte: (VAUGHAN; DAVIS; O'CONNOR, 1999)

2.1.2 Clinical Gait Assessment

Clinical gait assessment seeks to describe the way in which a person walks, whereas gait research aims to improve our understanding of gait. This may be required, if the aim is to document their current status, or it may be just one step in a continuing process, such as the planning of treatment or the monitoring of progress over a period of time. In general, clinical gait assessment is performed for one of three possible reasons: it may form the basis of clinical decision making, it may help with the diagnosis of an abnormal gait, or it may be used to document a patient's condition (WHITTLE, 2007).

The cyclic nature of human gait is a very useful feature for reporting different parameters (VAUGHAN; DAVIS; O'CONNOR, 1999). These parameters are used as basis measures for clinical gait assessment (KIRTLEY, 2006). There are hundreds of parameters that can be used to characterize gait, among them spatio-temporal gait parameters, ground reaction force, and muscle activity(VAUGHAN; DAVIS; O'CONNOR, 1999).

It is difficult on clinical observation to analyze these parameters, and to quantify their degree of deviation from normality. Such limitations led health professionals, engineers and scholars of the human movement to develop techniques for conducting study and gait analysis (SAAD; BATTISTELLA; MASIERO, 1996). Gait analysis is the systematic study of human walking, performed by collecting kinematic and kinetic data, and dynamic electromyography that describe and characterize it (CREA et al., 2014).

The spatio-temporal gait parameters are important functional measures that reflect the "vital signs" of gait. The main applications of these parameters, according to Kirtley

(2006), are:

- *Screening*, to detect elderly people at risk of falling;
- As a *performance* measure, to grade a patient's level of disability;
- *Monitoring* the efficacy of therapy, and
- *Normalization* of other gait measurements (in order to compare results from people walking at different speeds).

From the clinical point of view, the importance of human gait analysis lies in the fact that gait disorders affect a high percentage of the world's population and are key problems in neurodegenerative diseases. Study of human gait characteristics may be useful for clinical applications and may benefit the various groups suffering from gait-related disorders (HERRAN; GARCÍA-ZAPIRAIN; MÉNDEZ-ZORRILLA, 2014).

The gait cycle changes over time. According Kirtley (2006), natural walking speed remains relatively stable until about 70, as people age, it decreases by about 15% per decade. Balance slowly deteriorates, and this is reflected in the spatio-temporal gait parameters. Stance accounts for 59% of gait cycle at age 20, and 63% at age 70, with double support duration increasing from 18% to 26%. Furthermore, reduced stride length, reduced speed and increased double support time can be associated with fear of falling, rather than falling itself.

2.1.2.1 Spatio-temporal Gait Parameters

Herran, García-Zapirain e Méndez-Zorrilla (2014) presented many interesting spatio-temporal gait parameters used for Clinical Gait Assessment. Table 1 shows some these parameters.

Figure 4 shows some spatio-temporal gait parameters cited on Table 1 through the foot placement on the ground.

The *step duration* is the temporal difference between the moment of each hell contact, meanwhile the *cycle time* is the temporal difference between the hell contact of the same foot. The division of step length by step duration resulted in *step velocity*. The *walking speed* is the division of the performed distance by the time taken to complete the trajectory. Lastly, the *cadence* is the number of steps taken in a given time, the usual units being steps per minute (WHITTLE, 2007).

2.1.2.2 Center of Mass (CoM) and Margin of Dynamic Stability

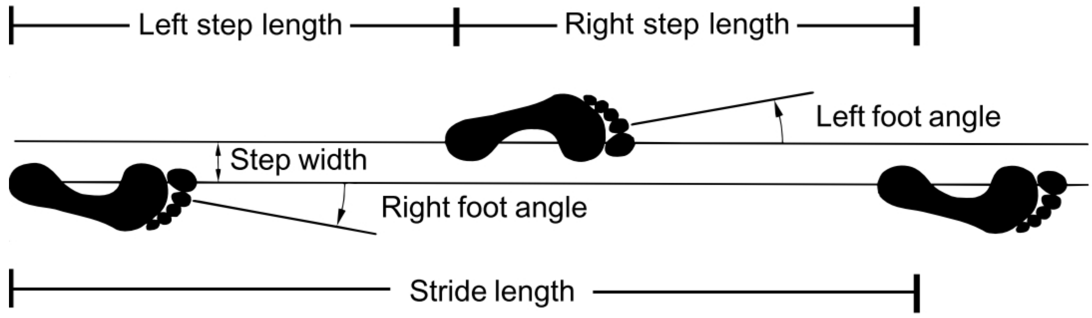
Center of mass (CoM) is a point equivalent of the total body mass in the global reference system (GRS) and is the weighed average of the CoM of each body segment in

Table 1 – Spatio-temporal gait parameters used for Clinical Gait Assessment.

	Spatio-temporal gait parameters
Length	Step length Step width Stride length
Time	Step duration Stance time Swing time Cycle time
Velocity	Step velocity Walking speed Cadence

Fonte: (HERRAN; GARCÍA-ZAPIRAIN; MÉNDEZ-ZORRILLA, 2014)

Figure 4 – Terms used to describe foot placement on the ground.



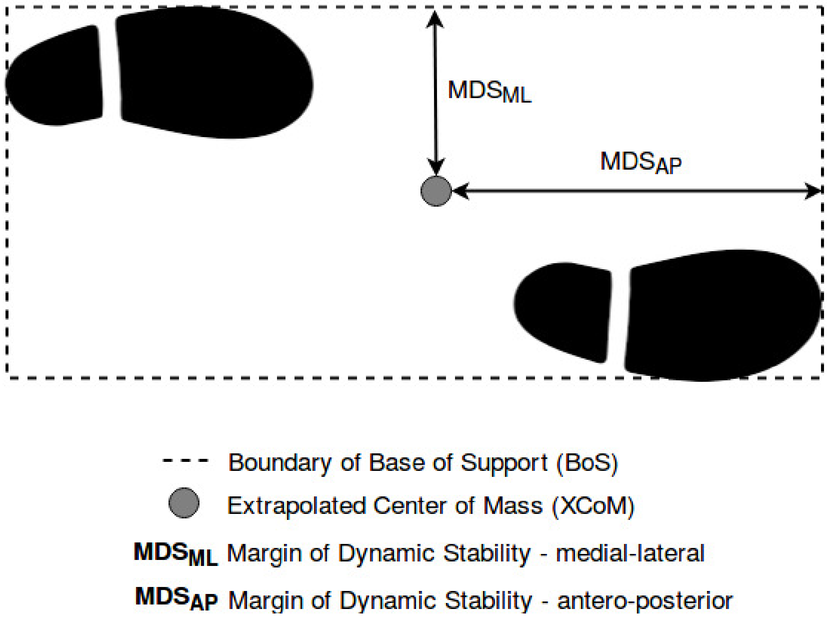
Fonte: (WHITTLE, 2007)

3D space. It is a passive variable controlled by the balance control system. The vertical projection of the CoM onto the ground is often called the centre of gravity (CoG). Its units are metres (m) (WINTER, 1995).

Stable gait is achieved as a function of the CoM position and velocity at the moment of foot placement (LUGADE; LIN; CHOU, 2011). The condition for human stability is the confinement of the CoM in static situations or extrapolated center of mass (XCoM) in dynamic situations within the base of support (BoS). The CoM–BoS interaction is indicative of both static and dynamic balance control ability (GUO; XIONG, 2017), and there is more noticeable deviations in balance control in elderly people (LUGADE; LIN; CHOU, 2011).

Margin of dynamic stability (MDS) is determined by CoM or XCoM position relative to BoS boundaries (GUO; XIONG, 2017), as shown in Figure 5, and was used as a measure of balance (LUGADE; LIN; CHOU, 2011). MDS reflects the CoM-BoS interaction, and is influenced by voluntary changes in two gait parameters (step width and length) and can be increased by longer steps (larger anterior-posterior BOS) and wider steps (larger medial-lateral BOS) (GUO; XIONG, 2017).

Figure 5 – Margin of Dynamic Stability.



Fonte: Adapted from (SIVAKUMARAN et al., 2018).

2.1.3 Prehension

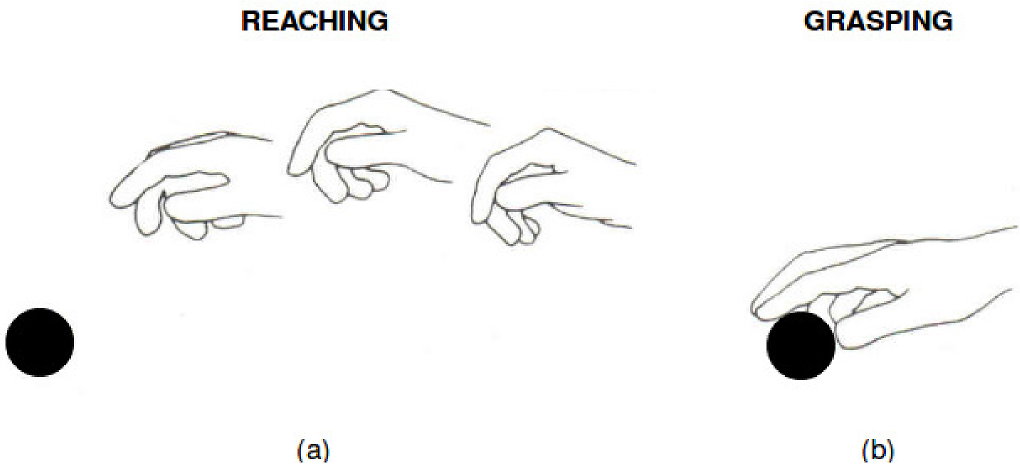
Prehension can be divided in two components: reaching or transporting (bringing the hand close to the object to be grasped), and grasping (ensuring that the object be enclosed) (MADALENA; MORAES, 2015; ZAAL; BOOTSMA; Van Wieringen, 1998), as shown in Figure 6. Reaching involves the control of more proximal muscles (such arm and forearm), whereas grasping involves the movement of more distal muscles (such fingers) (MADALENA; MORAES, 2015).

Human walking is a cyclical process and the gait phases can be clearly distinguished. In contrast, the reach and grasp movements are highly complex, and depend on many factors, such as position, orientation and purpose of the task. Due to its complexity, research into grasping still lags behind that of human gait and have been the scope of scientific interest for more than the past two decades (SUPUK; BAJD; KURILLO, 2011).

Butler et al. (2010) presents a propose of reach and grasp cycle, as shown in Figure 7.

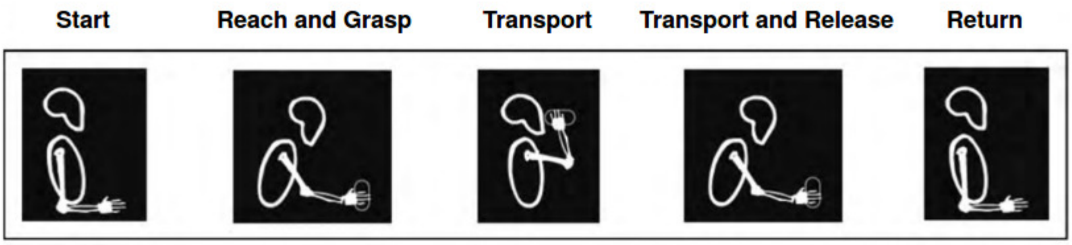
More specifically, the *reach-to-grasp* study is summarized in the start of the movement until the object be enclosed, excluding the *transport*, *transport and release* and *return* phases. Supuk, Bajd e Kurillo (2011) presents a propose of *reach-to-grasp* cycle, as cited below:

Figure 6 – Prehension movement divided by reach and grasp components. (a) Reach component. (b) Grasp component.



Fonte: (HOUGH; NEL, 2013).

Figure 7 – Reach and grasp cycle.



Fonte: (BUTLER et al., 2010).

- *Phase 1, hand acceleration*: Start the movement, acceleration of the wrist in order to reach the peak velocity, the fingers start pre-shaping and reached the peak tangential velocity.
- *Phase 2, hand deceleration*: Deceleration of the wrist, with deceleration peak in this phase. The focus is more on the fingers opening.
- *Phase 3, final closure of the finger*: The final closing of fingers around the object in order to obtain a stable grasp. At the onset of this phase the rate of hand closing is at its highest and then decreases toward zero-values.

Some interesting prehension parameters are divided in reaching and grasping parameters, and can be observed in Table 2.

Table 2 – Prehension parameters divided in reaching and grasping.

	Prehension parameters
Reaching	Movement time Peak wrist velocity Time-to-peak wrist velocity
Grasping	Peak grip aperture Time-to-peak grip aperture Peak grip aperture velocity Time-to-peak grip aperture velocity

Fonte: (HERRAN; GARCÍA-ZAPIRAIN; MÉNDEZ-ZORRILLA, 2014)

2.1.4 *Dual-task Paradigm*

Dual-task paradigm involves the execution of a primary task, which is the major focus of attention, and a secondary task performed at the same time (O'SHEA; MORRIS; IANSEK, 2002). Cognitive-motor and motor dual-tasks play important roles in daily life: walking while talking, using a mobile phone, carrying a bag or watching traffic (LIU et al., 2017).

The *dual-task paradigm* seen in the literature may not necessarily apply to gait (O'SHEA; MORRIS; IANSEK, 2002), however this paradigm is more applied to locomotion with others tasks, according literature. The principle of *dual-task paradigm* applied to gait (or *dual-task gait assessment*) is to compare task performance whilst walking and simoultaneously executing an attention-demanding task (BEAUCHET et al., 2009), since previous studies have indicated that performing two tasks simultaneously may negatively impact gait performance (LIU et al., 2017).

Studies of dual-task walking have used verbal/cognitive or manual tasks as second tasks (NORDIN et al., 2010). Thus, the *dual-task gait assessment* can be divided in two types: cognitive-motor and motor dual-task. Next sub-sections will approach the both types.

2.1.4.1 Cognitive-motor dual-task

Cognitive domains include visuoperception/visuospatial ability, speed of mental processing, memory, learning, and executive functions. As mentioned earlier in this paper, executive functioning refers to the ability to plan, initiate and monitor goal-directed behavior, with the flexibility to update goals when presented with new information, including attention, problem solving, working memory, verbal fluency, and set-shifting (MCKINLAY et al., 2010).

To assess the interactions between gait and cognition, the dual task paradigm has become the reference method (AUVINET et al., 2017). Cognitive decline is another cause of chronic disability in elderly people and well known to be a risk factor for falls, disability, and dementia. This cognitive decline increases with age: 25% of adults over 65 years have cognitive impairments (GRAHAM et al., 1997; AUVINET et al., 2017).

There is a direct relationship between cognitive impairment severity and increased gait abnormalities (HAUSDORFF et al., 2005), in both elderly people (BEAUCHET et al., 2014) and younger adults (KILLANE et al., 2014). The control of walking depends on shared brain networks dedicated to cognition and motor control (MONTERO-ODASSO et al., 2012; ROSANO et al., 2012; AUVINET et al., 2017). Gait abnormalities, or dual-task costs, such as slowing of gait, are interpreted as the increased cost of involvement of cortical attention processes while walking. The dual-task cost identified during a gait assessment may reveal subtle brain impairment and has been related to attention and executive function efficiency (MONTERO-ODASSO et al., 2012).

Many studies approach the use of *dual-task gait assessment* using cognitive-motor dual-task. In the Lamoth et al. (2011), the participants were asked to perform a letter fluency task in which the subject had to name as many words starting with a predefined letter "R" or "G". The number of different words was counted and the trunk accelerations in 3 orthogonal directions were measured. In general, all participants altered their gait pattern in response to dual tasking by decreasing walking speed and increasing stride time.

Other study involving cognitive-motor dual-task was Montero-Odasso et al. (2009). Participants walked the length of the mat while counting backward from one hundred by one aloud. Gait velocity, step length, stride length, step time, stride time and double support time were assessed using an electronic walkway system (GAITRite®).

In Auvinet et al. (2017) study, the patients divided in 4 subgroups (recurrent falls, memory impairment, gait instability and cautious gait) walked along a straight 30 meters corridor at their usual pace counting aloud backwards from 50 subtracting serial 1 second. Stride frequency and regularity, and walking speed were calculated by 3-D-acceleration sensor and a stopwatch.

2.1.4.2 Motor dual-task

Gait consists of highly preprogrammed movements and is thought to be regulated mainly at brain-stem, spinal, and cerebellar regions, with descending input from the cortex. Whereas some upper-extremity movements are more novel and are thought to require attention, visual guidance, and somatosensory feedback to control their performance, being mainly controlled by the motor cortical regions (O'SHEA; MORRIS; IANSEK, 2002).

With advancing age, reduced function of different systems used for walking may affect walking ability and increase risk of falling (NORDIN et al., 2010). Older adults with a history of falls (FOA) present some gait impairments and these changes in the walking pattern are even more evident when two motor tasks are combined (NORDIN et al., 2010; RINALDI; MORAES, 2016).

As seen in sub-section above, the inclusion of a cognitive task with the locomotion causes gait abnormalities. With an increase in task difficulty, such as motor task, older adults must allocate more attentional resources to walking to compensate for the reduction in sensory-motor control (RINALDI; MORAES, 2016).

Some studies approach the use of *dual-task gait assessment* using motor dual-task. Liu et al. (2017) used the dual-task gait assessment as training for patients after stroke (tasks: walking while carrying a tray). Speed, cadence, stride time and stride length were assessed. The patients presented significant improvements in gait speed and stride length after motor dual-task training.

Nordin et al. (2010) conducted different cognitive and motor tasks on the elderly people. The motor tasks were: carry a cup (a saucer with a coffee-cup) using one hand, carry a tray (a rectangular wooden tray) using both hands and carry tray-cup (the tray with the saucer and cup on top) using both hands. The gait parameters analyzed were: gait speed, step length, step width, step time and double support time. It was possible observe changing the mean step-width when carrying a cup, indicating that the ability to use sensorimotor resources in a flexible manner is related to a decreased fall risk.

Lastly, Rinaldi e Moraes (2016) conducted older adults with or without history of falls to perform gait combined with prehension. Spatio-temporal gait parameters, center of mass and margin of dynamic stability, and prehension parameters were analyzed. They found a decrease in step length and speed with the addition of the prehension task, concluding that older adults (both groups) adopted a more conservative strategy to allow them to allocate more attention to the grasping task and avoid errors.

2.2 Motion Analysis Systems

From the variables trajectory and the time course spent to execute the movement, kinematic indicators of structural importance for gait evaluation are observed: linear and angular variations of position, linear and angular velocities, center of gravity velocity, segments and joints velocities, variations of the movement accelerations, time of reaction and time of movement, and others variables to be selected according to the purposes of the movement analysis (AMADIO; SERRÃO, 2007).

Also, kinetic indicators of human movement are observed, biomechanics aspects related to forces that cause the observed movement, such as the repercussions on the analyzed phenomenon. The ground reaction force investigation during stance phase of locomotion movements, as well as the distribution of dynamic pressure on the plantar surface, brings important knowledge about the form and characteristics of mechanical overload on the human locomotor apparatus and its behavior during movement (AMADIO; SERRÃO, 2007).

Thus, motion analysis systems are developed to do acquisition of the kinematic and kinetic parameters, according to the application. This section will be present some of these systems and their advantages and disadvantages.

2.2.1 Imaging System

Vision-based motion analysis involves extracting information from sequential images in order to describe movement (COLYER et al., 2018). There are many types of imaging systems that could be used. In this sub-section will be approached three main types: *Manual Digitization* (COLYER et al., 2018), *Automatic Marker-based Systems* and *Markerless Motion Analysis Systems* (WEBSTER, 2015; COLYER et al., 2018).

2.2.1.1 *Manual Digitization*

Manual digitization was a motion measurement technique used for many decades. Prior to digital technologies, cine film cameras were traditionally used. With the advent of video cameras (transition of tape-based to digital), cine cameras have become essentially redundant in the field of biomechanics. Regardless of the technology used to capture motion, manual digitizing requires the manual localization of several points of interest in each sequential image from each camera perspective (COLYER et al., 2018).

One of the primary advantages of manual digitizing is that the attachment of markers is not necessarily required. Furthermore, manual digitization remains a valuable tool particularly in sports biomechanics as it allows analysis of movement in field (COLYER et al., 2018).

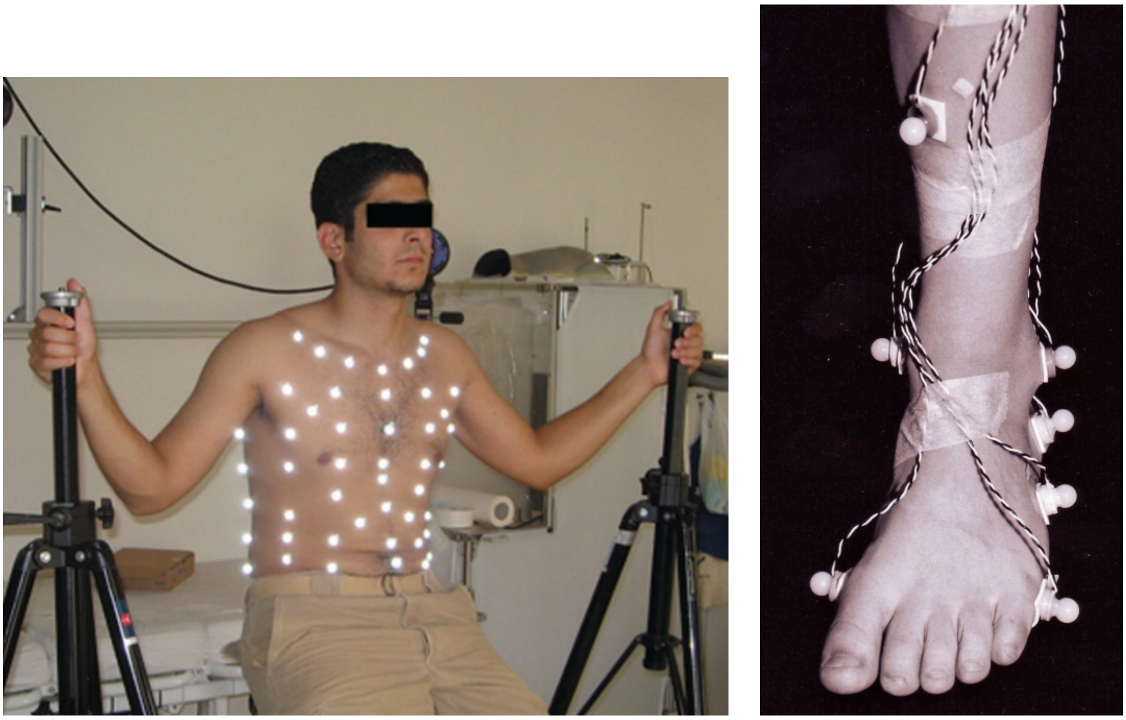
Certain drawbacks remain including the fact that manual digitizing is a notoriously time-consuming and laborious task, and is liable to subjective error. These limitations motivated the development of automatic solutions made available by the emergence of more sophisticated technologies (COLYER et al., 2018).

2.2.1.2 *Automatic Marker-based Systems*

A large number of commercial automatic optoelectronic systems now exist for the study of human movement. The majority of these utilize multiple cameras that emit invisible infrared light, and markers that reflect this infrared back to the cameras and allow their 3D position to be deduced (COLYER et al., 2018). These markers are mounted over predetermined anatomical landmarks: bony prominences, joint axes, limb axes, and others (WEBSTER, 2015).

Two types of markers are currently employed with these systems: passive retroreflective marker and active marker, as shown in Figure 8. The passive marker is made of lightweight spheres and do not require power packs. The reflected light is captured by the cameras and digitized by the system. Light is usually supplied by strobes of light-emitting diodes (LEDs) arranged surrounding each camera lens or placed near each camera. cameras are equipped with optical filters, selective of light in the infrared spectrum ($\lambda \approx 860nm$) (WEBSTER, 2015).

Figure 8 – Types of markers. (a) Passive markers. (b) Active markers.



(a) (b)

Fonte: (WEBSTER, 2015).

The second type of markers is an actively illuminated (optoelectric) marker. In these systems, the light-emitting diode markers are pulsed at a predetermined frequency, allowing higher sampling rates, an increased number of markers per unit area, and frequency-coded data sorting. However, active markers require that the subject carries a power pack, which may change the movement (WEBSTER, 2015).

Most of the problems of these systems are the physical and/or psychological constraints that attached markers impart on the participant, influencing movement execution. These drawbacks can limit the utility of marker-based systems within certain areas of sports biomechanics and rehabilitation, and have driven the exploration of potential markerless solutions.

2.2.1.3 *Markerless Motion Analysis Systems*

Markerless methods are not yet in widespread use within biomechanics, however the technology is under rapid development with modern computer vision algorithms improving the robustness, flexibility and accuracy of markerless systems (COLYER et al., 2018).

The typical image processing based system use threshold filtering to convert the video images of gait into black and white, the pixel count to compute the number of light and dark pixels or removes the background of the image via *background segmentation*. Another advanced method used with systems that employ image processing is *depth measurement*, also called *range imaging* (HERRAN; GARCÍA-ZAPIRAIN; MÉNDEZ-ZORRILLA, 2014). A depth map is an image where each pixel, instead of describing color or brightness, describes the distance of a point in space from the camera (COLYER et al., 2018).

The devices most commonly use one of two technologies: structured light or time-of-flight (ToF). Structured light devices sense depth through the deformations of a known pattern projected onto the scene, while ToF devices measure the time for a pulse of light to return to the camera. The most common devices which use structured light and ToF are the *Kinect* and *Kinect v2* sensors, respectively, which are provided with body tracking software designed for interactive systems (HERRAN; GARCÍA-ZAPIRAIN; MÉNDEZ-ZORRILLA, 2014; COLYER et al., 2018).

Beside that, there are others technologies, such as camera triangulation (or stereoscopic vision) and infrared thermography. The stereoscopic vision is based on the creation of a model through the calculation of similar triangles between the optical sensor, the light-emitter and the object in the scene. Whilst the infrared thermography uses the surface temperatures to create visual images of an object (HERRAN; GARCÍA-ZAPIRAIN; MÉNDEZ-ZORRILLA, 2014; COLYER et al., 2018).

2.2.2 Inertial Sensors

An alternative approach to motion analysis techniques involves the use of Inertial Measurement Units (IMUs) attached to the body for the purpose of examining segmental accelerations and angular velocities during the movement (KAVANAGH; MENZ, 2008). Since IMUs are sourceless, compact and light, they have been a popular choice for motion tracking (ZHOU et al., 2008).

The current benefits of using accelerometers to assess movement include: the low cost compared to more commonly used gait laboratory equipment; testing is not restricted to a laboratory environment; accelerometers are small which enables subjects to move relatively unrestricted; a variety of accelerometer designs offer diversity of dynamic range and sensitivity (KAVANAGH; MENZ, 2008); and others.

The combination of different inertial quantities permits the detection of the first contact of foot with the ground, which represents an important index for the assessment of healthy status of a subject during locomotion, by means of the estimation of the foot orientation. Moreover, IMU systems allow researchers to compute spatio-temporal gait parameters, that are stride length, cadence, etc., other than gait phases (TABORRI et al., 2016). Figure 9 shows Technaid IMUs (Technaid, Spain) coupled to the body for movement analysis.

Figure 9 – IMUs coupled to the body.

Fonte: Technaid, Spain.

Some studies use IMUs to evaluate the gait disorders in the older adults. Auvinet et al. (2017) evaluated the gait performance (under single and dual task) of one hundred and three elderly patients. They used a 3-D acceleration sensor and calculated the stride frequency, stride regularity and walking speed.

Pierleoni et al. (2015) developed a wearable fall detection device in which is incorporated a Magnetic, Angular Rate, and Gravity sensor (MARG) to overcome the limitation of a single accelerometer. MARG is the combination of IMUs (acelerometer, gyroscope and magnetometers) with altitude and heading. This device combines information from 3-axis accelerometer, 3-axis gyroscope and 3-axis magnetometer.

2.2.3 Force Platform

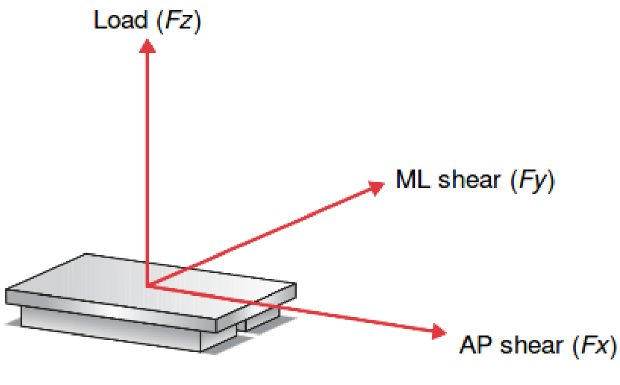
The force platform technique is one of the tools most widely applied in assessing postural balance in a quantitative way (PIIRTOLA; ERA, 2006). Force platforms provide an evaluation and comparison between normal and abnormal gaits. Although the evaluation of ground reaction forces, generated during gait, is but one of many ways in which locomotion can be studied, they have become an increasingly important part of the understanding of the biomechanics of gait (BESANCON et al., 2003).

The individual load and the shear components of the ground reaction force vector (direction and magnitude beneath the foot) can be measured using a force platform. The electrical output signals may be processed to produce three components of force (vertical, mid-lateral and antero-posterior directions), the two coordinates of the center of pressure and the moments about the vertical axis (WHITTLE, 2007).

This is a precision instrument using either strain gauges or piezo-electric quartz crystals to convert force into electric signals (KIRTLEY, 2006). This instrument has contributed greatly to the scientific study of gait and is now standard equipment in gait laboratories (WHITTLE, 2007).

Force platforms are very reliable and accurate devices, thanks to their very sensitive and high-frequency sensors and can be used for both static and dynamic studies, such for assessing balance, posture and gait. Figure 10 presents the Force Platform from EMG System.

Figure 10 – Force Platform. (a) Three components of force measured by Force Platform. (b) EMG System Force Platform.



(a)

(b)

Fonte: EMG System, Brasil.

Some studies use force platforms for evaluation of the older adults. Piirtola e Era (2006) did a systematic review of literature to search for and critically review the findings of prospective studies where force platform equipment (measuring changes in center of pressure as an indicator of balance) was used in predicting falls among elderly populations.

Furthermore, Condron, Hill e Physio (2002) developed a study to analyze whether the incorporation of a cognitive task while performing balance measures on a force platform would result in better discrimination between healthy young adults, healthy older adults, and older adults with a mild increase in risk of falling than measures without the cognitive task.

2.2.4 Sensorized Insole

When a high portability is needed, or measurement of pressures at foot plantar surface is required, in-shoe systems appear to offer the best trade-off in order to perform gait analysis. In-shoe systems can be used to record plantar pressure distributions with a sensorized insole within a shoe (CREA et al., 2014).

The availability of a low-cost, easy-to-use, portable system capable of collecting valid long-term data in gait analysis could provide important contributions to clinical practices supporting professional caregivers (for reporting and decision-making in hospitals or in outdoors environments), and to home-care practice to support patients for self-rating (for instance, in the home environment or during everyday activities) (MARTÍNEZ-MARTÍ; PALMA; CARVAJAL, 2014).

However, their use is limited to applications that do not need extremely precise measurements (CREA et al., 2014). Moreover, the placement of the sensors on patients with pathological gaits affects the accuracy and reliability and the wire connections can decrease the system service life (TABORRI et al., 2016).

Commercial products include the F-scan System (Tekscan, Inc., USA), the Novel Pedar System (Novel Inc., USA), and the Biofoot/IBV (Universidad Politécnica de Valencia, Spain), all of which capture dynamic in-shoe temporal and spatial pressure distributions that are used for dynamic gait stability analysis, gait detection, and altered gait characteristics during running (MARTÍNEZ-MARTÍ; PALMA; CARVAJAL, 2014).

Figure 11 presents an example of a sensorized insole from Tekscan, EUA.

Sheridan et al. (2003) used a sensorized insole to measure the ground reaction force and, consequently, determine the stride time. Through this, Sheridan et al. (2003) evaluated the interference of cognitive function and divided attention in the gait in Alzheimer's disease.

And Hausdorff, Rios e Edelberg (2001) used a sensorized insole to measure the gait rhythm on a stride-to-stride basis to test the hypothesis that increased gait variability predicts falls among community-living older adults attending an outpatient clinic.

Figure 11 – Sensorized insoles. (a) F-scan (Tekscan, Inc., USA). (b) Novel Pedar System (Novel Inc., USA)

Fonte: Produção do próprio autor.

3 Materials and Methods

This Chapter shows the materials and the techniques used to develop the proposed system. Details about the data acquisition and synchronization will approached, besides the data analysis. In addition, the softwares and the packages used will be presented and discussed about their advantages and drawbacks.

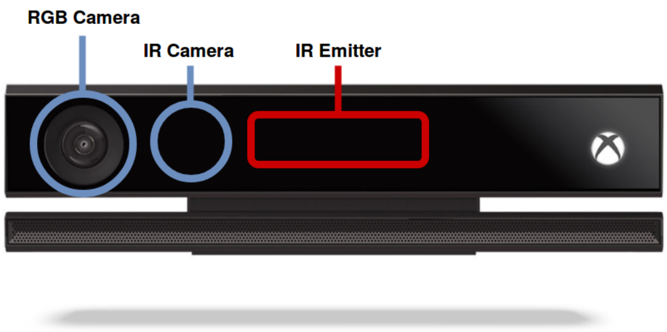
3.1 Materials and System Architecture

The used equipments were camera-based sensors. The following sections will approach the system architecture for development of this work, besides the informations about the sensors used.

3.1.1 *Kinect v2*

The *Kinect v2* is a 3D sensor produced by Microsoft[1], it is composed by a RGB camera (resolution of 1920 × 1080 pixels), an infrared camera (resolution of 512 × 424 pixels) and an infrared emitter, as shown in Figure 12. The sensor is based on depth measurement method and the time-of-flight (ToF) technology, approached in Chapter 2. In addition, it is based on the intensity modulation technique (CARUSO; RUSSO; SAVINO, 2017).

Figure 12 – *Kinect v2* sensor with cameras and emitter positions.



Fonte: Produção do próprio autor.

The field of view is 70° horizontally and 60° vertically, and the depth detection range is 0.5 to 4.5 meters. According to Microsoft's specifications, each *Kinect v2* sensor requires a dedicated USB 3.0 controller, thus each sensor has to be connected to a dedicated computer (MÜLLER et al., 2017).

[1] https://www.microsoft.com

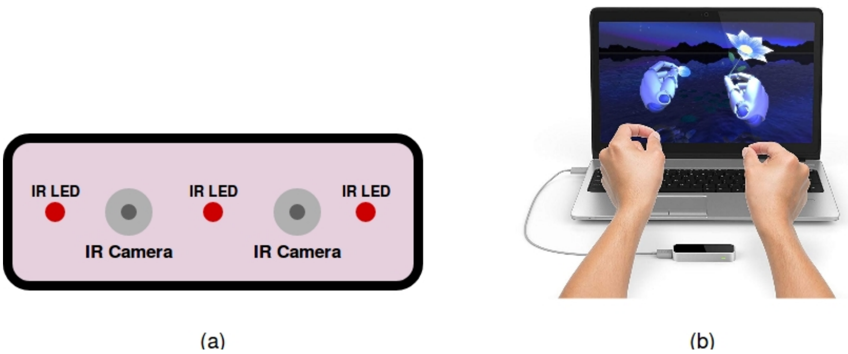
Through the *Kinect v2* software development kit (SDK) Microsoft provides color and infrared data streams, depth images, body index images and the skeleton information for every tracked person (25 joints at 30 Hz) (MÜLLER et al., 2017).

Furthermore, the *Kinect v2* SDK provides the tools and APIs, both native and managed, that is needed to develop Kinect-enabled applications for Microsoft Windows, including embedded libraries for skeleton tracking (SCANO et al., 2017). The most important data streams for the purpose of motion tracking are the color, depth and skeleton streams (CARUSO; RUSSO; SAVINO, 2017). For application on others Operating Systems, other SDKs are used. OpenNI 2.0 software and NiTE2.2 API (Application Programming Interface) are examples for application on Linux. These software and API will be detailed in the next section.

3.1.2 *Leap Motion Controller*

The *Leap Motion Controller*[2] uses infrared (IR) imaging to determine the position of predefined objects in a limited space in real time. Three separate IR LED emitters are used in conjunction with two CCD cameras (GUNA et al., 2014), as shown in Figure 13. The stereographic camera arrangement complements the IR LEDs to prevent losing track of objects within the field of view and, while, creates a 3D interaction space (CURIEL-RAZO et al., 2016).

Figure 13 – *Leap Motion Controller*. (a) Schematic. (b) *Leap Motion Controller* connected to a computer.



Fonte: Produção do próprio autor.

The *Leap Motion Controller* enables to analyze the objects observed in the device's field of view. The controller's field of view is an inverted pyramid centered on device's cameras and is 150°. The effective range of the controller extends from approximately 25 to 600 millimeters above the device (GUNA et al., 2014) and the sensor's accuracy in fingertip position detection is approximately 0.01 millimeters (WEICHERT et al., 2013).

[2] https://developer.leapmotion.com

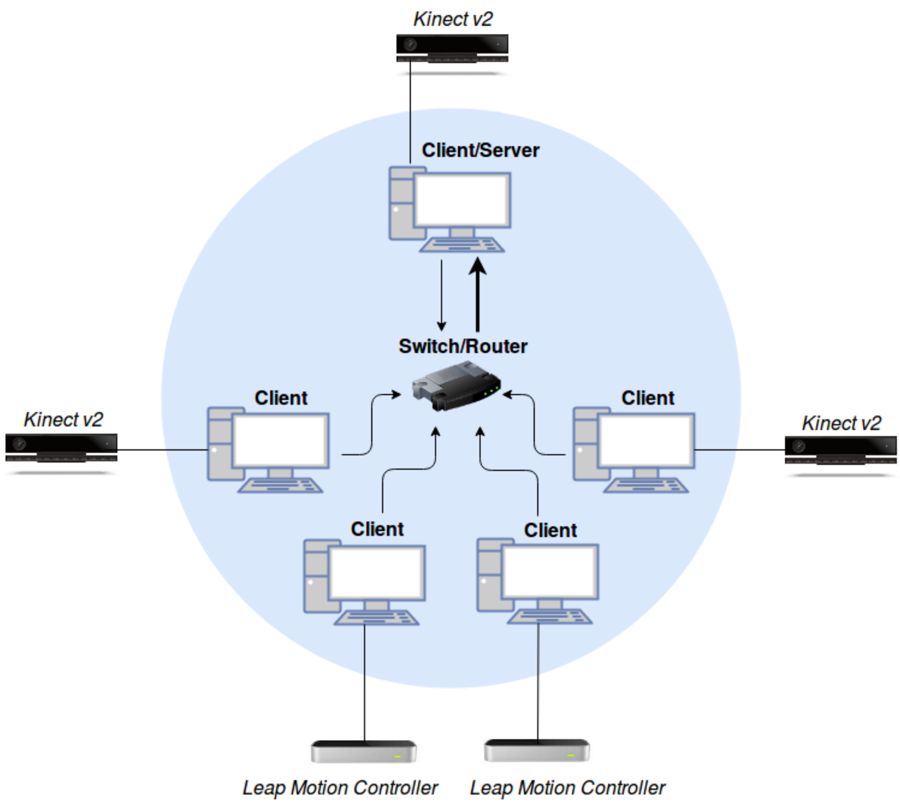
A challenge with the *Leap Motion Controller* is to maintain accuracy and fidelity of detection when the hands do not have direct line of sight with the controller (POTTER; ARAULLO; CARTER, 2013). Wachs et al. (2011) analyzed that tracking complications arise from occlusions, cluttered environments, and rapid motions that causes motion blur. Overcome these challenges allows good recognition accuracy to be achieved.

The *Leap Motion Controller* SDK contains two basic libraries that define the API to the Leap Motion tracking data. One library is written in C++, the second is written in C. It recognizes hands, fingers, tools, gestures, and motion, and is available on Windows, Mac OS X and Linux.

3.1.3 System Architecture

The proposed system consists of two different camera-based sensors: three *Kinect v2* and two *Leap Motion Controller* devices. Each sensor is connected to a different heterogeneous computer. The computers are divided in four clients and one client/server, and they are interconnected on a network switch/router. It was designed to support n devices as long as they are connected to different CPUs and graphic cards. Figure 14 shows the system architecture.

Figure 14 – System Architecture.

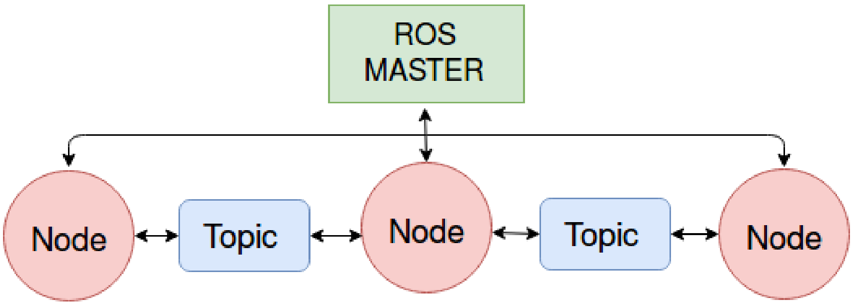


Fonte: Produção do próprio autor.

3.2 Data Acquisition and Synchronization

The network protocols management is performed by Robot Operating System[3](ROS). ROS is a robotics middleware (collection of software frameworks for robot software development) designed for robotic applications. It is a collection of tools, libraries, and conventions that aim to simplify the task of creating complex and robust robot behavior across a wide variety of robotic platforms. For this work, ROS has the advantage of providing a software middleware for the development of autonomous systems. The messages management of ROS is based on *nodes* and *topics*. A *node* is a process that performs computation and *topic* is the transport in which *nodes* exchange messages, as shown in Figure 15.

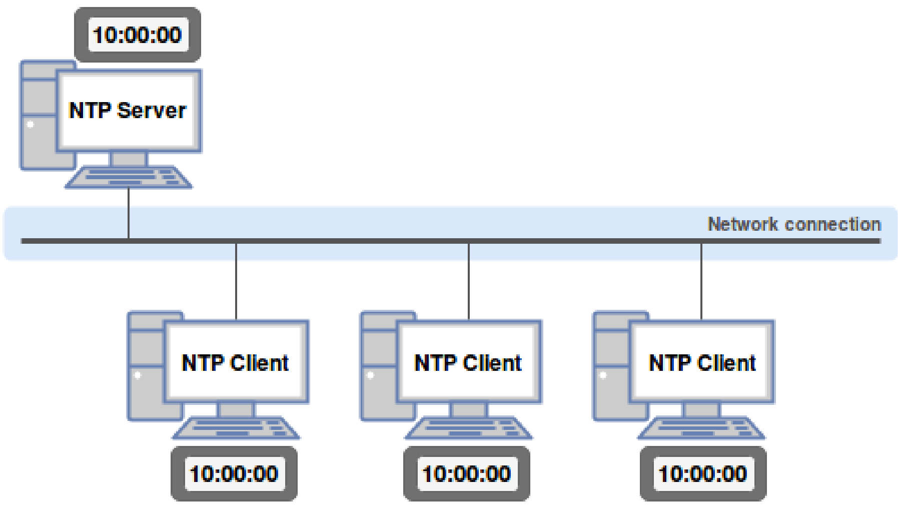
Figure 15 – ROS messages management by *nodes* and *topics*.



Fonte: Produção do próprio autor.

For synchronization of the computers, the Network Time Protocol (NTP) was used. NTP is a network protocol that synchronizes the server and client clocks, as shown in Figure 16.

Figure 16 – NTP protocol.



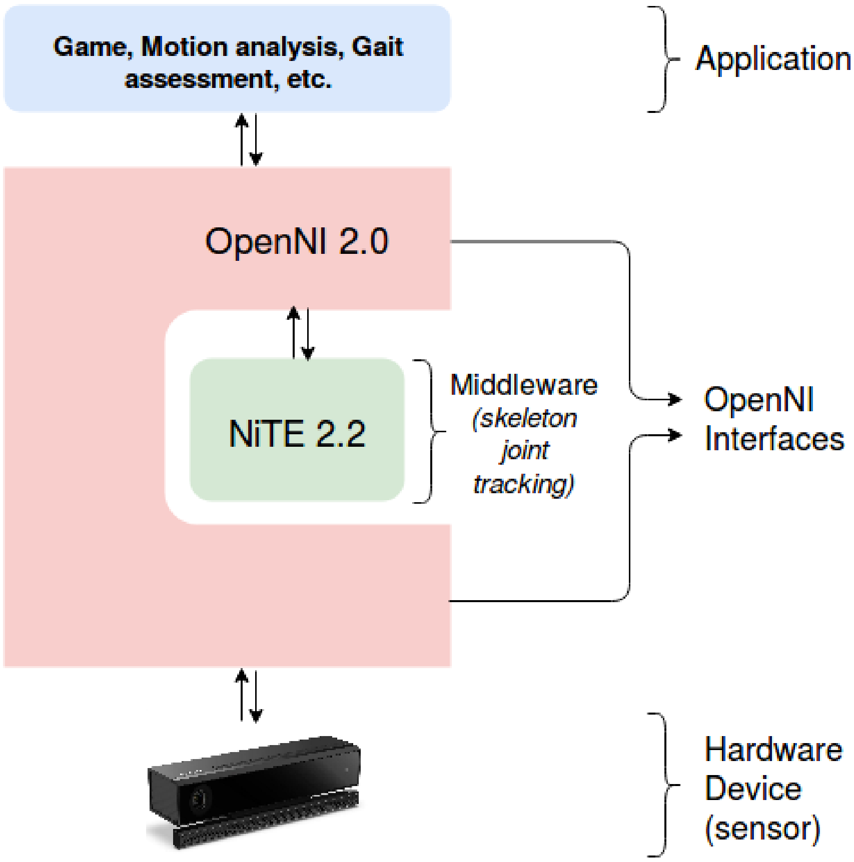
Fonte: Produção do próprio autor.

[3] http://www.ros.org

3.2.1 *Skeleton Joint Tracking*

For applications in operating systems different of Windows, the *Kinect v2* SDK is not available. Due to this system being developed in Linux, each *Kinect v2* estimates the user skeleton joint tracking through NiTE2.2 API, released by PrimeSense. NiTE2.2 is a middleware component that allows for skeleton and gesture detection and their algorithms perform functions such as scene analyzer (separation of users from background) and accurate user joint tracking, through OpenNI interfaces, as shown in Figure 17.

Figure 17 – OpenNI concept.



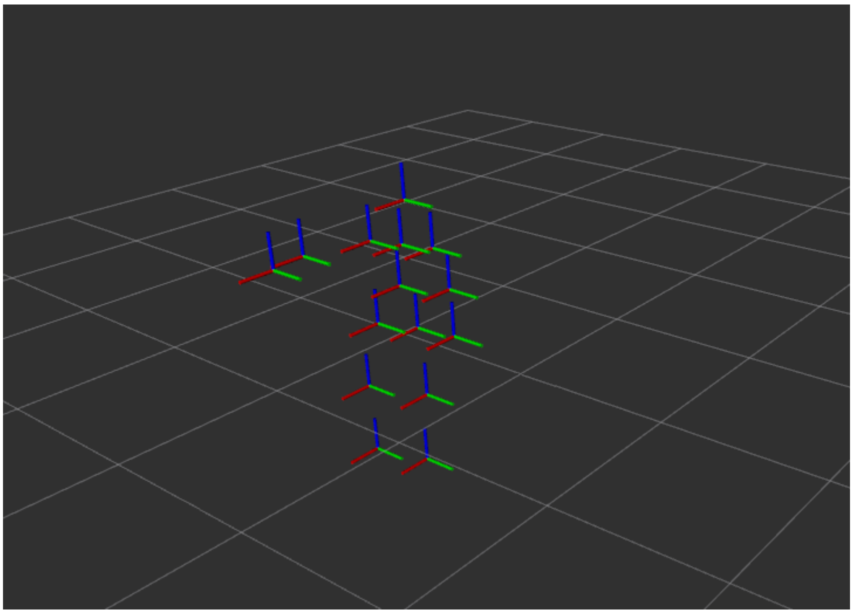
Fonte: Adapted from [4]

The OpenNI 2.0 API provides access to PrimeSense compatible depth sensors. It allows an application to initialize a sensor and receive depth, RGB, and IR video streams from the device. It provides a single unified interface to sensors created with depth sensors. OpenNI also provides a uniform interface that third party middleware developers can use to interact with depth sensors. Applications are then able to make use of both the third party middleware, as well as underlying basic depth and video data provided directly by OpenNI.

4 https://github.com/OpenNI/OpenNI/blob/master/Documentation/OpenNI_UserGuide.pdf

The joint positions provided by the clients (15 joints as shown in Figure 18), are merged through a fusion algorithm (Kalman Filter (KHALEGHI et al., 2013)) performed by the system server (CARVALHO, 2018). Data fusion with Kalman filtering has been studied by several researchers focused on using a multi-Kinect setup (MOON et al., 2016).

Figure 18 – *Skeleton joint tracking* (3-D) in Rviz environment.



Fonte: Produção do próprio autor.

3.2.2 *Hand Joint Tracking*

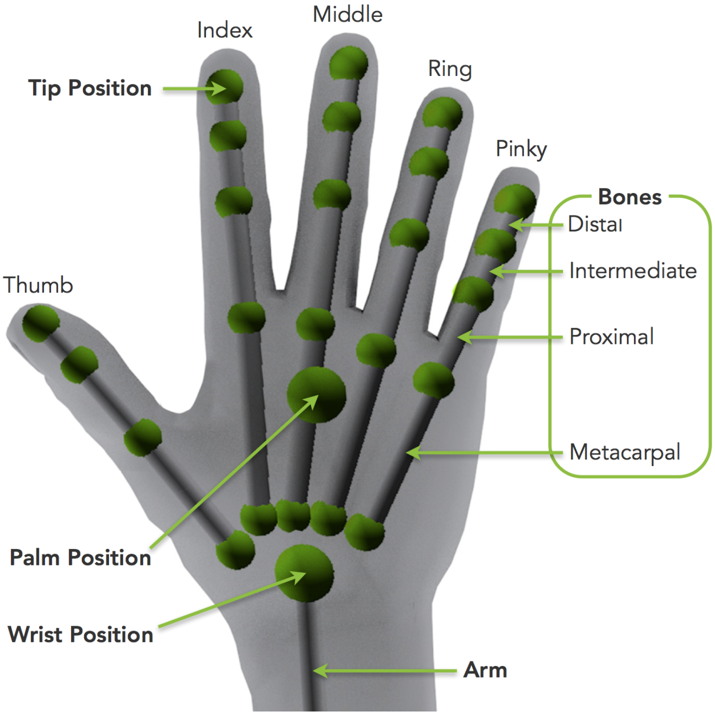
Due to the *Leap Motion Controller* SDK being supported by Linux, for application of this work the SDK was used. In addition, for hand tracking in ROS, it was necessary a ROS driver, *leap_motion package*.

The hand model provides information about the identity, position, and other characteristics of a detected hand, the arm to which the hand is attached, and lists of the fingers associated with the hand. Figure 19 shows the hand model through *Leap Motion Controller* SDK.

In addition, the *Leap Motion Controller* provides information about each finger on a hand. When a finger is not visible, the finger characteristics are estimated based on recent observations and the anatomical model of the hand. Fingers are identified by type name, i.e. thumb, index, middle, ring, and pinky.

The variables obtained through SDK are: direction and normal vectors of palm center, the palm position, Euler angles of the wrist, the tip direction of each finger and the position of each joint (tip, distal, intermediate, proximal and metacarpal) of each finger (thumb, index, middle, ring and pinky).

Figure 19 – *Skeleton joint tracking.*



Fonte: Adapted from [5]

3.3 Data Analysis

The proposed data were based on Rinaldi e Moraes (2016), in which gold standard marked-based cameras (VICON) were used to acquire gait and prehension parameters. Thus, this work aims to apply the procedures presented by Rinaldi e Moraes (2016) and based on the data analyzed by them, using the markerless system developed in this work.

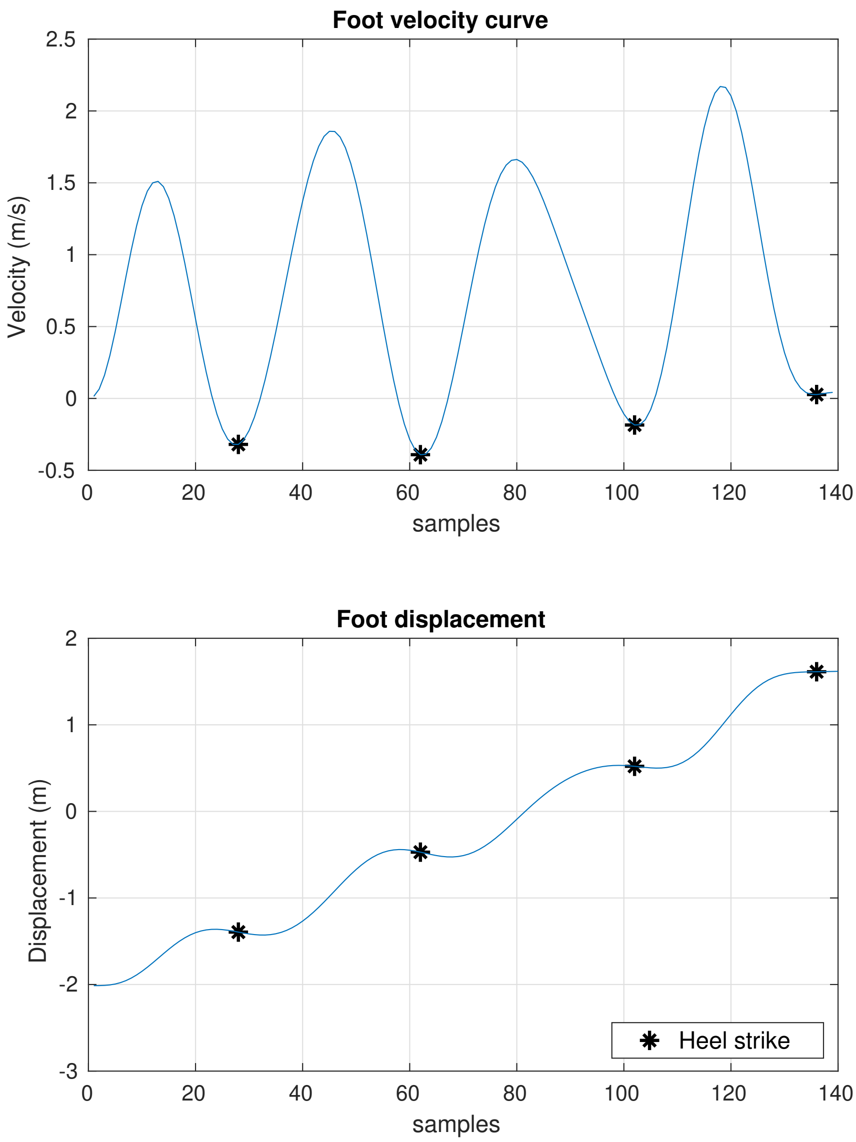
The analyzed data was divided in two groups: gait and prehension parameters. The gait parameters are subdivided in spatio-temporal gait parameters and center of mass (CoM) velocity. The spatio-temporal parameters consist the step and stride lengths. The CoM velocity was analyzed in three differents steps: the step at the moment of grasping (N), the before step (N-1) and next step (N+1). In addition, the CoM average velocity also was assessed. The prehension parameters were subdivided in reaching and grasping variables. The grasping variables consisted peak grip aperture, time-to-peak grip aperture, peak grip aperture velocity and time-to-peak grip aperture velocity. The reaching variables consisted movement time, peak wrist velocity and time-to-peak wrist velocity. Next subsections will show the procedure to estimate each data.

[5] http://blog.leapmotion.com/getting-started-leap-motion-sdk/hand-hierarchy/

3.3.1 Spatio-temporal gait parameters

In order to estimate the step and stride lengths, it is necessary to define the moment of each *heel strike*. The *heel strike* was estimated as being the lower values of the velocity curve (the valleys) of each foot in the antero-posterior (AP) direction, as shown in Figure 20.

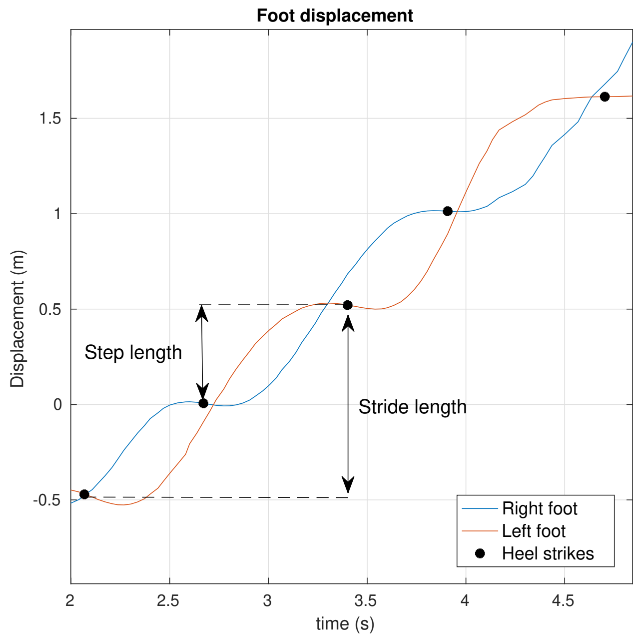
Figure 20 – *Heel strike* estimate.



Fonte: Produção do próprio autor.

Thus, the step and stride lengths were estimated as the difference between the *heel strikes*. The *stride length* is the difference between the *heel strikes* of the same foot and the *step length* is the difference between the *heel strikes* of the subsequent feet, as shown in Figure 21.

Figure 21 – *Step and stride lengths.*



Fonte: Produção do próprio autor.

3.3.2 Center of mass (CoM) velocity

The CoM velocity was calculated through the derivative of CoM displacement curve in AP direction, as shown in Equation 3.1 and Figure 22.

$$vCoM_{AP}[n] = \frac{CoM_{AP}[n] - CoM_{AP}[n-1]}{sample\_time} \tag{3.1}$$

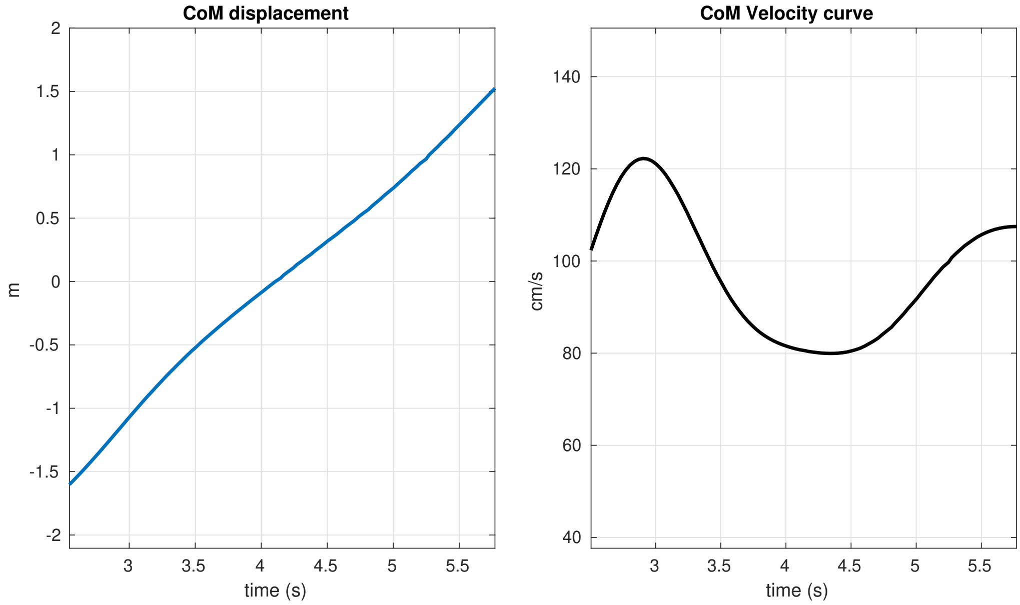
3.3.3 Prehension parameters

The prehension parameters are divided in two: reaching and grasping variables. The reaching variables depend on the fingers trajectory, while the reaching variables depend on the wrist trajectory.

3.3.3.1 Grasping variables

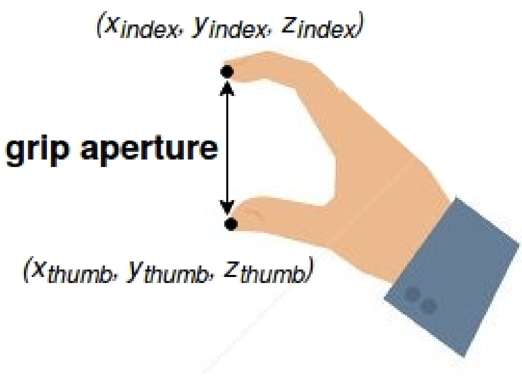
The grasping variables are: peak grip aperture, time-to-peak grip aperture, peak grip aperture velocity and time-to-peak grip aperture velocity. In order to estimate the grasping variables, it is necessary to define the grip aperture curve. This parameter is calculated as being the Euclidean distance of the thumb and index tip positions, as shown in Figure 23 and Equation 3.2.

Figure 22 – CoM displacement and velocity curves.

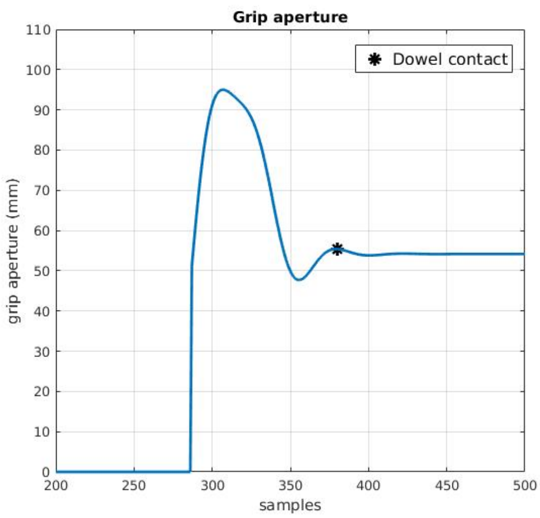


Fonte: Produção do próprio autor.

Figure 23 – Grip aperture. (a) Illustrative diagram. (b) Grip aperture curve.



(a)



(b)

Fonte: Produção do próprio autor.

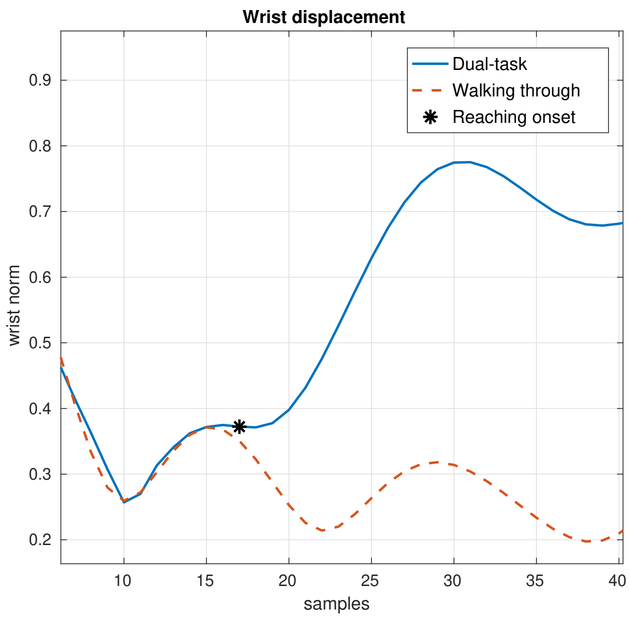
$$grip\_aperture = \sqrt{(x_{thumb} - x_{index})^2 + (y_{thumb} - y_{index})^2 + (z_{thumb} - z_{index})^2} \qquad (3.2)$$

Through this curve, it is possible obtain the *peak grip aperture*, the maximum aperture value. The grip aperture velocity curve is the derivative of the grip aperture curve, and the *peak grip aperture velocity* is the maximum value of this curve. The *dowel* *contact* is the moment when the hand grasp the object. The *time-to-peak grip aperture* and the *time-to-peak grip aperture velocity* are the time of ocurrence of peak grip aperture and peak grip aperture velocity, respectively. These two parameters are adjusted to *movement time*, by percentage (%), approached in the next sub-section.

3.3.3.2 Reaching variables

The reaching variables are: movement time, peak wrist velocity and time-to-peak wrist velocity. Firstly, the wrist trajectory was calculated through the wrist joint displacement in the AP direction obtained by *Kinect v2* system. In order to estimate the movement time, it is necessary to define two moments: *reaching onset* and *dowel contact*. The *reaching onset* was calculated as the first deviation of the wrist trajectory when compared with "walking through" profile. This deviation was estimated by the correlation of two wrist curves (walking through and dual-task), as shown in Figure 24. The *reaching onset* was defined when the correlation was lower than 0.98.

Figure 24 – *Reaching onset.*



Fonte: Produção do próprio autor.

The *dowel contact* was presented in the sub-section above. Thus, the *movement time* is the temporal difference between *reaching onset* and *dowel contact*.

The wrist velocity is calculated as the derivative of the wrist displacement. Thus, the *peak wrist velocity* is the maximum value of the wrist velocity curve, and the *time-to-peak wrist velocity* is the time of occurence of this peak, adjusted to *movement time*, by percentage (%).

4 Experimental Validation

This Chapter presents the experimental validations proposed in this work. The first experimental validation was performed in order to compare the *Kinect v2 system* with a commercial system. The next experimental validations approach the evaluation of *Leap Motion Controller system* in static and dynamic scenarios and different conditions, using one or two devices for dynamic scenario. And lastly, a clinical validation was applied on fallers and non-fallers adults.

4.1 *Kinect v2* System Validation with a Commercial System

In order to validate the used system to analyze the gait, a protocol was applied in which the participants walked on a straight line with inertial sensors coupled to the body. The aim of this protocol is to compare the system with a commercial sensor, *Xsens* (Xsens Technologies B.V., Netherlands), evaluating the stride length parameter. The validation protocol is shown in Figure 25.

This protocol was executed by 15 older adults and for each participant, two trials were performed. The *Xsens system* was used as reference method, and through this the error module was calculated.

4.2 *Leap Motion Controller* Validation

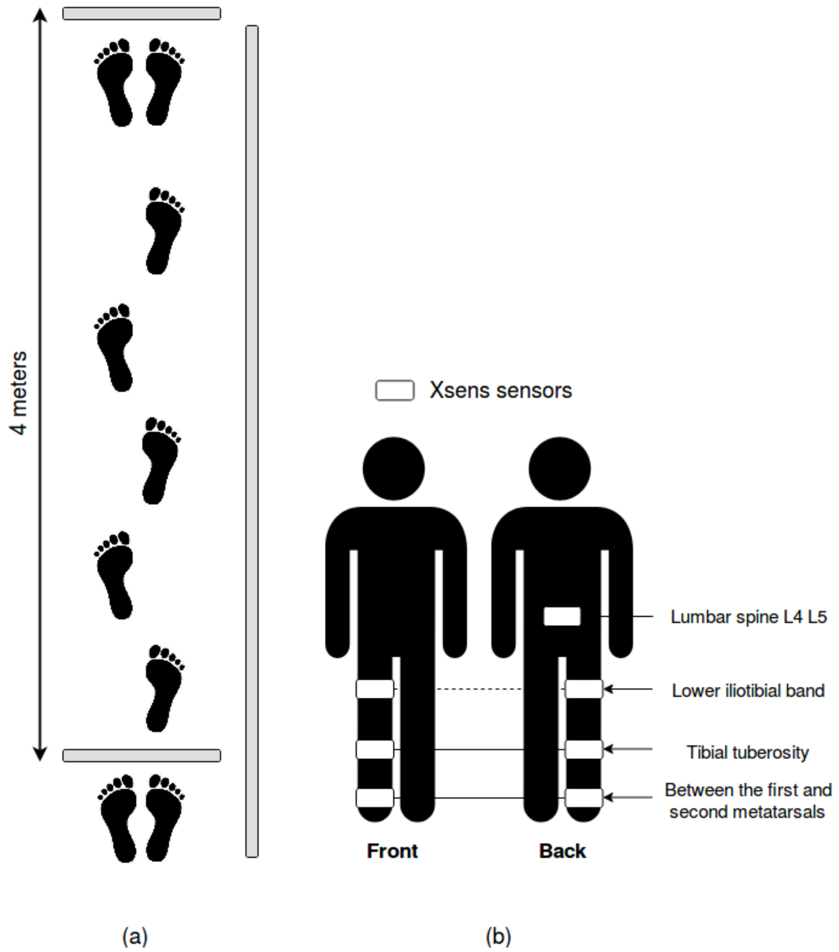
Two tests were proposed in order to evaluate the reliability of the *Leap Motion Controller*. The first test was static and the second was dynamic. The next subsections will explain these tests.

4.2.1 Preliminary Tests

The preliminary tests were performed with the sitting participant, only moving the right hand. The purpose of this protocol was to validate the sensor data with predetermined elements. Three objects with known lengths were used and only one Leap Motion Controller captured the data, as shown in Figure 26.

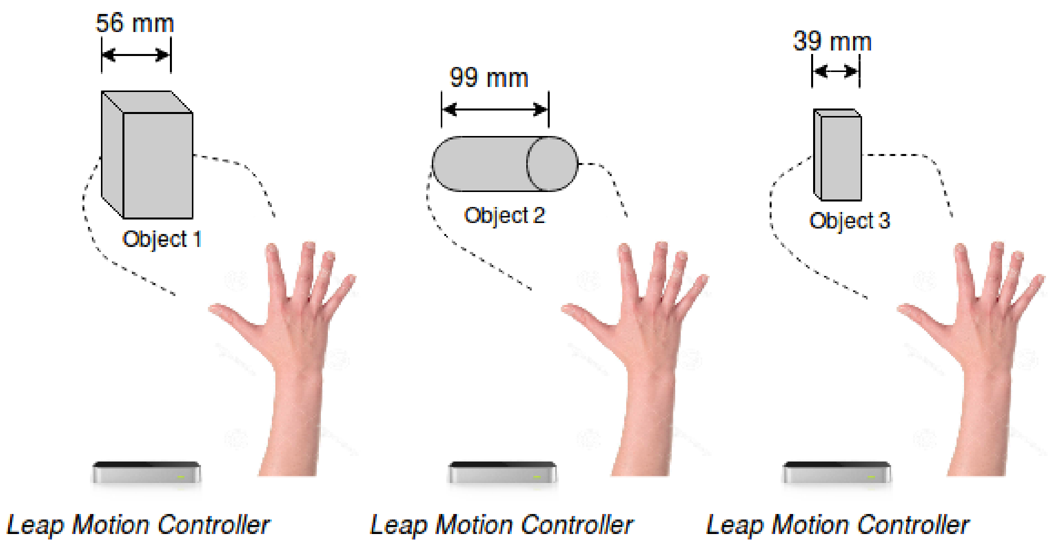
The grip aperture was estimated through the Euclidean distance between the index and thumb fingers. The goal of these tests was to compare the final estimated grip aperture of these fingers (when the object was grasped) with the objects lengths, in each case. In addition, the peak grip aperture, the time-to-peak grip aperture and the peak grip aperture velocity were evaluated. These tests were performed by one people.

Figure 25 – Validation protocol with Xsens. (a) Experimental protocol. (b) Sensors locations.



Fonte: Produção do próprio autor.

Figure 26 – Preliminary tests with Leap Motion Controller.

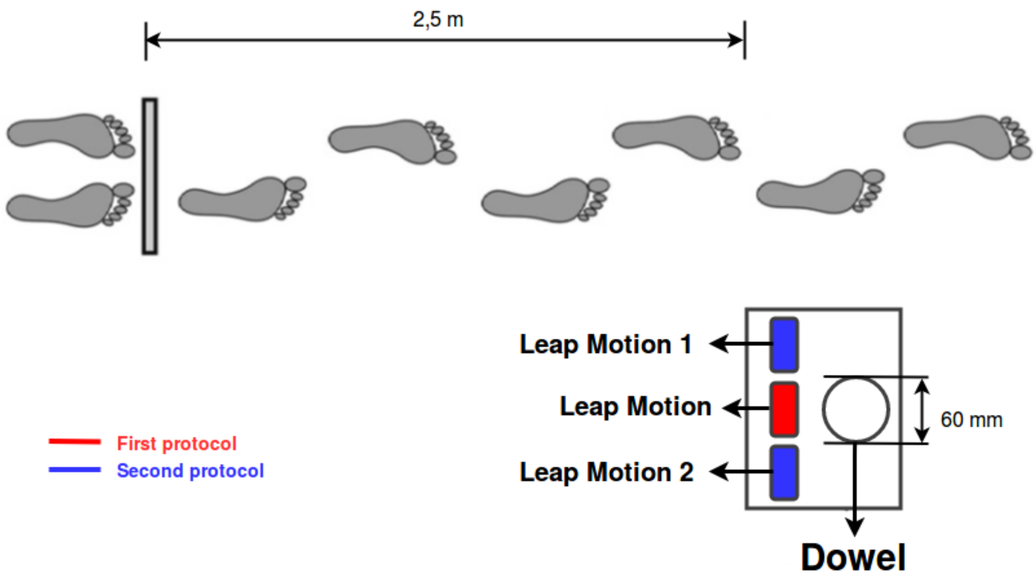


Fonte: Produção do próprio autor.

4.2.2 Dual-task Tests

The dual-task tests were performed with the participants walking and grasping the object simultaneously. These tests were divided in two protocols, in which the first one consisted in using only one Leap Motion Controller for data acquisition and the second one consisted in using two sensors, as shown in Figure 27. The purpose of these protocols was to evaluate the sensor response when the prehension task was performed with the moving participant and analyze if it is necessary to use more sensors.

Figure 27 – Dual-task tests with Leap Motion Controller.



Fonte: Produção do próprio autor.

Five young adults (3 females and 2 males, 24.0 ± 3.7 years old, 165 ± 4 cm height and 65.3 ± 10.5 kg mass) participated in these tests, as presented in Table 3. 5 trials were performed for each participant.

Table 3 – Informations about the participants of Dual-task tests.

Participants	Age (yrs)	Height (cm)	Weight (kg)
1	21	161	61
2	29	169	77
3	22	163	53
4	27	170	70
5	22	173	56

Fonte: Produção do próprio autor.

4.3 Clinical Validation

Twenty older adults participated in this work. They were distributed in two groups (n=10): older adults with no history of falls (Older adults, OA) (6 females and 4 males, 67.2 ± 4.9 years old, 160 ± 9 cm height and 68.6 ± 13.8 kg mass), and older adults who experienced at least one fall in the 12-month period before the data collection (Faller older adults, FOA) (6 females and 4 males, 68.0 ± 4.4 years old, 160 ± 9 cm height and 77.7 ± 15.7 kg mass), as shown in Table 4.

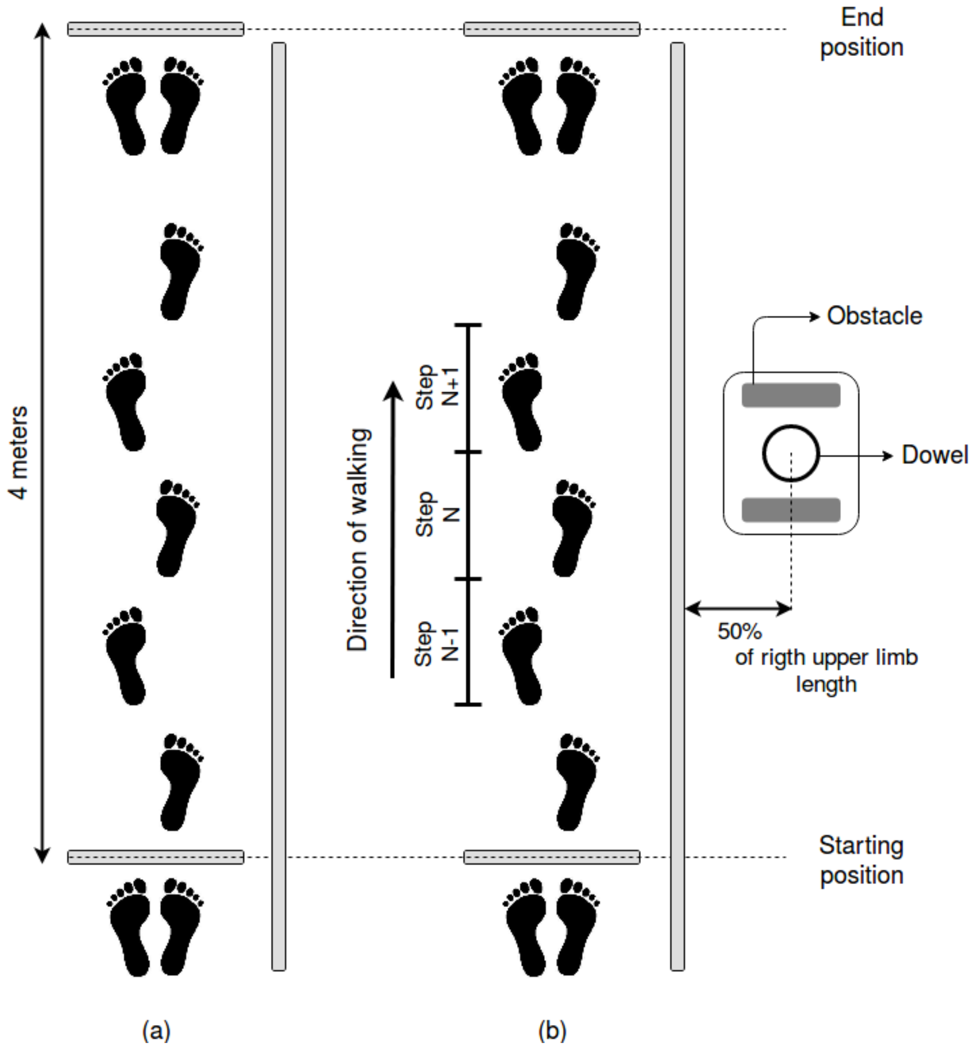
Table 4 – Informations about the participants of Clinical Validation.

	Participants	**Age (yrs)**	**Height (cm)**	**Weight (kg)**
Fallers	1	70	150	72.0
	2	65	152	64.2
	3	74	153	63.0
	4	74	162	62.0
	5	60	156	109.0
	6	69	151	75.2
	7	68	177	82.0
	8	63	174	99.0
	9	69	162	80.8
	10	68	158	69.8
Non-Fallers	1	72	151	67.0
	2	69	146	61.6
	3	74	166	79.1
	4	61	170	84.4
	5	65	153	82.0
	6	68	163	51.2
	7	64	172	87.0
	8	62	165	61.4
	9	63	164	64.2
	10	74	153	48.5

Fonte: Produção do próprio autor.

The experimental protocol, based on Rinaldi (2015), was divided in two conditions: walking through and gait combined with prehension, as shown in Figure 28. In the first condition, the participants should walk straight, at their self-selected pace, from the starting position to the end position. In the second condition, the participants should reach and grasp the object during the walking task. The object was placed on a support built especially for this study to allow for height adjustment. The support's height was adjusted to the participants' greater trochanter height and located approximately 2 m from the starting position, and the object was positioned laterally with a distance corresponding to 50% of the participant's right upper limb length. Participants performed three trials, completely randomized, for each condition.

Figure 28 – Experimental Protocol applied in fallers and non-fallers older adults.



Fonte: Produção do próprio autor.

5 Results and Discussion

This Chapter presents the results of the proposed experimental validations. Such the Chapter 4, the results and discussion were divided in the following sections about each experimental validation, besides the general discussion about all the results.

5.1 *Kinect v2* System Validation with a Commercial System

According to the experimental protocol presented in the Chapter 4, two trials were performed for each participant. Table 5 shows the stride length means and errors obtained in the two trials.

Table 5 – Stride length means by *Kinect v2 system* and *Xsens system*.

	Right stride			**Left stride**		
Older adults	Kinect v2 (cm)	Xsens (cm)	Error module (%)	Kinect v2 (cm)	Xsens (cm)	Error module (%)
1	97.98	98.75	0.77	102.13	97.47	4.79
2	107.51	111.03	3.17	116.10	108.72	6.78
3	114.49	103.53	10.98	105.01	106.56	1.46
4	104.49	101.24	3.22	103.04	98.14	4.99
5	112.89	108.76	3.79	107.70	108.97	1.16
6	105.27	105.83	0.52	107.10	101.87	5.14
7	121.72	116.59	4.40	125.07	116.17	7.66
8	107.76	102.20	5.44	100.87	98.03	2.89
9	115.02	111.07	3.57	117.37	107.04	9.64
10	99.83	103.97	3.99	100.72	94.50	6.58
11	124.64	118.25	5.40	131.08	128.00	2.40
12	107.79	120.28	10.39	120.94	125.49	3.63
13	106.90	100.19	6.70	107.16	99.30	7.91
14	115.15	111.67	3.12	104.68	104.74	0.06
15	127.69	117.80	8.40	124.10	117.72	5.42

Fonte: Produção do próprio autor.

Results showed a good correlation with a commercial system, presenting a mean error of 4.81% (4.92% for right stride length and 4.70% for left stride length). The higher error was 10.98% (10.96 mm), while the lower error was 0.06% (0.06 mm).

Through these results, the *Kinect v2 system* proved to be reliable, because the low errors when compared to a commercial system. In addition, the system was considered feasible for performing the clinical validation, considering that this protocol follows the pattern of the clinical validation, in the same conditions.

5.2 Leap Motion Controller Validation

5.2.1 Preliminary Tests

According to the experimental protocol presented in the Chapter 4, three tests were performed, one for each object with predetermined length. Table 6 shows the results obtained in these tests.

Table 6 – Prehension parameters of the preliminary tests.

Object	Object length (mm)	Peak grip aperture (mm)	Time-to-peak grip aperture (s)	Peak grip aperture velocity (mm/s)	Final grip aperture (mm)
1	56	74.0	1.27	58.3	53.8
2	99	112.7	2.69	41.8	99.2
3	39	70.4	1.76	39.9	39.2

Fonte: Produção do próprio autor.

The error module (em %) was calculated through the results of final grip aperture, as shown in Table 7. Figure 29 shows the grip aperture curves for each test. Results showed errors up to 3.9% (2.2 mm) and an error mean of 1.5%. The second and third tests presented error of 0.2 mm, revealing good performance of the sensor.

Table 7 – Grip aperture error (%).

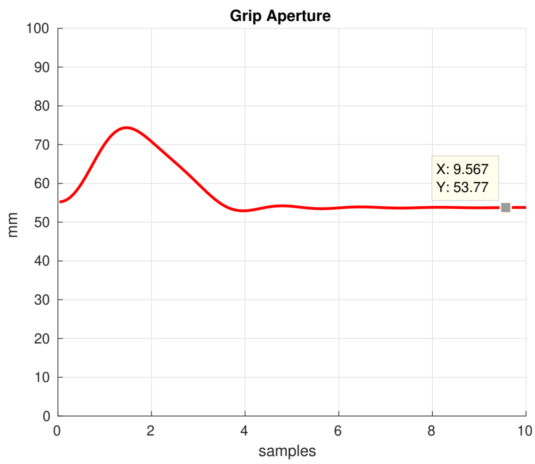
Object	Object length (mm)	Final grip aperture (mm)	Error module (%)
1	56	53.8	3.9
2	99	99.2	0,2
3	39	39.2	0,5

Fonte: Produção do próprio autor.

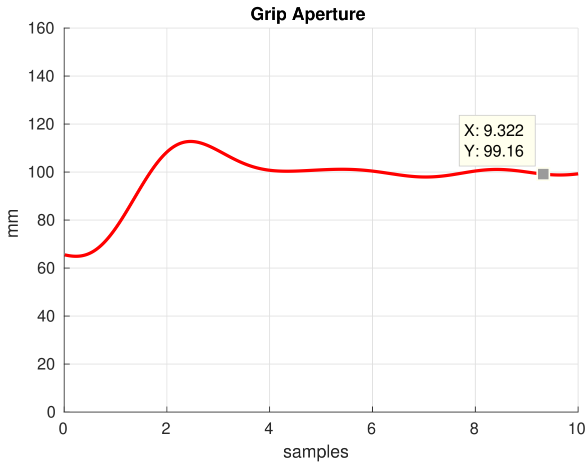
Through the results, it was possible to notice that there is an higher aperture before grasping the object, and after this peak the aperture stabilizes and is approximately equal to the object length, for each test. There is no initial aperture pattern, since it was not proposed a initial aperture. Thus, the hand began free in all the tests.

The preliminary tests were performed under static condition within the workspace, and were extremely important in the initial phase for the continuation of the work. Results showed that it was possible to obtain the hand, segments and finger joints 3-D positions, and thus obtain the prehension parameters, with low errors, presenting feasibility in the use of the *Leap Motion Controller*.

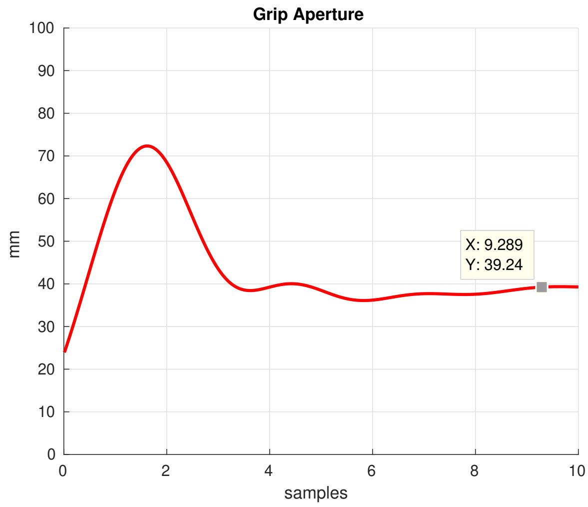
Figure 29 – Grip aperture curves. (a) Object 1. (b) Object 2. (c) Object 3.



(a)



(b)



(c)

Fonte: Produção do próprio autor.

5.2.2 Dual-task Tests

According to the Chapter 4, two protocols were defined. 25 tests (5 tests for each subject) were performed for each protocol. Table 8 shows the prehension parameters obtained during both protocols for each subject.

Results showed tests of the first protocol that obtained small errors, such as the subject 1's third test, which obtained 3.33% error, and showed tests with big errors, such as the participant 2's first test, which obtained 98.33% error. Five tests had acquisition failure.

Through the tests, the final grip aperture mean was 56 mm and the standard deviation, 33 mm, and the error module mean was 47.75%.

Figure 30 shows the hand grip aperture curves in three tests for the first protocol.

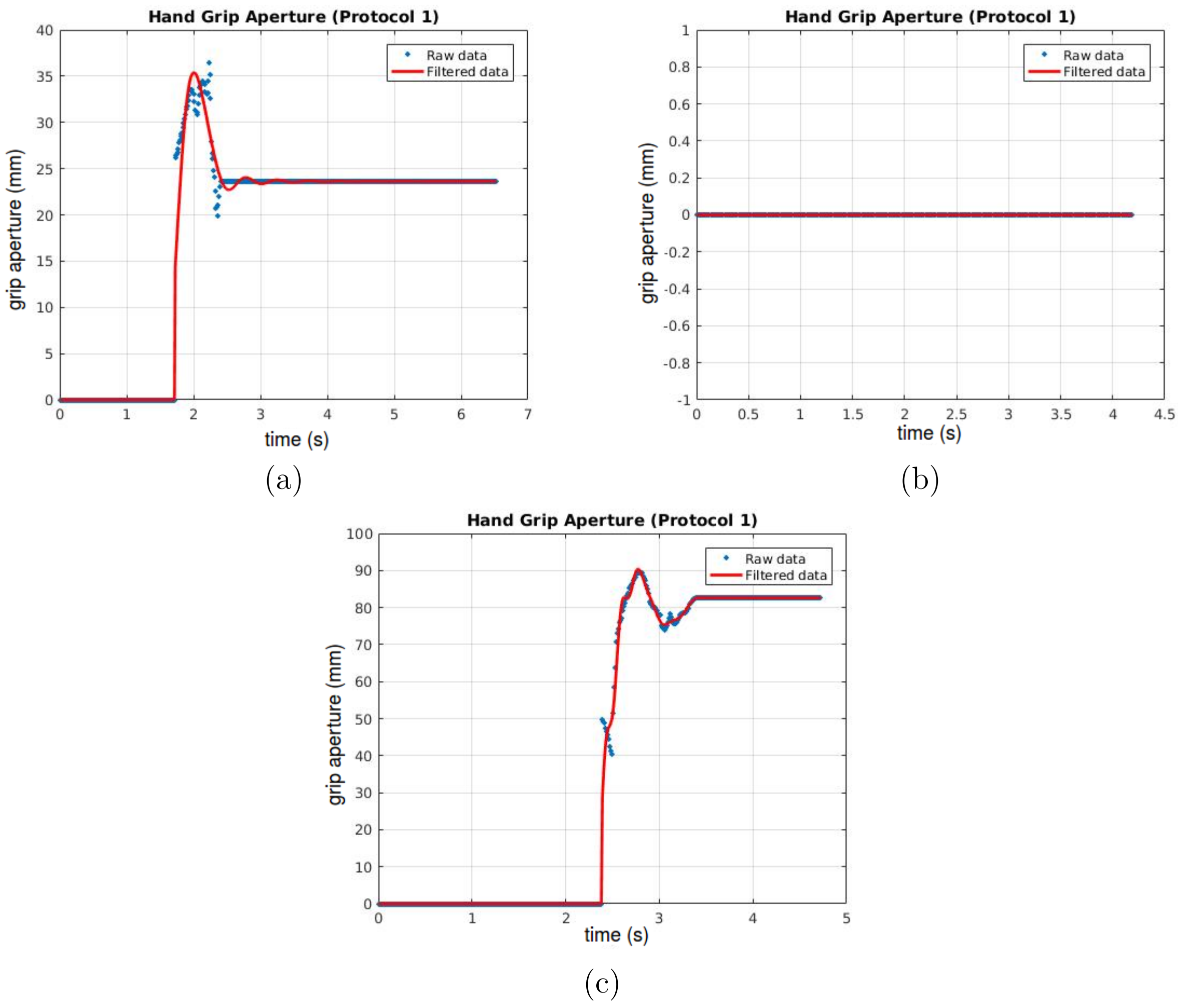
Table 8 – Results of each protocol for each subject.

	Tests	**Peak grip aperture (mm)**		**Final grip aperture (mm)**		**Error module (%)**	
		P1	P2	P1	P2	P1	P2
S1	1	97	85	24	57	60.00	5.00
	2	80	76	80	65	33.33	8.33
	3	72	78	62	65	3.33	8.33
	4	87	80	44	52	26.67	13.33
	5	80	64	19	46	68.33	23.33
S2	1	120	60	119	54	98.33	10.00
	2	92	104	88	42	46.67	30.00
	3	117	97	42	81	30.00	35.00
	4	119	70	20	51	66.67	15.00
	5	118	72	24	60	60.00	0
S3	1	98	97	84	83	40.00	38.33
	2	35	101	24	57	60.00	5.00
	3	92	92	82	61	36.67	1.67
	4	101	103	86	79	43.33	31.67
	5	84	92	77	72	28.33	20.00
S4	1	-	94	-	94	-	56.67
	2	-	86	-	47	-	21.67
	3	-	83	-	61	-	1.67
	4	106	81	9	73	85.00	21.67
	5	74	94	7	65	88.33	8.33
S5	1	-	119	-	82	-	36.67
	2	-	97	-	78	-	30.00
	3	94	100	78	69	30.00	15.00
	4	72	104	68	76	13.33	26.67
	5	90	94	82	65	36.67	8.33
	Mean ± SD			**56 ± 33**	**65 ± 13**	**47.8**	**18.9**

Fonte: Produção do próprio autor.

The first curve is the subject 3's second test and has 60% error. The second curve is the subject 4's first test and had acquisition failure. Lastly, the third curve is the subject 5's last test and has 36.67% error.

Figure 30 – Grip aperture curves of experimental protocol 1. (a) 60% error curve. (b) Acquisition failure. (c) 36.67% error curve.



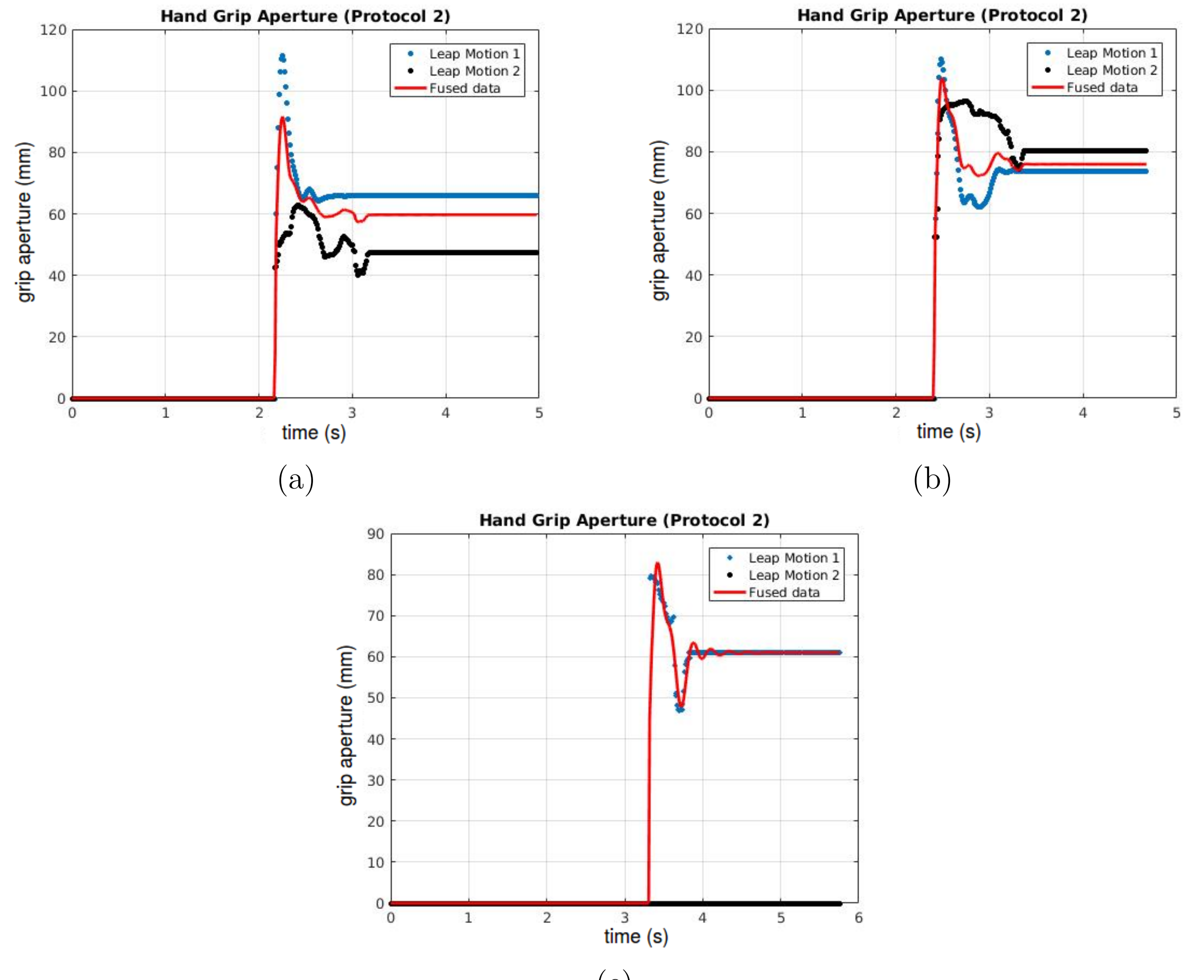
Fonte: Produção do próprio autor.

Results of the first experimental protocol showed acquisition failure in 5 tests and inconsistency of most obtained parameters. Moreover, the obtained standard deviation were high, which is 50% of the mean values presented in Table 8, showing low repeatability, also, the error module mean was high (47.75%), showing limitations of the use of the Leap Motion Controller in dynamic applications such as the one presented in this work. The results revealed the inconsistent performance of the controller in the dynamic scenario, as Guna et al. (2014) has shown in his work.

Figure 31 shows the hand grip aperture curves in three tests for second protocol. The red curve is fused data (Kalman filter (KHALEGHI et al., 2013)). The first curve is the subject 3's third test and presents an error of 1,67%. The second curve is the subject

5's fourth test and presents an error of 20%. The third curve is the subject 4's third test and presents an error of 1,67%. In last curve, the Leap Motion Controller 2 failed, however the Leap Motion Controller 1 performed the acquisition and obtained a result with low error.

Figure 31 – Grip aperture curves of experimental protocol 2. (a) 1.67% error curve. (b) 26.67% error curve. (c) 1.67% error curve with *Leap Motion* 2's acquisition failure.



Fonte: Produção do próprio autor.

Results showed 100% acquisition success for the second protocol. Furthermore, results showed tests of the second protocol that obtained small errors, such as the participant 2's last test, which obtained an aproximate error of 0%.

Through the tests, the final grip aperture mean was 65 mm and the standard deviation, 13 mm, and the error module mean was 18,87% for second protocol. The standard deviation and the error module mean of second protocol was much lower than first one, showing greater accuracy and repeatability.

Results of the second experimental protocol showed no acquisition failure, since when one Leap Motion Controller was not able to detect correctly the hand landmark, the other usually performed the acquisition correctly. Thus, 100% acquisition was ensured. Furthermore, the error mean and standard deviation of the second protocol was smaller than the first protocol, showing that the system is more accurate and presents measures more consistent.

This experimental validation presented a technique to combine two *Leap Motion Controller* devices to improve the reliability of data acquisition for dynamic tests. Since one difficulty of the Leap Motion Controller is non-constancy acquisition in the dynamic scenario, two Leap Motion devices are employed for the field of view increase and of the chances of acquiring the data. In addition, the data is fused, which result in an increase of accuracy. Results showed that the use of two Leap Motion Controller decrease the acquisition errors in dynamic tests and reduces the chances of acquisition failures.

5.3 Clinical Validation

According to the Chapter 4, two protocols were divided in two conditions, "walking through" and "dual-task". Twenty older adults (n=10) performed one test for each protocol. Results were divided in spatio-temporal gait parameters, CoM velocity and prehension parameters. In the next sub-sections, this parameters will be presented and discussed.

5.3.1 Spatio-temporal gait parameters

Interaction between groups:

Figure 32 shows the step and stride length during the *"walking through"* condition. The right step and stride length are 54.58 cm and 111.92 cm for FOA, and 57.61 cm and 118.19 cm for OA. While the left step and stride length are 56.47 cm and 111.59 cm for FOA, and 59,92 cm and 120.51 cm for OA. The biggest difference between FOA and OA was in the left step (3.45 cm) and left stride (8.92 cm).

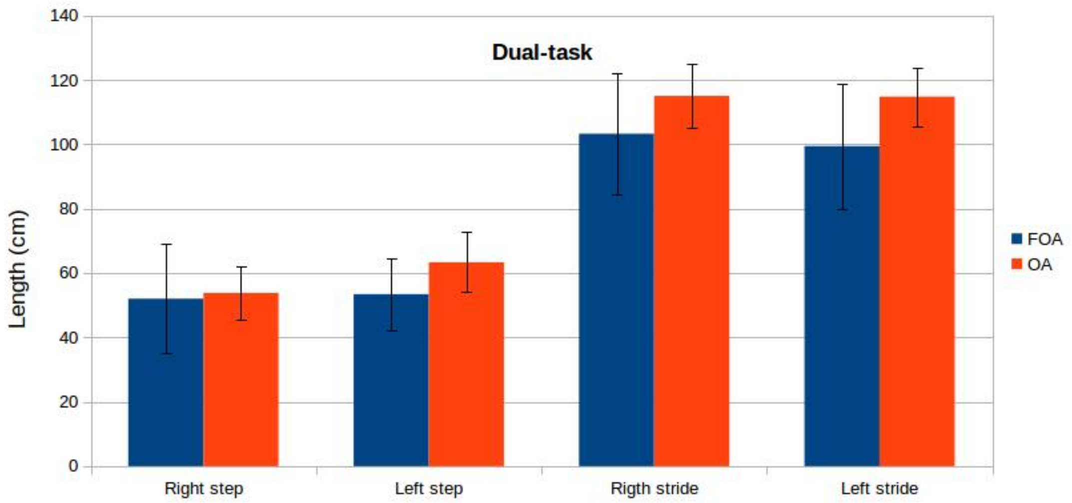
Both step and stride length are lower in the FOA group. This shows that even in the easiest task, the FOA are more cautious during the walking task. According to Rinaldi e Moraes (2016), the FOA present some gait impairments, such as stride length and an increase in gait variability.

Figure 33 shows the step and stride length during the *"dual-task"* condition. The difference between FOA and OA is higher than the *"walking through"* condition. The right step and stride length are 51.99 cm and 103.23 cm for FOA, and 53.76 cm and 115.00 cm for OA. While the left step and stride length are 53.40 cm and 99.42 cm for FOA, and 63,31 cm and 114.77 cm for OA. The most significant difference between FOA and OA was in the left step (9.95 cm) and left stride (15.35 cm).

Figure 32 – Step and stride length during *"walking through"* condition.

Fonte: Produção do próprio autor.

Figure 33 – Step and stride length during *"dual-task"* condition.



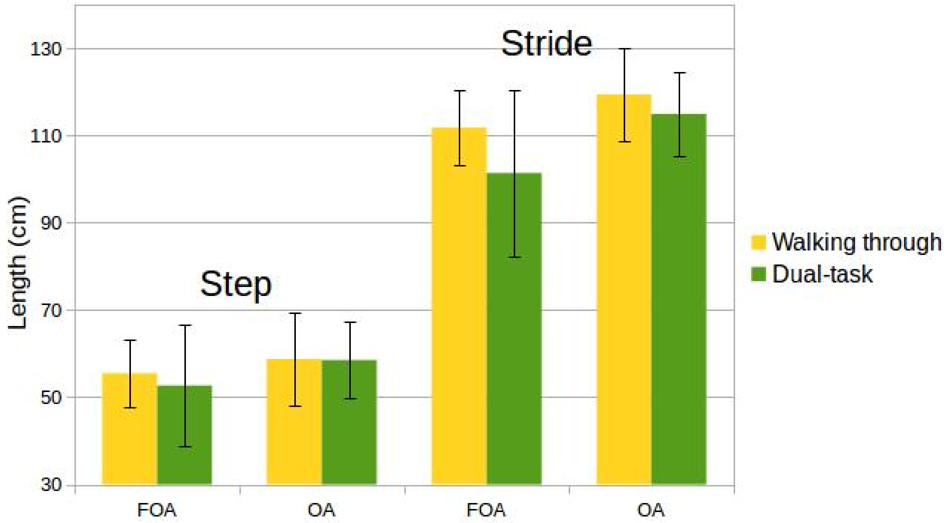
Fonte: Produção do próprio autor.

In the *"dual-task"* condition, the step and stride length differences between the groups are higher. This is due to the changes in the FOA walking pattern are more evident when two motor tasks are combined (RINALDI; MORAES, 2016). Hall et al. (2011) suggested that the more difficult the secondary task is, the greater the impact on gait performance. In addition, the FOA standard deviation is also higher, which means greater variability in this group, presenting FOAs with more difficulty than others.

Interaction between conditions:

Figure 34 shows the interaction between conditions for each group, through the step and stride length average. It is evident that the addition of the other motor task does not effectively affect the step and stride length of the OA group, presenting almost the same step length average (**WT:** 58.77 cm / **DT:** 58.54 cm) and difference up to 4.46 cm in the stride length average (**WT:** 119:35 cm / **DT:** 114:89 cm) between the two conditions.

Figure 34 – Interaction between conditions for step and stride length.



Fonte: Produção do próprio autor.

The FOA group presented difference between the two conditions in the step length average (**WT:** 55.51 cm / **DT:** 52.69 cm). The difference is even more evident in the stride length average, with value up to 10.45 cm (**WT:** 111:78 cm / **DT:** 101:33 cm). The high standard deviation is related to high step and stride length variability, since there is a big difference between lengths because the step at the moment of the prehension.

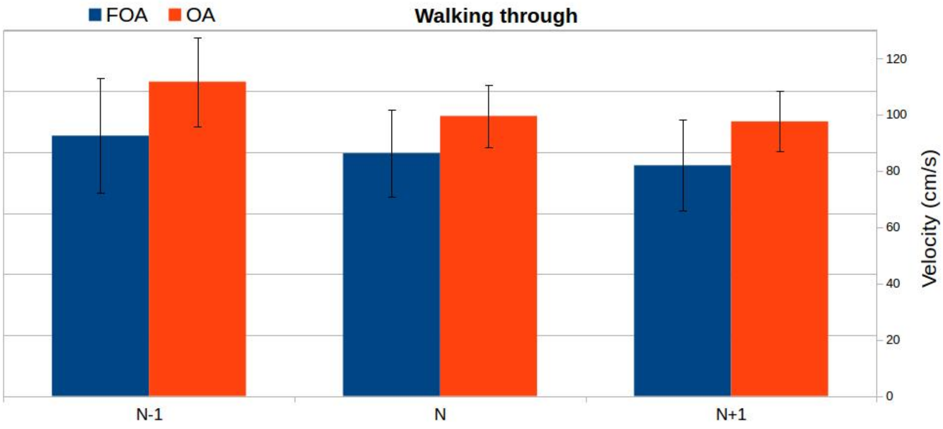
5.3.2 Center of mass (CoM) velocity

Interaction between groups:

Figure 35 shows the CoM velocity in the steps (N-1, N and N+1) during the *"walking through"* condition. Both group performed the three steps with low standard deviation in the velocities (**FOA:** 5.31 cm/s ; **OA:** 7.65 cm/s). The higher variation in FOA group is 11.41% between the (N-1) and (N+1) steps and in the OA group is 12.65%, also between the (N-1) and (N+1) steps. It reveals that both groups start with high velocity and decrease it as they are close to the end position, in the *"walking through"* condition.

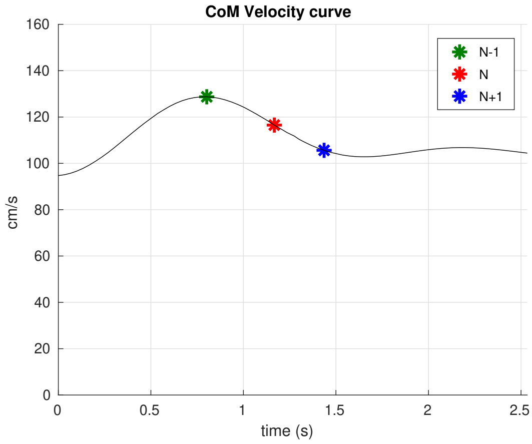
Figure 36 shows the CoM velocity of one of each group in the *"walking through"* condition. Both presented a decrease of the velocity from the start to end.

Figure 35 – CoM velocity during *"walking through"* condition for each step.

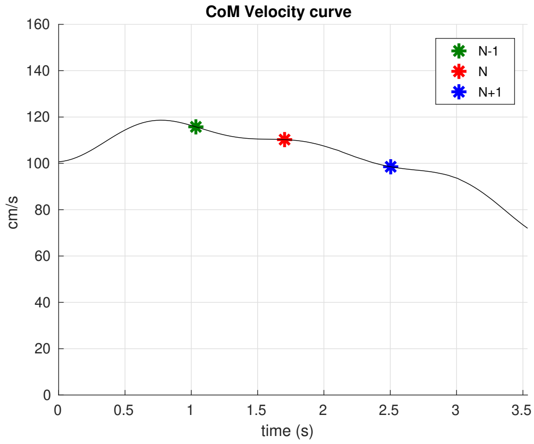


Fonte: Produção do próprio autor.

Figure 36 – CoM velocity curves in the *"walking through"* condition. (a) FOA. (b) OA.



(a)



(b)

Fonte: Produção do próprio autor.

Figure 37 shows the CoM velocity in the steps (N-1, N and N+1) during the *"dual-task"* condition. It is noticeable that the FOA have a more significant decrease of the CoM velocity in the step N (at the moment of the prehension) when compared to the OA group. The FOA group presented a CoM velocity standard deviation of 9.34 cm/s, while the OA group presented 1.75 cm/s.

In addition, the biggest variation in the FOA group is 21.75% between the (N) and (N+1) steps and in the OA group is 3.92%, between the (N-1) and (N) steps. It reveals that the FOA group start with more attention and have a lower velocity in the step N, and after the prehension moment, the velocity increase again (**N-1:** 76.98 cm/s ; **N:** 62.24 cm/s ; **N+1:** 79.54 cm/s). While the OA group keep the velocity almost equal in the three steps (**N-1:** 89.47 cm/s ; **N:** 85.97 cm/s ; **N+1:** 87.81 cm/s).

Figure 37 – CoM velocity during *"dual-task"* condition for each step.

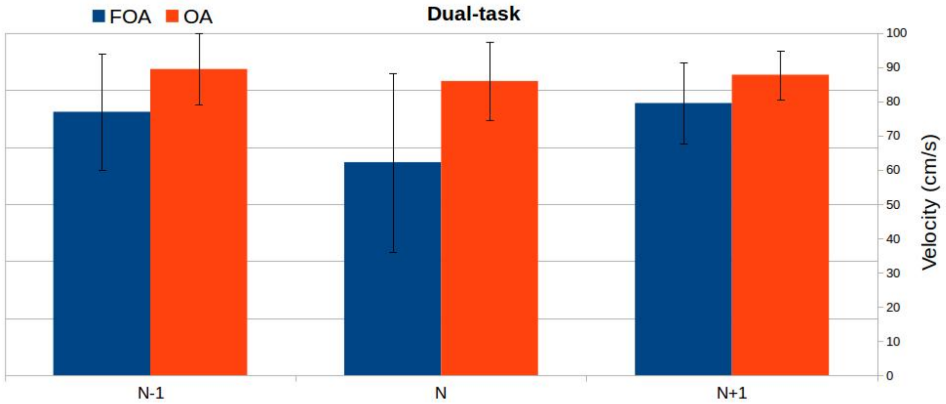
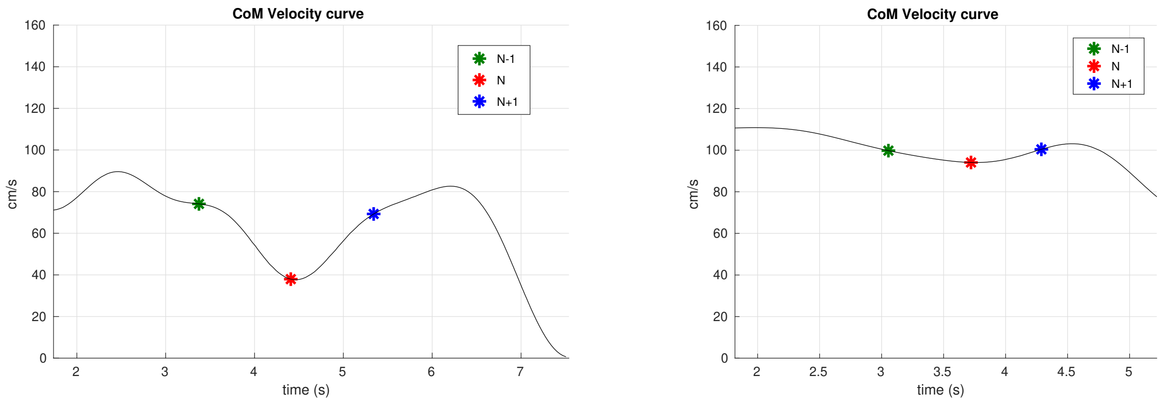


Figure 38 shows the CoM velocity of one of each group in the *"dual-task"* condition. It is noticeable that the CoM velocity variation between the steps is higher for the FOA, presenting the velocity decrease of 43.73% in the step **N** (prehension step). While for the OA, the decrease is 5.70%.

Figure 38 – CoM velocity curves in the *"dual-task"* condition. (a) FOA. (b) OA.



(a) (b)

Fonte: Produção do próprio autor.

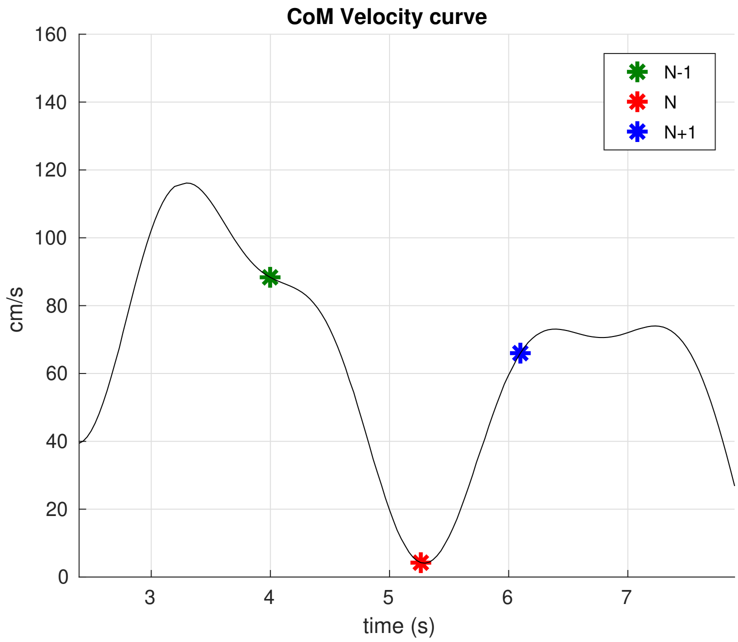
It is important to emphasize that one FOA almost stopped to perform the prehension task. This reveals a more conservative walking strategy and decoupled the combined task, as shown in Rinaldi (2015). Figure 39 shows the CoM velocity curve of this FOA.

Figure 40 shows the CoM mean velocity during all the walking. In both conditions the CoM mean velocity of the FOA are lower when compared to the OA group. This result is similar with previous studies (RINALDI; MORAES, 2016; TOEBES et al., 2012).

Interaction between conditions:

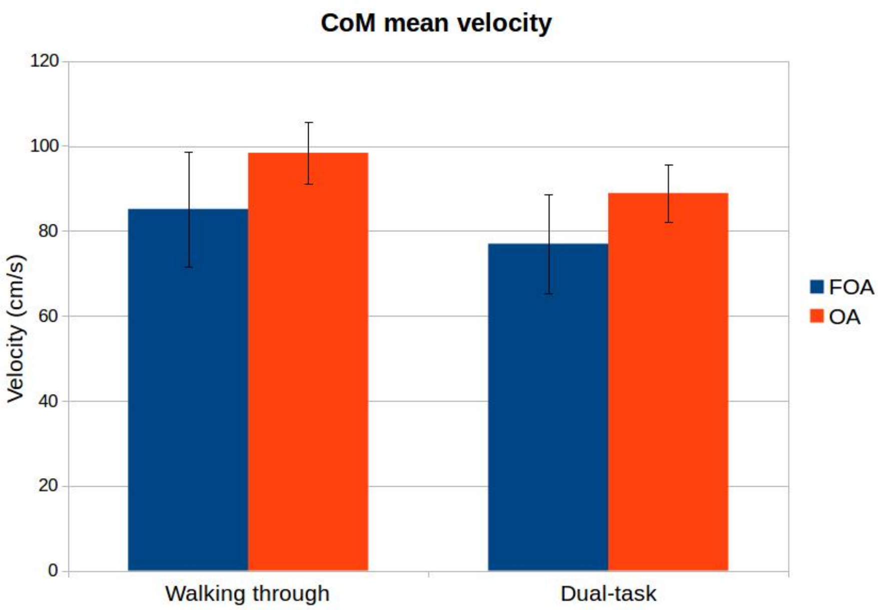
Table 9 and Figure 41 show the CoM mean velocity of each group in each condition. The *"walking through"* condition presented higher CoM velocity average in the steps (up

Figure 39 – CoM velocity curve of a faller older adult during the *dual-task* condition.



Fonte: Produção do próprio autor.

Figure 40 – CoM mean velocity during both conditions for each group.



Fonte: Produção do próprio autor.

to 111.68 cm/s for FOA in the step **N-1**). The higher CoM velocity variation between conditions of the FOA group is 27.84% (step **N**), while the variation of the OA group is 19.88% (step **N-1**).

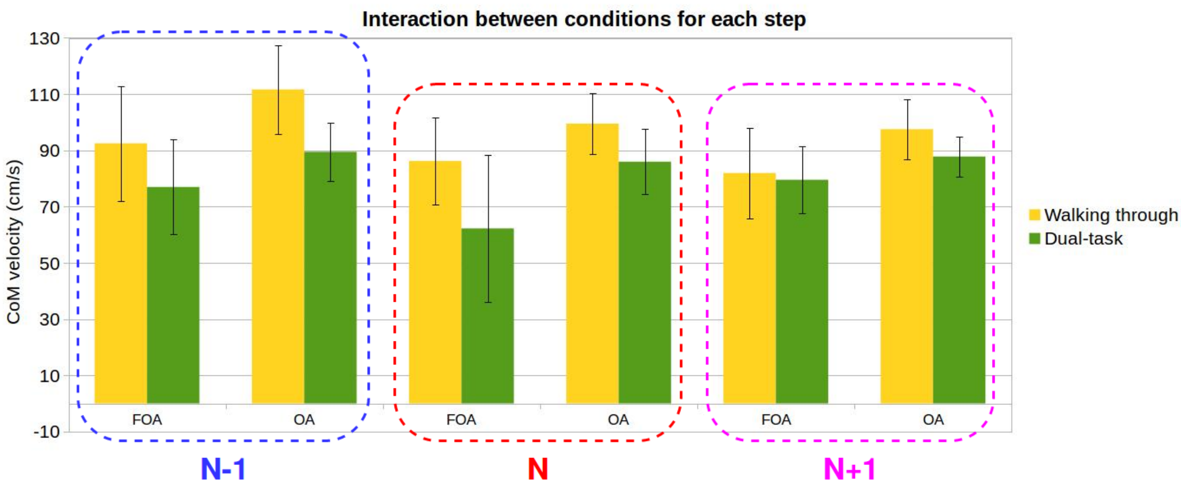
It reveals that the *"dual-task"* condition affect more the velocity in the FOA group,

Table 9 – CoM velocities for both groups in each step and each condition.

Groups	**FOA**			**OA**		
Steps	N-1	N	N+1	N-1	N	N+1
Walking through	92.52	86.25	81.96	111.68	99.51	97.56
Dual-task	76.98	62.24	79.54	89.47	85.98	87.81

Fonte: Produção do próprio autor.

Figure 41 – Interaction between conditions for each group during steps (N-1, N and N+1).



Fonte: Produção do próprio autor.

since the lower velocity is at the prehension moment, showing a difficulty to keep the walking speed and the attention to grasping an object.

5.3.3 Prehension parameters

The prehension parameters were divided in reaching and grasping variables. The reaching variables are: movement time, peak wrist velocity and time-to-peak wrist velocity. While the grasping variables are: peak grip aperture, time-to-peak grip aperture, peak grip aperture velocity and time-to-peak grip aperture velocity.

Reaching variables

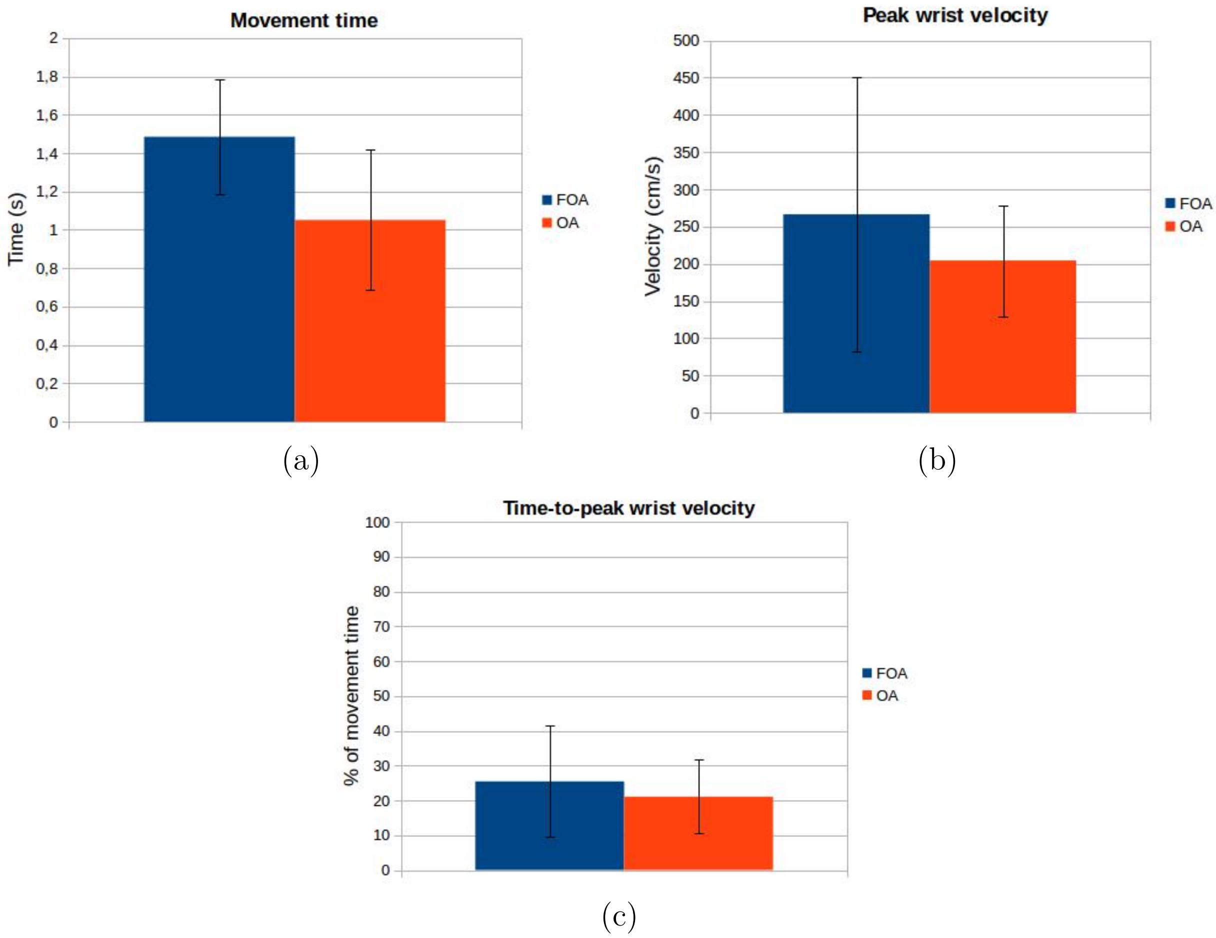
Table 10 and Figure 42 present the reaching variables for both groups. It is possible to observe that the FOA group presented slower movement time (0.44 seconds higher than the OA group). This slowness suggests that FOA need more time to gain sensory information to accomplish the manual task successfully, as has been suggested for walking tasks (CHAPMAN; HOLLANDS, 2007), as shown in Rinaldi e Moraes (2016). Figure 43 shows the CoM velocity with the steps (N-1, N and N+1), *reaching onset* and *dowel* *contact* moments of two older adults (FOA and OA).

Table 10 – Reaching variables.

Group	**Movement time (s)**	**Peak wrist velocity (cm/s)**	**Time-to-peak wrist velocity (%)**
FOA	1.49 ± 0.30	266.37 ± 183.78	25.46 ± 16.01
OA	1.05 ± 0.37	204.34 ± 74.70	21.08 ± 10.62

Fonte: Produção do próprio autor.

Figure 42 – Reaching variables for each group. (a) Movement time. (b) Peak wrist velocity. (c) Time-to-peak wrist velocity.



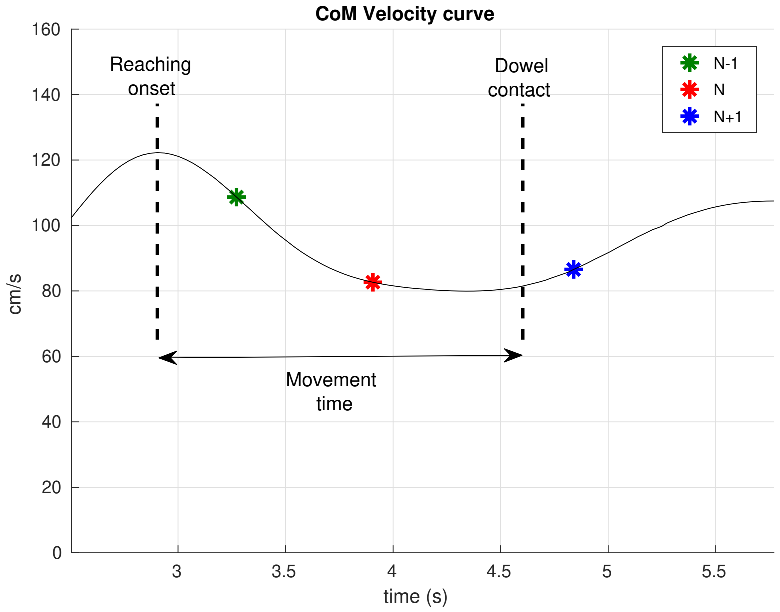
(a) (b)

(c)

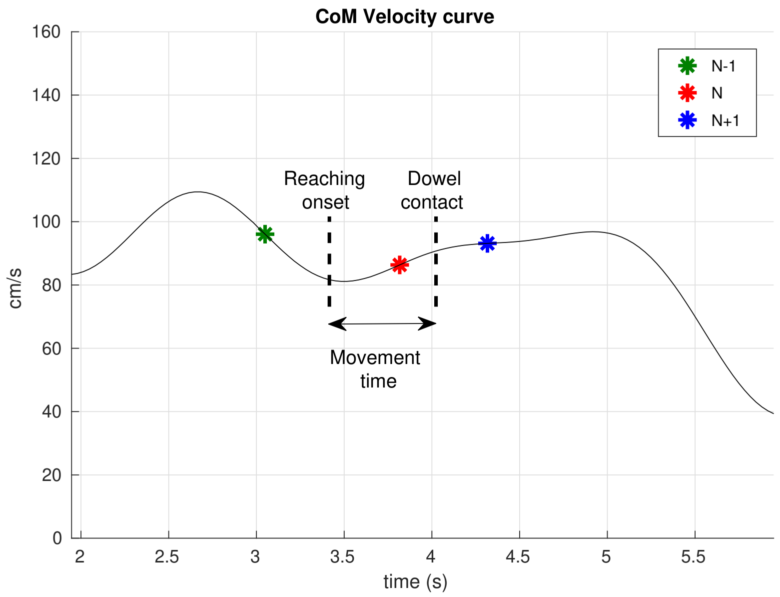
Fonte: Produção do próprio autor.

In addition, the peak wrist velocity average is also higher for FOA group. However, it is due to two FOAs that presented values much higher than the peak wrist velocity average (57.41% and 167.62% higher). Excluding these two FOAs of the group, the new peak wrist velocity average is 191.44 ± 72.05 cm/s, which is lower than the OA group (204.34 ± 74.70 cm/s), as shown in Rinaldi e Moraes (2016). However, this variation of the

Figure 43 – CoM velocity curves in the *"dual-task"* condition with *reaching onset* and *dowel contact* moments. (a) FOA. (b) OA.



(a)



(b)

Fonte: Produção do próprio autor.

groups (6.31%) is not enough when compared to the standard deviation (35.26%), which makes this parameter not effective for comparison between FOA and OA group. Besides this, the time-to-peak wrist velocity parameter presented low variation between groups (4.38%), which also makes this parameter not effective for comparison between FOA and OA group.

Grasping variables

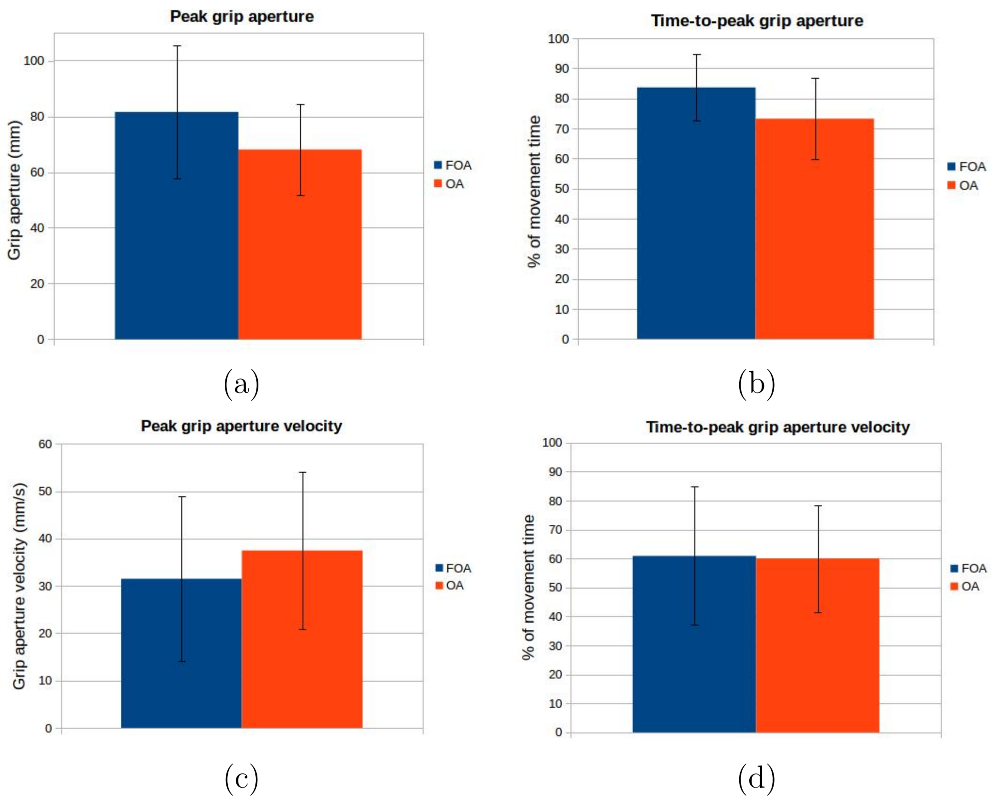
Table 11 and Figure 44 present the reaching variables for both groups. The peak grip aperture average for FOA group was higher than OA group, as expected since the FOA group presents a conservative performance in the tasks. The variation between the peak grip aperture average of the groups is 13.48 mm (16.53%). Different from Rinaldi e Moraes (2016), the peak grip aperture parameter was affected by group.

Table 11 – Grasping variables.

Group	Peak grip aperture (mm)	Time-to-peak grip aperture (%)	Peak grip aperture velocity (mm/s)	Time-to-peak grip aperture velocity (%)
FOA	81.56 ± 24.02	83.64 ± 10.96	31.50 ± 17.37	60.92 ± 23.85
OA	68.08 ± 16.22	73.23 ± 13.56	37.47 ± 16.67	60.06 ± 18.45

Fonte: Produção do próprio autor.

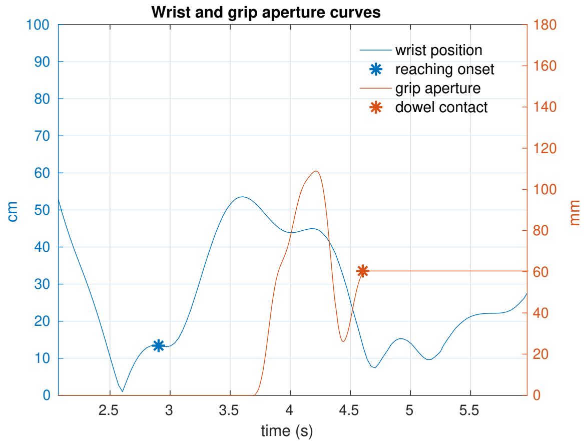
Figure 44 – Grasping variables for each group. (a) Peak grip aperture. (b) Time-to-peak grip aperture. (c) Peak grip aperture velocity. (d) Time-to-peak grip aperture velocity.



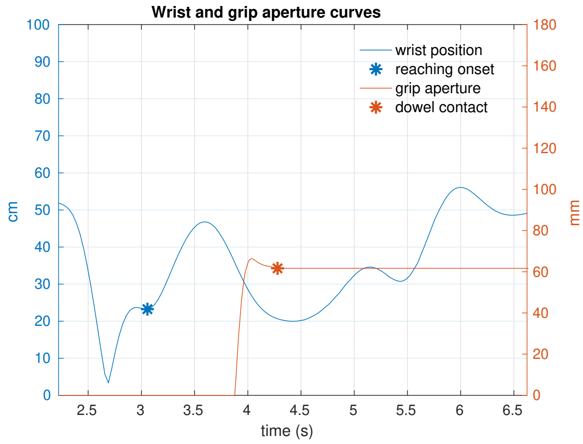
Fonte: Produção do próprio autor.

Figure 45 shows the wrist and grip aperture curves, showing the *reaching onset* and *dowel contact* moments, besides being able to see the peak grip aperture moment.

Figure 45 – CoM velocity curves in the *"dual-task"* condition with *reaching onset* and *dowel contact* moments. (a) FOA. (b) OA.



(a)



(b)

Fonte: Produção do próprio autor.

The time-to-peak grip aperture average was higher for FOA group. This result was not expected, since the FOA group was expected to anticipate more the peak grip aperture when compared to OA group. This result may indicate that the FOA group change the control of grasping when performing the gait, allocating more attention for dynamic stability control. Consequently, the time to adjust the hand configuration decreased, which indicates a prioritization of the primary task (walking) (RINALDI; MORAES, 2016).

The peak grip aperture velocity and the time-to-peak grip aperture velocity parameters were not conclusive, since the variation was low (15.93% and 0.85%, respectively) when compared to their standard deviation values (44.42% and 23.85%, respectively).

5.3.4 General Discussion

Throughout this clinical validation was possible to notice the change of some parameters in the interaction between the FOA and the OA groups, besides the interaction between *walking through* and *dual-task* conditions.

When comparing the groups, it was possible to observe variations in the step and stride length, and CoM mean velocity in the walking task in both conditions. All these parameters were lower in FOA group, showing the adoption of a more conservative walking strategy. In the *dual-task* condition, the changes in the walking pattern were even more evident, due to the influence of the combination of two motor tasks. In addition, there is variation also in some prehension parameters, such movement time, peak grip aperture and time-to-peak grip aperture. The movement time was slower for FOA group, showing that FOA need more time to divide the attention to perform two tasks simultaneously. The time-to-peak grip aperture happened later when compared to OA group, which may suggest that the FOA group change the control of grasping when perform the gait, allocating more attention for dynamic stability control and, consequently, decreasing the time of the hand configuration.

When comparing the conditions, it was possible to observe that both groups were influenced due to the addition of the other motor task to the walking task. The step and stride length, and the CoM mean velocity decreased during *dual-task* condition for the both groups. For the FOA group, the CoM velocity abruptly decreased in the step **N** (prehension moment), when compared to *walking through* condition.

Comparing the results obtained in this work with Rinaldi e Moraes (2016), it was possible to observe similarities. Table 12 shows a comparative analysis of each parameter through the interaction between groups, obtained by each work. It is important to emphasize that the sample characteristics of Rinaldi e Moraes (2016) (**FOA:** 155 cm and 65.7 kg / **OA: 154 cm and 59.9 kg**) are different from the sample characteristics of this work (**FOA:** 160 cm and 77.7 kg / **OA: 160 cm and 68.6 kg**), besides the fall level from FOA group. Therefore the data were compared by variation percentage between the groups.

Table 12 – Comparative between this work and Rinaldi e Moraes (2016).

	Comparative works	**This work**		**Rinaldi and Moraes (2016)**	
	Groups	FOA	OA	FOA	OA
Gait parameters	Step length	A decrease for dual-task when compared to walking through (3 cm).	Almost the same for both conditions (58 cm).	A decrease for dual-task when compared to walking through (11 cm).	Almost the same for both conditions (58 cm).
	CoM mean velocity	A decrease for dual-task when compared to walking through (30%), with adults almost stopping.	A decrease for dual-task when compared to walking through (20%).	A decrease for dual-task when compared to walking through (60%), with adults almost stopping.	A decrease for dual-task when compared to walkingthrough (30%).
Reach and Grasp parameters	Movement time	1.49 seconds	1.05 seconds	1.57 seconds	1.21 seconds
	Peak wrist velocity	FOA lower than OA, after exclude 2 outliers (6.31%)		FOA lower than OA (12.62%)	
	Time-to-peak wrist velocity	It was unaffected by group		It was unaffected by group	
	Peak grip aperture	FOA higher than OA (16.53%)		It was unaffected by group	
	Time-to-peak grip aperture	83.64%	73.23%	73.2%	86.8%
	Peak grip aperture velocity	OA higher than FOA (20%)		OA higher than FOA (15.93%)	
	Time-to-peak grip aperture velocity	It was unaffected by group		37.2%	47.3%

Fonte: Produção do próprio autor.

6 Conclusions and Future Works

This work presented a development of alternative camera-based system for *dual-task* analysis in older adults. Since the commercial motion capture systems are high cost or invasive and may disturb the movement performance, accessible cameras were used in this work. *Kinect v2* and *Leap Motion Controller* devices were integrated in order to analyze the *dual-task paradigm* in faller and non-faller older adults.

Firstly, experimental validations were performed in order to evaluate the feasibility and reliability of the selected sensors. The *Kinect v2 system* was compared to a commercial system and presented low variations. While the *Leap Motion Controller system* was evaluated in static and dynamic scenarios and different conditions, using one or two devices for dynamic scenario. It was possible to improve the results through the use of two devices, increasing the chance of acquisition and improving the accuracy merging the data.

Lastly, a clinical validation was applied on twenty older adults (n=10), divided in two groups (faller and non-faller older adults), in order to assess the walking and prehension performance, evaluating the interaction between groups and between conditions.

Results showed similarity with previous studies. History of falls affected the walking and prehension movements. FOA exhibited a more conservative strategy in the walking task, and allocated more attention for dynamic stability control when two motor tasks were combined, however, in *dual-task* condition the perfomance was even more affected. Besides this, the addition of the motor task influenced the performance of both groups.

The system architecture allows the integration of more sensors to expand this application to others scenarios. Future works involve the application of these protocols in more fallers and non-fallers older adults, and possibly older adults with some pathology. The addition of more parameters and more conditions will be studied, beside the improvement of materials and methods. The goal is to make this system as an assessment tool for fall risk prediction.

6.1 Contributions

The main contribution of this dissertation is the development of an alternative non-wearable system, with lower-cost sensors when compared with the commercial motion capture systems, for use in the analysis of balance and cognitive impairments through *dual-task paradigm* in older adults.

Other contribution involve the use of this system for research of Postgraduate Program in Physical Education (UFES). The research also assess faller and non-faller older adults, however involve more conditions and parameters.

6.2 Publications

The work presented in this dissertation originated the following publications:

Conference Proceedings:

- **Avellar, L. M.**, Loureiro, T., Rinaldi, N. and Frizera, A.. Proposta de uma Rede Heterogênea de Sensores para Análise de Marcha combinada com Tarefa de Preensão. Proceedings of the IX Congreso Iberoamericano de Tecnología de Apoyo a la Discapacidad - IBERDISCAP 2017, 101–107. Bogotá, Colombia, 2017.
- **Avellar, L. M.**, Rinaldi, N., Bastos, T., and Frizera, A.. Development of multisensor system for dynamic analysis of prehension tasks. Proceedings of the XXVI Congresso Brasileiro de Engenharia Biomédica - CBEB 2018. Búzios, Brasil, 2018.

Journal:

- Valencia-Jimenez, N., Leal-Junior, A., **Avellar, L.**, Valencia-Vargas, L., Caicedo-Rodríguez, P., Ramírez-Duque, A. A., Lyra, M., Marques, C., Bastos, T. and Frizera, A., A Comparative Study of Markerless Systems Based on Color-Depth Cameras, Polymer Optical Fiber Curvature Sensors, and Inertial Measurement Units: Towards Increasing the Accuracy in Joint Angle Estimation. Eletronics, v.8, n.173, p.1–21, 2019.

It is important to emphasize that there is an accepted work entitled “Development of camera-based system for analysis of dual task in elderly: gait combined with prehension” and will be presented at the International Workshop of Assistive Technology - IWAT 2019 on February 18-20. The authors are **Avellar, L. M.**, Valentino, J., Frizera, A. and Rinaldi, N..

In addition, it is also important to emphasize that there is an accepted work entitled “Analysis of center of mass velocity during dual-task in fallers and non-fallers elderly: gait combined with prehension task during avoidance of an obstacle” and will be presented at the International Society of Posture & Gait Research World Congress - ISPGR 2019 on June 30 - July 4. The authors are Rinaldi, N., Valentino, J., **Avellar, L. M.** and Frizera, A..

Some works were published about other topics and as a consequence of the interaction with other researchers during the development of this work.

Conference Proceedings:

- **Avellar, L. M.**, Leal-Junior, A. G., Botelho, T. and Frizera, A.. Sistema embarcado para análise de parâmetros cinemáticos e cinéticos do pé durante a marcha. Proceedings od the XIII Simpósio Brasileiro de Automação Inteligente - SBAI 2017, 1153–1158. Porto Alegre, Brasil, 2017.
- Leal-Junior, A. G., **Avellar, L. M.**, Frizera, A. and Pontes, M. J.. Aplicação de fibras ópticas poliméricas em um sistema portátil para detecção de eventos da marcha. Proceedings of the VI Encontro Nacional de Engenharia Biomecânica - ENEBI 2018. Águas de Lindóia, Brasil, 2018.

Journal:

- Leal-Junior, A. G., Frizera, A., **Avellar, L. M.** and Marques, C.. Polymer Optical Fiber for In-Shoe Monitoring of Ground Reaction Forces During the Gait. IEEE Sensors Journal, v.18, n.6, p.2362–2368, 2018.
- Leal-Junior, A. G., Frizera, A., **Avellar, L. M.** and Pontes, M. J.. Design considerations, analysis, and application of a low-cost, fully portable, wearable polymer optical fiber curvature sensor. Applied Optics, v.57, n.24, p.6927, 2018. ISSN 1559-128X. Disponível em: <https://www.osapublishing.org/abstract.cfm?URI=ao-57-24-6927>.

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