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Validation of EMG Envelope for The Clinical Assessment of Muscle Activity Peak During Walking

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Introduction

Electromyography (EMG) examines the combined electric signal that muscles emit when they contract under the supervision of the nervous system. The EMG signal characterizes the anatomical and physiological distinctive characteristics of muscles and reflects the number and firing pattern of muscles fibers involved during contraction. There are two main approaches to the acquisition of EMG signal: surface EMG (sEMG) and intramuscular EMG (IEMG). IEMG involves inserting a needle electrode into the muscle of interest to get a precise and accurate measurement of the electrical activity of individual muscle fibers. However, IEMG is an invasive procedure, which may be uncomfortable for the patient. sEMG consists of placing electrodes on the skin overlying the muscle of interest to measure the electrical activity of the overall muscle. This method is non-invasive and relatively easy to perform, making it preferable for clinical applications [1]. Thus, the sEMG technique supplies relevant information on the activity of particular muscles and muscles group that are becoming progressively important in many application areas, including clinical, biomedical, prosthetics, rehabilitation, man-machine interactions and more [2],[3]. One of the main fields of application is the gait analysis [4],[5]. The use of sEMG signals in combination with kinematic and kinetic data can provide a deep comprehension of the physiology of this complex task and a more comprehensive understanding of the underlying factors contributing to gait abnormalities. This information can help guide rehabilitation interventions and improve outcomes for individuals with gait impairments.

Linear envelope (LE) is one of the most commonly demodulation technique used to extract information from the observed EMG waveform. It is typically extracted when the amplitude analysis of muscular activation is required [6],[7],[8]. LE is indeed a filtered version of the signal that leads to gradual amplitude variations, making the envelope a simple and commonly used approach in clinics to detect the EMG signal trends. The computation of the linear envelope involves the combination of rectification process and low pass filtering: the raw signal of interest undergoes a first rectification, in which all the negative values are converted into positive, ensuring a better interpretation of the signal. Then, a low pass filter with a cut-off frequency is applied: scientific literature shows that cut-off frequencies ranging between 3-25 Hz are more appropriate to estimate meaningful EMG profiles [9] (in the present study a cut-off frequency of 5 Hz has been adopted). In recent years, the analysis of the EMG-signal envelope, and in particular its peak, has gained increasing attention for the characterization of muscle activity during walking. In various studies, indeed, the peak of the envelope is used as a significant parameter to assess any possible deviation in the muscle activity during gait [10] and other motor tasks [11]. However, further studies on similar topics reported conflicting findings in terms of muscle activation detection: Kwon et al. [12], for instance, showed

early activation of gastrocnemius lateralis and tibial anterior muscles, whereas Akashi et al., [13] reported the largest visible movement during midstance and midswing periods. It was also reported that walking causes continuous EMG linear envelopes to display dynamics that are strongly periodic but slightly aperiodic, similar to those seen in walking kinematics [14], [15]. Thus, techniques that were previously used to describe the dynamic stability of walking kinematics are also appropriate for describing the dynamic stability of muscle activations, such as EMG.

Although the LE peak is widely used in the clinical setting, its reliability to characterize muscle activation during human walking is still subject of debate. The computation of LE, indeed, requires filtering out all the high-frequency components creating very smooth signal trends. This may affect the procedure of LE peak identification and the consequent computation of the timing of the actual activation. Thus, a detailed analysis of the suitability and reliability of this approach is desirable. The objective of the present work is to test the accuracy provided by LE peak in the identification of muscle activity during walking of healthy subjects. Many algorithms have proven to be effective in assessing muscle activation: among these, some threshold-based methods were carried out by comparing the raw EMG signal, or some of its extracted variables, against a set threshold value [16], [17], [18], [19]. However, these kinds of approaches resulted to be not adequate to detect completely both onset/offeset contraction durations. For this reason, they are only used for the onset estimation. Furthermore, more sophisticated methods have been adopted to determine thresholds based on a local sEMG signal portion's frequency content [20]. In particular, a novel algorithm has been recently developed to characterize muscle activation in the time-frequency domain through the application of the continuous wavelet transform (CWT) analysis [21]. In order to favor a more detailed knowledge of the neurophysiological processes behind muscle activations, the reliability of this algorithm has been tested with satisfactory outcomes for the analysis of the electromyographic signals both in the time and frequency domain. In the present study, this CWT-based algorithm has been chosen to test the reliability of LE peak by direct comparison in a large number of strides. This validation was carried out by quantifying the number of times when LE peaks fall within the activation intervals detected by the CWT algorithm (true positives) and by quantifying the error when this does not happen. Moreover, in order to further quantify the performance of LE peak in a more detailed way, LE-peak approach was tested also versus smaller intervals characterized by higher signal energy.

Chapter 1- Surface Electromiography

1.1 Skeletal Muscle

Skeletal muscles are the voluntary muscles responsible for body movements posture. These muscles are attached to bones via tendons and connections between muscles allow for coordinated movements. The three layers of connective tissue that surround and compartmentalize the muscle fibers, within each skeletal muscle, give the muscle its structure (*Figure 1.1*). The epimysium, a sheath of dense connective tissue around each muscle, allows the contraction and powerful movement, while preserving the structural integrity of the muscle itself. The muscle can move independently because the epimysium divides it from nearby tissues and organs [22].



Figure 1.1: The skeletal muscle anatomical structure [22].

Each skeletal muscle contains bundles of muscle fibers, called fascicles, that are encased in a middle layer of connective tissue, known as *perimysium*. This fascicular arrangement enables the nervous system to cause a particular movement of a muscle, by activating a portion of the muscle fibers within a fascicle. The *endomysium*, a thin layer of collagen and reticular fibers in connective tissue, surrounds each muscle fiber inside each fascicle. The endomysium, which encircles the extracellular matrix of the cells, aids in the transmission of force from the muscle fibers to the tendons.

1.2 Muscle Fibers and how they work

To better acknowledge the functionality of muscles, it is useful to analyse a muscle from a macroscopic level to a microscopic one. Macroscopically, muscle fibers are arranged and typically recognized by their direction of pull, line of action, their origins and insertions. However, a deeper examination at this arrangement, reveals that the muscle is really made up of compartments. Some muscles, in fact, actually consist of a number of smaller compartments that run parallel to one another or slightly in the opposite direction. Each muscle compartment contains muscle fibers, as visible in *Figure 1.1*: these fibers may be clustered together into narrow subcompartments separated by a thin septum of connective tissue, that holds the muscle cells together in their parallel arrangements. It is possible to divide these individual fibers into groups of myofibrils, which are tiny, hairlike strands. Microscopically, myofibrils appear to be coiled in light and dark bands. Each myofibril consists of aggregates of myosin and actin filaments.

Skeletal muscle fibers can be significantly larger than other cells, with lengths of up to 30 cm and diameters of up to 100 μ m. These cells are able to produce the large quantities of proteins and enzymes, required to maintain their normal functionality. Skeletal muscle fibers include nuclei, as well as other cellular organelles, like mitochondria and *endoplasmic reticulum* (ER), responsible for the transport of eukaryotic cells and protein folding. Calcium ions are stored, released, and retrieved by the specialized smooth endoplasmic reticulum known as the *sarcoplasmic reticulum* (SR) and the basic contactile unit from which all muscles are derived is known as *sarcomere*.



Figure 1.2: The composition of muscle cells, muscle fascicles, muscle fiber, myofibril, myofilaments, sarcomere, thick and thin filaments.

1.2.1. The Sarcomere

A sarcomere is the basic structural and functional unit of a muscle. It is composed of repeating units of thick and thin filaments arranged in a specific pattern in a myofibril, which is the smallest contractile unit of the muscle fiber. The thick filaments, which are made up of the protein myosin, are arranged in the center of the sarcomere, while the thin filaments, which are made up of the protein actin, are attached to the Z lines at the ends of the sarcomere. When a muscle contracts, the thin filaments slide past the thick filaments, causing the sarcomere to shorten. The length of a sarcomere determines the length of a muscle fiber and its optimal length for muscle contraction is when there is maximum overlap between the thick and thin filaments, which results in maximum tension generated by the muscle fiber. The overlapping myosin and actin filaments extend from one Z line to the next Z line within the muscle; these dark lines (*Figure 1.3-b*) reflect the attachment of actin fibers within the sarcomere. There are areas where only actin locates (**I bands**), areas where only myosin locates (**H bands**), and areas where myosin overlaps the actin fibers (**A bands**).

The lighter I band regions contain thin actin filaments anchored at the Z-disks by a protein, known as α -actinin. The thin filaments continue into the A band toward the M-line and overlap with regions of the thick filament. The A band is dark due to the presence of thicker myosin filaments. The A band is characterised by myosin filaments that cross at the center of the sarcomere and extend towards the Z-disk, that serves as an anchoring site for contractile proteins. During the muscle contraction Z-lines are closely bound together. The H zone, located in the middle of the A band, has a slightly lighter

color, because it only contains the portion of the thick filaments that does not overlap with the thin filaments.



Figure 1.3: a) Schematic representation of the principal constitutive elements of the sarcomere. The A-band comprises myosin filaments crosslinked at the centre by the M-band. Thin actin-containing filaments are tied at their pointed end at the Z-disk and intertwine with the thick filaments in the A-band. b) Electron micrograph of a longitudinal section of a muscle [23].

1.3 Physiology of the muscle Fiber

The muscle is a tissue constantly immersed in an ionic medium. At regular intervals, the transverse tubular system interrupts the *sarcolemma*, the cell membrane that sorrounds a skeletal muscle fiber, as visible in *Figure1.4*. In some places, the transverse tubules (**T-tubules**) run longitudinally, connecting with other T-tubules and the sarcoplasmic reticulum (SR) network. T-tubules are fundamental components for delivering the action potential deeply and transversely into the myofibrils, where it can fully activate the entire muscle fiber.



Figura 1.4: T-tubule structure and relationship to the sarcoplasmic reticulum in skeletal muscle.

Under resting conditions, a voltage gradient exists across the muscle fiber membrane such that the inside of the fiber lies about -90 mV with respect to the outside. The voltage gradient arises from the different concentrations of sodium (Na⁺), potassium (K⁺), and chloride (Cl⁻⁾ and other anions across the membrane. Under resting conditions, the concentration of Na⁺ is relatively high outside the membrane and relatively low inside the fiber. Instead, the concentration of K⁺ is relatively low outside the membrane and relatively high on the inside of the muscle fiber. The size of the resting membrane potential is slightly more positive in slow-twitch fibers. The greater positivity arises from the enhanced Na⁺ permeability, the higher intracellular Na⁺ activity is in slow-twitch fibers than in fast-twitch fibers [24].

1.3.1 Muscle Fiber Action Potential

Muscle fibers are excitable tissues: when the muscle fiber is depolarized by about 10 mV or more, the membrane potential responds in a stereotypical and predictable manner, producing the so called **muscle fiber action potential (MFAP)**, or simply action potential. The action potential generated at the neuromuscular junction extends along the muscle fiber, in both directions, from the neuromuscular junction. In the first phase of the action potential, Na⁺permeability increases and Na⁺ rushes into the cell, ultimately reversing the polarity of the cell so that the cell is temporarily about 10 mV positive. As the Na⁺ permeability increases, so does the membrane permeability to K⁺, and it is the outflow of K⁺ that ultimately results in the return of the membrane potential to its resting state (*Figure 1.5*). The sodium permeability has a significant influence on how long the action potential lasts. Following the nerve or muscle impulse, there is a refractory period during which the membrane's

excitability declines. The membrane becomes completely refractory and all of the Na⁺ channels close for a brief period of time, rendering the membrane incapable of generating an action potential regardless of the magnitude of the excitatory stimulus.



Figure 1.5: Muscle fiber action potential mediated by the alterations in membrane permeability [24].

The membrane potential returns to baseline, following the main portion of the action potential, with a very slow time course. This makes it challenging to calculate the length of the MFAP in clinical and quantitative studies. The action potential of the muscle fiber is reflected in this slow afterwave as being in the negative phase. It appears to be caused by the T-tubule system repolarizing. Near the neuromuscular junction, recordings yield more pronounced afterwaves. Since the slow afterwave has frequency characteristics in the 2 to 40 Hz range, high-pass filtering the EMG signal at higher frequencies will reduce the slow afterwave's appearance. At the neuromuscular junction, brief potentials are released at random intervals. These highfrequency spikes, known as **end-plate potentials** (**MEPPs**), can occasionally be seen on the electromyogram. If the electrodes are placed close to an end-plate zone, the MEPP spikes will be much more noticeable because the spikes degrade significantly with length.

1.3.2 Muscle Fiber Conduction Velocity

The characteristics of the MFAP, as it travels along the muscle fiber, have a significant impact on the electromyogram. **Muscle fiber conduction velocity** (**MFCV**), which ranges from 2 to 6 m/s, is comparatively slow compared to nerve conduction velocity rates, which can reach 100 m/s. To calculate MFCV, a variety of methods are available (*Figure 1.6*).



Figura 1.6: (a) By inserting two stimulating electrodes into the muscle and measuring the response at a predetermined distance, it can be determines the muscle fiber conduction velocity (MFCV). (b) As an alternative, two or more channels can be recorded during voluntary activity from the skin's surface, and MFCV can be calculated using cross-correlation techniques.

However, from a physiological point of view, MFCV depends on a few properties of muscle fibers [25]:

- Intramuscular milieu: MFCV decreases with higher extracellular K⁺ concentrations and decreases with lower intracellular pH in soleus and extensor digitorum longus muscles, but is unaffected by Na⁺ concentration and extracellular pH. One of the main causes of MFCV decline with fatigue is the decrease of MFCV with low pH values.
- **Temperature:** muscle temperature affects muscle fiber conduction velocity, which typically rises with higher temperatures and falls with lower temperatures.
- **Muscle fiber diamenter and morphology**: a linear relationship between the fibers' diameter and conduction velocity (CV) occurs, as well as a correlation between the upper arm's circumference and the CV. Conduction velocity also rises with the motor unit's recruitment

threshold, peaking near the end-plate region and falling off near the tendon.

- **Muscle Length:** The CV is anticipated to decrease with muscle fiber lengthening (stretching) due to a reduction in the fiber's effective diameter. In human muscle, the MFCV also drops as muscle lengthens. Because superficial muscle fibers experience a greater change in muscle length during passive stretching than deeper muscle fibers, the effect is seen more dramatically in superficial muscle fibers than in deeper fibers.
- **Fiber Type:** fast-twitch muscle fibers have a higher CV than slow-twitch muscle fibers. Vastus lateralis fiber type and MFCV measured during maximal effort contraction were found to be correlated with each other, indicating that fiber type could be accurately predicted from a noninvasive measurement of CV.
- **Muscle Fatigue:** MFCV can actually rise during light exercise, most likely as a result of temperature rises, muscle swelling, or modifications to other membrane properties.

1.4 The Motor Unit

The fundamental level of nervous system organization of the muscle is the **motor unit** (**MU**), together with its associated alpha motor system, that is, the lower motor neuron, its axon, and the muscle fibers it innervates. A **MU** refers to a collection of muscle fibers and motor neurons (*Figure 1.7*).



Figure 1.7: Motor unit and muscle fibers linked by neuromuscolar junctions.

Motor neurons are responsible for transmitting signals from the brain and spinal cord to the muscle fibers, causing them to contract. A motor unit consists of a single motor neuron and all the muscle fibers that it innervates. This means that when the motor neuron sends a signal, all the muscle fibers in that motor unit will contract together. The number of muscle fibers in a motor unit can considerably vary in the human body, depending on the muscle being innervated and the fine-tuning required for its movement. For instance, a motor unit in the eye might only contain a few muscle fibers, while a motor unit in the leg muscles might contain hundreds or even thousands of muscle fibers.

When a nerve action potential travels down the axon to the neuromuscular synapse, it releases *acetylcholine* (ACh), which breaks down the muscle's ionic barrier and transmits the signal via the transverse tubules to the rest of the body. As a result, the muscle contracts and the motor unit action potential is created. From the motor end plate to the tendinous attachments at both ends, the depolarization travels in both directions [26].

1.4.1 Motor Unit Action Potential

Since a single motoneuron innervates multiple muscle fibers, when a motoneuron fires, numerous muscle fibers discharge almost simultaneously. A **motor unit action potential** (**MUAP**) is produced as a result of the combined activity of all these muscle fibers (*Figure 1.8*) [25].



Figure 1.8: The surface electromyogram is composed of the algebraic sum of all motor unit action potentials.

Individual MFAPs are added together at the recording site, both temporally and spatially, in order to determine the MUAP's amplitude. A motor unit's muscle fibers may have lengthy terminal axon segments, or "twigs" of axon. These support the later MUAP components and may lead to intricate, spike-filled MUAP shapes. Whereas, the MUAP may have a short duration and a high amplitude if

the terminal axon segments are equal in length and all muscle fibers within the motor unit fire simultaneously.



Figure 1.9: The generation of motoneuron APs (motor unit action potentials) is the first step in the process of recording MUAPs.1) The motoneuron APs arrive at each muscle fiber end plate.2) Muscle fiber APs are generated. 3) The summation of all individual muscle fiber APs generates a motor unit AP. 4) A motor unit AP can be recorded with appropriate electrodes and amplifiers in (5).

1.4.2 Muscle Contraction Cycle

Muscle contraction is the process by which muscles generate force to produce movement, mantain posture and perform other mechanical work. Contraction occurs when muscle fibers shorten and generate tension, pulling on the tendons that attach muscles to bones.

Muscle contraction, as prevolusly introduced, is initiated by the nervous system, which sends action potentials to the muscle fibers via motor neurons, causing he release of calcium ions. The latter, then bind to proteins within the muscles, known as troponin and tropomyosin, causing conformational change that allows the muscle fibers to contract.

The energy for the contraction process is obtained from the ATP molecules (adenosine triphosphate), supplied mostly by the mitochondria [27]. This cyclical process is known as cross-bridging and it is schematized in *Figure 1.10*. The cross-bridging cycle consists of four main stages:

1) ATP binds to myosin which, strongly bound to actin (state of rigor), it decreases the affinity by causing disassociation.

2) The process of hydrolysis of ATP takes place, thus providing the energy necessary for rotation of the myosin head and attack on actin, forming crossbridges.

3) The myosin head flexes itself, rotating toward the center of the sarcomere that leads to *power stroke*.

4) As myosin head bind ATP, the crossbridges detach from actin.



Figure 1.10: Muscular contraction cyclic mechanism.

1.5 EMG tool

Despite its obvious potential as a non-invasive measure of muscle activity, surface Electromiography (sEMG) is currently an underused clinical tool in rehabilitation medicine. However, sEMG signals can be used to extract essential data regarding the patterns and characteristics of muscle activation. The clinical decision-making process may be facilitated by using this information, which can offer an objective, quantitative method of evaluating muscle function, movement patterns, and local muscle fatigue. The further interpretation of EMG signals implies the understaning of how muscles produce bioelectrical signals [28].

1.6 The source of Electromyographic Signal

The amount of energy produced by the muscle is extremely small and is expressed in *microvolts*, or millionths of a volt. The received signal must be amplified using extremely sophisticated and sensitive equipment in order to be seen and heard. A surface electromyograph is essentially nothing more than a highly sensitive voltmeter: early surface electromyography (sEMG) amplifiers were in fact

susceptible to contamination from other electromagnetic energy present in the recording environment. Consequently, sEMG recordings were frequently made in a "copper room." These spaces occasionally consisted only of copper screens. using copper. The electrical noise in the room was captured by the screen and sent to "ground," removing it from the recording environment.

The source of the sEMG signal is the motor unit action potential (MUAP), as introduced in the prevoius Chapter. Action potentials are given off by each of the motor units activated during a given contraction. In any given recruitment pattern, populations of motor units are activated in an asynchronous pattern, which provides the possibility of smooth movement. It is the sum of the activity that constitutes the volume conducted signal, which is picked up at the electrodes and amplified by the SEMG instrument. A block diagram of the different parts of the SEMG instrumentation is shown in *Figure 1.11*.



Figure 1.11: Block diagram of sEMG instrumentation [25].

1.7 Surface Electrodes

According to the investigation to be carried out, different types of electrodes are used: *surface electrodes* for superficial muscles, in direct contact with the skin and are non invasive; needle electrodes for deeper muscles are required, in order to guarantee a direct contact with the target muscle group and require an invasive investigation. The most widely used surface electrodes are Ag-AgCl type gels, stickers, and disposables. Depending on the distance at which electrodes are placed, they usually have a single circular shape button or two. The AgCl-coated silver disk's conductive area must be no larger than 1 cm. They can be further divided into *active* and *passive* electrodes; the active electrodes (*Figure 1.12-A*) have already the pre-amplification circuit integrated, which allows to

reduce the input noise, such as stray voltages due to capacitive couplings or to the movement of the electrodes. While the passive electrodes require an external amplification circuit (*Figure 1.12-B*).



Figure 1.12: A) Active electrode; B) Single and dual electrodes.

Needle electrodes are made of thin, strong, flexible wires, typically alloys of platinum, silver, nickel and chromium. Depending on the muscle to analyze, they will have one tip more or less long (from 25 to 70 mm) and thick (average diameter ranging between 0.30 and 0.45 mm) [29].



Figure 1.13. C) Concentric needle electrodes; D) Screw subdermal needle electrodes.

The muscle's energy, once it has made it to the skin, is sensed by the electrodes. The skin's surface is delicately positioned at the sensing electrode's interface. For instance, the amount of moisture in the skin, the amount of superficial skin oil, and the density the dead-cell layer can affect the skin's impedance, which is also known as resistance to a direct current (DC). Furthermore, an electrolytic medium is frequently used to act as a cushion between the electrode's surface and the skin's surface. If no electrolyte is used, the skin senses the presence of the foreign object (the electrode) and

eventually begins to produce sweat, thereby providing its own electrolytic medium. In sEMG, it's essential to maintain a skin electrode site impedance as low as possible: commonly, this is done by vigorously abrading the skin with an alcohol pad. The impedance at the electrode site should be less than 5,000 to 10,000 Ohms for research purposes.



Figure 1.14: Left: Motor unit activity represented visually in relation to the recording electrodes (left). Right: interface of skin impedance and input of sEMG preamplifier (Ii = input impedance of the preamplifier; Is = impedance of the skin).

Input impedance and skin impedance must be matched in specific ways, as schematically depicted in *Figure 1.14*. The muscle energy that has reached the electrode-skin interface is essentially absorbed by the preamplifier's input impedance (Ii), which then serves as a foundation for amplifying the small signal. In order to absorb the energy that it wants to measure, the sEMG amplification system emits a known input impedance. The skin's impedance (Is) must be lower than the preamplifier's input impedance at the electrode-skin interface should be 10 to 100 times greater than the input impedance of the sEMG preamplifier.

1.8 EMG Amplification

The amplifier stage is one of the most complex parts of a diagnostic machine: after the muscle action potential has passed the electrode-skin interface, the processes of *differential amplification* and *common mode rejection* are applied (Block scheme-*Figure 1.11*). The potential differences between the electrodes are detected by the differential amplification, which also eliminates outside interferences. Amplification is also necessary for the amplitude of the detected EMGs to match the dynamic range of the A/D converter (typical dynamic range: from ± 2.5 V to ± 10 V) and it has the task to amplify the signal and make it viewable onto the screen. In order to prevent potential power

line interference, due to unbalanced impedance in the electrode-skin interfaces, amplifiers for sEMG recording should have high input impedance and high "**Common Mode Rejection Ratio**" (**CMRR**), to make sure that any common mode voltages, seen by individual surface electrodes, are cancelled [30].

1.9 Filtering Process

Once the sEMG signal has been boosted by the differential amplifiers, it undergoes the first level of processing, known as filtering. High-frequency noise can be caused by signal conduction in the nerves and from interferences derived from electronic instrumentation (computers, mobile devices etc.), that can be removed through a *Low-Pass Filter* (LPF) (*Figure 1.15*). The cut-off frequency (f_c) for low-pass filters is around 400-450 Hz [31].



Figure 1.15: Trend of a Low Pass filter response.



Figure 1.16: 1st Order Low Pass filter passive electric scheme.

The low frequency noise can be caused by the DC offset of the amplifier, sensor movements on the skin and temperature variations and it can be removed through the usage of a *high-pass filter* (HPF) (*Figure 1.17*). The cut-off frequency (f_c) for High-Pass Filters is around 15-20 Hz.



Figure 1.17: Trend of a High Pass filter response.



Figure 1.18: 1st Order High Pass filter passive electric scheme.

The cut-off frequency equation for the circuits in *Figure 1.16* and *Figure 1.18* is given by eq. (1):

$$fc = \frac{1}{2\pi RC} \tag{1}$$

However, a 2nd order low pass filter can be more effective as compared to a 1st order one. A second order high pass filter can also be designed. An effective design is employed by using an active electronic component [32], that allows to isolate the filter from the rest of the circuitry. The design adopts two first order filters in series and is facilitated by an operational amplifier. The circuit is showed in *Figure 1.19*. For this circuit, if $R_1 = R_2$; $C_1 = C_2$ then f_c is given as eq. (1). R_3 and R_4 are optional and are required for separate gain settings as:

$$A_0 = 1 + \frac{R_4}{R_3} \tag{2}$$

Using a 2nd order filter is recommended as they provide a roll-off of 40 dB/dec as compared to 20 dB/dec provided by 1st order filters [33]. A 2nd order low pass filter can be more effective as compared to a1st order one. It can be designed by cascading two 1st order filters attached to an operational amplifier (*Figure 1.20*). A 2nd order low pass filter is again recommended as compared

to a 1st order one for the same reasons mentioned above.



Figure 1.19: 2nd Order High Pass Filter circuitry.



Figure 1.20: 2nd Order Low Pass Filter circuitry.

For $R_1 = R_2$ and $C_1 = C_2$, the cut-off frequency of the circuit in *Figure 2.10* is the same as that of Eq. (1). R_3 and R_4 are optional as they are required for separate gain settings as given in Eq. (2). The majority of sEMG instruments have a 60-Hz notch filter. A notch filter is a type of band reject filter that has an extremely steep slope and a width that is typically very narrow (59–61 Hz). This is done in order to remove any electrical noise (60 Hz) from the recording environment that is too strong for the common mode rejection scheme to handle. It specifically rejects any energy with a frequency in the range of 59 and 61 Hz. However, these filters are not so reliable, in fact, if the noise levels are too high, the filter will quickly become saturated.

1.10 A/D Converter Resolution

The digitization process of the analog signal is carried out with an Analog to Digital Converter (ADC): the conversion of a signal from an analog voltage to a digital signal (*A/D conversion*) is necessary before it can be displayed and analyzed in a computer. The higher the resolution the more voltage levels are used to digitize the amplitude of analog signals. The resolution of A/D converters is defined by dividing its dynamic range by its number of levels (N). N is given by 2B = N, where B is the number of bits. For instance, the smallest measurable amplitude by an A/D converter with 12bits and $\pm 2.5V$ dynamic range is 1.22 mV (for example 5 V/212 levels) [30]. Very small signals may need a higher amplification to achieve a better amplitude resolution.



Figure 1.21: A/D conversion and D/A conversion of a signal.

A/D Sampling Rate

The other important technical aspect is the choice of an appropriate *Sampling Rate*. The sampling rate is the frequency expressed in Hertz (Hz) at which the ADC samples the input analogue signal. In general, the higher the frequency content of a signal and the higher the sampling rate required to faithfully reproduce it, the faster the rate at which a signal changes. In order to accurately "translate" the complete frequency spectrum of a signal, the sampling rate at which the A/D board determines the voltage of the input signal must be at least twice as high as the maximum expected frequency of

the signal. The sampling Nyquist theorem explains this relationship: sampling a signal at a frequency that is too low causes aliasing effects.



Figure 1.22: Example of Aliasing effect in time domain. The two signals have the same values at the sampling instants, even if their frequencies are different. The actual signal is higher than twice the sampling frequency [34].

Chapter 2- Gait Analysis

2.1 Introduction to Gait Analysis

Gait analysis (GA) is the systematic study of human locomotion to determine the biomechanics of movement. It involves the analysis of how an individual walks or runs, with the goal of identifying any abnormalities or deviations from normal movement patterns. Gait analysis is commonly used in physical therapy, orthopedics, and sports medicine to evaluate and treat individuals with movement disorders, injuries or disabilities. The analysis can be done through visual observation or through the use of specialised equipment such as force plates, motion capture systems, and electromyography.

The assessment of the characteristics of posture, movement and gait, as well as their variations with respect to a normal condition, can be useful for the diagnosis of particular pathologies, in addition to the planning and control of specific rehabilitation treatments. The observational GA is in fact a useful clinical tool, which is performed by a simple visual observation of the human or animal walking. Although subjective, this method allows to identify gait variations during the main phases of position and oscillation. The movement produced during a walk is very complex and consists of a refined interaction between different muscles and joints. Hence, GA, within the study of walking and locomotor behavior, includes not only the observation and measurement of movement, body mechanics, but also the activity of the muscles involved [35].

Furthermore, GA allows also a detailed evaluation of the effectiveness of the treatment carried out on the patient and the quantitative monitoring of patients' movements leads to measure the effects produced by a given pharmacological, surgical, rehabilitative technique. The walking is therefore a complex activity that acts as a resultant of external and internal forces connected to each other: the movement of the limbs provides repetitive sequences associated with the interaction between the various segments of the lower limbs and the total body mass. During walking, our musculoskeletal system performs specific functions, such as:

- Generation of a propulsive force.
- Stability mantainance.
- Shock absorption due to impact with the ground at each step.
- Conservation of energy in order to minimize the muscle effort.

In fact, in the advancement of the body, one limb acts as a support while the other proceeds to the next support; later the two limbs switch roles and both feet are in contact with the ground during the

transfer of body weight from one limb to the other. A single sequence of these functions is called a **gait cycle**.

2.2 Gait Cycle: Three- dimensional and Cyclic

The anatomical planes are hypothetical planes used to divide the body into sections and describe the location of structures in human anatomy. The analysis of human gait can be treated threedimensionally, as shown in *Figure 2.1*: the main movements during gait take place in the three primary planes of the human body, which are the sagittal, coronal (or frontal) and transverse. The sagittal plane (lateral or Y-Z plane) divides the body into sinister and dexter (left and right) sides and it is probably the most important one, where much of the movement occur. The coronal plane (frontal or Y-X plane) divides the body into dorsal and ventral (back and front) sections; it also separates the anterior and posterior portions. Finally, the transverse plane (axial or X-Z plane) divides the body into superior and inferior (head and tail) sections. It is typically an horizontal plane through the center of the body and it is parallel to the ground.



Figure 2.1: Anatomical Planes in a Human; The three basic planes are: sagittal, coronal, and transverse. The coordinate system with the Z-axis (from front to back), the X-axis (from left to right) and the Y-axis (from up to down) is used to describe a human in the anatomical position.

There are two fundamental requirements for walking:

- I. Periodic movement of each foot from one position of support to the next.
- II. Sufficient ground reaction forces, applied through the feet, for body support.

No matter how distorted the pattern is due to underlying pathology, these two components are required for any type of bipedal walking to occur. The cycle of human gait is exemplified by this repetitive leg movement. *Figure 2.2* depicts a wheel turning from left to right with the highlighted spoke pointing vertically downward. By convention, the beginning of the cycle is initiated at 0%. As the wheel continues to move from left to right, the highlighted spoke rotates in a clockwise direction. At 20% it has rotated through 72°, and for each additional 20%, it advances again 72°. When the spoke returns back to its original position (vertically downward), the cycle is assumed complete at 100%. This analogy of a wheel can be applied to the nature of human gait [36].



Figure 2.2: A wheel in motion illustrates the circular nature of forward movement [36].

Walking representations are normally constrained to a single cycle, with the assumption that successive cycles are all about the same. Although not purely accurate, this assumption is a good approximation for the majority of people.

2.3 Gait Phases

The gait cycle is a time interval or sequence of motion occurring between two successive events, from heelstrike to heelstrike of the same foot, during the walking pattern. The gait cycle consists of two fundamental periods: **stance** and **swing**, tipically defined as gait phases. The stance identifies the whole period of contact between foot and ground (*Figure2.3*), while, the swing is the time when the foot is lifted from the ground for the advancement of the limb: the swing begins when the foot leaves the ground.



Figure 2.3: Division of the gait cycle: the first part highlights the stance phase starting from the heel initial-contact, the second identifies swing phase starting from toe-off.

The stance phase covers the 60% of the gait cycle and can be subdivided into:

- *First double-leg support*, both feet are in contact with the ground.
- Single-leg stance, the left foot is swinging and the right foot is contact in with the ground.
- Second double-leg support, both feet are in contact with the ground.

At an average walking speed, the double-leg support represents the 10% of the entire gait cycle but decreases with increased walking speed and ultimately tends to disappear as one begins to run. At slower walking velocities the double-leg support times are greater. Single-leg stance includes the 40% of the normal gait cycle. The muscles that are active during the stance phase act to prevent buckling of the support limb. These include the tibialis anterior, the quadriceps, the hamstrings, the hip abductors, the gluteus maximus, and the erector spinae. Whereas, the **swing phase** is described when the limb is not weight bearing and represents 40% of a single gait cycle. It is subdivided into three phases: *initial swing* (acceleration), *midswing*, and *terminal swing* (deceleration).

2.4 Gait Events

The gait cycle is typically divided into eight events or periods, five during stance phase and three during swing. These events are based on the movement of the foot, as illustrated in *Figure 2.4*. The following are the stance phase events according to conventional nomenclature:

- Initial contact: initiates the gait cycle, going from 0 to 2%, as the foot, projecting forward, strikes the ground with the heel (*heel strike*). This event represents the point at which the body's centre of gravity reaches its lowest position: the hip is flexed, the knee is extended, the ankle is neutral and the posterior leg is at the end of the stance phase.
- 2. Loading response: is the time when the plantar surface of the foot touches the ground and ranges from 2 to 10% of the gait cycle. The heel is used as a fulcrum, as the knee begins to flex, in order to absorb the impact. The opposite leg is in its pre-swing phase.
- 3. **Midstance**: ranges from 10 to 30% of the gait cycle and begins with the take-off of the contralateral foot (toe-off) and ends when the foot it is entirely supported by the heel, metatarsal and toe bones. This event occurs when the swinging foot proceeds the stance foot and the body's centre of gravity reaches its highest position.
- 4. **Terminal stance**: extends from 30 to 50% of the gait cycle and ends when the contralateral limb touches the ground. The limb has passed the vertical and the body begins to fall forward, the knee bends slightly under its weight and the center of gravity (COG) is lowered.
- 5. **Preswing**: goes from 50 to 60% of the gait cycle starts with the *heel-off* of the limb of interest and terminates the stance phase as the foot, of the same limb, leaves the ground (*toe-off*).



Figure 2.4: Percentage analysis of the human gait cycle in correspondance of each phase and event [37].

The swing phase is divided into three events, as follows:

- 6. **Initial swing**: ranges from 60 to 70% of the gait cycle and begins as soon as the foot leaves the ground and the subject activates the hip flexor muscles to accelerate the leg forward.
- 7. **Midswing**: ranges from 70 to 85% and it occurs when the foot passes directly beneath the body, coincidental with midstance for the other foot.
- 8. **Terminal swing:** covers the last interval of the gait cycle, from 85 to 200%, and describes the activity of the muscles as they slow the leg and stabilize the foot in preparation for the next heel strike.

2.5 Energy Conservation during Gait

There are three main instances during gait where energy is expended. For instance, shock absorption at heelstrike, propulsion during push-off, when the center of gravity (COG) is projected upward and forward, and controlling forward motion during deceleration toward the end of the swing phase are all examples of this.

The center of mass (COM) of a human is situated approximately in the middle of both hip joints, just anterior to the second sacral vertebra. When a body moves in a straight line, with the COM not veering up or down or side to side, the minimal amount of energy is needed. Normal gait would allow for such a straight line if a man's lower limbs ended in wheels rather than feet. However, our COM deviates from the straight line in vertical and lateral sinusoidal displacements. Evidently, this is not the case, and as a result, our COM deviates from the line in both vertical and lateral sinusoidal displacements.

Regarding vertical displacement, the COM moves forward in a rhythmic upward and downward motion. In correspondance of midstance the point reaches its highest position, and the time of double support is where the point is lowest. In an adult male, there is typically a vertical displacement of 5 cm. Regarding lateral displacements, the pelvis shifts to the weight-bearing side as weight is moved from one leg to the other. An approximate 5 cm side-to-side displacement results from the COM oscillating; at midstance, the lateral limits are attained.

2.5.1 The Six Determinants of Gait

The six determinants of gait are kinematic characteristics of gait that are intended to reduce the vertical displacement of the body center of mass (COM), thereby lowering the energetic cost of locomotion. As described in [38], to prevent the COM from deviating from the "compass gait," in

which the COM moves in a circular arc, movements like pelvic rotation and knee flexion during the stance phase are coordinated. These six determinants are all integrated during each gait cycle and they work to reduce the body's COM's displacement and flatten the arc in the vertical and horizontal (lateral) planes, lowering the energy expenditure. The overall result is described as a smooth, sinusoidal translation of the COM through space.

First determinant: pelvis rotation in the horizontal plane (*Figure 2.5*). As a result, the swinging hip can advance more quickly than the stance hip. During midstance, the pelvis rotates posteriorly during midstance and anteriorly on the swinging limb. It reaches its peak right before heelstrike with a total pelvic rotation of 3–5 degrees to each side. For the same amount of hip flexion in the leading leg and hip extension in the trailing leg, pelvic rotation also results in a longer stride length. As a result, it permits longer steps without significantly altering the COM displacement.



Figure 2.5 Rotation of the pelvis in the Horizontal plane during swing phase. The rotation is needed in order to reduce the angle of flexion and extension of the hip and to minimize the excursion of COM (vertical movement of the hip).

Second determinant: pelvic tilt (or pelvic obliquity) in the frontal plane. Pelvic tilt is controlled by the hip abductors of the stance hip as the pelvis on the swing leg is lowered. The pelvis moves 4-5 degrees away from the stance leg and toward the swing leg during normal gait. During single limb support, this pelvic dip reduces the COM's horizontal displacement (*Figure 2.6*).



Figura 2.6: Pelvis tilts about an anteroposterior axis, raising at first one side and then the other side; the hip of the swing leg decreases, and this reduces the vertical displacement of the COM.

Third determinant: knee flexion, which reduces the vertical displacement of the COM, is the third factor. This happens in midstance when the eccentric quadriceps contract, causing the knee to flex to about 15° , where it stays until the foot is flat on the ground. With each step, these first three determinants reduce the vertical displacement by one inch (*Figure 2.7*).

Four and Fifth determinants: they are associated with the control of the knee-ankle-foot motion. During the first part of the stance phase, this coordinated movement leads to eccentric control of the knee flexion and plantar flexion of the ankle. By preventing sudden changes in the lowest part of the COM arc, these factors help to create an arched pattern rather than a smooth, sinusoidal curve (*Figure 2.7*). These five gait determinants are the movements to reduce the excursion of the center of gravity.



Figure 2.7: The third, fourth and fifth determinants of gait are concerned with adjusting the effective length of the leg, by lengthening it at the beginning and end of the stance phase and shortening it in the middle, to keep the hip height as constant as possible.

Sixth determinant: it occurs with the lateral pelvic movement and it is the side-to-side oscillation or lateral sway that happens with every step. This describes how the COM moves in the horizontal plane. To provide stability during the stance phase, the pelvis shifts over the supporting foot. The base of support determines how much sway will occur. The feet can be closer together during forward movement thanks to normal knee valgus between the femur and tibia, which helps to minimize the amount of pelvic shifting necessary for stability. If the feet are positioned far apart (left image, *Figure 2.8*), large lateral movements of the center of gravity would be necessary to maintain balance. If the feet are close to each other , this reduces the size lateral movements. The reduction in lateral acceleration and deceleration leads to a muscular energy decrease, keeping the base narrow, little lateral movement is needed to maintain balance.



Figure 2.8: The sixth determinant of gait is about side by side movement; if the gait width is larger, the COM moces more in lateral direction (left image). If the gait width is smaller the lateral movement is reduced.

2.5.2 Distance Measures

Gait asymmetries, related to step spacing and stride lengths, can be measured by viewing gait from a spatial perspective. The stride length, foot angle, step width, and step length are some common spatial gait parameters used to describe a gait pattern. *Figure 2.9* shows how a collection of footprints can yield helpful distance information. The length between the heels from one heel strike to the next on the same side is the definition of stride length, which is the distance covered by a person during one cycle. One stride length is made up of two step lengths (left plus right). In normal subjects, one stride length is made up of the two step lengths (left plus right). Normal subjects will have roughly equal step lengths on both sides, but some patients may present an asymmetry between the left and right sides.



Figura 2.9: typical spatial gait parameters.

Step width, which is the mediolateral distance between the feet and has a value of a few centimeters for healthy subjects, is another helpful parameter. The degree of external or internal rotation of the lower extremity during the stance phase can be determined by looking at the angle of the foot in relation to the line of progression.
2.6 Biomechanics of the Ankle Joint

The ankle joint complex is constituted by the lower leg and foot and it generates the kinetic linkage that enables the lower limb to interact with the ground, a crucial component for gait and other daily activities. The ankle's bony and ligamentous structure allows gait to perform with a high level of stability, despite preserving high compressive and shear forces during gait.

2.6.1 Anatomy of the ankle

The twenty-six individual bones of the foot and the long bones of the lower limb combine to form thirty-three joints in the foot and ankle. The ankle joint complex comprises the *talocalcaneal* (subtalar), *tibiotalar* (talocrural) and *transverse-tarsal* (talocalcaneonavicular) *joint (Figure 2.10)*.



Figura 2.10: Ligaments of the ankle joint complex. (A) Central view, (B) Anterior view. (AITFL= anterior tibiofibular ligament; IOL=interosseous ligament; PITFL= Posterior tibiofibular ligament) [39].

Talocalcaneal Joint:

The calcaneus is the most posterior and the strongest bone of the foot, supplying attachment for the Achilles tendon, and it is positioned inferiorly to the talus. The anterior part of the calcaneus is where the talus lies. The components of the *anterior talocalcaneal joint* on the inferior part of the talus are convex, and on the superior part of the calcaneus are concave, while the components of the *posterior talocalcaneal joint* on the inferior part of the talus are concave, and on the superior part of the talus are concave, and on the superior part of the talus are concave, and on the superior part of the talus are concave, and on the superior part of the calcaneus are concave, and on the superior part of the talus are concave. This morphology permits inversion and eversion of the ankle [40].

Many ligaments are present to attach the two bony surfaces. The interosseous talocalcaneal ligament, which is a thick ligament that extending from the articular parts of the inferior talus to the superior surface of the calcaneus, is the main connection between the two. Furthermore, the lateral talocalcaneal ligament and the anterior talocalcaneal ligament also contribute to the connection of this joint, but they are very weak.

Tibiotalar joint:

The tibiotalar joint forms the junction between the distal tibia and fibula of the lower leg and the talus. The talus is constrained by the malleoli of the tibia and fibula, which makes the joint work as a hinge joint and primarily aids in the plantar- and dorsiflexion motion of the foot. The tibiotalar joint's conforming geometry is thought to be effective in the stability of the joint. In the stance phase, the joint geometry alone provides resistance to eversion, otherwise, stability comes from the soft tissue structures. The medial collateral ligaments (or deltoid ligaments), which support the medial part of this ankle joint, are essential for preventing valgus stresses and eversion motion in the joint. The deltoid ligament is fan-shaped and is made up of the anterior and posterior tibiotalar ligaments, known as the tibionavicular ligament and the tibiocalcaneal ligament (*Figure 2.10-A*).

The lateral ligaments that reduce inversion of the joint, leading to a limitation of varus stresses and reduction of rotation are: the anterior and posterior talofibular ligaments and the calcaneofibular ligament (*Figure 2.10*).

Transverse-tarsal joint:

it consists of the calcaneocuboid joint, also known as Chopart's joint, which connects the calcaneus and cuboid, and the junction between the talus and navicular, where the talar head articulates with the posterior aspect of the navicular anteriorly (*Figure 2.11*).

The transverse tarsal joint and subtalar joint are classified as belonging to the same functional unit, because they have a common axis of motion. This joint also helps the foot move in an eversion-inversion pattern.



Figure 2.11: Transverse Tarsal Joint (Chopart's Joint).

2.6.3. Muscles of the ankle

The twelve extrinsic muscles, which arise in the leg and insert in the foot, are responsible for most of the motion in the foot and ankle (*Figure 2.12*). There are four compartments that accomodate these muscles:

-The anterior compartment is characterised by four muscles: the *tibialis anterior*, the *extensor digitorum longus*, the *extensor hallucis longus*, and the *peroneus tertius*. The tibialis anterior and the extensor hallucis longus produce dorsiflexion and inversion of the foot. The peroneus tertius produces dorsiflexion and eversion of the foot. The extensor digitorum longus produces only dorsiflexion of the foot.

-The lateral compartment is characterised by two muscles: the *peroneus longus* and the *peroneus brevis*, which produce plantarflexion and eversion of the foot.

-The posterior compartment is characterised by three muscles: the *gastrocnemius*, the *soleus*, and the *plantaris*, which lead to plantarflexion of the foot.

-The deep posterior compartment is characterised by three muscles: the *tibialis posterior*, the *flexor digitorum longus*, and the *flexor hallucis longus*, which produce plantarflexion and inversion of the foot.



Figure 2.12: Muscles commonly associated with the ankle.

2.6.2 Ankle and Foot Motion

The main movement related to the ankle joint complex are: *plantar- and dorsiflexion*, which occur in the sagittal plane; *ab-/adduction*, which occur in the transverse plane and *inversion-eversion*, which occur in the frontal plane, as depicted in *Figure 2.13* [40]. When these movements are combined across the subtalar and tibiotalar joints, the three-dimensional supination and pronation motions are produced. The combination of plantarflexion, inversion, and adduction during supination results in the sole to face medially. Dorsiflexion, eversion, and abduction help position the sole facing laterally during pronation.



Figure 2.13: 3D illustration of the ankle joint complex motions.

The axis of rotation of the ankle joint complex in the sagittal plane is formed around the line passing through the medial and lateral malleoli (*Figure 2.14-A*). The coronal plane axis of rotation takes place around the intersecting point between the malleoli and the long axis of the tibia in the frontal plane. The transverse plane axis of rotation takes place around the long axis of the tibia intersecting the midline of the foot (*Figure 2.14-B*).

The subtalar joint's axis is also oblique, running from back to front and forming an angle of about 40 degrees with the anteroposterior axis in the sagittal plane and a 23-degree angle with the midline of the foot in the transverse plane. The subtalar joint produces multiple motions during plantar and dorsiflexion, resulting in pronation and supination, similarly to the tibiotalar joint.



Figure 2.14: A) Axis of rotation of the ankle joint complex in the sagittal and frontal planes. The axis of rotation for the dorsiflexion and plantarflexion is represented by the black dashed line. The intersecting point (red dot) is the point of rotation for inversion and eversion. B) Axis of rotation of the ankle joint complex in the transverse plane. The intersecting point (red dot) is the point of rotation for inversion [40].

Gait analysis is a frequently employed method for measuring the complex kinematics and kinetics of the ankle joint. During a normal gait cycle, the stance phase can be subdivided into three sub-phases based on the sagittal motion of the ankle; i) the *heel rocker*; ii) the *ankle rocker* and iii) the *forefoot rocker*. The heel rocker phase starts at heel strike, where the ankle assumes a plantarflexed position pivoting around the calcaneus, until the foot is flat on the ground ending the heel rocker phase.

The *ankle rocker* phase occurs when the ankle moves from plantarflexion to dorsiflexion during which the shank (tibia and fibula) rotate forward around the ankle leading to progress the body forward. In the *forefoot rocker* phase, the foot rotates when the calcaneus lifts off the ground and the the ankle begins to plantarflex until maximum plantarflexion is achieved at toe-off, when enough power is generated for the leg to initiate the swing phase.



Figure 2.15: The three mechanisms of the ankle and foot that produce shank kinematic during stance phase of the gait cycle.

In order to better investigate these three phases, Perry et al. [41] proposed representations of gait analysis data related to the ankle joint complex kinematics, kinetics and power as shown in *Figure 2.16-2.17*. Motion of the ankle occurs primarily in the sagittal plane, with plantar- and dorsiflexion arising mainly at the tibiotalar joint.



Figure 2.16: The ankle range of motion (ROM) in normal gait cycle.

Each gait cycle includes two periods of plantar flexion and dorsiflexion. The ankle is positioned at a 90° angle at the beginning of stance. The foot flexes its plantar surface by 10° as the heel is loaded. After that, the movement is reversed and slowly moves to 10° of dorsiflexion. During the double-

stance phase, when the limb is being rapidly unloaded, plantar flexion resumes at this point and progresses to 20° by the end of stance. Immediately after toe-off, the foot is quickly raised to neutral dorsiflexion and kept there throughout the swing [41].



Figure 2.17: representation of the ankle complex rotation in the frontal and transverse planes; the shaded areas represents ± 1 standard deviation.



Figura 2.18: representation of the ankle moments in the sagittal plane (left) and ankle power in the sagittal plane. The shaded areas represents ± 1 standard deviation.

The evaluation of the ankle moment during gait (*Figure 2.18 left*) shows a dorsiflexion moment at heel strike as the eccentric contraction of the dorsiflexors regulates the foot's rotation on the ground and avoids the foot from hitting the ground. A plantarflexor moment occurs during the second phase as the ankle dorsiflexors contract eccentrically to allow the shank to advance over the foot. The plantar flexion moment continues in the third phase, with the plantar flexors contracting concentrically towards toe-off. Ankle kinetic patterns assume a constant profile, but with larger magnitudes as walking speed rises.

When the main muscles, controlling the ankle joint complex, are either producing or absorbing power during gait, ankle power tends to vary (*Figure 2.18 right*). The negative values, shown in the first part, reflect the contraction of the plantarflexors during the heel and ankle rocker phases, which absorbs power. The plantarflexors, which are responsible for the lower limb's ability to push the body

forward and toward toe-off, generate their maximum joint power at about 50% of the gait cycle, during the forefoot rocker phase.

2.6 Muscular Activations during Gait

The human body is made up of numerous interconnected segments, joined by muscles, that work interactively to produce motion. In order to comprehend the movement, it is significant to examine the coordination and contribution of the involved muscles during activity.

Walking can be viewed from the perspective of a functional study of the subject, as the mechanism by which each subject sets his or her own muscle chains in motion. The entire body is involved in the gait, which is didactically divided into a *Passenger Unit* and a *Locomotor Unit*.



Figura 2.19: Functional division of the human body [41].

The head, neck, trunk, and arms are collectively referred to the *Passenger Unit*, since they are transported, rather than directly participating in walking. During normal walking, the muscular activity of the neck and trunk has the exclusive function of maintaining the neutral vertebral alignment with postural changes. Whereas, the lower limbs and the pelvis represent the anatomical segments that form the *Locomotor Unit:* each limb is responsible for supporting the *Passenger Unit* allowing it to progress forward.

A complex sequence of muscle contractions is required to walk. During the gait cycle, the muscles must contract precisely at the appropriate timing, in order let the deambulation progress with the proper sequence of rhythmic and alternating motions. *Figure 2.20* highlights the main muscle

activation patterns during activity: The gluteus maximus and hamstrings function as hip extensors. The hamstrings are active at the IC to prevent hyperextension of the knee. The quadriceps are knee extensors, that help the controlling of knee flexion. The iliopsoas is a hip flexor and active during the initial and mid-swing phase. Tibialis anterior are active all over the swing phase and the loading response, in order to control the ankle plantarflexion during the loading response and initial swing and maintain the ankle dorsiflexion during the late swing phase. Triceps surae are active during late mid-stance and terminal stance to control dorsiflexion during the corresponding periods [42].



Figure 2.20: Prevalent muscle activity patterns during gait cycle revealed through Electromiography.

As previously introduced, the gait cycle is the term also used to describe the series of hip, knee, and ankle movements that occur during walking, resulting in intricate muscle interactions between the agonist-antagonist group. An ideal walk cycle is actually created when a muscle or group of muscles contracts, relaxes, and goes through concentric and eccentric contractions in a smooth, interdependent manner, under the effect of an external force or moment.

According to *Perry et.al*, the muscles that are mostly involved in the motor gesture are: tibialis anterior (TA), gastrocnemius lateralis (GA), soleus (Sol), vastus lateralis (VL), biceps femoris (BF) and rectus femoris (RF). Their features and typical activation intervals can be detected at the temporal level of gait cycle, in terms of %GC [41].

Tibialis Anterior

It is s the largest of four muscles found in the anterior compartment of the leg and it originates on the lateral aspect of the tibia and inserts into the plantar aspect of the bone wedge-shaped, at the base of the first metatarsal bone. This muscle is primarily responsible for dorsiflexion and inversion of the foot. The tibialis anterior is characterised by two activations during the gait cycle: the first ranges

between 0-10% and the second between 60-100%. In the first activation, the activity of the tibialis anterior increases significantly in intensity during the load response and then rapidly decreases and coming to an end at the start of the phase mid-stance (*Figure 2.21*). In the initial contact the tibialis anterior stabilizes the ankle joint by slowing the speed of plantar flexion. The second activation begins in the pre-swing phase and becomes more intense during the initial swing, in order to accomplish foot lift from the ground.



Figure 2.21: Activity of the tibialis anterior muscle during gait cycle. Intensity given as a percentage of the value manual muscle test maximal (%MMT) and expressed as a function of the percentage of GC. The black area indicates the activity pattern of most individuals. The light gray area indicates the least frequent activations. The Vertical bars designate gait phase subdivisions (N = number of samples included in the data) [41].

Gastrocnemius

It is a considerably large muscle positioned posteriorly within the leg and it is also the most superficial among the muscles of the leg, forming the bulk of the calf. Together with the solues muscle, the gastrocnemius is a component of three-headed muscles trio, known as triceps surae. Together, they perform a variety of fundamental motions like running, jumping, and walking. The gastrocnemius acts primarily as a plantar flexor: its functionality consists of extending the foot, internally rotating it thanks to the flexion of the ankle, and participate also to the flexion of the leg, being a biarticular muscle. With its contraction it also participates in the initiation of pre-swing and in the elevation of the heel from the ground. Gastrocnemius activity, as depicted in *Figure 2.22*, reveals a small delay in the onset of about 5% of the gait cycle; at the initiation of the gait cycle is achieved. Once the increment in intensity has stopped, an equally quick decline phase occurs, lasting until the initiation of pre-swing phase.



Figure 2.22: Activity of the gastrocnemius muscle during gait cycle. Intensity given as a percentage of the value manual muscle test maximal (%MMT) and expressed as a function of the percentage of GC. The black area indicates the activity pattern of most individuals. The light gray area indicates the least frequent activations. The Vertical bars designate gait phase subdivisions (N = number of samples included in the data).

Soleus

The soleus muscle is a wide flat leg muscle located on the back of the leg. It extends from just below the knee to the heel and lays immediately deep to the gastrocnemius. These two muscles, together with the plantaris muscle, are components of superficial posterior compartment calf muscles group. Soleus' contraction results in strong plantar flexion: due to its crucial function as an antigravity muscle, it also enables us to maintain an upright posture. The soleus muscle initiates its contraction at the end of the load response phase: it increases quickly and continues throughout midstance, aiming to slowing down the speed of advancement of the tibia. The onset of terminal stance causes a significant amplitude increase in muscle activity, which continues until a peak at 45% of the gait cycle is achieved (*Figure 2.23*). At this point, the soleus' action rapidly decreases until it reaches zero at the start of the pre-swing phase.



Figure 2.23: Activity of the soleus muscle during gait cycle. Intensity given as a percentage of the value manual muscle test maximal (%MMT) and expressed as a function of the percentage of GC. The black area indicates the activity pattern of most individuals. The light gray area indicates the least frequent activations. The Vertical bars designate gait phase subdivisions (N = number of samples included in the data).

Vastus Lateralis

The vastus lateralis muscle is found on the lateral side of the thigh. This muscle is the largest of the quadriceps which includes rectus femoris, vastus intermedius, and vastus medialis. The quadriceps work together with the knee and hip to support movement, strength, and stability. Walking, running, and jumping are commonplace activities that they power and support. It has the activity of extension and stabilization of the knee joint. In the gait cycle it reaches a peak of 25%, corresponding to the load response phase and ends in the stance phase with a corresponding peak of almost 60% of the gait cycle (*Figure 2.24*).



Figura 2.24 Activity of the vastus lateralis muscle during gait cycle. Intensity given as a percentage of the value manual muscle test maximal (%MMT) and expressed as a function of the percentage of GC. The black area indicates the activity pattern of most individuals. The light gray area indicates the least frequent activations. The Vertical bars designate gait phase subdivisions (N = number of samples included in the data).

Biceps Femoris

The biceps femoris muscle is located on the back of the thigh. This muscle forms the semitendinosus and semimembranosus muscle group, collectively known as the hamstrings. The biceps femoris muscle extends all the way from the ischial tuberosity to the proximal end of the fibula, by crossing the knee joint and the hip joint. Biceps femoris performs flexion and external rotation at the knee joint, as well as extension and external rotation at the hip joint. It is activated in the final part of the intermediate swing, at 82% of the gait cycle, with a peak of 92% in the initial part of terminal swing, in order to control hip flexion. Its activity on knee flexion prevents excessive hyperextension due to the moment of the tibia on the femur that has finished extension. Subsequently, the activity of the muscle reduces, but it remains active for the entire duration of the response to the load (10%), providing a balancing force.



Figure 2.25: Activity of the hamstring muscles during gait cycle. Intensity given as a percentage of the value manual muscle test maximal (%MMT) and expressed as a function of the percentage of GC. The black area indicates the activity pattern of most individuals. The light gray area indicates the least frequent activations. The Vertical bars designate gait phase subdivisions (N = number of samples included in the data).

Rectus Femoris

The rectus femoris is a fusiform muscle that consists of two heads and originates from two sites on the ilium. It is located on the anterior compartment of the thigh and belongs to the "four-headed" muscle unit of the quadriceps femoris. The rectus femoris is a bi-articular muscle that acts at the level of two joints as hip flexor and hip extensor knee. Its activity is very restricted in terms of time and intensity: in fact, the activation interval occurs between 56% and 64% of the gait cycle, in the late pre-swing and early swing phases. The contraction of this muscle collaborates to perform the leg extension in the push action, along with the other quadriceps heads, and it also guarantees the flexion of the thigh to make possible the lifting of the limb forward.



Figure 2.26: Activity of the rectus femoris muscle during gait cycle. Intensity given as a percentage of the value manual muscle test maximal (%MMT) and expressed as a function of the percentage of GC. The black area indicates the activity pattern of most individuals. The light gray area indicates the least frequent activations. The Vertical bars designate gait phase subdivisions (N = number of samples included in the data).

Chapter 3- Wavelet Analysis

The majority of signals are characterised by time domain representation. Signal analysis, which transforms the time signals using an analysis function, can be used to understand more about the time signals. The most well-known technique for determining the frequency information of a time signal is the Fourier transform (FT), which only retrieves back the global frequency content of a signal. Therefore, only stationary and pseudo-stationary signals can benefit from the Fourier transform. For signals that are highly non-stationary, noisy, or aperiodic, the Fourier transform does not provide satisfactory results. The wavelet analysis, instead, is a recent technique of analysis that, in contrast to Fourier analysis, which uses long waves for the analysis function, primarly utilizes short wavelets. The wavelet analysis is a suitable alternative for many applications, since it has a number of significant advantages over the Fourier transform. In recent years, wavelet analysis has expanded rapidly in both usage and fields of application [43].

3.1 General Application

To comprehend the wavelet transform (WT), it is necessary to firstly understand the Fourier transform (FT). The short time Fourier transform (STFT) is the most effective way to explain the change from the Fourier transform to the wavelet transform. The STFT determines the Fourier transform of a windowed portion of the signal and shifts the window over the signal itself. There are several ways to perform wavelet analysis, including continuous wavelet transforms, discretized continuous wavelet transforms, and true discrete wavelet transforms. With increased public awareness of the analysis method, wavelet analysis applications are becoming more widespread. Thus, the range of applications includes fields such as science, engineering, medicine, and also finance. This chapter provides an introduction into the wavelet analysis.

3.2 Multiresolution Analysis

Multiresolution analysis (MRA) refers to fragment a signal into components, which produce the original signal exactly when added back together. To be useful for data analysis, it is necessary to know how the signal is decomposed. The decomposition of a signal is in fact essential for its applicability in data analysis. By using the MRA approach, it is possible to examine a signal at

different frequencies with different resolutions. A schematic representation showing the resolution change is displayed in *Figure 3.1*.



Figure 3.1: Multiresolution Time-Frequency plane [44].

The time-frequency-plane contains rectangular blocks, usually known as *Heisenberg cells*. The size of each cell is determined by the uncertainty principle. The principle deomonstrates the required trade-off, in the calculations, between the accuracy in time and in frequency that affect consequently the accuracy of the phase calculation. Better resolution in the time domain triggers a diminution in the frequency resolution and viceversa [45]. As depicted in *Figure 3.1*, narrow rectangles are used to represent high frequencies that provide a precise localization in time (short time intervals, high frequency information). Large rectangles are used to represent the low frequencies that provide precise localization in frequency (high time intervals, low frequency information). Thus, the Wavelet Transform (WT) is a tool mainly used for signal decomposition by frequency and location. This technique measures the relative temporal changes in the frequency content of non-stationary signals, without losing the resolution in time and frequency. The time-frequency plane, used for the WT method, is also known as *scalogram*, that will be discussed later.

The wavelet analysis determines the correlation between the signal under examination and a wavelet function $\psi(t)$. The matching between the signal and the analyzing wavelet function is determined

separately for different time intervals, generating a two-dimensional representation. The analyzing wavelet function $\psi(t)$ is also recognized as the 'mother wavelet'.

3.3 Wavelets

In contrast to the Fourier transform, the analyzing function of the wavelet transform can be selected with more flexibility, without the need of using sinusoidal waveforms. Wavelet transforms are able to break signals down into oscillations localized in space and time. Whereas Fourier transforms break down signals into oscillations that persist over the entire sequence. A wavelet function $\psi(t)$ is a small wave, which must be oscillatory in some way to discriminate between different frequencies [43]. The wavelet contains both the analyzing shape and the window: an example of a possible wavelet, known as the Morlet wavelet, is shown in *Figure 3.2*.



Figure 3.2: Morlet wavelet representation (p= frequency parameter; σ = decay parameter) [46].

An analyzing function $\psi(t)$ is categorised as a wavelet function if the following mathematical requirements are satisfied [43]:

• A wavelet must have finite energy:

$$E = \int_{-\infty}^{\infty} |\psi(t)|^2 dt < \infty$$
(3)

The energy E is the integrated squared magnitude of the analyzing function $\psi(t)$ and must be less than infinity.

The following requirement must be satisfied if Ψ(f) is the Fourier transform of the wavelet ψ(t):

$$C_{\psi} = \int_0^\infty \frac{\left|\hat{\psi}\left(f\right)\right|^2}{f} \, df < \infty \tag{4}$$

This condition means that the wavelet has no zero-frequency component ($\Psi(0) = 0$), (i.e. the mean of the wavelet $\psi(t)$ must equal zero) and it is known as the admissibility constant. The value of C_{ψ} depends on the chosen wavelet.

• For complex wavelets the $\Psi(f)$ must be both real and vanish for negative frequencies values.

4.3.1 Mother Wavelet Families

Wavelet transformation provide solutions to the time-scale analysis by separating signals into a superposition of shifted and scaled versions of the mother wavelet. The waveforms referred to as "mother wavelets" have a rapid decay or a finite-length oscillating shape. To fit the signal to be analyzed, they are translated and scaled. There are various mother Wavelet types, and during analysis, the waveform to be used is selected according to how closely it resembles the input signal. The following representations show some examples of Wavelet families.



Figure 3.3: Mother Wavelet families illustrations.

4.4 Continuos Wavelet Transform

Wavelets are characterized by two basic properties: *scale* and *location*. Scale (or dilation) determines how stretched or compressed a wavelet is. This property relates to frequency as defined for waves. Whereas, location determines where the wavelet is positioned in time (or space).

The continuos wavelet transform (CWT) is defined as the sum over time of the signal multiplied by the scale and the shifted version of the wavelet function $\psi(t)$:

$$CWT_{x(a,b)} = \int x(t) \psi_{a,b}^*(t) dt \quad a \neq 0$$
(5)

where $\psi_{a,b}^*$ is the Mother Wavelet.

The CWT calculation is usually performed by taking discrete values for the scaling parameter *a* and the shifting parameter *b*. The resulting wavelet coefficients are also called wavelet series. For analysis purposes only, the discretization can be done arbitrarily, however if reconstruction is required, the wavelet restrictions mentioned in Section 4.3 become important.

$$\psi_{a,b}(t) = \frac{1}{\sqrt{a}}\psi\left(\frac{t-b}{a}\right)$$
(6)

The parameter a, in the expression above, sets the scale of the wavelet. If its value is decreased, the wavelet will look more compressed. This in turn can capture high frequency information. Conversely, increasing the value of a will stretch the wavelet and capture low frequency information (*Figure 3.4*). Thus, the variation in scale a changes not only the central frequency (f_c) of the wavelet, but also the window length.



Figure 3.4: Mother Wavelet representation as function of scale a coefficient variation [44].

Whereas, the parameter b defines the location of the wavelet. Decreasing b will shift the wavelet to the left. Increasing b will shift it to the right, as demonstrated in *Figure 3.5* below. Location is essential because unlike waves, wavelets are only non-zero in a short interval. Thus, shifting operation is performed in order to delay (or anticipate) the position of mother wavelet in time. Additionally, when examining a signal, we are interested in both the oscillations and the locations where those oscillations take place.



Figure 3.5: Mother Wavelet representation as function of b shift coefficient variations [44].

The outcomes obtained from the CWT are coefficients expressed as function of scale and position. These coefficients are obtained with a process that follows some steps: STEP 1: Comparison between the mother Wavelet and a section (window) at the beginning of the original signal.



Figure 3.6 CWT step 1 [44].

STEP 2: Computation of the Wavelet Coefficient C, representing how closely the "mother" signal is correlated with this section of the analyzed signal. The higher C is, the higher is the similarity.



STEP 3: Shifting of the mother wavelet to the right and repetition of steps 1 and 2 for the entire signal.



Figure 3.8: CWT step 3 [44].

STEP 4: Scaling (stretch/compress) of the parent Wavelet and repetition of steps 1 and 3.



Figure 3.9: CWT step 4 [44].

The discretized CWT of a general signal x(t), analyzed with the Morlet wavelet, is shown in *Figure* 3.10, in which a contour plot of the wavelet coefficients is shown.



Figure 3.10: Left side: x(t) is sampled at a frequency of 1 kHz. From 0.1 s to 0.3 s the signal is a sine with a frequency of 45 H. At 0.4 s the signal shows a sinusoid with a frequency of 250 Hz which changes to 75 Hz at 0.5 s. From time 0.7 s up to 0.9 s there are two superposed sinusoids of 30 Hz and 110 Hz. Right side: CWT Contour Plot.

Large scales correspond to low frequencies and small scales to high frequencies. The CWT in *Figure 3.10* gives a good frequency resolution for high frequencies (small scales) and a good time resolution for low frequencies (large scales). The CWT performs a multiresolution analysis by contraction and dilatation of the wavelet functions.

4.5 Multiresolution Filter banks

The CWT performs a multiresolution analysis, allowing the analysis of a signal at various frequencies and resolutions. For high frequencies (low scales), which last briefly in time, a good time resolution is preferred. For low frequencies (high scales) a good frequency resolution is more important. The CWT has a time-frequency resolution, as prevolusly shown in *Figure 3.1*. Another solution is the discrete wavelet transform (DWT), obtained because of this multiresolution technique exploiting filter banks. It's important to note that the discretized version of the CWT is not equal to the DWT: the DWT adopts filter banks, while the discretized CWT uses discretized versions of the scale and dilatation axes.

A filter bank represents a set of filters that split the signal into frequency bands. An input signal x(k) enters the *analysis bank* and is filtered by the *Low Pass Filter* (L(z)) and *High Pass Filter* (H(z)), which separate the frequency content of the input signal in frequency bands of equal width. The output of the filters both contain half the frequency content, but the same number of samples as the input signal. The two outputs together contain the same frequency content as the input signal, but the amount of data is doubled. Thus, downsampling by a factor two ($\downarrow 2$), is applied to the outputs of the filters. Using the *synthesis filter bank*, it is possible to recreate the original signal (reconstruction).



Figure 3.11: Filter bank with two level channel.

The low-pass and high-pass filtering branches of the filter bank return the *approximations* and *details* of the signal x(k). The filter bank can be enlarged to an arbitrary level, according to the desired resolution. The coefficients $c_1(k)$, in *Figure 3.12 (a)* represent the lowest half of the frequencies in x(k), downsampling doubles the frequency resolution. In the second level, the outputs of L(z) and H(z) double the time resolution and decrease the frequency content, for instance the width of the window is increased. After each level, the output of the high-pass filter represents the highest half of the frequency resolution of the low-pass filter of the previous level, this leads to a pass-band. The time-frequency resolution of the analysis bank of *Figure 3.12(a)* is similar to the resolution represented in *Figure. 3.1.* For a specific set of filters L(z) and H(z), this structure is known as discrete wavelet transform DWT, the filters are called wavelet filters.



Figure 3.12: Processes of analysis and reconstruction in a three level filter bank.

4.6 Wavelet Denoising

in the context of wavelets, 'denoising' refers to the process of minimizing noise without degrading the signal. The wavelet transform's time-frequency-amplitude matrix is used for denoising. Wavelet denoising is based on the idea that the unwanted noise will be separated from the desired signal by their frequency ranges. Most commonly in scientific measurements, the desired signal components are typically located at low frequencies, while the noise is typically found at high frequencies. The process is influenced by both the selection of wavelet type and by a positive integer number, called the wavelet "level"; the higher the level, the lower is the frequency divider between signal and noise.



Figure 3.13: Schematic representation of the WT denoising process.

Any signal that needs to be de-noised must first be subjected to the wavelet transform before being entering into the decomposition process. Signal can be divided into groups of coefficients using the wavelet transform at various frequency levels. After the application of the first stage, it is possible to gather pertinent data regarding the signal characteristic. The next step is to choose the ideal threshold values and apply them to this set of coefficients so that unwanted data can be removed: the definition of the threshold value is a necessary step in thresholding uses. Thresholding is applied according to two methodologies [47]:

Hard thresholding is the setting elements to zero, when their absolute values are below the threshold itself.



Figure 3.14: Hard threshold application [44].

Soft thresholding is the lowering of the coefficient that are abov the threshold. Although it requires more computatational effort, soft thresholding provides better denoising performance.



Figure 3.15: Soft thresholding application.

Thus, by thresholding the wavelet coefficients using a thresholding function that includes the advantages of both hard and soft thresholding, EMG signal noise is decreased.

4.7 Scalogram Function

The scalogram function represents the maximum location of energy in the time-frequency domain and it is defined as the square of the absolute value of the CWT coefficients W_x (Eq.7)[48]:

$$P_W x (a, b) = |W_x(a, b)|^2$$
(7)

The main information that can be extracted from the scalogram function is the location in time of the maximum energy density, interpreted as the range of time in which the sEMG signal achieved its maximum energy value, i.e. the region of the gait cycle in which the muscle is primarily recruited. Further details are foiund in the the frequency location of the maximum energy density, interpreted as the frequency band where the EMG signal shows the maximum frequency content. This region of frequency varies from muscle to muscle. The most common frequency band in all muscles could be

ranged between 70 and 160 Hz. An example of 3D ans 2D representations of CWT scalogram are shown in *Figure 3.16*:



Figure 3.16: 3D representation (A) and 2D representation (B) of the scalogram of the tibialis anterior signal.

Chapter 4- Material and Methods

As stated in the previous sections, surface electromyography (sEMG) is known as a non-invasive approach specifically suited for activity monitoring muscles during dynamic forms of movement, i.e. walking. The electromyography (EMG) measurement and analysis can be used in the walking exercise to precisely locate the position of the force and the muscle contraction, further exploring and revealing the characteristics and essence of the walking movement. To obtain a space/time characterization at the muscle level during the walk, it is necessary to go and evaluate the main events on gait, such as the instant of contact to the ground (stance) and the consequent height from the ground (swing). Additionally, the assessment of muscle-recruitment timing, which influences the neuromuscolar performance, extracted from EMG becomes suitable in many application fields, particularly in the context of clinical gait analysis.

4.1 Dataset

The recorded signals data set, involved in the present study, includes 31 healthy adults (17 females and 14 males), acquired in the Analysis Laboratory of the movement of the Marche Polytechnic University, Ancona, Italy. The selected participants comprehend the following averaged characteristics:

- Age: 24.2 ± 1.9 years;
- Height: 171.4 ± 10.3 cm;
- Weight: 59.6 ± 10.9 kg;
- Healthy BMI range: $18.5 \text{ kg/m}^2 25 \text{ kg/m}^2$.

Underweight, overweight, obese people and subjects affected by any pathological condition, joint pain, or undergone orthopedic surgery, that may have affected the lower limbs mechanics, were discarded in the present dataset. The research was undertaken in compliance with the ethical principles of the Helsinki Declaration and was approved by an institutional expert committee. The data are freely accessible by consulting the archive public medical research data, *PhysioNet* [49], [50], [51].

4.2 Signal Acquisition

Relevant data of the subjects under investigation were gathered from 16-channels EMG system obtained by sensors placed on the patient's body. The sEMG signals acquisition process consisted in applying differential electrodes to the left and right lateral gastrocnemius and tibialis anterior muscles of the participants, following the SENIAM recommendations for the orientation and position of the electrodes on the muscles with respect to the tendons and fiber direction [52]. On the other hand, the signals relating to foot-to-floor contact, were detected by three basographic sensors placed under the heel, on the first and on the fifth metatarsal head of each foot and have been processed to segment the signal into steps. After the sensors location, the subjects were asked to walk barefoot, on their own natural gait for five minutes, following a figure eight trajectory (*Figure 4.1*), which implies natural deceleration, acceleration and inversion. The signal acquisition takes about 5 minutes per subject, where in the first 5 seconds, approximately, the subjects stand in an orthostatic position before starting to walk.



Figure 4.1: Illustration of the traveled path in the conducted experiment.

4.2.1 Equipment Features

- *Electrogoniometer*: it was placed on the lateral side of the limb, to measure the angle of the knee joint in the sagittal plane.
 - Precision: 0.5 degrees.
- *Pedals*: have been glued under the heel, first and fifth metatarsals of the foot.
 - \circ Surface: 1,21 cm²;
 - Activation Force: 3N.
- *sEMG probes*: single differential with fixed geometry (Ag/Ag-Cl disc).
 - Electrode diameter: 0.4 cm;
 - Electrodes distance: 0.8 cm;

- Gain: 1000;
- High-Pass-Filter: 10 Hz;
- Input Impedance: 1.5 Ohm;
- o CMRR>125 dB.



Figure 4.2.: On the left: sEMG probes on GL and TA. On the right: foot switches positioned under the foot.

4.2.2 Signal Pre-processing

The analysis initiates with the processing of the basographic signals of each subject, obtained from the foot switches positioned under the right (left) foot. This procedure allows to partition the steps in order to classify the principal gait phases. The first step consists into the quantization of the raw basographic signal (raw_f), originally partitioned into 8 levels, by computing the least significant bit (lsb) by applying the following computation:

$$lsb = \frac{max_f - \min_f}{8}$$
(3)

Where max_f and min_f represent the maximum and minimum value of raw basographic signal.

The value of *lsb* (*Least Significant Bit*) defines the spacing of the eight intervals into which the raw signal has been divided. The 8-level quantization of the raw basographic signal thus obtained is shown in *Figure 4.3*. After, the signal undergoes a further quantization to reduce the basographic partition

into 4 levels. Each level corresponds to a specific phase of the foot-floor contact: flat foot contact, heel strike, push- off and swing, as depicted in *Figure 4.4*.



Figure 4.3: Raw Basographic signal into 8 levels quantization.



Figure 4.4: Raw Basographic signal into 4 levels quantization of the first step selected.

The analysis proceeds with the selection of the initial and final steps: since the examination is carried with healthy controls, it is possible to identify the beginning of the step with the initiation of the '*heel strike*' phase and the end with the initiation of the '*heel strike*' phase of the next step. In particular, the

analysis related to the tibialis anterior was conducted by segmenting the gait starting from the swing phase, leading to a signal distribution shift. Once the steps have been identified, the signal of the muscle of interest (specifically, the adopted muscle signals for this discussion are right lateral gastrocnemius and tibialis anterior) undergo a filtering process with a Butterworth between 20 and 450 Hz. Those pre-processing steps were applied to remove both motion artifacts and high frequency noise from each muscle signal.

Subsequently, a further processing through two distinct methodologies based on *continuous wavelet transform* (CWT) and *linear envelope* and are applied. Signal filtering was implemented using MATLAB_R2022b.

To reduce the signal noise and also to provide the energy localization in time-frequency, the wavelet transform was applied by using a mother wavlet belonging to the wavelets family Daubechis of order 4 with 8 levels of decomposition (db4), as depicted in *Figure 4.5*.



Figure 4.5: illustration of 'db4' Mother Wavelet map.

This option is specifically chosen since Daubechies wavelets, whose shape similarly resembles that of motor unit action potentials, are appropriate for detecting variations in the signal. CWT of sEMG(t) is defined in the following Eq. (8):

$$CWT_{sEMG}(a,b) = \int sEMG(t)\psi_{a,b}^*(t) dt \quad a \neq 0$$

Where the mother wavelet is defined in Eq. (9):

$$\psi_{a,b}(t) = \frac{1}{\sqrt{a}} \psi \left(\frac{t-b}{a}\right) \tag{9}$$

(8)

with *a* is the scale parameter and *b* the translation parameter [16].

The multiscale decomposition of the signal into elementary components, the adjustment of the detail coefficients with respect to a specified soft threshold, and then the signal reconstruction, are the foundations of CWT denoising. This step is accomplished by using the denoising algorithm in Wavelet Matlab Toolbox and using in the denoising process the soft threshold, also known as the Donoho threshold. Within the multiscale decomposition, the signal is fragmented into elementary components with a frequency content that increases as a function of the scale increment.

4.3 Linear Envelope

Detailed information is already present in the raw EMG recording. Activations and deactivations, as well as other qualitative evaluations, can be directly derived from the raw signal and provide a crucial first step toward the understanding of the neuromuscular control used to carry out motor task. To increase the validity and reliability of the results, when quantitative analysis is the research goal, it is necessary to implement some specific EMG signal processing stages. The EMG interference "pattern" is naturally irregular since the current values of the motor units recruited are constantly changing inside the motor units and the overlapping of their action potentials is arbitrary. This abovementioned problem results in the fact that a raw EMG signal cannot be reproduced a second time with its precise form. To address this issue, the non-reproducible part of the signal is minimized by applying digital smoothing algorithms outlining the average development trend of the signal.

The linear envelope is one of the most frequently employed method to extract data from the waveform of the sEMG signal. By using this processing technique, the frequency content of the signal can be significantly reduced, and it also facilitates its interpretability and the identification of muscle activations. To obtain the linear envelope, the electromyographic signals of each control undergo the rectification process, so that the signal assumes positive polarity, and then consequent filtering with a Low-Pass Butterworth Filter at 5Hz cut-off frequency is applied to reduce artifacts from the signals. Then, the timing of LE peak is computed for every single stride.



Figure 4.6: The steps involved in linear envelope detection of the EMG signal. Upper part, the raw EMG signal is shown. In the middle, the first step is the full-wave rectification. Bottom part, the last step is the low-pass filtering. To the left of the EMG waveforms the engineering schematization of the main steps is represented on the left of the EMG wavefroms [24].

4.4 Muscle activation intervals

The wavelet coefficients are determined by scale and position, and from these, the scalogram coefficients are also determined. The scalogram, defined as the square of the absolute value of the CWT coefficients, is used to identify the localization of the sEMG signal energy in the time-frequency domain [48]. The time location of the maximum concentration of energy density can be interpreted as the time interval during which the muscle is mostly recruited. Instead, the frequency location of the maximum concentration of energy density corresponds to the frequency band where the sEMG signal exhibits the highest frequency content. Following the coefficients computation, the conversion from the x-axis in samples into the percentage of the gait cycle (%GC) is performed. Subsequently, the calculation of the maximum peak value reached with the signal energy density was performed, by setting arbitrarily a threshold on the signal intensity.

Different threshold levels (1%, 25%,50%, 75%, 90%) were set and for each of this threshold the activation intervals were assessed in order to identify activation intervals with different level of energy, increasing from 1% to 90%. The portion of the signal under the threshold is discarded. The assumption behind is that the onset and offset instants of muscle activations can be identified as the beginning and end of the time intervals where the scalogram is greater than a percentage of the peak value of the energy density defined by the threshold. If the scalogram exceeds the threshold value then activation occurs, otherwise not [21]. Therefore, all the time instants (in terms of gait cycle percentage) inside each activation interval are saved together with the corresponding frequency values. At this stage, activation intervals whose time distance is less than the 3% of the gait cycle are

considered to belong to the same muscle activation. When the threshold is equal to 1%, the activation intervals which last for less than 3% of the GC are removed because they are assumed to be false activations. This step was taken because the control of joint mobility is unaffected by time intervals of muscle recruitment lasting less than 30 ms [21]. For 25%, 50%, 75%, 90% this restriction was removed because by increasing the threshold, the intervals tend to narrow significantly, and the information content of the signal is then reduced. Inside each activation interval, the peak of the envelop and its corresponding time instant were identified. To assess the accuracy of the activation recognition of the intended muscles during walking, quantification errors was determined in samples and milliseconds. Errors are computed following three approaches: the first error is the computation of the distance between the time instant when the peak of the envelope occurs and the mean value of the activation interval. The second error is the distance between time instant when the peak of the envelope occurs and its closest extremity of the activation interval. The third error is the distance between the peak of the envelope and the peak of the sEMG original signal, irrespective of the threshold applied. In addition, the information about the belonging/non-belonging of the envelope peak to the considered activation interval was saved. A table containing all the selected steps (100 steps), the onset and offset time instant as well as the minimum and maximum frequency values for each muscle activation, the maximum peak activation in correspondence of the %GC and the three quantification errors.

Chapter 5- Results

The analysis of the present work was conducted on the main lower-limb muscles, gastrocnemius lateralis (GL) and tibialis anterior (TA), for the identification of the major activations over the 3100-stride dataset through the LE peak computation. The signals obtained from both muscles of each subject were subjected to filtering processing (Butterworth 20-450 Hz), to eliminate any artifacts and high-frequency noise components, as reported in *Figure 5.1*. In order to minimize noise and provide the localization of energy in time-frequency domain, the CWT was further applied. The respective wavelet denoised signal outcome of both muscles is depicted in *Figure 5.2*



Figure 5.1: Gastrocnemius Lateralis (GL) and Tibialis Anterior (TA) sEMG signal representations under filtering process.



Figure 5.2: Gastrocnemius Lateralis (GL) and Tibialis Anterior (TA) sEMG signal representation under wavelet denoising.

Subsequently the LE of the signals was computed over the rectified signals, as *Figure 5.3* and *Figure 5.4* show as follows:


Figure 5.3: Rectification and enveloped Gastrocnemius lateralis (GL) signal representation.



Figure 5.4: Rectification and enveloped Tibialis Anterior (TA) signal representation.

The CWT was also applied to identify muscle activation intervals based on the different degrees of activation, which revealed the concentration of the signal density. These thresholds were respectively set at 1%, 25%, 50%, 75%, 90%. The activation intervals are represented by rectangular shapes in which the ON/OFF time instants of muscle activations are identified in terms of percentage of the gait cycle. The width of these intervals varies according to the degree of the threshold applied. It is

possible to graphically observe the linear envelope of an arbitrarily selected step (step 10) of subject 1 in GL muscle in *Figure 5.5.* For a threshold set at 1% the LE peak falls within the activation interval, of amplitudes ON/OFF: [26.9%- 48.6%]GC. The maximum peak occurance is at 38.2% of the GC. The same occurs for the threshold set at 25 %, in which the LE peak falls within the activation interval, of amplitude [34.8%-43.7%] GC. For a threshold set at 50% the activation interval has an amplitude of [36.1-37%]GC, which implies that the LE peak doesn't belong to the current temporal window. As a consequensce, the increase of the threshold leads to a less suitable usage of the LE technique, despite it is able to reliably detect muscle activation. The same situation occurs for the thresholds set at 75% and 90%, in which the activation intervals have an amplitude of [36.8%-36.9%]GC and [40.9%-41.1%]GC.



Figure 5.5: Gastrcnemius Lateralis (GL). Rectified Signal, LE with muscle activation intervals, maximum peak of activation in Subject 1. The rectangular shapes highlight the different settings of threshold levels: green=Th:1%, yellow=Th:25%, purple=Th:50%, lightblue=Th:75%, black=Th:90%). On the right: zoomed representation of the left figure.

For the second analysis, it is possible to graphically observe the LE of an arbitrarily selected step (step 8) of subject 1 in TA muscle in *Figure 5.6*. For a threshold set at 1% the LE peak falls within the activation interval, of amplitudes ON1/OFF1:[0.7%-:17.7%]GC and ON2/OFF2: [31.7%-43.5%]. The maximum LE peak occurance is at 40.4% of the GC. Differently occurs for the threshold set at 25% where the amplitude of the interval is [41.1%-42.4%] GC. Additionally, the same condition prevoiusly stated occurs for the threshold set at 50%, in which the activation interval has an amplitude of [41.3-42.1%]GC. This outcome implies that the belonging of the LE peak to the current temporal window doesn't happen. The same situation occurs for the thresholds set at 75% and 90%, in which the activation intervals have an amplitude of [41.4%-42%]GC and [41.9%-42%]GC. The rectified

sEMG signal permits to distinguish visually the zones of greater muscular activity and to confirm their correspondence with those distinguished by the CWT calculation and by the LE peaks.



Figure 5.6: Tibialis Anterior (TA). Rectified Signal, LE with muscle activation intervals, maximum peak of activation (subject 1). The rectangular shapes highlight the different settings of threshold levels: green Th:1%, yellow Th:25%, purple Th:50%, lightblue Th:75%, black Th:90%). On the right: zoomed representation of the left figure.

Subsequetly, the 3-D scalogram representantion is used to identify the localization of the sEMG signal energy in the time-frequency domain. The various threshold levels, set as a percentage of the maximum value of the scalogram, are represented on a plane by different colored bars in the following figure.



Figure 5.7: Scalogram 3-D representation with the different levels of set thresholds, for GL and TA. Each bar divides the signal in different levels of energy. The color-levels represents the logarithm of the squared modulus of the wavelet coefficients (red: condition observed when the max percentage of energy is reached, dark blue: condition observed when the minimum percentage of energy is reached).

To test the reliability of the approach used in this work, statistical graphs are reported below to indicate the average of the quantified errors with the respective standard deviation, for each level of threshold, calculated with both approaches on the 31 subjects. The first approach is the computation of the error that calculates the distance between the time instant when the LE peak of occurs and the mean value of the activation interval; while the second approach calculates the distance between time instant when LE peak occurs and its closest extremity of the activation interval. The third error is the distance between LE peak and the peak of the sEMG original signal applied, irrespective of the threshold applied. The corresponding mean values for GL are respectively 66.1 ± 199.2 (as function of samples) and 33.0 ± 99.6 (milliseconds); while for TA the mean values are 157.4 ± 281.8 samples and 78.3 ± 141.0 ms (*Figure 5.10*).



Figure 5.8: Mean (SD) of the quantified errors computed on all subjects in the GL The overall results are expressed in function of samples(right) and ms (left).



Figure 5.9: Mean (SD) of the quantified errors computed on all subjects in the TA muscle. The overall results are expressed in function of samples(right) and ms (left).

for each threshold level two colors are represented, differentiated by the approach used in the error quantification. In the 1st approach for GL muscle at threshold 1%, the average values of the error and the corresponding SD in samples are equal to 62.8 ± 68.6 and 31.4 ± 34.3 in ms; for threshold 25% the values are 20 ± 21.4 samples and 10 ± 10.7 ms; for the threshold at 50% the values are 24.9 ± 42.9 samples and 12.5 ± 21.5 ms; for threshold set at 75% the corresponding values are 35.8 ± 94.0 samples and 17.9 ± 47.0 ms; finally, for threshold at 90% the values are 44.8 ± 112.4 samples and 22.4 ± 56.2 ms. For what concerns the 2nd approach for GL muscle at threshold 1%, the average values of the error and the corresponding SD in samples are equal to 191.6 ± 63.9 and 95.8 ± 31.9 in milliseconds; for threshold 25% the values are 68.9 ± 41.9 samples and 34.5 ± 21.0 ms; for the threshold at 50% the values are 32.4 ± 44.3 samples and 16.2 ± 22.1 ms; for threshold set at 75% the corresponding values are 32.4 ± 44.3 samples and 14.7 ± 46.0 ms; finally, for threshold at 90% the values at 75% the corresponding values are 36.2 ± 110.3 samples and 18.1 ± 55.2 ms.

On the other hand, in the 1st approach for TA muscle at threshold 1%, the average values of the error and the corresponding SD in samples are equal to 93.2 ± 117.3 and 46.6 ± 58.7 in milliseconds; for threshold 25% the values are 21.9 ± 21.8 samples and 11.0 ± 10.9 ms; for the threshold at 50% the values are 40.4 ± 95.6 samples and 20.2 ± 47.8 ms; for threshold set at 75% the corresponding values are 92.3 ± 209.8 samples and 46.1 ± 104.9 ms; finally, for threshold at 90% the values are 121.6 ± 254.0 samples and 60.8 ± 127.0 ms. For what concerns the 2nd approach for TA muscle at threshold 1%, the average values of the error and the corresponding SD in samples are equal to 172.4 ± 89.4 and $86.2\pm$ 44.7 in ms; for threshold 25% the values are 47.9 ± 37.5 samples and 24.0 ± 18.7 ms; for the threshold at 50% the values are 38.0 ± 92.6 samples and 19.0 ± 46.3 ms; for threshold set at 75% the corresponding values are 83.0 ± 207.8 samples and 41.5 ± 103.9 ms; finally, for threshold at 90% the values are 112.5 ± 252.6 samples and 56.2 ± 126.3 ms.



5.10: Mean (SD) of the quantified errors computed on the distance between the original sEMG peak and LE peak on all subjects in the GL and TA muscles. Left: error expressed in samples, right: error expressed in ms.

To conclude, the final graphical representation in *Figure 5.11* shows the final percentages of quantified errors for the same thresholds in both muscles.



Figure 5.11: Error percentage in correspondance of each level of threshold both for GL and TA.

The corresponding percentage values identified in the analysis of the GL on all the subjects are respectively: for threshold at 1% the LE peak belongs correctly to the intervals; therefore, it is not able to identify muscle activation in 0.03% of the steps. For a threshold of 25% the muscle activation is not identified in 4.4% of the steps; for a threshold of 50% the muscle activation is not correctly identified by LE peak in 24% of the steps; for thresholds at 75% and 90% the muscle activations are not identified in 50% and 64% of the steps.

Regarding the analysis of the TA muscle on all the subjects, the corresponding percentage values of the quantified error are respectively: for threshold at 1% the LE peak belongs correctly to the intervals; therefore it is not able to identify muscle activation in 0.1% of the steps. For a threshold of 25% the muscle activation is not identified in 13% of the steps; for a threshold of 50% the muscle activation is not correctly identified by LE peak in 41% of the steps; for thresholds at 75% and 90% the muscle activations are not identified in 64% and 72% of the performed steps.

Chapter 6- Discussion and Conclusion

The linear envelope (LE) is a processing method for the EMG signal widely used in clinics and rehabilitation contexts to detect the main muscular activities during specific motor tasks. The LE consists of a low-pass filter that provides a smooth curvilinear trend of the EMG signal under examination. Therefore, this type of signal representation guarantees an EMG signal pattern that is much easier to interpret during the motor gesture phases, compared to the typical non-uniform signal shapes. The peak of LE is typically adopted to identify muscle activations during gait [10]. The aim of the present work of thesis was to test the reliability of the LE peak of the sEMG signal for the estimation of the muscular activations during walking in healthy subjects. Specifically, validation of LE peak was performed versus reference activation intervals computed by means of a recent CWTbased algorithm run on sEMG signal from gastrocnemius lateralis (GL) and its antagonist muscle, i.e., the tibialis anterior (TA) [21]. This validation was performed by counting the instances in which LE peaks fall within the activation intervals identified by the CWT algorithm and quantifying the error when that condition does not occur. The first relevant result lies in the fact that all the LE peaks fall within the activation intervals detected by the CWT-based algorithm (Figure. 5.11). This allows to appreciate as the LE peak is a reliable tool for the identification of the main muscle activation during able-bodied walking. However, the present analysis was designed to go further.

LE peak represents the maximum value of signal energy assessed by the envelope. In order to test the accuracy of LE peak in identifying the timing of the zone of sEMG with high levels of signal energy, increasingly selective thresholds levels were set in the CWT-based algorithm (25%, 50%, 75%, 90%). Increasing the selectivity of the threshold reflects in reducing the duration of the intervals and increasing the energy signal level included in each interval, i.e., intervals detected with 90% threshold are shorter and identify only that part of the signal characterized by a signal level higher than 90% of signal peak value. A further aim of this study was just to quantify the possible loss of accuracy of LE peak when the activation intervals are reduced to include only high levels of signal energy. What we would expect from this analysis is a reduction of detection accuracy for highest threshold levels. Indeed, Figure 5.11 shows as the number of times when LE peaks do not fall within the activation intervals detected by the CWT algorithm (false negative) rapidly increases with the increase of the threshold level up to 90%. This outcome suggests that the performance of LE peak worsens when used to identify the timing of heigh energy levels of the signal. Specifically, for thresholds equal to 50% a false negative value is detected in 25% of the strides for GL and over 40% for TA. For threshold equal to 75% and 90%, false negative is even more than 50% of the total strides. This is due to the fact that the increment of the threshold level minimizes the amount of selected signal, resulting in an

erroneous identification of the muscle activation interval.

Moreover, the graphs in *Figure 5.8*, *Figure 5.9* show the average errors (and standard deviation) computed with both approaches on all the subjects for each level of threshold. The first approach was the computation of the error defined as the distance between the time instant when the LE peak occurs and the mean value of the activation interval; while the second approach calculated the distance between time instant when LE peak occurs and its closest extremity of the activation interval. The third error was the distance between LE peak and the peak of the sEMG original signal, irrespective of the threshold applied. Through the support of this CWT-based algorithm it was possible to predict the presence of an error of several milliseconds effected by the LE, which depedended on the performed step and signal intensity.

In fact, from the graphs in the *Figure 5.8* and *Figure 5.9*, it can be seen that, although with the 1% threshold the LE peaks fall within the activation interval, the average errors calculated with both approaches are greater than that with all the other thresholds tested. Whereas, the average errors (ms) increase with increasing of threshold value (from 25% up to 90%), confirming that the LE peak seems to be inadequate to characterize high energy zones of the EMG signal. Furthermore, as the threshold increases, the error values show a high variability (represented by a high standard deviation value) since the identification of the intervals depends particularly on the selected step and on the signal intensity which can vary from subject to subject.

In conclusion, the results of this work suggest that LE peak could be adopted to reliably identify the number of muscle activations during a complex task as human walking. However, when a more refined analysis is required, as for example the identification of the timing of EMG-signal high energy, the use of LE envelope should be considered carefully because it could introduce a relevant inaccuracy in the identification of the muscle activation in terms of both number of detected activations and error in identifying their timing.

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