

# Experimental investigation of the inversion/eversion motion for the design of a two degrees of freedom ankle prosthesis

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ABSTRACT

Bipedal walk is a very complex gait. Lower limb amputees know this very well since it is challenging to recreate healthy ankle motions with a robotic ankle prosthesis. It always leads to some comfort and stability reduction, which often limit the daily life activities participation for amputees.

In order to propose some guidelines that can improve the design of two degrees of freedom ankle prosthesis, this master thesis focuses on the analysis of the inversion/eversion movement of a healthy ankle.

In a first step, the normal life outside conditions were characterized to be able to recreate similar situations inside the lab. Therefore, an electromechanical device has been developed to measure ground angles that are reached during a normal walk in the street as well as the ankle inversion/eversion angle during some daily life movements. The device is designed to fulfill the specifications needed to allow accurate measures without disturbing the user.

These preliminary tests enable us to acquire experience in the experimental domain. They allow us to identify which kinds of gait use the ankle inversion/eversion degree of freedom. These results were used to select the moves analyzed further in the lab.

The second part of this study focuses on the in-lab measurements of the inversion/eversion of 4 gaits selected on base of the outside conditions observations: turning step, lateral step, 15° inclined plane step and 30° inclined plane step. The torque of inversion/eversion in the ankle is computed for these different gaits in order to analyze the behavior of a healthy ankle in the frontal plane. The results reveal that the inclined plane steps use more the inversion/eversion movement than the normal walk. The behavior of the ankle in the frontal plane appears to be correlated with the inclination of the ground. In the turning and lateral steps, the inversion/eversion movement was comparable to that in the normal walk. The determination of the characteristic impedance shows a linear profile for all gaits and was very comparable between the two inclined plane tests. The computed power involved in the frontal plane shows a correlation with the sagittal plane.

Finally, some guidelines are proposed for the future design of an assistive prosthetic device targeting the eversion/inversion motion, in order to have the best two degrees of freedom robotic ankle prosthesis as possible. The characteristic impedance being linear, a semi-active or even a passive system may be possible.



## ALGORITHM S DEVELOPED DURING THIS WORK

### *Algorithm name*

### *Details*

#### *Step detection*

This algorithm was entirely developed during this thesis. Described in Algorithm 2 in section 3.5.2 (page 31), it detects a stance phase with a high reliability using only an accelerometer.

#### *Remove outliers*

This algorithm removes an imprecision of the optoelectronic system called "marker jump". Based on statistics, this algorithm detects an outlier in a set of measurements and replaces it with an interpolated value. It is described in Algorithm 3 in section 5.2.1 (page 47).

#### *Foot contact detection*

Based on existing literature this algorithm detects the different contacts of the foot with the ground. The difference with the existing one is that ours can be used out of a treadmill. This iterative algorithm is presented in Algorithm 4 in section 5.2.2 (page 48).



## LIST OF ABBREVIATIONS

### *Markers*

|                |                          |
|----------------|--------------------------|
| $M_{RHeel}$    | Right heel               |
| $M_{RMeta5}$   | Right metatars 5         |
| $M_{RMeta1}$   | Right metatars 1         |
| $M_{RToe}$     | Right toe                |
| $M_{LHeel}$    | Left heel                |
| $M_{LMeta5}$   | Left metatars 5          |
| $M_{LMeta1}$   | Left metatars 1          |
| $M_{LToe}$     | Left toe                 |
| $M_{RExtMal}$  | Right external malleolus |
| $M_{RIntMal}$  | Right internal malleolus |
| $M_{RTibProt}$ | Right tibia tuberosity   |
| $M_{LExtMal}$  | Left external malleolus  |
| $M_{LIntMal}$  | Left internal malleolus  |
| $M_{RExtKnee}$ | Right external knee      |
| $M_{RIntKnee}$ | Right internal knee      |
| $M_{LExtKnee}$ | Left external knee       |
| $M_{LIntKnee}$ | Left internal knee       |
| $M_{C7}$       | C7 vertebra              |
| $M_{L5}$       | L5 vertebra              |

### *Special points*

|     |                    |
|-----|--------------------|
| CoR | Center of rotation |
| AoR | Axis of rotation   |
| CoP | Center of pressure |

### *Notations*

|            |                                         |
|------------|-----------------------------------------|
| $A_{ie}$   | Inversion/eversion ankle's angle        |
| $A_{Foot}$ | Angle of the foot in the inertial frame |
| DOF        | Degree of freedom                       |



|          |                                                                                         |           |
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Two million years ago, the *Hominidae* started to control bipedal locomotion. This wonderful evolution allows to liberate the upper-limbs and so to manipulate tools while walking.

Nowadays, bipedal locomotion seems so natural that most people have a hard time to imagine the complexity of this gait. However, lower limb amputated people realize how difficult and how ambitious it is to try to reproduce the walk with a robotic prosthesis.

## 1.1 Context

A hundred years ago, there was the First World War. There was a lot of *gueules cassées*, which is a French term for the soldiers that were injured or amputated. These unfortunate people were repaired the best they can with prostheses in wood, metal or porcelain. However, few of them recovered a normal life. A century later, the advances in bionics and prosthesis suggest to the mutilated in Iraq and Afghanistan the hope not to relive the trauma of their elders.

The war is not the only reason of amputation, it can affect more people than one might think. A study was made in order to evaluate the limb amputation and it has been shown that most amputations occurred in patients over 60 years and that the diabetes is the reason of amputation from 25 to 90% of the cases (depending on the country). Table 1.1 present the amputation causes in function of the country. The proportion of amputations associated with diabetes varied from between 20% (major amputations in women in Vincenza) up to 91% (major amputations in women in Navajo) [1].

In 2007, there was a prevalence rate of people treated for diabetes of 3.95% in France, which represents 2.5 million individuals. The diabetes epidemic continues to progress beyond the last previsions of experts [2]. This high prevalence rate in France but also in many places in the world explained why there are more than 1.6 million amputees in the USA<sup>1</sup> [3], 8.8 (per 100 000 population per year) in Vicensa (Italy) or 3.6 (per 100 000 population per year) in Madrid (Spain) [1].

---

<sup>1</sup>Both lower limb and upper limb but the majority is lower limbs (and so the ankle).

|                   |       | Men    |          |           |     | Women  |          |           |     |
|-------------------|-------|--------|----------|-----------|-----|--------|----------|-----------|-----|
|                   |       | Trauma | Diabetes | Infection | PVD | Trauma | Diabetes | Infection | PVD |
| Leeds, UK         | Major | 4      | 40       | 57        | 83  | 10     | 36       | 58        | 70  |
|                   | Minor | 10     | 46       | 60        | 57  | 0      | 29       | 33        | 26  |
| Leicester, UK     | Major | 0      | 48       | 33        | 88  | 2      | 36       | 17        | 81  |
|                   | Minor | 5      | 57       | 43        | 62  | 0      | 41       | 12        | 59  |
| Madrid, Spain     | Major | 0      | 80       | 60        | 93  | 0      | 60       | 40        | 100 |
|                   | Minor | 8      | 69       | 77        | 100 | 0      | 80       | 80        | 80  |
| Middlesbrough, UK | Major | 10     | 44       | 54        | 77  | 9      | 36       | 50        | 73  |
|                   | Minor | 0      | 59       | 71        | 65  | 5      | 43       | 38        | 62  |
| Montgomery, USA   | Major | 3      | 64       | 14        | 72  | 2      | 62       | 9         | 60  |
|                   | Minor | 2      | 60       | 32        | 89  | 1      | 51       | 26        | 68  |
| Navajo area, USA  | Major | 10     | 90       | 85        | 51  | 0      | 91       | 100       | 65  |
|                   | Minor | 6      | 81       | 88        | 16  | 10     | 86       | 90        | 17  |
| Newcastle, UK     | Major | 10     | 33       | 37        | 71  | 15     | 23       | 31        | 62  |
|                   | Minor | 14     | 36       | 54        | 64  | 0      | 29       | 29        | 50  |
| Ilan, Taiwan      | Major | 50     | 41       | 53        | 56  | 16     | 53       | 84        | 74  |
|                   | Minor | 57     | 38       | 43        | 43  | 25     | 75       | 75        | 75  |
| Tochigi, Japan    | Major | 3      | 33       | 49        | 55  | 0      | 41       | 22        | 30  |
|                   | Minor | 0      | 64       | 27        | 36  | 20     | 100      | 40        | 20  |
| Vicenza, Italy    | Major | 0      | 61       | 22        | 78  | 4      | 20       | 4         | 88  |
|                   | Minor | 7      | 60       | 60        | 60  | 12     | 50       | 38        | 62  |

**Table 1.1:** Conditions associated with first ever major and minor amputations (values are percentages and more than one condition may be present per patient. PVD = Peripheral Vascular Disease [1].

Table 1.2 presents re-amputation and mortality rates in function of the initial amputation level and presence or absence of diabetes. Among those with diabetes, which are more frequent accordingly to Table 1.1, persons with initial amputation at the toe, foot or ankle and bi-lateral levels were most likely to experience at least one re-amputation surgery during the 12-month after the surgery (37.4%, 40.1% and 38% respectively) [4].

The cause of re-amputation could be the comorbidity factor, which is the poor quality of life after amputation. The study of Pell J.P. et al. [5] shows that amputated people have problems with mobility, social isolation, lethargy, pain, sleep and emotional disturbance [5].

| Initial Amputation Level | Reamputations (%) |       |             | Mortality (%) |
|--------------------------|-------------------|-------|-------------|---------------|
|                          | None              | One   | Two or More |               |
| All persons              |                   |       |             |               |
| Toe                      | 65.1              | 24.0  | 10.9        | 22.8          |
| Foot and/or ankle        | 61.2              | 28.8  | 10.0        | 28.5          |
| Transtibial              | 77.0              | 19.7  | 3.3         | 35.5          |
| Transfemoral             | 86.0              | 12.0  | 2.0         | 50.4          |
| Bilateral                | 70.6              | 17.7  | 11.7        | 53.0          |
| Diabetics                |                   |       |             |               |
| Toe                      | 62.6*             | 25.3* | 12.1*       | 22.7          |
| Foot and/or ankle        | 59.9*             | 29.0  | 11.1*       | 27.1*         |
| Transtibial              | 75.4*             | 20.6* | 4.0*        | 34.0*         |
| Transfemoral             | 82.4*             | 14.4* | 3.2*        | 50.2          |
| Bilateral                | 61.5*             | 23.1* | 15.4*       | 46.2*         |
| Nondiabetics             |                   |       |             |               |
| Toe                      | 77.0              | 17.5  | 5.5         | 24.5          |
| Foot and/or ankle        | 67.7              | 27.7  | 4.6         | 35.4          |
| Transtibial              | 83.3              | 16.2  | 0.5         | 41.5          |
| Transfemoral             | 91.1              | 8.5   | 0.4         | 50.5          |
| Bilateral                | 100.0             | 0.0   | 0.0         | 75.0          |

**Table 1.2:** Twelve-month re-amputations and mortality among medicare beneficiaries undergoing amputation, by level of initial amputation [4]

|                                            |           |
|--------------------------------------------|-----------|
| Average hours/day using prosthetic (range) | 12 (8-12) |
| Use of walking aid                         |           |
| No                                         | 8         |
| Yes                                        | 3         |
| Difficulty with usual activities           |           |
| Considerable                               | 0         |
| Some                                       | 2         |
| Slight                                     | 4         |
| None                                       | 5         |
| Modify activities for fear of falling      |           |
| No                                         | 7         |
| Yes                                        | 4         |

**Table 1.3:** *Questionnaires for prosthetic feet user : 11 people answered [6]*

Fogelberg et al. made a study in order to increase the understanding of the impact of a prosthetic foot on participation in the activities of daily life. Table 1.3 shows some characteristic of amputated people, in which six lower-limb amputated people on 11 have some problems with usual activities and that 4 of these 11 amputees has to change their activity for fear of falling.

The result of this study shows that foot prostheses are well suited for a forward walking on even ground. However, "standing tolerance, sideways or backwards movement, traversing uneven ground or slopes, and various surfaces constituted significant challenges for people using prosthetic feet" [6].

Knowing that amputated people have a gait speeds 30 to 40% slower than healthy walkers and expend up to 30% more metabolic energy [7] and knowing how many people are affected, make their lives better and more enjoyable is laudable. This master thesis will focus on the inversion/eversion movement of the ankle (movement in the frontal plane, defined in Figure 1.4 on page 5) in order to reduce the pain, the fear of falling and to increase the mobility with an ankle prosthesis which adapts itself to the ground. This could reduce the number of re-amputations or at least it could increase the quality of life of amputated people in terms of comfort and mobility.

This thesis focus on this degree of freedom (DOF) also because 8 to 50 percent of the walking movements in daily life are sidestepping [8], which require motion in the frontal plane. In addition, the ground is rarely perfectly flat. The foot is inherently compliant, adapting to the ground inclination to have a good adherence. For a prosthesis, this means that allowing the inversion/eversion movements could be very useful in that case.

This frontal plane DoF will be investigated through the characterization of the outside conditions and the inversion/eversion movement analysis. The first one will consist of measurements of the ground angles to be able to recreate the outside condition inside a lab. The inversion/eversion movement analysis will be performed in-lab thanks to the outside conditions determined by the first part in order to characterize the inversion/eversion of a healthy subject both kinematically and dynamically.

This last part will be made in collaboration with the ULg, in the Laboratory Human Motion Analysis (*LAMH*)

## 1.2 Human walking gaits

This section aims to explain on the main lines the biomechanic of the walk gaits. As presented in Figure 1.1, there are different phases in the walk. The stance phase corresponds to the phase on which the foot touches the ground, from the Heel contact to the toe off moments. The swing phase corresponds to the entire time where the foot does not touch the ground, from the toe off to the heel contact events.

The entire phase is composed of the stance phase and the swing phase. This cycle is composed of single support (SS) and double support (DS) moments (Figure 1.1).

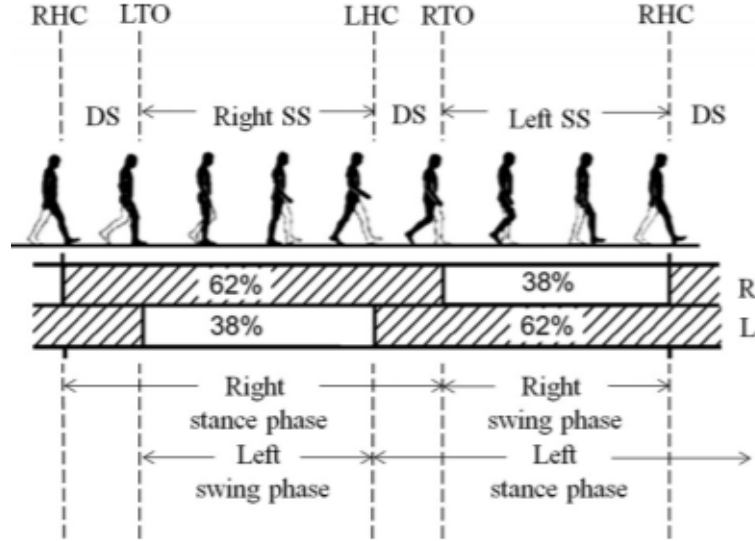


**Figure 1.1:** Walk gait cycle and phases. *RHC = Right Heel Contact. LTO = Left Toe Off. LHC = Left Heel Contact. RTO = Right Toe Off. SS = Single Support. DS = Double Support* [9]

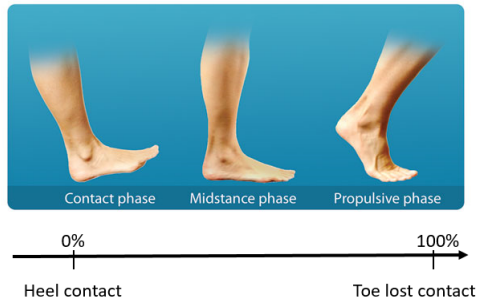


**Figure 1.2:** Stance phase definition [10]

For the analysis, the stance phase is the most important. Figure 1.2 present the definition of this particular phase.

At 0% of the stance phase the heel just hits the ground and at 100% of the stance phase the heel has no contact with the ground anymore.

The name of human planes are depicted in Figure 1.3. There is the sagittal plane which is the main plane involved in the walking gaits and that is implemented by most of prosthetic feet. The frontal plane is the one that is investigated in this thesis and finally the transverse plane which are not involved in this work.

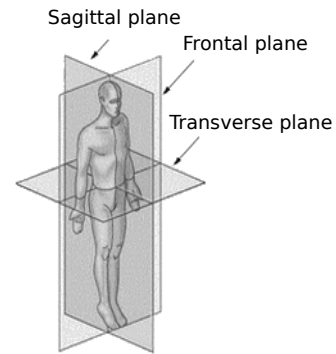


**Figure 1.3:** Definition of sagittal, frontal and transverse planes [11]

For each plane (Figure 1.3), there is an associated movement of the foot. Figure 1.4 presents the definition of the plantar flexion/dorsiflexion movements (sagittal plane), the inversion/eversion movements (frontal plane) and the external/internal rotations (transverse plane).

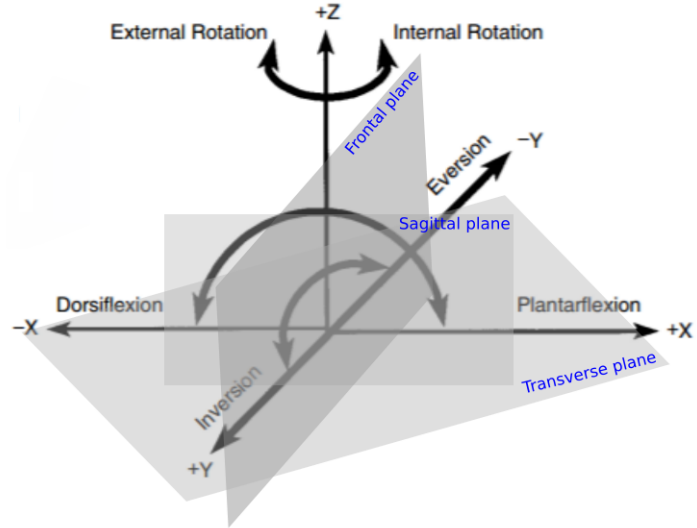


**Figure 1.4:** *Defintion of the inversion/eversion, plantar flexion/-dorsiflexion and internal rotation/external rotation movements of a foot. Inversion/eversion are in frontal plane (y-z plane), internal/external rotations are in transverse plane (x-y plane) and plantar flexion/dorsiflexion are in the sagittal plane (x-z plane) [12]*

This thesis will focus on the inversion/eversion and plantar flexion/dorsiflexion movements. The Figure 1.5 is a more illustrative representation of these two movements. The conventions of these angles are the following : the inversion movement is a positive angle while the eversion is a negative one. In the same way, the dorsiflexion is a positive angle while the plantar flexion is a negative one. These conventions will be recalled in the beginning of the section that use them.

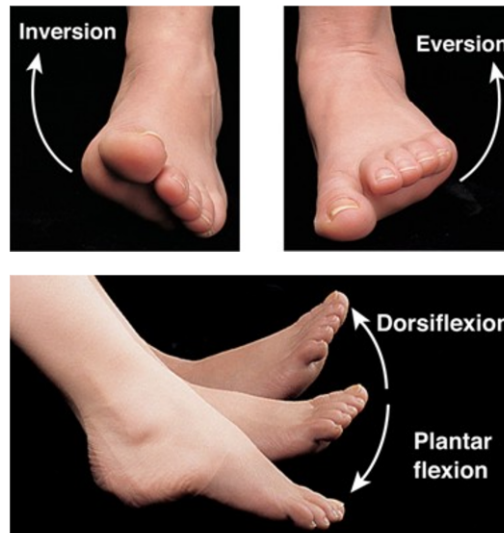


**Figure 1.5:** *Illustration of the inversion/eversion and plantar flexion/dorsiflexion movements [13]*

### 1.3 Project objectives

The main objective of this thesis is the characterization of the inversion/eversion motion during daily life activities to infer whether this degree of freedom should be embedded in a prosthesis. To achieve this, the project was conducted in two steps, first the daily life conditions characterization and secondly

highlight important behaviors of the ankle in these daily life conditions.

The aim of the first part is to create an out of the lab device measurement. It will have different purposes such as determine which kinds of movements seems to use a lot the inversion/eversion movement, its range of motion and how to recreate an out-lab environment.

In the second steps, once data of the device are collected, the results will be used in order to recreate the out-of-the-lab conditions in a laboratory and to properly conduct an experimental campaign. The purpose is be to highlight important behaviors of the ankle while walking in an outside environment with more accurate and richer measurements. The added-value of the frontal motion of the foot will be determined within these experiments.

Finally, the best kind of mechanism it can be suggested in order to implement the DoF allowing motions in the frontal plane on the basis of our results will be discussed. This section will be a guideline for further work, in which the properties that should be fulfilled by a mechanism in the light of our experimental results are listed.

## **1.4 Reading guidelines**

These reading guidelines will help the reader to understand the structure of this work. It will briefly describe the content of each chapter.

### **Chapter 2 - State of the art**

This chapter aims to review the literature on what is already done about feet prosthetic. Then it focus on two DoFs ankle prosthesis.

### **Chapter 3 - Kinematic measurements during daily life activities**

This chapter described the experimental device. We present how we have chosen this design, how it works and what are the technical specifications. It is also here that we expose how we have post-treated the collected data.

We also explain what were the preliminary tests and their experimental protocol.

### **Chapter 4 - Results of the measurements in daily life context**

This chapter includes the results obtained with the out-of-the-lab measurement device and their discussion. It is also in this part that we justify the kind of movements we explain how we designed the protocol for the second experiment (in-lab), in order to reproduce out-lab environments in the lab.

### **Chapter 5 - Ankle dynamics in-lab experimentation; Experimental protocol & Data processing**

This chapter present first the experimental protocol of the in-lab measurements. We review the required equipment, the method and the data interpretation. We get into the data processing of the in-lab measurement. The difference with the first experiments is the precision but mostly the possibility to compute the dynamic part of the movements.

After this, we explain the dynamic model and the dynamic computations. The end of this chapter gives the equations allowing us to compute the dynamic part of the results, which are in the next chapter.

This entire part and the newt one was possible thanks to ULg, who has accepted the collaboration for the *LAMH* laboratory.

## **Chapter 6 - Ankle dynamics in-lab experimentation; Results & Discussion**

The results of the in-lab measurement are analyzed and discussed here. First, we make a validation of the model by comparing some results to the literature. Then we analyze the kinematics (angle and angular speed) and the dynamic (torque, characteristic impedance and power) of each tests we performed.

## **Chapter 7 - Guidelines for the mechanical design of a two degree of freedom robotic ankle prosthesis**

To go further, we present preliminary efforts towards the mechanical design of an ankle prosthesis with two degrees-of-freedom. From our results, we are able to introduce some guidelines to design a better 2 DoFs foot prosthesis. The purpose of this part is to investigate the optimal design (active, passive or semi-active) that should be implemented.

## **Chapter 8 - Conclusion, limitations and perspectives**

All the limitations of our model will be given in this chapter. The perspectives will also be presented in order to improve the model in a further work. This chapter also summarizes all the work in a few lines. It answers the objectives we stated in the introduction chapter.



## 2.1 Overview of the lower limb prosthesis

The prosthetic foot is nothing new. Its evolution is a long story from its primitive beginning to its sophisticated present, to the exciting visions of the future. This story began about 1500 B.C with the Egyptians. They were the early pioneers of prosthetic technology with wood prosthesis. The first prosthesis ever discovered was a toe prosthesis found in Egypt and dated to between 950-710 B.C.E. (Figure 2.1) [14]. Then, in the Renaissance (1400s to 1800s), the first more complex mechanism appears [15].



**Figure 2.1:** *First lower limb (toe) prosthesis ever discovered in Egypt and dated to between 950-710 B.C.E. [14]*



**Figure 2.2:** *Prosthesis with articulated feet thanks to tendon [15]*

In 1800, James Potts designed a prosthesis composed of wood for the shank and metal for the knee joint. This prosthesis had an articulated foot that was controlled by catgut<sup>1</sup> tendons from the knee to the ankle [15].

Until the 80s, the purpose of a prosthetic foot was to restore basic gait such as walking. But in order to improve the lives of amputees, the prosthetic feet were significantly improved such as it allows to walk more fluidly, reduce the metabolic energy consumption and allow more complicated gait.

<sup>1</sup>Catgut is a rigid rope made from sheep intestines.

Today, there are three main categories of prosthetic feet (Figure 2.3): the conventional feet (CF), the energy-storing-and-returning feet (ESR feet) and the bionic feet [16].

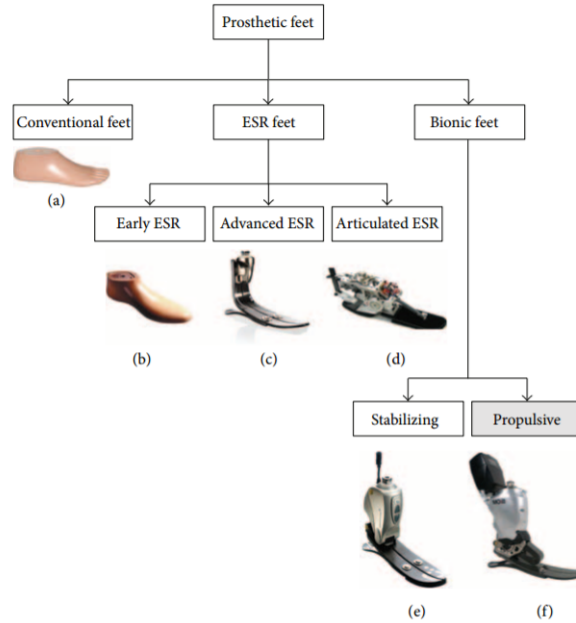


**Figure 2.3:** Categorization of today's prosthetic feet with for example the SACH foot (a), the SAFE foot (b), Össur's Flex-Foot (c), the CESR foot (d), the Össur's Proprio Foot (e) and iWalk's Powerfoot (f) [16].

### 2.1.1 Conventional feet

The conventional feet will not be described into details as it is the simplest type of non-articulated foot. The most common prosthesis of this type is the SACH foot, which is the acronym of "Solid Ankle Cushioned Heel". It refers to a compressible heel wedge that provides "pseudo-plantar flexion". It provides mid-stance stability but little lateral movement.

The SACH prosthesis is used (especially in low-income country) because it is the most robust and the less expensive of all feet prosthesis [17].

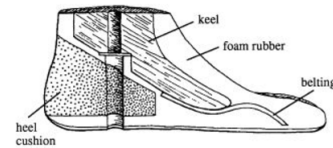


**Figure 2.4:** Postoperative SACH foot [18].

### 2.1.2 Energy Storing and Returning prostheses (ESR feet)

An energy-storing-and-returning prosthetic feet is capable of storing energy during stance phase and returning it to the amputee to assist in forward propulsion in late stance.

The advantages of ESR prosthetic feet compared to conventional prosthetic feet has been studied by Graham LE et al. [19]. It is shown that while wearing the Vary-Flex (Figure 2.6 D), the subject is significantly more likely to walk faster in the gait lab and to perform more equal steps. Subjects also had more symmetric peak ankle dorsiflexion angles in late stance and there was significantly more, at least three times [19], prosthetic ankle plantar flexion power at push-off.

The first ESR foot was the Seattle foot (in 1981). It incorporates a flexible keel inside a shell of polyurethane. Other manufacturers followed this idea (Figure 2.5).

This kind of prosthesis lost a lot of energy due to friction and is dissipated in heat and sound. This is why a completely different prosthetic foot concept is commercially available since 1987 [17].

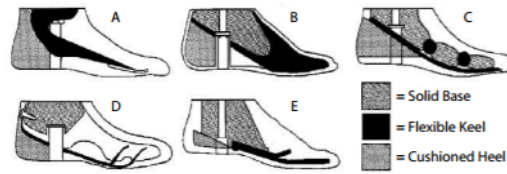


**Figure 2.5:** *Energy-storing-and-returning feet. (A) Seattle Foot. (B) Dynamic Foot. (C) STEN Foot. (D) SAFE Foot. (E) Carbon Copy Foot [17].*



More advanced ESR feet (Figure 2.6) are different from less advanced ESR feet as it allows the entire prosthesis, rather than solely the foot part, to flex, store, and return energy to the amputee.

They are composed of a flexible 100% carbon fiber shank and a heel spring [17].

**Figure 2.6:** *Advanced ESR feet. (A) Flex-Foot Axia. (B) LP-Ceterus. (C) Talux Foot. (D) VariFlex. (E) Re-Flex VSP, (F) ModularII. (G) Flex-Spring. (H) Sprinter. (I) Advantage DP. (J) Pathfinder [17].*

All these prostheses are passive prosthesis. It means that no power is added to the mechanism in order to facilitate the amputee's movements. Semi-active ESR prostheses are also available. Even if no power is added in the mechanism, some power is used to control it. For instance, the energy that a spring stored can be released at the right time. Two of them will be reviewed.

Finally an example of an active foot prosthesis will be shown. In this case, a motor injects power in the mechanism.

The AMP 1.0 foot prosthesis (Figure 2.7) aims to deliver an extended ankle push-off using a specific planetary gearbox arrangement and locking mechanisms.

It was tested on a transfemoral amputee and it appears that the extended push-off is not really felt. Furthermore, the prosthesis is too heavy, according to the subject [20].

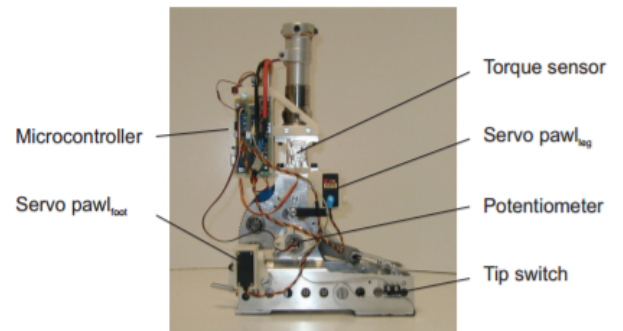


**Figure 2.7:** *Prototype of the semi-active AMP 1.0 feet prosthesis [20]*

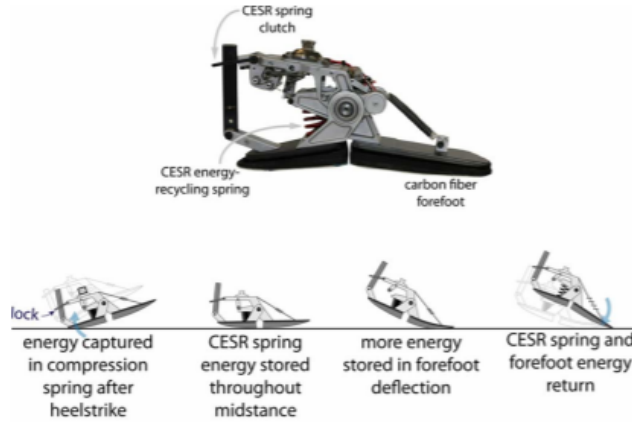


**Figure 2.8:** *CESR prosthesis prototype* [21]

The Vanderbilt Powered Ankle Prosthesis prototype shown in Figure 2.9, has a mass of 2.29 Kg including the battery and electronics. This kind of prosthesis is not a semi-active prosthesis anymore because there is a motor that can generate up to 110 Nm as peak torque. However, it is still an ESR prosthesis since there are parallel spring in the foot which engages in a predetermined ankle angle and supplements the motor output with additional plantarflexive torque [22].

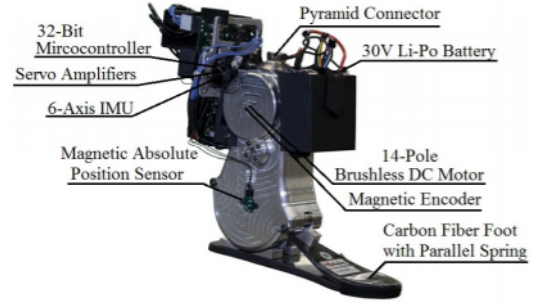


**Figure 2.9:** *Vanderbilt Powered ankle prosthesis prototype* [22]

### 2.1.3 Bionic feet

Bionic foot is defined as a mechanical device with one or more active components worn by transtibial amputated [23]. Some examples are shown in Figure 2.10.

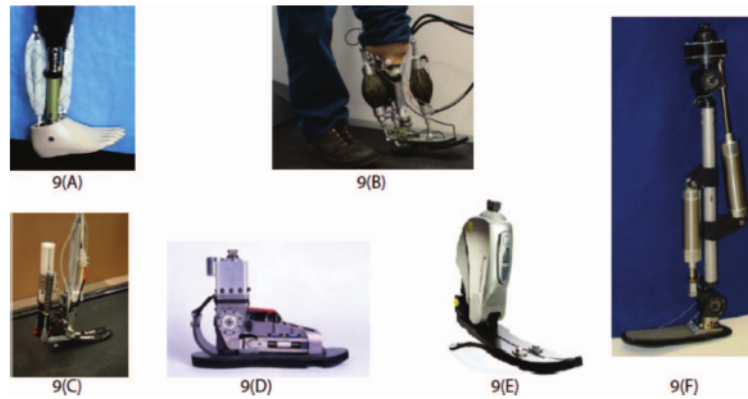


**Figure 2.10:** *Bionic feet. (A) Transtibial prosthesis powered by McKibben artificial muscles. (B) Transtibial prosthesis powered by PPAMs. (C) SPARKy. (D) Electrically driven foot of MIT. (E) Proprio foot. (F) Powered transfemoral prosthesis of Vanderbilt University* [23].

They can be categorized based on their actuating principle as electrically driven devices and pneumatically driven devices. They can either provide stabilization (help for the stability in slopes and slippery surfaces)

or propulsion [23]. Two examples of stabilizing bionic ankle prosthesis are presented in Figures 2.11.



**Figure 2.11:** *Raize (left) and Elan (right) bionic foot electrically driven [24, 25]*

#### 2.1.4 Most advanced active lower limb prosthesis

The bionic foot prostheses are still on the research stage. Despite some are already on the market, there is a lot of research about them. The improvement in the bionic prosthesis begins only. This section shown and briefly described the mechanical architectures of the most advanced prostheses that present a promising future.

##### Four-Bar foot prosthesis

The four-bar foot prosthesis is presented in Figure 2.12. This prosthesis is based on a four-bar mechanism used in conjunction with a motor and a spring to restore normal human ankle moments during the stance phase of gait. While not fully develop, this mechanical design seems to demonstrate a dramatic departure from prostheses previously developed. A prototype has already been developed and tested [26].



**Figure 2.12:** *Four bar foot prosthesis prototype [26]*

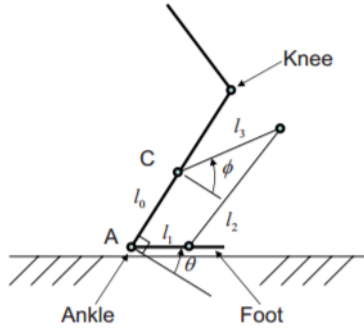


**Figure 2.13:** Model of the four-bar prosthesis configuration where  $C$  is the location of the spring/motor and  $l_0 - l_3$  are the links of the four-bar mechanisms [26]

Through this kind of implementation (Figure 2.13), the spring and damper were placed on joint  $C$  instead of joint  $A$ . This is a better position in a volumetric point of view. An advantage is that the four-bar design is only reliant on the mechanism's length ratios [26].

### Pantoe prosthesis

The Pantoe prosthesis is depicted in Figure 2.14. It is a design of a new powered ankle-foot prosthesis with compliant ankle and a segmented foot. In order to make the walking gaits of the amputees more stable and fluid, the foot is segmented and there is a toe joint. There is two series-elastic for the ankle that aim to provide enough torque tolerance shocks.

The weight is 1.47Kg (not including the battery of about 1Kg), which comparable to the weight of a human ankle [27].

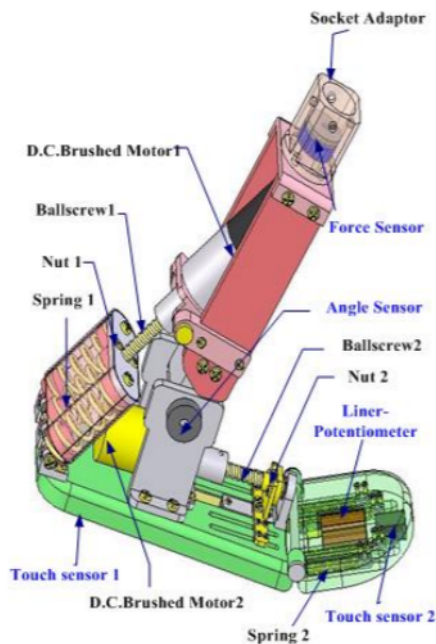


**Figure 2.14:** The CAD models of PANTOE 1 with compliant joints and a segmented foot (left) and the prototype of PANTOE 1. It is made of aluminum-alloy [27]

### The MIT prostheses

The Massachusetts Institute of Technology (MIT) has developed a different kind of prosthesis. Some of them are presented in Figure 2.15. They are prototypes that lead to the design of some ankle prostheses

such as the PowerFoot One [16].



**Figure 2.15:** *Example of bionic prosthesis develop by the MIT [16]*

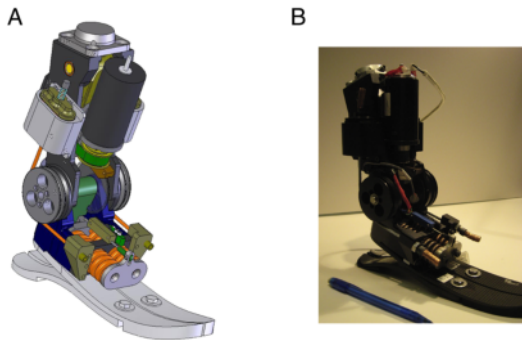


**Figure 2.16:** *Powered ankle-foot prosthesis. (A) CAD model and (B) the physical prototype [28]*

The aim of the MIT prosthesis shown in Figure 2.16 is to study amputee-machine interaction. The particularity of this new design is that the mechanical impedance can vary during the stance period of walking, similarly to a normal human ankle.

Furthermore, this design provided a sufficiently large instantaneous power output and torque to allow the amputee to walk fluidly. The main advantage of this design is his low weight to power output ratio [28].

The basic architecture of the mechanism is a physical spring configured in parallel to a high power, force-controllable actuator and a series elasticity. The five principal components of the mechanism are the following : high power DC motor, a transmission, a series spring, a uni-directional parallel spring and a foot made of carbon composite [28, 29].

### Pneumatically actuated ankle prosthesis

Several advanced design of pneumatically actuated prostheses exist (Figure 2.17). There are the Washington's variable stiffness prosthetic ankle (a), a PPAM (pleated pneumatic artificial muscle) actuating prosthesis from the *Vrije Brussel Universiteit* (b), a rocket-like propellant ankle-foot from the *Alabama* and *Vanderbilt* University (c), a sleeve muscle actuation prosthesis designed by Zheng and Shen and finally an ankle prosthesis from *Vanderbilt* University [16].

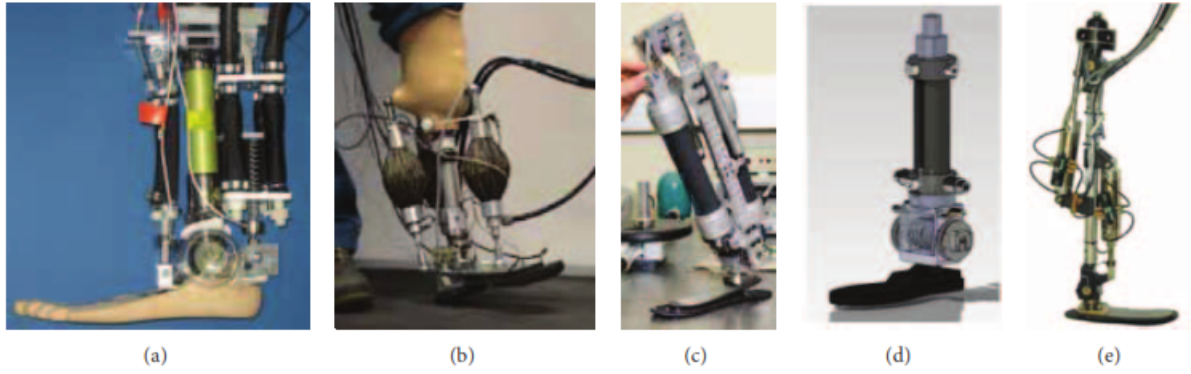


**Figure 2.17:** *State of the art in pneumatically actuated ankle prosthesis [16]*

### SPARKY 1 and 2

The SPARKY prosthesis is developed in *Arizona State University*. SPARKY means Spring Ankle with Regenerative Kinetics). It uses a robotic tendon actuator and a 150W brushed DC motor. In a walk, it is able to provide until 100% of the push-off power while maintaining intact gait kinematics.

There are 3 prototype of SPARKY prosthesis. The first one is able to store and release approximately 16J of energy per step. However, a not damaged ankle (for an 80Kg walking at 0.8Hz) needs almost 36J [16, 30, 31].

The second prototype is an amelioration of the first one. In fact, the second prototype was built with a lighter and more powerful roller screw transmission and brushless DC motor.

The third prototype has an incredible improvement since it was designed to control both the plantar flexion-dorsiflexion and the inversion-eversion movements. But this prototype has not been built to the author's knowledge. Section 2.2 will discuss more about this prototype [16, 32].

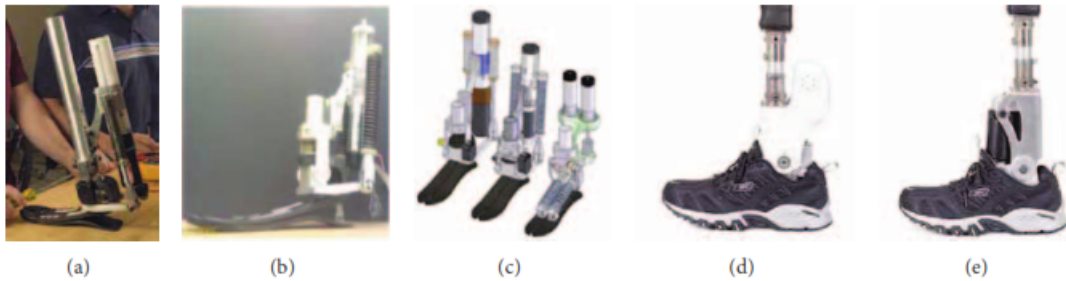


**Figure 2.18:** *Ankle-foot prototypes within the SPARKY project at Arizona State University (USA) [16]*

## 2.2 Two degrees of freedom prosthesis

As described in point 2.1.4, there is not a lot of existing 2 degrees of freedom prosthesis. However, one of the most famous is the sparky3 prosthesis (Figure 2.19). This prosthesis is a two actuated degree of freedom robotic ankle and is designed for people that require a more active lifestyle, thus without being limited by their robotic prosthesis. The prosthesis is not designed for every amputee: the purpose of this prosthesis is to perform tasks such as running and jumping. People who are most interested in this prosthesis are military amputees or amputees who want to recover more than just their walking ability [32].

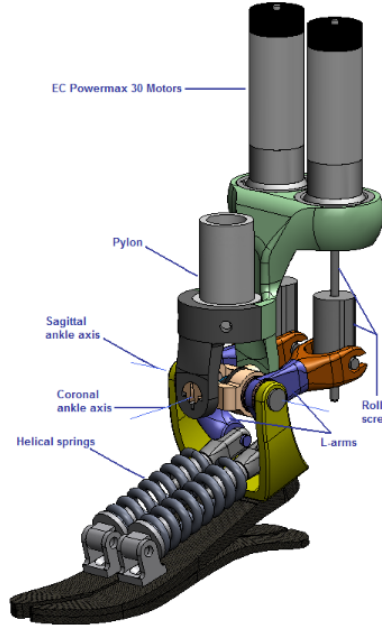


Figure 2.19: *Sparky 3 ankle prosthesis* [32]

It is also important to know that these two degrees of freedom ankle prosthesis seem to be merely an electromechanical design for now. The only thing reported by Bellman RD et al. about a controller is that this second degree of freedom and dual actuators add an element of complexity to design an appropriate controller [32]. After some research, nothing about a potential controller was found.

## 2.3 Review of the literature exploring inversion/eversion movements

This section review the literature about the inversion/eversion movements in a foot prosthesis. There are several published experiments that characterized this movement [7, 12, 33, 34].

An example of result for the analyze of the inversion/eversion movement for the stance phase is presented in Figure 2.20. The study was made on 153 healthy participants. During these tests, the angular displacement of the calcaneus, navicular, and first metatarsal relative to the tibia was measured while the subject walked along a walkway.

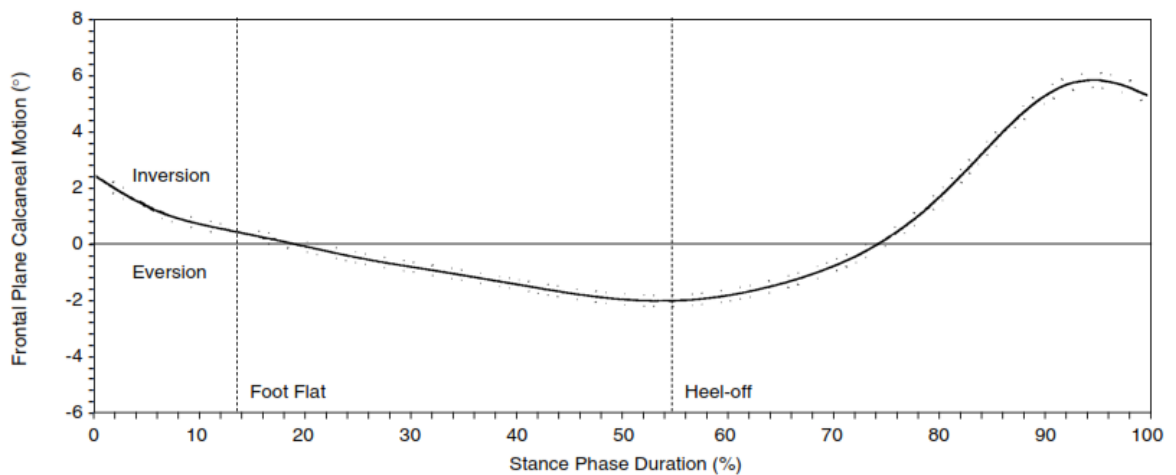


Figure 2.20: *Example of results for an inversion/eversion movement analysis* [12]

These documents ([7, 12, 33, 34]) shows that the data are always taken in laboratory conditions. In that case, one can easily imagine that the degree of freedom of plantar flexion/dorsiflexion is much more

considerable than the inversion/eversion movements on a flat ground.

From the paper of Ficanha [7], Some characteristics that could be interesting to look into was found, but they are not sufficient to design a prosthesis. Indeed, it is difficult to find the characteristic impedance which is the torque as a function of the inversion/eversion angle, the torque, the speed and the position as a function of the time in an inclined ground case. It will help to anticipate the reactions of the ankle on an inclined plane and it is difficult to determined if a passive system could be adapted without them.

In this article, the available data were the angle and the torque of the inversion/eversion movements for a straight line and a side cutting step for both the right and left ankles. These tests were made on 10 participants who repeated them 5 times each.

After contacting the author of the Ficanha's paper [7], we have access to data. For instance, we have computed the speed of the inversion/eversion movement on the basis of the angular position of the degree of freedom (see Figure 2.21). We have also access to the torque involved in the inversion/eversion movement (Figure 2.22).

We will use this data to compare them with our result in order to validate our model.

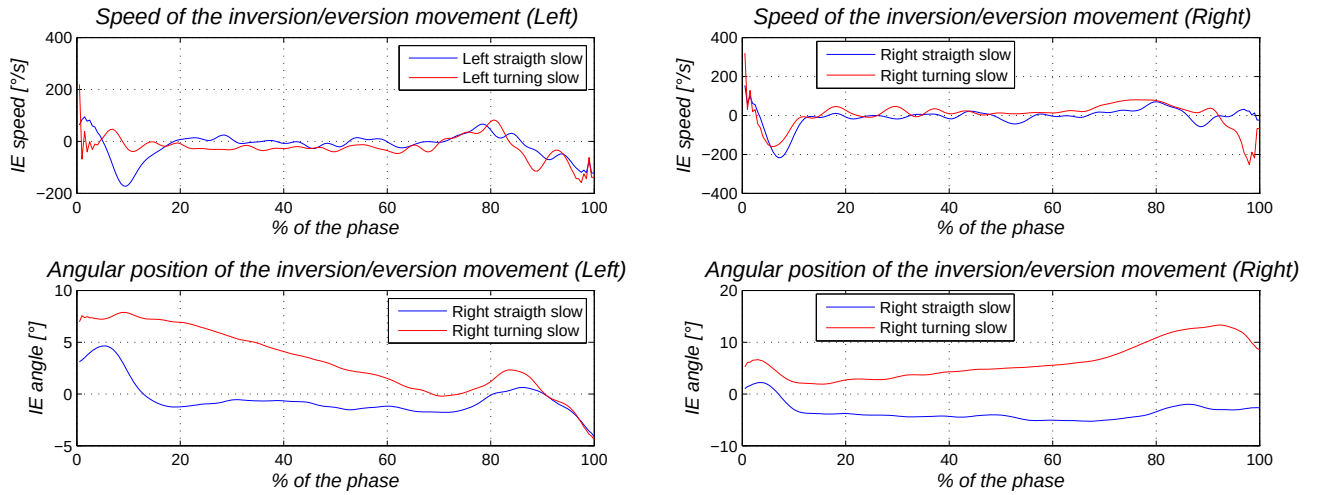


**Figure 2.21:** Angular position and speed of the left ankle (left) and the right ankle (right) [7]

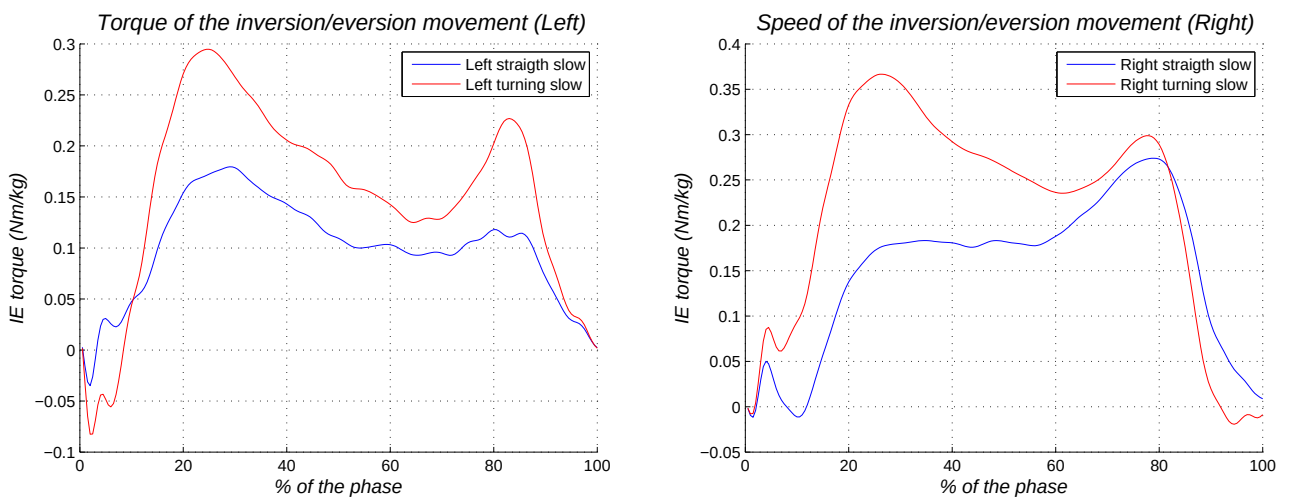


**Figure 2.22:** Torque of the left ankle (left) and right ankle (right) [7]

## 2.4 Conclusion

The field of bionic prostheses is in full expansion. There are a lot of extraordinary things that are done to improve the daily-life of amputated people. Some advanced prostheses are developed to be more efficient or to increase the comfort.

The purpose of this work will be to open doors about a topic that is quite original: characterization of the inversion/eversion movement in a healthy ankle. The aim of this work is to allow an optimal design of a two degree of freedom ankle prosthesis.

The experimental study closest to us is the Ficanha experiments [7]. We will complete these experiments because we want to characterize the frontal degree of freedom in outside conditions. This is the reason for the first part of the work : the experimental device that will help us to recreate outside conditions and select the most restrictive gaits for the inversion/eversion movement.



## CHAPTER 3

# KINEMATIC MEASUREMENTS DURING DAILY LIFE ACTIVITIES

As described in chapter 2, measurement of inversion/eversion profiles out of the lab is an innovative approach. The objectives are to evaluate scientifically the necessary degree of freedom range needed to design an ankle prosthesis.

To achieve this objective, our idea was to build a specially designed device allowing the measurement of these parameters in a daily-life context. The collected data will enable us to determine the range of motion of this movement and which kinds of gaits required the most of this degree of freedom. They will also give the possibility to recreate the external range conditions inside a lab which are needed for further experiments.

This section aims to describe the device we have made to measure this inversion/eversion angle outside the lab. The first part will concern the choice of the optimal device. It will directly be followed by the mechanical design and all the specifications of this device.

### 3.1 Measure of the roll and pitch angles of the ground

With a simple accelerometer (Figure 3.1) fixed on the shoes, we can detect each step and the entire stance phase. At this moment, we are able to compute the angles thanks to gravity and the accelerometer output. It will be combined with one of the two mechanisms that aims to measure the inversion/eversion angle presented in the next section.

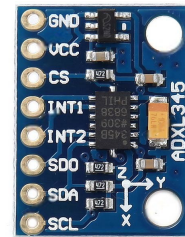


Figure 3.1: Accelerometer

### 3.2 Measure of the ankle's angle in the frontal plane

The mechanism to measure the ankle's angle in the frontal plane is presented in Figure 3.2. It is composed of three main parts: a sliding part fixed on the tibia, a 3D printed part and a pulley fixed on the latter. The pulley is attached to each side of the ankle by a belt. We will refer to this mechanism as the pulley mechanism.

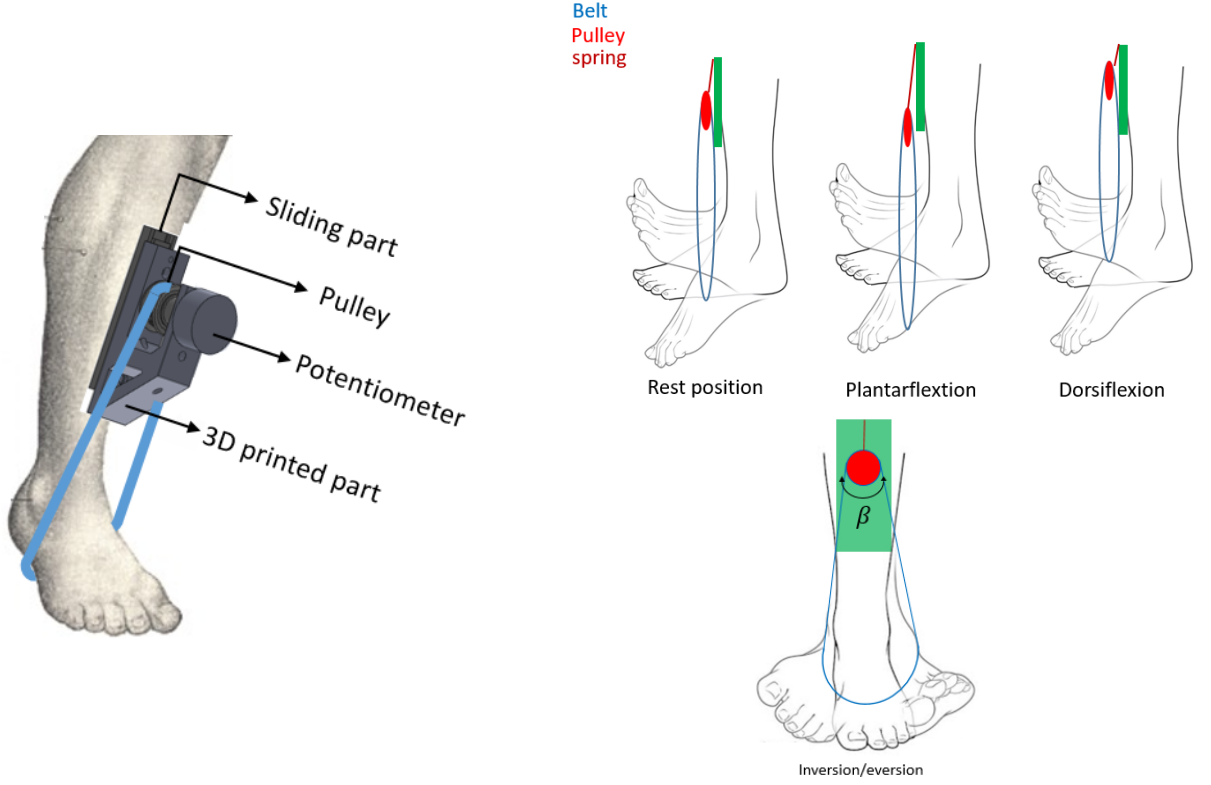


**Figure 3.2:** *Pulley mechanism: Schematic view (left) and operation schematic views (right) [35]*

When the ankle makes plantar flexion or dorsiflexion movements, the belt is stretched thanks to a spring attached to the top of the 3D printed part. When the ankle makes inversion or eversion movements, the belt rolls on the pulley and we can measure it with the potentiometer fixed on this pulley. Schematic representation of these behaviors are depicted in Figure 3.2 (right parts).

### 3.3 Design of the outside measurement device

In this section we will compute all parameters needed for an optimal design. It will help us to determine the kind of spring and sliding part that is necessary.

The only constrain for the sliding part is that the range of motion of this part should at least be as high as the vertical displacement of the belt due to the sagittal motions of the foot. To compute it, we will take the worst case by assuming a  $75^\circ$  of plantar flexion/dorsiflexion ( $\beta$  angle) range of motion [36] and that the belt is placed at 4 cm of the center of rotation (CoR). So, we have  $l_2 = 4\text{cm}$  in Figure 3.3.

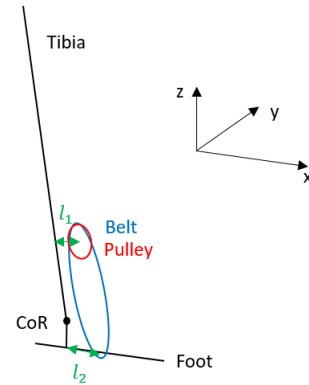


**Figure 3.3:** *Schema used to design the outside measurement device*

The vertical displacement ( $D$ ) is given by:

$$D = l_2 \sin(\beta) \quad (3.1)$$

which is equal to 3.8 cm 8cm in our case. The sliding part has to be able to move at least 3.8cm vertically. This is why we have chosen a sliding part of 4.5cm on the *Misumi* site which is the one that correspond the most to the constraint.

Now that we have the sliding part sized, we know the maximal spring stroke and the dimensioning of the spring is possible. This mechanical part will ensure to the belt to be all the time tightened. If it was not the case, the device would not work at all since it is the movements of the belt that moves the pulley and so enable us to record the inversion/eversion movements.

Therefore, the spring should be as stiff as possible to have as much as possible forces in the belt. But, on the other hand, we cannot disturb the motion of the foot. This is why the stiffness of the spring cannot be as high as we want.

To avoid disturbing the plantar flexion/dorsiflexion movements, the constraint on the spring is that it should not imply a torque on the ankle that is greater than 1% of the torque in a human ankle in a normal walk, which is about 128.6 Nm (for a 75Kg subject) [7].

The angle  $\beta$  has still a range of motion of  $75^\circ$ . The reference being the rest position, we have  $\beta = 15^\circ$  in dorsiflexion and  $\beta = -60^\circ$  in plantar flexion [36].

The torque induced by the spring ( $\tau$ ) has the shape presented in Figure 3.4. It can be computed with the following relation :

$$\tau = l_2 * \cos(\beta) * F_{spring} \quad (3.2)$$

where  $F_{spring}$  is the vertical force of the spring and is given by  $F_{spring} = D * k = k * l_2 * \sin(\beta - \gamma)$  in which  $D$  is the vertical displacement and  $\gamma = (15 * \frac{\pi}{180})$ . The  $15^\circ$  component is due to the reference of the angle. As we want to have the spring in his rest position in the maximal dorsiflexion position, we have to subtract  $15^\circ$  to  $\beta$ .

Therefore, we have the following equation for the torque induced by the spring :

$$\tau = \cos(\beta) * l_2^2 * k * \sin(\beta - \gamma) \quad (3.3)$$

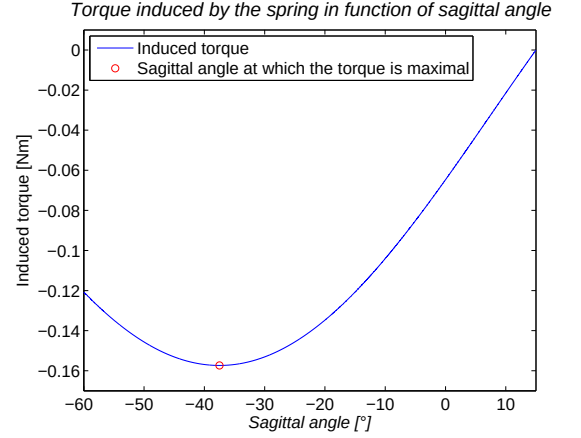


**Figure 3.4:** Shape of the torque induced by the spring with  $k = 100$

The sagittal angle at which there is the maximal torque induced by the spring can be found by computing the derivative of equation 3.3. We have :

$$\frac{d\tau}{d\beta} = 0.05^2 [-\sin(\beta) \sin(\beta - \gamma) + \cos(\beta) \cos(\beta - \gamma)] \quad (3.4)$$

$$= 0.05^2 \cos(2\beta - \gamma) \quad (3.5)$$

When we equal equation 3.5 to zero, we found that the sagittal angle  $\beta$  of the maximal torque is  $-37.5^\circ$ . From this, we can apply the constraint by equaling the equation 3.3 with 1% of the torque in a normal ankle for a walk (112 Nm). We have:

$$112 \cdot 0.01 = l_2^2 \cdot k \cdot \cos(\beta) \sin(\beta - \gamma) \quad (3.6)$$

in which  $\beta = -37.5 \frac{\pi}{180}$ .

We finally found the stiffness of the spring :

$$k = \frac{112 \cdot 0.01}{l_2^2 \cos(\beta) \sin(\beta - \gamma)} \quad (3.7)$$

In our case, this value is equal to  $1276.99 \left[ \frac{N}{m} \right]$ . The selected spring has stiffness of  $1226 \left[ \frac{N}{m} \right]$  because it is the closest one we have found.

### 3.4 Description of the outside measurement device

The mechanism is composed of the accelerometer mechanism and the pulley mechanism. We can see the two parts of this mechanism in Figure 3.5, which are the mechanism itself (left) and a box containing the electronic part (right). There is a switch on the box in order to switch between the different modes of the device (Calibration, Initialization and Measuring or waiting phases) and two leds in order to indicate in which mode is the device. These modes will be described in the section 3.4.2.

All the connections visible in the electronic part (Figure 3.5 (right)) are presented in annex B



**Figure 3.5:** *Mechanical part (left) and electronic part (right) of the optimal mechanism*

#### 3.4.1 Details of the device

##### Micro controller

For this device, we obviously need a micro controller. We can compare them (Table 3.1) to choose the best one.

| MCU              | Frequency      | RAM            | Power supply    | price |
|------------------|----------------|----------------|-----------------|-------|
| Rapberry Pi      | 0.7 to 1.2 GHz | 256 Mo to 1 Go | micro-USB 5V    | 40€   |
| Arduino Mega2560 | 16 MHz         | 8 KB           | 5V USB or 7-12V | 10€   |
| Arduino Uno      | 16 MHz         | 2 KB           | 5V USB or 7-12V | 6€    |

**Table 3.1:** *Comparison between different micro controllers*

We have chosen the Arduino Mega2560 because the Arduino Uno may have an insufficient RAM and the Arduino Mega2560 is only 4€ more expensive. Rapberry Pi has a huge frequency and RAM, but we do not need such memory and frequency, especially considering a price four times greater. Thus the best compromise is the Arduino Mega2560.

##### Accelerometer

The accelerometer is the ADXL345 because it is one of the most commonly used with Arduino, which communicates with the accelerometer through a I2C communication. The accelerometer has different

modes :  $\pm 2g$ ,  $\pm 4g$ ,  $\pm 8g$  or  $\pm 16g$ . We will use the  $\pm 2g$  mode to have a higher sensitivity as possible (3.9 mg/LSB) (see the accelerometer datasheet in annex [M](#)).

## Pulley

The pulley just has to fulfill three conditions :

1. The pulley must be notched to ensure that the belt will not slip.
2. The pulley should be as small as possible. Indeed, if the pulley has a large radius, the small displacement of the belt will not induce a big rotation of the pulley and thereby not allowing us to detect small movements.

### 3.4.2 Operation of the device

In order to use this measuring tool, we just have to know that there are four main phases:

1. Calibration
2. Initialization
3. Measuring or waiting phase
4. Errors

A block diagram of these phases is depicted in Figure [3.6](#).

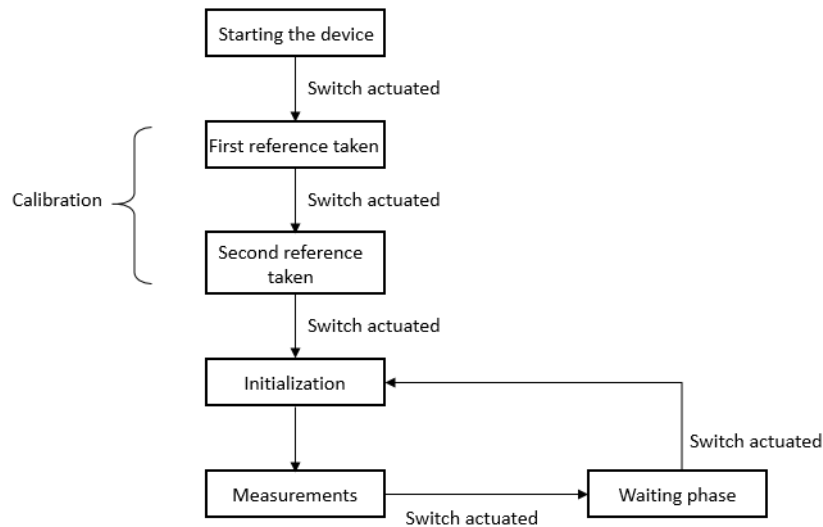


**Figure 3.6:** Block diagram of the phases sequence of the device

## Calibration

When turning the device on, it will automatically begin with the calibration. The purposes of this phase are to initialize all the variables (like the amount of data we will record before registering them on the SD card, leds, etc.) and to calibrate the device.

The potentiometer cannot rotate indefinitely. As the pulley is fixed on it, the pulley cannot turn indefinitely either. To avoid the pulley being on one of its extreme positions (fully turned on the left or fully turned on the right), the device will check if the pulley can rotate about one half turn on the left and on the right. This stage is very important because we cannot see the position of the pulley and if it is in an extreme case, all the measurements will be wrong because the pulley will not be able to rotate as it

should.

We will then calibrate the device to be able to transform a certain rotation of the pulley into a certain rotation of the ankle. This operation is quite simple and is described below :

- Both the green led and the red led are ON
- We place the ankle of the user on the calibration device in the first position (Figure 3.7)
- We activate the switch
- We place the ankle of the user on the calibration device in the second position (Figure 3.8)
- We activate the switch
- The red led should be ON while the green one should be OFF

The calibration device can be seen in Figure 3.7 and 3.8.



**Figure 3.7:** *Calibration device in the first position*



**Figure 3.8:** *Calibration device in the second position*

### Initialization

After calibrating the device, the user can place himself on the starting line of the circuit (located on a flat and straight place) and activate the switch. Data will not be recorded immediately. Indeed, the device will first record the initial ankle's angle of the user to have a zero angle. It will also record the roll and pitch angles of the ground to know the rotation of the accelerometer with respect to the ground and hence to avoid having to place the accelerometer precisely. This is why the starting line should be on a flat and straight surface.

When the initialization is done, the device will automatically begin to record data and the user can move. To know when the initialization is done, the user will have to follow the following operations:

- Only the red led is ON (just after the calibration phase)
- The user is ready on a straight and flat surface and he activates the switch: both green and red leds are ON
- The red led automatically shifts to OFF
- The user can begin his circuit

### Measuring or waiting phase

When the user has calibrated the device and the initialization is done, they can walk and record data. But if they would like to stop, they can just reactivate the same switch, the red led will turn back ON and the device will stop registering the data. It will, however, keep the calibration in memory.

If the user would like to begin another circuit, they can do it by activating the same switch. They just have to pay attention to the fact that they will begin again in the initialization phase, meaning they have to be on a flat and straight surface. It also means that if the device has moved or if the user has changed, the calibration will not be redone and the data could end up corrupted. So this operation can be used only for the same user who has kept the device on their ankle.

## Errors

There is only one reason to get an error. This happens if the user forgets to put the SD card in or if the SD card is not well placed (or moved during the test). In this case, the red led will twinkle and the data will not be recorded anymore. The user has no choice, he has to shut down the device and try again. However, the previous data will still remain on the SD card.

### 3.4.3 Technical specification

We will describe here the main characteristics of the device, i.e. the frequency, the power supply, the Repeatability, the range, the sensitivity, and the offset. Table 3.2 present the specification we want the device to reach in order to be efficient.

| <i>Specification</i>          | <i>Value</i>             | <i>Justification</i>                                                                                                                                                                                                                                                                |
|-------------------------------|--------------------------|-------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------|
| Frequency                     | $\geq 10$ Hz             | Winter use a 5-6 Hz low pass filter [37]. The walk gait could not be higher than this frequency but we want to take as much as possible measurements in a step. This is why the the frequency should be as high as possible.                                                        |
| Power supply                  | Through USB (5V)         | A computer or a small portable battery could be used for outside measurement.                                                                                                                                                                                                       |
| Repeatability (pulley)        | std $< 5^\circ$          | Standard deviation (std) of $5^\circ$ is a reasonable limit for an outside measurement device.                                                                                                                                                                                      |
| Repeatability (accelerometer) | std $< 2^\circ$          | Less than $2^\circ$ of standard deviation (std) could be reached because the accelerometer is more precise than the pulley mechanism.                                                                                                                                               |
| Hysteresis effect             | $< 5^\circ$ for 60 steps | For a bigger outside run, we will have an hysteresis effect (described below). As 60 steps is a lot for one recording and the repeatability of the pulley mechanism has a standard deviation of $5^\circ$ , the hysteresis effect should not be higher than $5^\circ$ for 60 steps. |
| Range (pulley)                | $180^\circ$              | As it is known that the inversion/eversion movement has a lower range of motion than $180^\circ$ , the pulley could measure the entire range of motion if it have this range.                                                                                                       |
| Range (accelerometer)         | $> 1g$                   | As the angles are measured on the basis of the gravity, the accelerometer must at least have a range of $1g$                                                                                                                                                                        |

**Table 3.2:** *Minimal specifications of the device to be efficient*

## Frequency

The frequency of the human walk is maximum 10 Hz. That means that we have to record at a frequency of 20 Hz minimum to respect the Shannon theorem that says that the sample frequency has to be at least two times greater than the continuous signal frequency to have all the information. As we would like to have a greater frequency than the minimum frequency, we will try to increase it as much as possible. To understand how we could do it, we have to understand how the device work.

Opening a file and register the data on the SD card take a lot of time. This is why the device will keep in RAM a certain amount of data (named  $\delta$ ) before register on the SD card.

If we register on the SD card every 100 data, we can reach a frequency of about 180 Hz, which is very high for this application. However, every 100 data, the device has to stop recording data to write on the SD card, which takes more than 0.3 seconds. As the frequency is about 180 HZ, this means that the device will stop during 0.3 seconds every 0.6 seconds. This is really not good.

To avoid this phenomenon, we can first reduce the frequency to avoid writing too much on the SD card. We can also try to reduce the time writing on the SD card by reducing the amount of data kept in RAM.

After a lot of experimental tests, we have found the best amount of data to keep in mind and the best delay included on the device to have the best performances. We are able to reach 30 Hz, which is really sufficient for this application, and only 0.15 seconds writing on the SD card when keeping in mind 50 data, which means that every 1.67 seconds, the device will stop 0.15 seconds to write on the SD card.

## Power supply

We can supply the device by using a USB cable connected to a 5 Volts portable battery or to a computer (which could be placed on a backpack to wear it easily). We can also use a jack cable connected to a battery with a voltage between 7 and 12 Volts.

## Repeatability

The repeatability is the reproduction of the same experience with the same method, the same person and the same equipment. Two special tests were made to measure the repeatability of the device. The purpose of the first one is to highlight the repeatability of the ankle's angle (inversion/eversion) repeatability and the second one is to highlight the ground angles (roll and pitch) measurement repeatability.

For the first test, the device was calibrated and then the subject has reached six times the same eversion angle. The standard deviation of the test is  $1.52^\circ$  with a mean of  $15.65^\circ$  (Figure 3.9). This is a higher repeatability than expected.

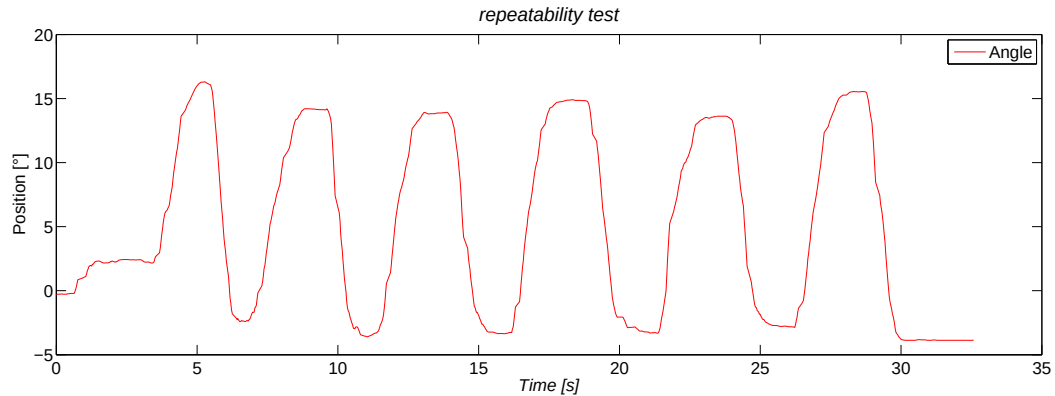


**Figure 3.9:** *Inversion (negative)/eversion (positive) ( $^\circ$ ) measured by the device after the ankle reach 6 times the same eversion position in order to highlight the repeatability of the pulley system*

The second test is of the same kind as the first one. The device is calibrated and the subject will walk on an inclined plane of  $30^\circ$  or on a flat ground every two steps. The results of this experiment are depicted in Figure 3.10. The mean of the of the inclined plane steps is  $27.5^\circ$  with  $2.19^\circ$  as standard deviation. The mean of the even ground steps is  $0.026^\circ$  and the standard deviation is  $0.67^\circ$ . As it is measured when walking outside, this is enough precision for the device.

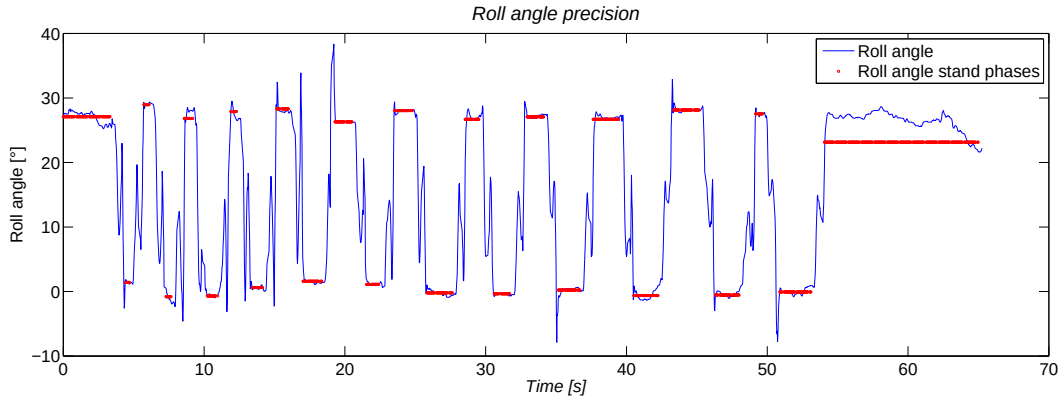


**Figure 3.10:** *Roll angle(°) measured by the device when ankle is walking on a flat and on a 30° plane alternatively every two steps in order to highlight the repeatability of the accelerometer system*

We have a small kind of hysteresis effect for the pulley system which is due to the movements of the belt onto the foot. Figure 3.11 represents a test composed of 60 steps, there is a small difference between the first measurement of the angle and the last one while the first and the last position are exactly the same. It is interesting to note in this figure the difference between the beginning and the end. We have lost 3.5° which is quite correct because 60 steps is very big for a test. We can accept this effect because the precision is 3° and we will not perform tests with more than 60 steps.

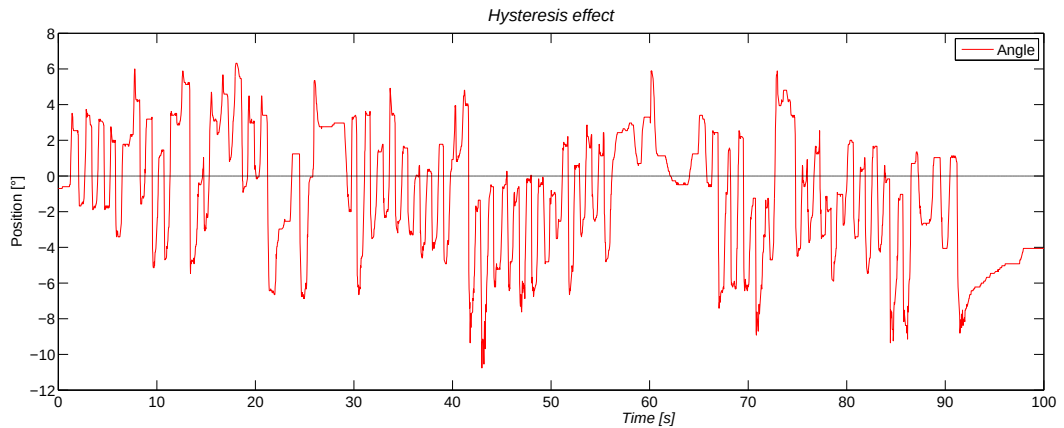


**Figure 3.11:** *Hysteresis effect observed when measuring position (°) with the device when performing a 60 steps walk*

### Range

The range of the pulley is 245 °.

The range of the accelerometer depends on the mode in which it is used. We have the following modes and ranges associated with the mode :

- Mode 16g : range =  $\pm 16g = 32g$
- Mode 8g : range =  $\pm 8g = 16g$
- Mode 4g : range =  $\pm 4g = 8g$
- Mode 2g : range =  $\pm 2g = 4g$

In this application, we only use the 2g mode to have the higher sensitivity as possible (see accelerometer datasheet in annex M).

## Sensitivity

The sensitivity of the accelerometer depends on the mode in which it is used. The supply voltage being 3.3 Volts, we have the following results table (Table 3.3):

| Mode | Bits | Sensitivity [bits/g] | Sensitivity [Volts/g] |
|------|------|----------------------|-----------------------|
| 2g   | 10   | 256                  | 0.825                 |
| 4g   | 11   | 256                  | 0.4125                |
| 8g   | 12   | 256                  | 0.206                 |
| 16g  | 13   | 256                  | 0.103                 |

**Table 3.3:** *Sensitivity of the accelerometer (annex M)*

## Offset

For the accelerometer, we have two kinds of offsets : the first offset is the usual offset in any accelerometer. It comes from the elements that are free to move and these elements are extremely sensitive to mechanical stresses. This leads us to a very small offset.

The second kind of offset in our situation comes from the position of the accelerometer. We cannot place it exactly horizontal or vertical. This means that we will not have the same measure between the different people who use the device.

To avoid these offsets, we will just calibrate it. Once the device is placed on the subject, we take a reference measure and we subtract it to all measure (as we explain in section 3.4.2). It allows us to avoid the offsets and the misplacement of the accelerometer.

## Resolution

The resolution of the accelerometer is 3.9 mg/bit. This means that the lowest acceleration the accelerometer is able to detect is 3.9 mg. With this resolution, we are able to measure inclination changes fewer than 1° (annex M). This is sufficient for this application. The resolution of the angle measurement is about 0.239°/LSB.

## 3.5 Data processing

Once data are collected, they have to be treated in order to extract the result presented below. We will describe how this treatment is done to extract the results presented below.

### 3.5.1 Filtering

Data should first be filtered. Smoothing allows to highlight important patterns while leaving out things that are less important like the noise. We have chosen the moving average filter because it is easier to implement and we do not need something more complicated. The principle of a moving average filter is to replace each sample by the average of the N consecutive samples. To avoid a time shift, the samples are taken from an equal amount of data on either side of a central value. We can see the pseudo-code in Algorithm 1 and the code in the annex E. The result of such a filter with a size of 9 (chosen randomly for this figure) is depicted in Figure 3.12.

The real size used for our data has been chosen with the method described in the annex F and is equal to 3.

---

**Algorithm 1** Moving average algorithm

---

```
1: function FILTER( $X, N$ )
2:    $L \leftarrow \text{length}(X)$ 
3:    $V \leftarrow \text{zeros}(1, L)$ 
4:    $N_2 \leftarrow \frac{N}{2}$ 
5:   for  $i \leftarrow 1 : L$  do
6:     if  $i < L - N_2$  &  $i > N_2$  then
7:        $V(i) \leftarrow \frac{\text{ComputeSum}(X, i - N_2, i + N_2)}{N + 1}$ 
8:     else if  $i < N_2$  then
9:        $V(i) \leftarrow \frac{\text{ComputeSum}(X, i, i + 2N_2)}{N + 1}$ 
10:    else if  $i > L - N_2$  then
11:       $V(i) \leftarrow \frac{\text{ComputeSum}(X, i - 2N_2, i)}{N + 1}$ 
12:   $V(\text{end}) = V(\text{end} - 1)$ 
13:  return  $V$ 
14: function COMPUTESUM( $A, x_1, x_2$ )
15:    $\text{sum} \leftarrow 0$ 
16:   if  $x_2 \geq x_1$  then
17:     for  $j \leftarrow x_1 : x_2$  do  $\text{sum} \leftarrow \text{sum} + A(j)$ 
18:   return  $\text{sum}$ 
```

---

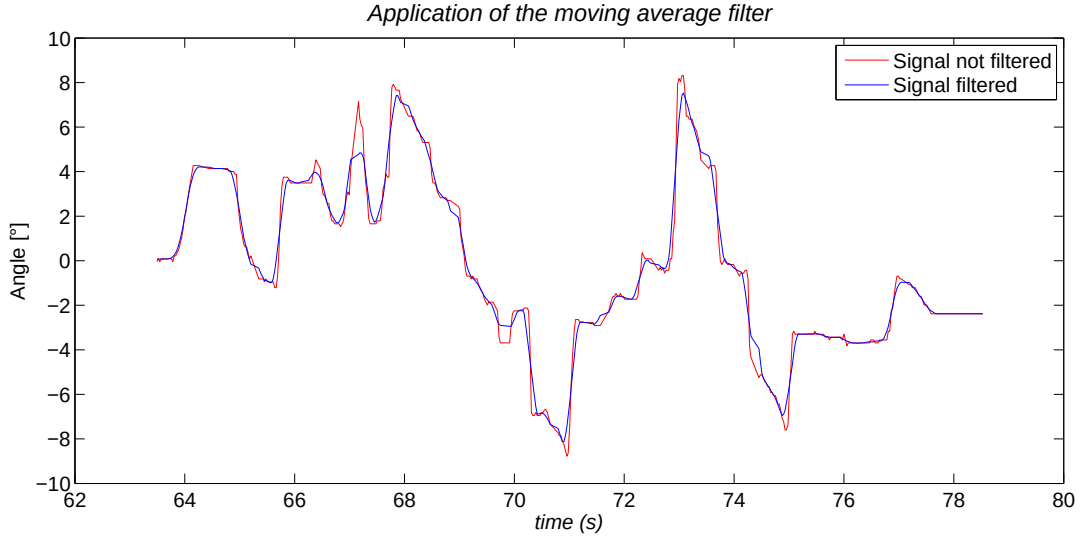


**Figure 3.12:** Comparison of a not filtered signal and the same signal filtered using the moving average filter with a size of 9

### 3.5.2 Detection of the steps

As the device should have the highest possible frequency, we could not detect the step directly in real time (which was done previously but the frequency was too low). So we had to record only the acceleration and detect each time there is a step in post processing. *Matlab* function to perform it is presented in annex C and the pseudo-code for better understanding in Algorithm 2.

The operating mode of this function is composed of 3 steps :

1. Initialize the vectors at zero with the appropriate length
2. Check if the actual sample and the 7 other consecutive samples of the acceleration norm respect the following condition :

$$\text{norm} < 1 + \epsilon \quad || \quad \text{norm} > 1 - \epsilon$$

where 1 is the gravity component (1g). The condition means that the foot is moving. If the norm is within  $[1 - \epsilon, 1 + \epsilon]$ , the condition is respected.

3. If they respect this condition, we say that this sample is on a stand phase sample and we can compute the angles of the ground. If it does not respect the condition, we say that this sample is not on a stand phase and we move to the next sample.

---

**Algorithm 2** Step detection algorithm

---

```

1: function DETECTSTEP( $a_x, a_y, a_z, \epsilon$ )
2:    $norme \leftarrow \sqrt{a_x^2 + a_y^2 + a_z^2}$ 
3:    $L \leftarrow \text{length}(a_x)$ 
4:    $nbrConsecutive \leftarrow 8$  ▷ Experimentally find
5:   for  $j \leftarrow 1 : L$  do
6:     if  $j < L - nbrConsecutive$  then
7:        $echec \leftarrow 0$ 
8:       for  $k \leftarrow 0 : nbrConsecutive$  do
9:         if  $norme(j+k) > 1 + \epsilon \quad || \quad norme(j+k) < 1 - \epsilon$  then
10:           $echec \leftarrow 1$ 
11:          break
12:       if  $echec == 0$  then
13:          $Roll \leftarrow (\text{atan2}(a_z(j), \sqrt{a_x(j)^2 + a_y(j)^2})) * 180.0 / \pi$ 
14:          $Pitch \leftarrow (\text{atan2}(a_x(j), \sqrt{a_y(j)^2 + a_z(j)^2})) * 180.0 / \pi$ 

```

---

Figure 3.13 presents the result of this method. The red curve is the norm of the acceleration and in blue all the samples that are detected as a stand phase. Eleven steps were performed during the test and we can see twelve times a detection of a stand phase. This is simply due to the rest position at the beginning.

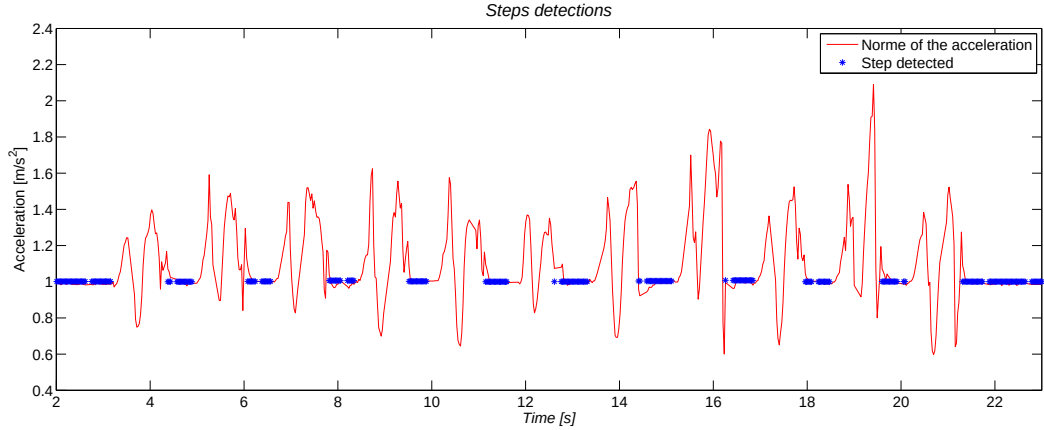


**Figure 3.13:** Detection of the stance phases using the Step detection algorithm (Algorithm 2)

### 3.6 Compute the ground angles

Now that the data are filtered and all the stance phases are detected it is possible to compute the ground's angles associated with each sample that is considered as a stance phase. In fact, as we compute these angles thanks to accelerations in different directions and that the accelerometer is on foot, we cannot measure these angles when the foot is moving.

The purpose of this section is to explain how we have computed the ground inclination using an accelerometer. We will consider just the roll angle (around the x axis) while the accelerometer is placed on the ground. There are three methods to compute the angles [38]. The first one uses only one axis of the

accelerometer, the second uses two axes and is more precise and the third one uses three axes and is the one used in this work because it is the most reliable.

### 3.6.1 One axis computation

To compute the tilt angle using one axis is the simplest way to perform it. As we can see on Figure 3.14, we just need to take the inverse sine function as following :

$$\Theta = \sin^{-1}(x)$$

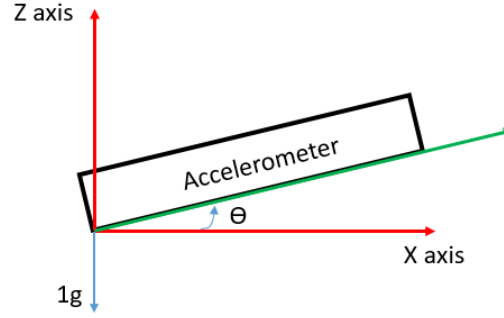


**Figure 3.14:** *Computation of an angle using one axis of the accelerometer*

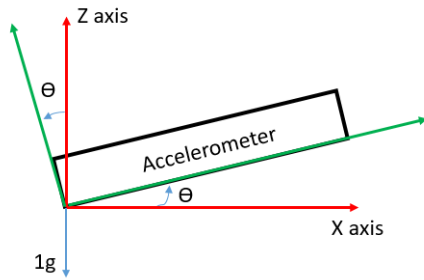
However, we cannot compute angles bigger than  $45^\circ$  without losses in sensitivity due to the nature of the sine function.

### 3.6.2 Two axis computation

The second method is more reliable and allows us to measure angles from  $-90^\circ$  to  $90^\circ$ . Based on Figure 3.15, we obtain the equation 3.9 :

Again, this would be less precise if the y axis of the accelerometer is not horizontal. If it is slightly turned in the y direction, some components of the vector from z axis will be lost in the y axis.

This is why we will use the third method.



$$\tan(\Theta) = \frac{x}{z} \quad (3.8)$$

$$\Theta = \text{atan}\left(\frac{x}{z}\right) \quad (3.9)$$

**Figure 3.15:** *Computation of an angle using two axes of the accelerometer*

### 3.6.3 Three axis computation

This method is the most accurate between these three methods. The equations are nearly the same as for the method with two axes but instead of dividing by one axis, we divide by the norm of the two others :

$$\Theta = \text{atan}\left(\frac{x}{\sqrt{(z^2 + y^2)}}\right) \quad (3.10)$$

With the equation 3.10, we can have the equation for the pitch and roll angles in the case of our measurement device :

$$\begin{aligned} Roll &= \text{atan}\left(\frac{y}{\sqrt{x^2 + z^2}}\right) \\ Pitch &= \text{atan}\left(\frac{x}{\sqrt{y^2 + z^2}}\right) \end{aligned}$$

### 3.7 experimental protocol of the preliminary tests

One of the applications of the outside device measurement is to perform some preliminary tests. They are two purposes of these tests.

The first one is to determine which tests will be interesting to investigate. Indeed, we already know that they are some movements that we perform a lot in the daily life. But we do not know if these movements use the inversion/eversion movement.

Obviously, we could have an idea and make the hypothesis that these movements use this degree of freedom but this is not a real scientist procedure.

The second reason is that we have to be prepared for the lab. If we come to the lab without having an experimental protocol or without knowing exactly what we will do, we could waste a lot of time. Or worse, we could miss some important thing.

From the paper [8], we know that we have to test the lateral step. But we would like to recreate external conditions, which are inclined planes and not stable ground. Also, we had thought that we could also test the turn step, which consists to take a sudden turn and the sit-to-stand motion, which is used every time we want to pick up something or each time we want to sit.

They were 6 subjects that perform these tests. The age was between 20 and 24 years old (both men and women).

**Turn step** The first test that the subject has performed is the turn step. To perform it, the participant has joined his feet and stands straight. Once the switch is activated (for the calibration of the device) and when the red led is off, the subject begins a normal step forward (a right step, with the device). Then, when he initiates the swing phase of his left foot, he changes suddenly of direction in the way that his new direction is  $45^\circ$  on the right of the initial direction. Figure 3.16 depicted this test.

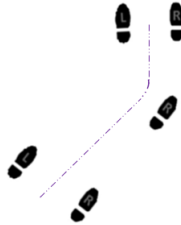


Figure 3.16: Layout of the turn test

**Lateral step** The second step consists in left lateral step with the device always on the right ankle. Once again, the subject will calibrate the device in his rest position. When the red led is off, he will perform a right step and then a lateral step. Figure 3.17 (left) represent the schematic of this move.

**Inclined plane step** The subject has to calibrate the device again by activating the switch in his rest position. Then, the subject will walk two steps, beginning with his right foot in order to put it on the inclined plane (of approximately  $15^\circ$ ) just in front of him. Figure 3.17 (right) schematizes the test.



**Figure 3.17:** *Schema of the lateral step (left) and the inclined plane (right) tests [7]*

**Unstable ground** The unstable ground is the same test as the inclined plane test except that it is not an inclined plane that is placed in front of the subject but an unstable ground. This one is simulated thanks to a fitness mat. We can see it in Figure 3.18 (left).

**Sit-to-stand motion** The sit-to-stand motion is the final preliminary test. The subject will just have to perform a sit-to-stand motion, which is like a squat. To help them, we place a chair and they just have to sit and get up once the device is calibrated. This test is represented in Figure 3.18 (right).

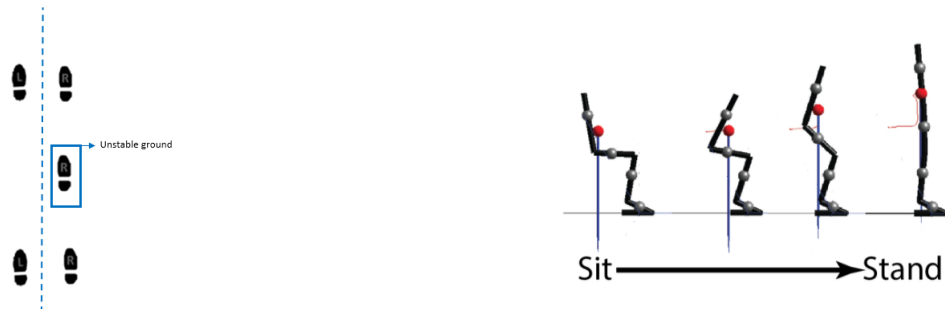


**Figure 3.18:** *Schema of the unstable ground (left) and the sit-to-stand motion (right) [39]*



## CHAPTER 4

### RESULTS OF THE MEASUREMENTS IN DAILY LIFE CONTEXT

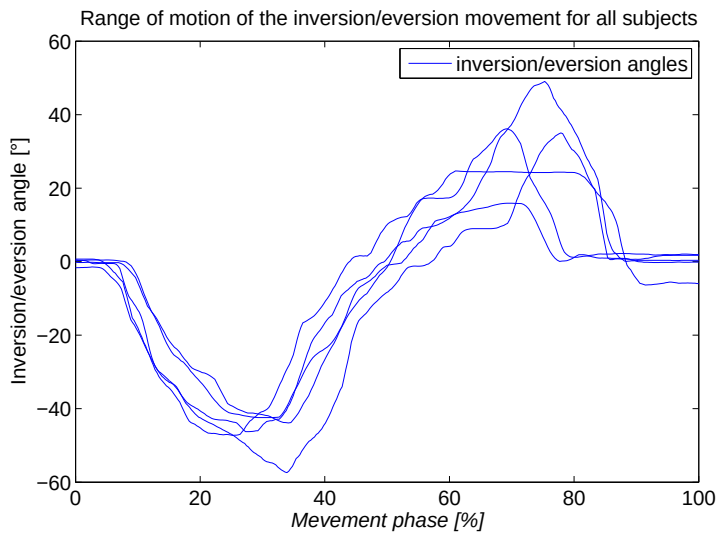
Through these outside experiments, we would like to establish some bases for the in-lab measurement and for the mechanical design of the ankle prosthesis.

First of all, we are now able to determine the range of motion of this degree of freedom. Indeed, this is quite important for the mechanical design. Then, we will measure the typical angle of the ground. If we want to recreate an external context, we have to know this context. Finally, we will perform some preliminary tests in a way to decide which tests should be done in the in-lab experiments.

For these preliminary tests, we have not used the time as abscissa but the movement phase. This means that when we are at 0% of the movement phase, we are at the very beginning of the test and when we are at 100% of the movement phase we are at the very end. This is why the reader should read the section 3.7 (at page 34) to understand these graphs correctly.

#### 4.1 Range of motion

The range of motion was evaluated with the help of six subjects. It was asked to do the maximum eversion angle they can do and then the maximum inversion angle. This test is a kind of validation of our device since the values can be compared to the literature.



As observed in Figure 4.1, the range of motion for the six subjects are very similar.

For the eversion movement (negative in Figure 4.1), the range of motion reaches almost 50° for one subject but the mean of the maximal range of motion for the eversion movement is 46.8°.

The inversion movement (positive in Figure 4.1) is much lower, we only reach a mean of 30.89°.

**Figure 4.1:** Range of motion of the inversion/eversion movement for the six subjects

These tests are conclusive because the range of motion found is  $77.69^\circ$  whereas the value found in literature is  $75^\circ$  [36].

## 4.2 Inclination of the ground

In this section, we will show the results of the inclination of the outside ground in order to recreate it in the lab. To realize these measurements, we have walked on the street without paying attention to the ground. However, we avoided the biggest obstacles. Indeed, the purpose of the ankle prosthesis will be that the user is able to walk on a normal street without loss in stability but not to walk on a steep mountain road.

As we explained in section 3.6, we have computed the ground angles thanks to the accelerometer. This one has registered the acceleration at almost 30 Hz all the time but we are interested in the result only when the foot is entirely flat on the ground.

Figure 4.2 presents these results. Each measure of the ground angle (so each step) is a single blue point in the graph.

We notice that it is not rare to have a step on a ground that has  $15^\circ$  of inclination. However, the  $30^\circ$  ground is reached but more rarely. It is such an extreme case.

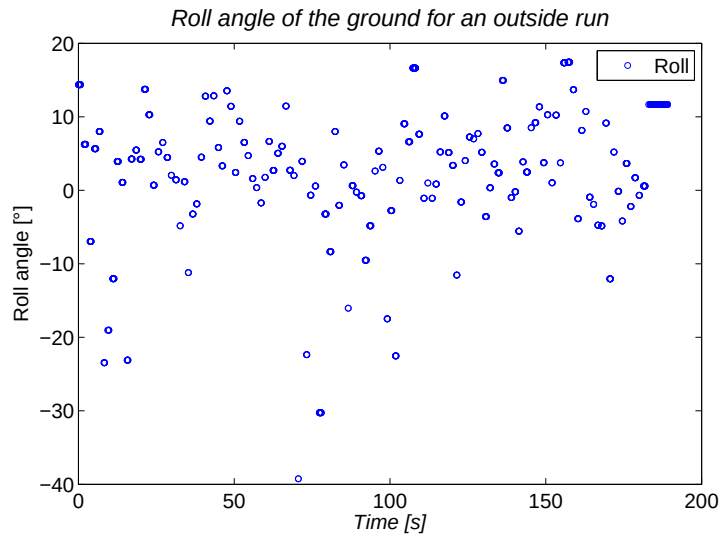


**Figure 4.2:** Angles of the ground measured on a normal road

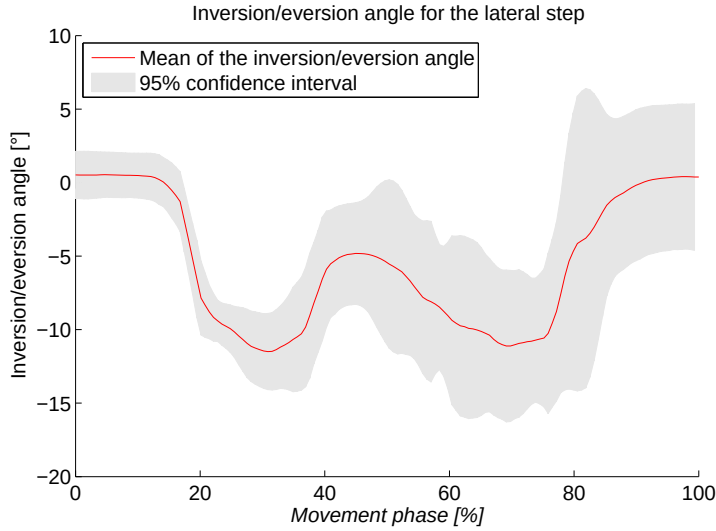
## 4.3 Preliminary tests results

As we have seen in the section, 3.7, some preliminary tests were performed to select the gait that uses the most of the inversion/eversion movement in the lab experiments.

As a reminder, we have performed the following tests : a lateral step, a turn step, an inclined ground step, a not stable ground step and the squat gait. We will determinate which test we should perform in lab with an optoelectronic system. In all of these graphs, the convention is that the negative angles are the eversion angles and the positive ones are the inversion angles.

### 4.3.1 Implication of the inversion/eversion movement in the lateral step

Figure 4.3 depicts the inversion/eversion angle along the lateral step gait. At the very beginning it is logical to find  $0^\circ$  as eversion angle because the subject is at rest as well as at the very end of the phase.



Between these two moments, there are two pikes of eversion angle up to  $6^\circ$ . This represents an eighth of the range of motion. However, as this range of motion is the extreme value, an eighth is already enough to say that the inversion/eversion movement in a lateral step would be interesting to analyze with the optoelectronic system.

**Figure 4.3:** *Inversion/eversion angle along the lateral step gait*

### 4.3.2 Implication of the inversion/eversion movement in turn step

The turn step seems to use less inversion/eversion degree of freedom (Figure 4.4). In fact, the mean of this angle reach only  $5^\circ$  of eversion angle.

However, we can notice that the 99.7% confidence interval is much larger than for the lateral step (Figure 4.3). This could mean that this movement is used in a different way between each subject.

We can notice something else by comparing the eversion angle of the turn step (Figure 4.4) with the eversion angle of the lateral step (Figure 4.3). Even if the turn step has a lower eversion angle magnitude, they both have the same kind of shape.

It means that the turn step is very similar to the lateral step for the inversion/eversion movement.

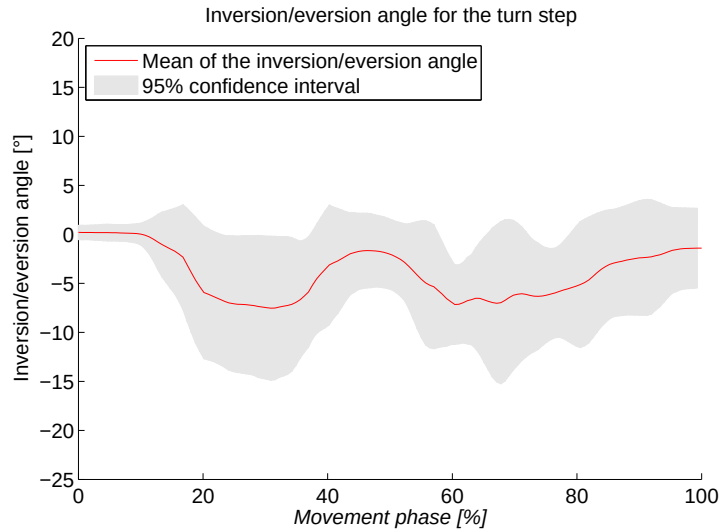


**Figure 4.4:** *Inversion/eversion angle along the turn step*

### 4.3.3 Implication of the inversion/eversion movement in the $15^\circ$ inclined plane test

The inversion/eversion angle of the  $15^\circ$  inclined plane (Figure 4.5) has a huge magnitude compared to the other tests. It reaches  $15^\circ$ , which is the inclination of the plane. Therefore, we can deduce that the foot is flat on the inclined plane and the tibia is vertical. It means that the preliminary test seems to confirm what we expected, which is that the inversion/eversion degree of freedom is mostly used for the inclined ground. Hence, the inclination will be our main concern.

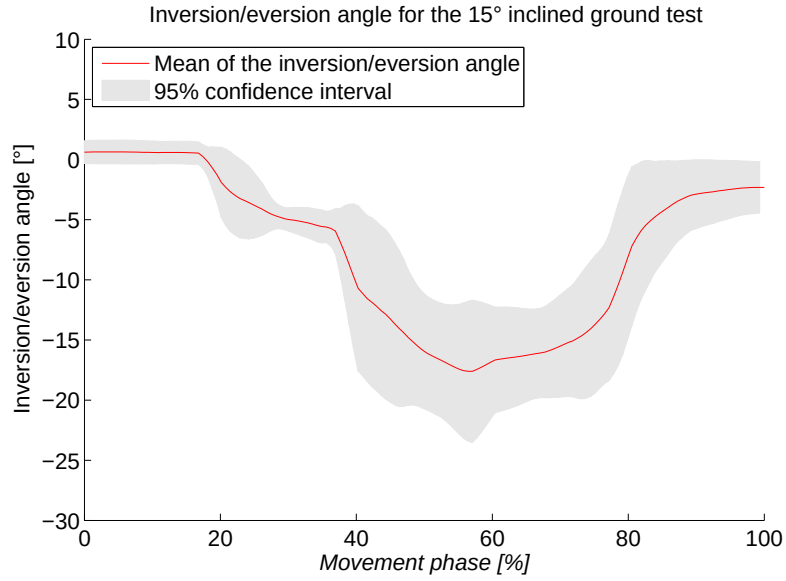


**Figure 4.5:** *Inversion/eversion angle along the 15° inclined plane step*

#### 4.3.4 Implication of the inversion/eversion movement in not stable ground

The purpose of the unstable ground test is to analyze how the ankle reacts when the subject has to keep his stability when walking on a ground that is not really stable.

Figure 4.6 shows the magnitude of the mean angle and it does not overpass the magnitude of the turn step. Furthermore, the confidence interval is lower than the one of the turn step.

We can explain the fact that the magnitude is not so high by the fact that the subject try to keep his stability. It means that he will just apply a torque on his ankle in a way not to have a too big inversion/eversion angle.

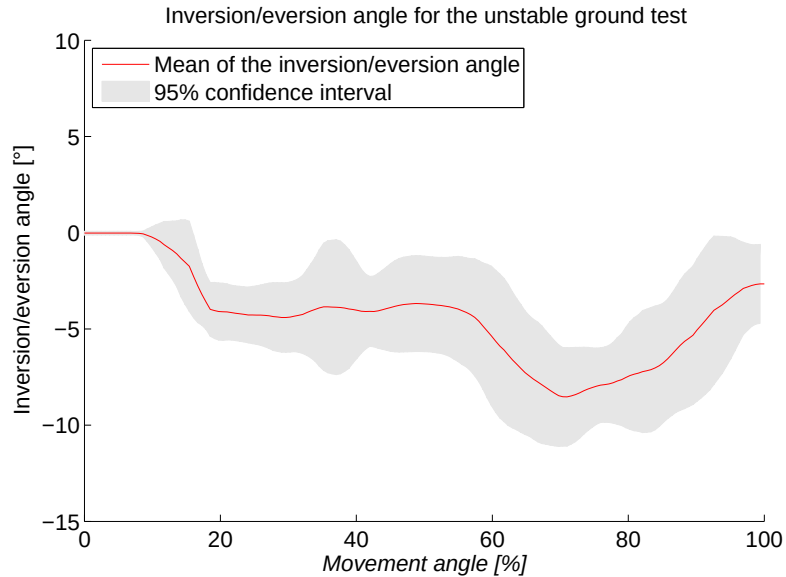
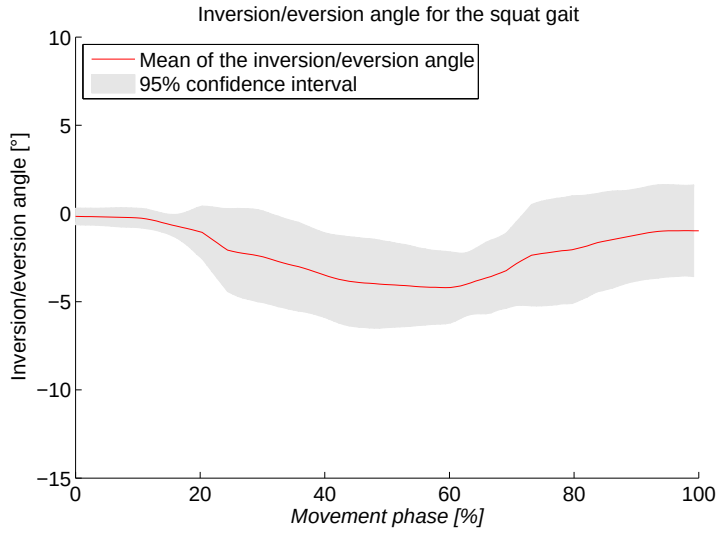


**Figure 4.6:** *Inversion/eversion angle along a step on an unstable ground*

### 4.3.5 Implication of the inversion/eversion movement in the squat gait



As expected, the squat gait uses almost not at all the inversion/eversion movement. This result confirms that it is unnecessary to perform this test in the lab with the optoelectronic system.

Figure 4.7: Inversion/eversion angle in a squat gait

## 4.4 Conclusion

After having analyzed the range of motion, the outside ground inclination and the preliminary tests, we are ready to prepare the in-lab measurements correctly in a daily life context.

We know that we have to test the 15° inclined plane because we noticed in section 4.2 that the ground on a typical road has a 15° inclination easily. Furthermore, section 4.3.3 shows us that the inversion/eversion movement is strongly depending on the inclination of the plane. It will be very interesting to analyze more precisely the kinematics and more importantly the dynamic of the ankle on an inclined plane. We will also test a more extreme case, which is the 30° inclined plane.

We have seen in section 4.3.5 that it is unnecessary to analyze the squatting motion in this work. The unstable ground case will also be dropped for two main reasons. The first one is that the inversion/eversion movement seems to be not really used in this kind of gait. It would be interesting to verify it in a dynamic point of view but there is a second reason. From the fact that we will use a force platform to find the torque in the ankle (as we will see in section 5), we are not able to put an unstable ground on it. Furthermore, finding the contact point correctly on the force platform will not be feasible, which is really important for the dynamic computation.

From these results, we might give up the turn gait step because it has the same shape as the lateral step but it has a lower magnitude. However, as this gait is one of the most used in the daily life [8], we will analyze it anyway.



## CHAPTER 5

# ANKLE DYNAMICS IN-LAB EXPERIMENTATION; EXPERIMENTAL PROTOCOL & DATA PROCESSING

For now, we have designed and programmed an outside measurement device. We also have post-treated the data and analyzed them. In this chapter, we will investigate the in-lab experiments with the results we obtained from the outside measurement device. Indeed, we are ready to recreate the daily life context in the lab and we know which tests we will execute.

This chapter will introduce the experimental protocol that we followed all along these in-lab experiments. Then, we will explain how we have post-treated the data on the way to get the results.

## 5.1 Experimental protocol

### 5.1.1 Required equipment

To fulfill this thesis, we will have to measure the position and the torque generated by the ankle in the frontal plane. In order to measure it, we will need the following equipment :

1. An optoelectronic analysis system (composed of cameras and markers)[40]. This system will be necessary to record the position of the ankle at any time. As it is human motion that is recorded, the measurement frequency should be at least 100 Hz. The operation of an optoelectronic system is based on infrared markers that are spotted by the cameras. It enable us to have the absolute position of each markers.
2. Two force platforms which will enable us to compute the reaction forces on the ground and then obtain the torque in the ankle via an appropriate multibody model of the walker body.
3. Some cones to make the obstacle race.
4. An inclined plane to measure the foot adaptation to a slope.

### 5.1.2 Methods

Some specific obstacle track (see Figures 5.1, 5.2 and 5.3) will be performed by the subjects. They represent scenarios encountered in daily life and will answer our objectives (added-value and passive or active mechanism). It will be composed of different parts : turn, sidestep and inclined ground. The subject will first perform the obstacle race normally. As the position of the ankle will be recorded with the optoelectronic system, the subject will have to put some markers on his ankle and other body segments to be able to record the movements [40]. Four cameras will be placed in the way so that we can take the measurements properly.

The group of subjects will be composed of men and women who are healthy, at least for their lower limbs. The age of subjects should be between 18 and 50 years. If the subject is younger, he may not have

reached its adult size, which could imply some differences in the gait and if the subject is older, he may also have some differences in his gait compared to a younger subject [33]. There were seven men and one woman involved in this study and they were all between 20 and 30 years. Each subject performs the three walking tests under two different conditions and they have repeated them three times. The three tests are the following:

- **Turn** The subject will walk in a straight line to a cone. He will then bypass it by changing his lead of  $45^\circ$  to the left, without stopping to walk during the transition (figure 5.1). The cone will be placed in such a way that the first two steps in the turning phase take place on the force platforms.
- **Sidestep** The subject walks straight ahead, steps aside, then walks straight ahead again. The side step shall be made so that the right foot of the subject is on the first force platform for the pulse and the left foot of the subject is placed on the second force platform for reception (Figure 5.2). Cones will be placed to allow the subject to visualize the location where the side step is to take place.
- **Inclined plane** An inclined plane of  $15$  or  $30^\circ$  in the frontal plane will be placed on a force platform. The subject will walk along a straight line so that his left foot rests on the first force platform (without inclined plane) and his right foot rests on the second, containing the inclined plane. The subject will be asked to continue walking in a straight line (Figure 5.3).

The two conditions are the following :

- **Normal condition** The subject will perform the tests without any constraints.
- **Constraint condition** The subject will perform the tests with the degree of freedom allowing movements in the frontal plane locked, thanks to a special orthosis (figure 5.4). The aim of this condition is to evaluate the kinematics and dynamic consequences of a locked ankle in the frontal plane, which corresponds to the configuration of a majority of current bionic prostheses.

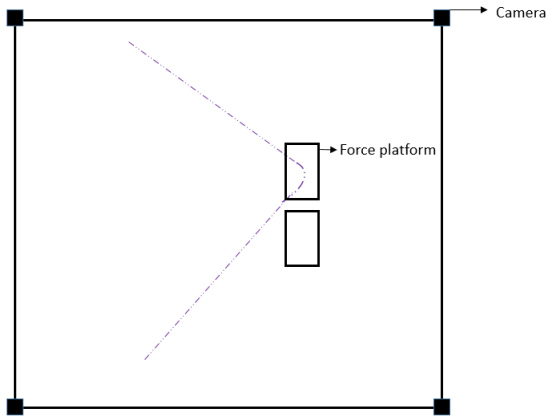


**Figure 5.1:** *Layout of the turn test*

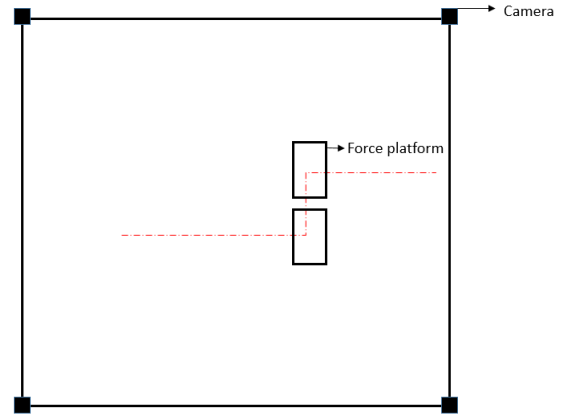


**Figure 5.2:** *Layout of the sidestep test*

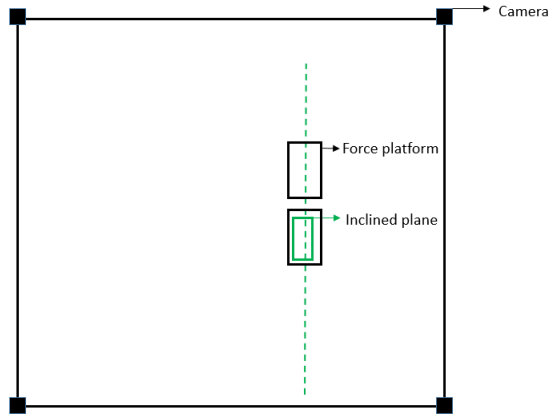


**Figure 5.3:** *Layout of the inclined plane test*

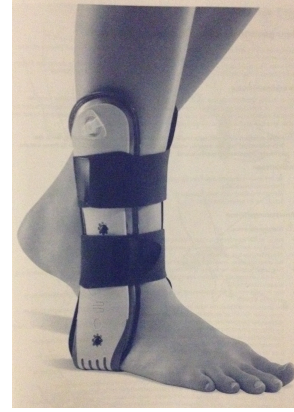


**Figure 5.4:** *Orthosis*

In all trials, the participants will be free to adopt the most comfortable walking speed and the order of the trials is randomly chosen.

As it is difficult to move the force platform, we must install them in order to never have to move them for each test. Figures 5.1, 5.2 and 5.3 present the way we do that. Subjects will also perform a normal walk test, in which they just have to walk along a straight line at the most comfortable speed. The purpose is to be able to compare the real tests to a normal case.

For all of these tests, 28 markers will be posed on the body of the subject in order to measure the position of the different parts of the body over time<sup>1</sup>. That data will make possible to calculate the articular forces, and in particular of the ankle, both in the sagittal and frontal plane.

The markers will be positioned on anatomical landmarks traditionally used in the walk analysis and depicted in Figure 5.3 and Table 5.2.

An estimation of the total time needed for a subject to perform all tests is almost two hours. The details of this estimation are presented in Table 5.1.

| Action               | Required time [minutes] |
|----------------------|-------------------------|
| Installation         | 30                      |
| Each test            | 2                       |
| All tests            | 50                      |
| Install the orthosis | 15                      |
| Uninstall            | 10                      |
| Possible delay       | 15                      |
| Total                | 120                     |

**Table 5.1:** *Details of the time estimation*

---

<sup>1</sup>Video will be provide to the reader in order to visualize these tests if needed.

| All markers   |                         |
|---------------|-------------------------|
| Body segments | details                 |
| Foot          | Heel                    |
|               | Metatarsal 1            |
|               | Metatarsal 5            |
|               | Toe                     |
| Tibia         | External Malleolus      |
|               | Internal Malleolus      |
|               | Lateral                 |
|               | Tuberosity of the tibia |
| Femur         | External condyle        |
|               | Internal condyle        |
|               | Trochanter              |
| Pelvis        | Left ASIS               |
|               | Right ASIS              |
|               | Left PSIS               |
|               | Right PSIS              |
| Spine         | L5                      |
|               | C7                      |

**Table 5.2:** Positioning of the markers

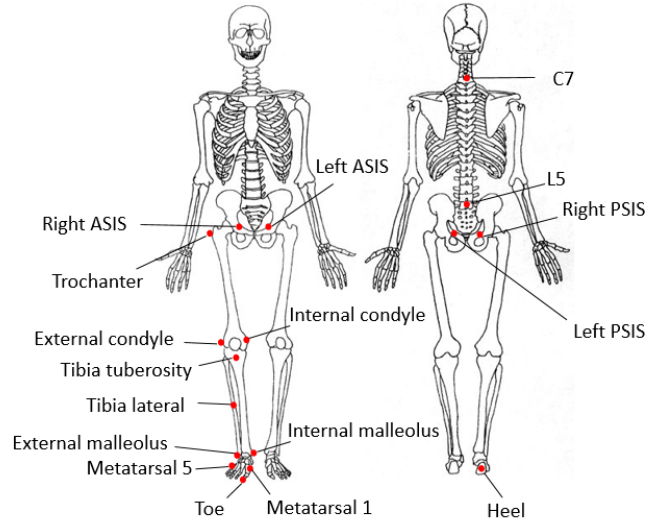


**Table 5.3:** Location of the 28 markers on the body (only 17 are represented because of the symmetry of the right and left legs) [41]

The purpose of the cutting step is to analyze the motion of the ankle in the frontal plane for a gait that is frequently used. Section 4 shows that the turn test could also do the job but this one implies other movements like rotation in the transversal plane (adduction/abduction) and the magnitude of the inversion/eversion movement is lower.

The inclined plane's objective is to analyze how the ankle reacts when walking on an incline plane, just as an outside ground. Does the foot just adapt itself to the inclined plane or there are huge torques in the ankle that are required to walk properly?

### 5.1.3 Data interpretation

Once we have the positions and the torques of the ankle in the frontal plane for each exercise, we can analyze them to find a particular characteristic impedance, which is the evolution of the torque in function of the inversion/eversion angle. It could have a particular curve, such as being linear and it could be deduced the behavior of the ankle (such as elastic control in the case of linear characteristic impedance).

We will analyze the power along the time. To obtain it, we will compute the derivative of the inversion/eversion position of the ankle to have the angular speed of the ankle in the frontal plane. We multiply it by the torque to obtain the instantaneous joint power.

$$P = v \cdot \tau$$

With the first two exercises, we hope to find that the energy needed to execute the inversion/eversion movement will be small compared to the sagittal plane. To compute the energy, we integrate the power:

$$W = \int P dt$$

We want to demonstrate which kind of mechanism allowing motions in the frontal plane (active, semi-active or passive) would have a real impact on the gaits of amputated people. To prove it, we will measure the acceleration of the participant's body and the angle of his body trunk in order to evaluate his stability.

## 5.2 Data processing

### 5.2.1 Remove outliers

As the basic principle of an optoelectronic system is to locate markers thanks to a camera, we could have interference, noise, or simply a marker that not be seen by a camera. This can generate two kinds of problems :

- **Noise** The noise problem is the same one as for every measurement device. If we have some noise on the collected data, we can just filter them to select the most important frequencies.
- **Marker jump** The biggest issue arises when a marker is detected a dozen centimeters from his real place for just one time step.

To minimize these problems, we have used four cameras to better detect all the markers. Unfortunately it is not enough and we have still some marker jumps.

As there are many tests and many subjects and as we could use it again for further research, we have developed a method to detect these outliers and replace them with an approximation of the real value (Annex D contains the *Matlab* code and the pseudo-code in the Algorithm 3 on page 48).

Firstly, we have to detect the outliers. To do this, we will find the coefficients of a polynomial  $P$  of degree 1 that fits best  $2N$  data around the time step  $t$  in a least-square sense. Once we have these coefficients, we will evaluate the polynomial at the same abscissa as the real points. An example is depicted in Figure 5.5 (on the left) where we show the real curve and all the regressions<sup>2</sup>. Now we can compare the real data with the data found by interpolating the regression corresponding to the time step  $t$  and compare them.

In statistics, for a data set  $x_1, x_2, \dots, x_N$ , a data  $x_i$  can be considered as an outlier if the data has a gap with the mean ( $\mu$ ) greater than three times the standard deviation  $\sigma$  which is the square root of the variance. The variance is given by [42]:

$$\sigma^2 = \frac{1}{N} \sum_i (x_i - \mu)^2 \quad (5.1)$$

So  $x_i$  has to fulfill the following condition to be an outlier :

$$x_i - \mu > 3 \cdot \sigma \quad (5.2)$$

In our situation, it makes no sense to use the mean because we do not have a data set  $x_1, x_2, \dots, x_N$  that should have the same value but we have a data set that evolves over time. However, we can use the interpolation of the regressions. So the variance of the equation 5.1 becomes

$$\sigma^2 = \frac{1}{2N+1} \sum_i (x_i - Int_i)^2 \quad (5.3)$$

where  $Int_i$  represent the  $i$ th value of the evaluations of the polynomial  $P$  for the time step  $t$ .

The condition for  $x_i$  to be an outlier of equation 5.2 becomes

$$x_i - Int_i > 3 \cdot \sigma \quad (5.4)$$

Accordingly to equation 5.4, if the value  $x_i - Int_i$  exceeds three times the mean error of the  $2N+1$  data, it is detected as an outlier and we replace it by the interpolated value. If not, we keep the real value.

---

<sup>2</sup>Actually, we have plotted only one regression every five regressions for a visibility purpose.

The outliers detected with this method can be observed in Figure 5.5 (on the middle) as well as the interpolated points in red and the real points in blue. We finally obtain the new data represented in red in Figure 5.5 (on the right) instead of the basic one in blue that contains marker jumps.

For our data, the best parameter was found experimentally to find all the correct outliers, without removing correct points. Many tests were done and the final parameter is a number of data interpolated at each time step of 21 ( $N = 10$ ).

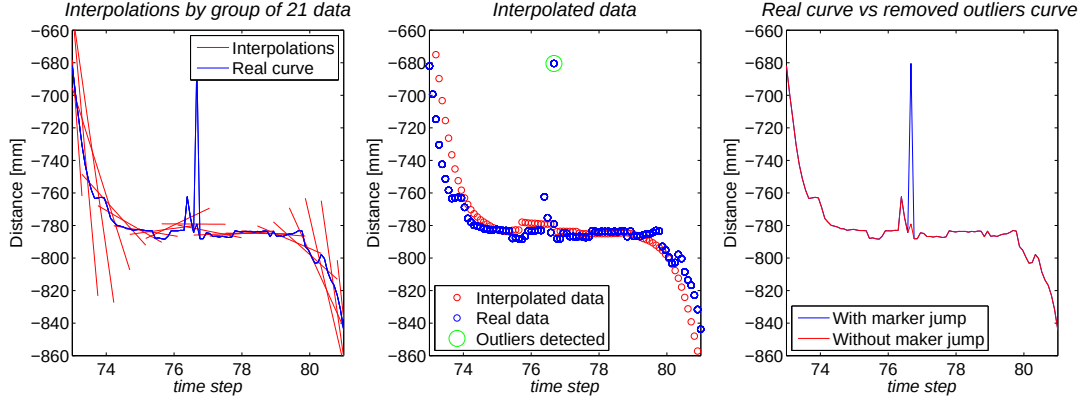


Figure 5.5: Example of removing outliers

---

**Algorithm 3** Remove outliers algorithm

---

```

1: function REMOVEOUTLIERS(AllMarkers)
2:    $see \leftarrow 10$ 
3:    $degrees \leftarrow 1$ 
4:   for  $k \leftarrow 1 : \text{length}(\text{AllMarkers})$  do
5:     for  $i \leftarrow 1 : 3$  do
6:        $marker = \text{AllMarkers}(k)_i$ 
7:        $x \leftarrow \text{linspace}(0, 100, \text{length}(marker))$ 
8:       for  $j \leftarrow 1 : \text{length}(x)$  do
9:          $f = \text{polyval}(\text{polyfit}(x(j-see : j+see), \text{fct}(j-see : j+see), degrees), x(j-see : j+see)))$ 
10:         $v(j) \leftarrow f(see + 1)$ 
11:         $Variance \leftarrow \text{mean}(\text{abs}(\text{fct}(j-see : j+see) - f).^2)$ 
12:        if  $\text{abs}(\text{fct}(j) - v(j, 1)) > 3 * \text{sqrt}(Variance)$  then
13:           $E(j) \leftarrow v(j)$ 
14:           $marker(j) \leftarrow v(j)$ 
15:        else
16:           $E(j) \leftarrow marker(j)$ 
17:         $\text{NewAllMarkers}(k)_i = marker$ 

```

---

Now that the outliers are removed, we can filter the new data to highlight important patterns in the data while leaving out the second problem which was the noise. As for the outside data in section 3.5.1, a moving average filter is used (see *Matlab* code in annex C). The annex F presents how the size of the filter was chosen.

## 5.2.2 Foot contact detection

We will explain in this section how we have detected the four following events: the heel strike, the toe strike, the heel off and the toe off. There are two main purposes of knowing these events.

The first one is to be able to detect the entire stance phase. Without it, we could not know where we are in the measurement. So this is the purpose of the heel strike and the toe off events detection.

The second reason is for a better understanding of the results. Thanks to the toe strike and heel off events detection, we have a better idea of where we are exactly in the stance phase.

It exists different ways to detect these moments based on the forces plat or on the position or speed of the markers. [43–45]. We have chosen a method that exists for the heel strike and the toe off events detection that use the force platforms [43] but we have created a new algorithm for the toe strike and the heel off events detection. We have taken the method based on the marker position in [44] but we had to modify it. In fact, this method only works for a run performed on a treadmill.

**Force plate method** In order to detect a heel strike event, we can use the force platforms. When the force in the z direction (vertical force) exceeds a threshold, we can assimilate it to the heel strike moment. The threshold could be 10N, as it is explained in [43].

A toe off event is determined in the same way except that the vertical force has to fall down below a 5N threshold. This difference in the threshold is explained by the fact that there is a more gradual reduction in force for the toe off moment.

However, as explained in [46], we can also use 5% of the weight of the subject as a threshold. This is what we have done for a more precise detection.

**Marker position method** Zeni J.A. et al. [44], described a method consisting of analyzing the trajectory in the x direction (direction of the walk) of a marker placed on the foot. As the subject walks on a treadmill, the graph of the foot marker versus the time is a sinusoidal curve. Hence, the peaks correlate to the time at the heel strike moment and the valleys coincide with a toe off event. Hence this method consists in detecting these maximums and minimums.

However, we are not on a treadmill and so we have to modify this algorithm. Figure 5.6 presents how we have done this for the heel off detection.

The modified method consists in taking the mean (dark) of the position of a foot marker in the z direction in the stance phase (blue). Then, we iterate on each time step in order to detect the index in which the value of the z direction of the marker is bigger than the mean times a threshold (1.2, which is experimentally found).

We repeat it but instead of taking the mean of the entire stance phase, we take the mean up to the point we detected just above (red).

We iterate until it reaches 25 iterations or if the previous index is the same as the new one. We experimentally observe that we do not need to iterate more than three times in general.

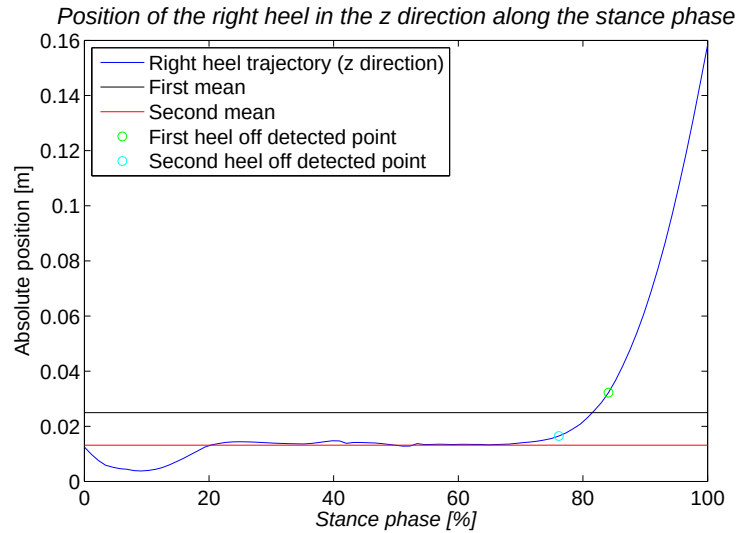


**Figure 5.6:** *Detection of the heel off and toe strike moment*

The pseudo-code of this function can be seen in the Algorithm 4 and the code is presented in the annex E.

---

**Algorithm 4** Foot contact detection algorithm

---

```
1: function FOOTCONTACTDETECTION(Heel, Toe, factor, iterMax)
2:   HeelMean  $\leftarrow$  mean(Heelz(1  $\rightarrow$  end))
3:   ToeMean  $\leftarrow$  mean(Toez(1  $\rightarrow$  end))
4:   done  $\leftarrow$  [0 0]
5:   for j  $\leftarrow$  1 : iterMax do
6:     for i  $\leftarrow$  1 : length(Heel) do
7:       if Heelz > factor · HeelMean & done(1) == 0 then
8:         newHeelIndex = i
9:         done(1)  $\leftarrow$  1
10:      if Toez < factor · ToeMean & done(2) == 0 then
11:        newToeIndex = i
12:        done(2)  $\leftarrow$  1
13:      HeelMean  $\leftarrow$  mean(Heelz(1  $\rightarrow$  newHeelIndex))
14:      ToeMean  $\leftarrow$  mean(Toez(1  $\rightarrow$  newToeIndex))
15:      if (newHeelIndex == oldHeelIndex) & (newToeIndex == oldToeIndex) then
16:        break
17:      oldToeIndex  $\leftarrow$  newToeIndex
18:      oldHeelIndex  $\leftarrow$  newHeelIndex
19:      done  $\leftarrow$  [0 0]
```

---

## 5.3 Dynamics model and torques computation

### 5.3.1 Modeling of body segments

In order to study the kinematics and the dynamics of the foot, each subject is represented with the same model but with different body segments length.

To get these segments length, a reference test was performed, in which the subject has to stand straight on a force platform in a position in which all markers are visible. The data of the markers and of the force plate is recorded for a few seconds in this position.

Thanks to this reference test, we can have the length of each segment of the body, particularly the length of the foot. But we also have a reference for the angles which is the rest position of the subject. As we will compute the dynamics, we also need the weight of the subject, which is measured during this test.

### 5.3.2 Angles computation

For both the kinematics and the dynamics analysis, we have to compute some angles. On the way to obtain the more precise computed angles, we have to involve as many as possible markers in the equations. In this part, we will present how we have computed the most important angle, which is the inversion/eversion angle. The other angles are computed in the same way.

All markers have three components: x, y and z. To denote the x component of the marker *Marker*, we will write *Marker*<sub>x</sub>.

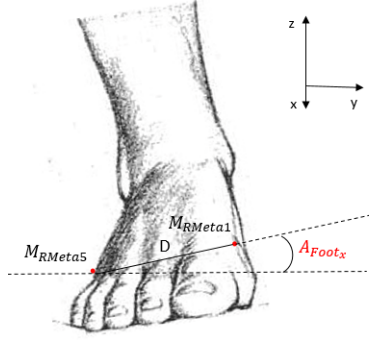


**Figure 5.7:** Schema used to compute  $A_{Foot_x}$  [47]

For the inversion/eversion angle, denoted  $A_{ie}$ , we must first compute the angle of the foot in the frontal plane, denoted  $A_{Foot_x}$  (because this rotation is around the x direction). This is performed thanks to two markers on the foot :  $M_{RMeta1}$  and  $M_{RMeta5}$  which are the markers on the metatarsal 1 and the metatarsal 5 respectively (Figure 5.7).

First of all, we compute the distance between the two markers  $M_{RMeta1}$  and  $M_{RMeta5}$  :

$$D = \sqrt{(M_{RMeta5_x} - M_{RMeta1_x})^2 + (M_{RMeta5_y} - M_{RMeta1_y})^2 + (M_{RMeta5_z} - M_{RMeta1_z})^2} \quad (5.5)$$

So we can compute the angle  $A_x$  with the following equation :

$$A_{Foot_x} = \text{asin} \left( \frac{M_{RMeta5_z} - M_{RMeta1_z}}{D} \right) - A_{Ref_{A_{Foot_x}}} \quad (5.6)$$

where  $A_{Ref_{A_{Foot_x}}}$  is the angle computed from the reference test, as explained in the section 5.3.1. It corresponds to the same angle ( $A_{Foot_x}$ ) but computed at rest, to have a reference. In this case, a  $0^\circ$  angle means that the foot is flat on an even ground.

The computation of the angle of the foot in the frontal plane does not depend on the direction of the walk (equation 5.6). But this is not the inversion/eversion angle. To obtain it, we have to compute the angle of the tibia in the frontal plane. The following markers are involved in the calculation :  $M_{RIntMal}$ ,  $M_{RExtMal}$ ,  $M_{RIntKnee}$  and  $M_{RExtKnee}$  which are the markers placed on the internal malleolus, external malleolus, internal knee and external knee respectively (Figure 5.8).

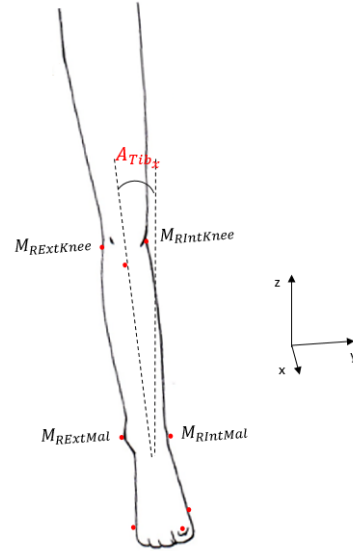


**Figure 5.8:** Schema used to compute  $Tib_x$  [48]

The tibia angle in the frontal plane ( $Tib_x$ ) is computed as follows :

$$A_{Tib1_x} = \text{atan2} (M_{RExtKnee_y} - M_{ExtMal_y}, M_{RExtKnee_z} - M_{ExtMal_z}) \quad (5.7)$$

$$A_{Tib2_x} = \text{atan2} (M_{RIntKnee_y} - M_{RIntMal_y}, M_{RIntKnee_z} - M_{RIntMal_z}) \quad (5.8)$$

$$A_{Tib_x} = \frac{A_{Tib1_x} + A_{Tib2_x}}{2} - A_{Ref_{Tib_x}} \quad (5.9)$$

where  $A_{Ref_{Tib_x}}$  is once again the reference of the tibia angle in the frontal plane obtained from the reference test (see section 5.3.1).

The  $\text{atan2}(X, Y)$  function is the four-quadrant inverse tangent in *Matlab* [49] and we have :

$$-\pi \leq \text{atan2}(X, Y) \leq \pi \quad (5.10)$$

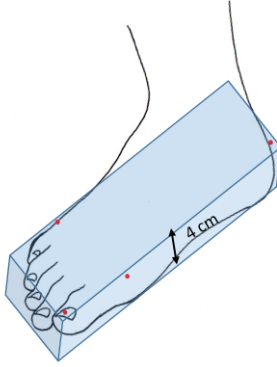
Finally, thanks to equations 5.6 and 5.9, we are able to compute the inversion/eversion angle :

$$A_{ie} = A_{Foot_x} - A_{Tib_x} \quad (5.11)$$

To conclude this part, we can notice that the equation of the inversion/eversion angle (equation 5.11) does not depend on the direction of the walk, which is important since we have different directions of the walk between each test. In fact, all angles have been computed in the same way, all independent of the direction of the walk.

### 5.3.3 Inertial modeling of the foot

For the computation of the torque, we will need the inertia of the foot. The model chosen is to compare the foot to a rigid box. Obviously, the dimensions of this fictive box are not the same between each subject. However, thanks to the reference test (see section 5.3.1), we are able to adapt the dimension of the box to the precise dimensions of the foot of each subject.



The width of the box is the distance between the markers on metatarsal 1 and metatarsal 5 in the y direction and the length is the distance between the marker on the heel and the marker on the toe in the x direction.

Concerning the thickness of the box, we cannot adapt it for each subject. Fortunately, this parameter does not variate a lot between each subject. Hence the value has been measured once and is put to 4cm. This model is depicted in Figure 5.9.

**Figure 5.9:** *Inertia model of the foot* [50]

Now that we have this model, we can compute the inertia moments thanks to the following equations:

$$i_x = \frac{m_f}{12} \cdot (b^2 + c^2) [Kg \cdot m^2] \quad (5.12)$$

$$i_y = \frac{m_f}{12} \cdot (a^2 + c^2) [Kg \cdot m^2] \quad (5.13)$$

$$i_z = \frac{m_f}{12} \cdot (a^2 + b^2) [Kg \cdot m^2] \quad (5.14)$$

$$(5.15)$$

where  $m_f$  is the mass of the foot and a, b and c are length, larger and thickness respectively.

Concerning the mass of the foot, we have used an anthropometric table [51]. We have found that the mass of the foot is 1.37% of the total mass of the subject.

### 5.3.4 Rotation center localization of the ankle

The determination of the center of rotation (CoR) or of the axis of rotation (AoR)<sup>3</sup> is important for the computation of the torque. We will explain our assumption in order to determine it.

<sup>3</sup>The straight line through all fixed points of a rotating rigid body around which all other points of the body move in circles.

The ankle is composed of the talocrural and the subtalar joints. The first one is the articulation formed between the talus and the tibia/fibula and the second is the articulation between the talus and the calcaneus (Figure 5.10) [52].

It has been proved that these two joints can be considered as a monocentric hinge joint for the walk and the run gaits [53]. In this work, we will use the same assumption as Ficanha paper [7], which is to say that the center of rotation of the foot is approximate as to be constantly the midpoint between the internal malleolus and the external malleolus.

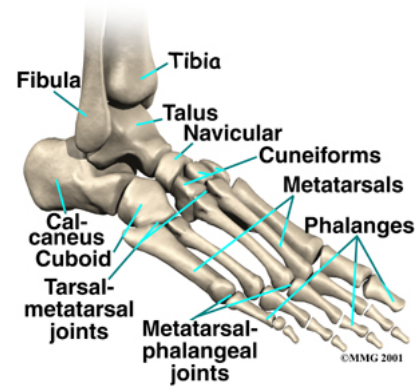


Figure 5.10: *Anatomy of the ankle* [54]

However, we could improve our model by using the method described in the annex K. This method is based on the computation of the center of rotation (CoR) at each time step and it could be more precise than the assumption we have done.

The principle of the method is to estimate the parameters of ball joints and hinge joints with a fixed axis of rotation (AoR)[55]. It is a least squares solution using all the data and without assuming rigidity but only that the markers keep the same distance from the CoR or AoR. We will here explain how to estimate the AoR.

### 5.3.5 Torque computation

As explained in the experimental protocol (section 5.1 on page 43), 28 markers were placed on each subject. But we have realized that the markers on the foot are enough to compute the torque in the ankle. The others are partially used for other purposes than the torque computation.

In fact, thanks to the force plate, we have access to the force reaction of the ground and the point of pressure of this force (see in annex J). So we have enough data by regarding only the foot and the reaction forces and not the rest of the body to find the articulation torque.

To find the torque we have the two equations 5.16 and 5.17. We can see the description of the terms in table 5.4. In the program, we had to use the framework of the foot which is explained in annex G.

$$\vec{W}_f + \vec{F}_{f/g} + \vec{F}_{t/f} = m_f \vec{a}_f \quad (5.16)$$

$$\vec{C}_{t/f} + \vec{M}_{W_f} + \vec{M}_{F_{f/g}} + \vec{M}_{F_{t/f}} = I_p \vec{\alpha}_p \quad (5.17)$$

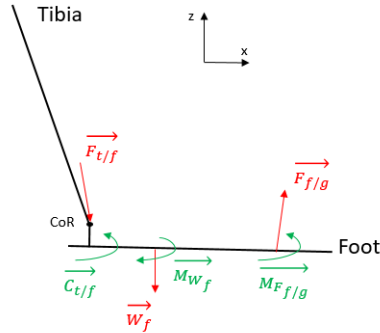
Each equation is actually three equations, one in each direction.

We will now compute these terms for the different directions in order to determine this articular torque.

| Notations     | Meanings                                                                     |
|---------------|------------------------------------------------------------------------------|
| $W_f$         | force due to the weight of the foot                                          |
| $F_{f/g}$     | Force of the foot to the ground                                              |
| $F_{t/f}$     | Force of the tibia (and the rest of the body) on the ankle                   |
| $m_f$         | Mass of the foot                                                             |
| $a_f$         | Acceleration of the foot (translation only)                                  |
| $C_{t/f}$     | Articular torque in the ankle                                                |
| $M_{W_f}$     | Torque due to the weight of the foot                                         |
| $M_{F_{f/g}}$ | Torque due to the force of the foot to the ground                            |
| $M_{F_{t/f}}$ | Torque due to the force of the tibia (and the rest of the body) on the ankle |
| $I_p$         | Angular moment of the foot                                                   |
| $\alpha_p$    | Angular acceleration of the foot                                             |

**Table 5.4:** Description of useful notations

We will first compute the sagittal torque in the ankle. The free body diagram of the tibia and the foot is depicted in Figure 5.11. All forces and torques involved in equations 5.16 and 5.17 are presented.



Concerning the forces, the only one we do not already know is the force  $F_{t/f}^z$ . Indeed, we know  $F_{f/g}^z$  thanks to the force plate and  $W_f^z$  thanks to the anthropometry table.

Hence, we directly obtain :

$$F_{t/f}^z = F_{f/g}^z - W_f^z$$

**Figure 5.11:** Free body diagram of the tibia and the foot used for the torque computation

Regarding the torque computation (Equation 5.17), we can know all term but we have to compute them. The first term we will develop is the torque due to the weight of the foot.

Thanks to the computed  $\beta$  angle (Figure 5.12, computed thanks to the section 5.3.2) and the distance  $L$  between the center of rotation and the center of mass (section 5.3.4), we are able to compute the following equation (Equation 5.18) for each time step:

$$M_{W_f} = L \cdot \cos(\beta) \cdot W_f^z \quad (5.18)$$

Now, we can develop the torque due to the force of the foot to the ground. This torque can be decomposed in two parts : the part due to the z component of the reacting force and the part due to the x component of the reacting force. Hence, based on the schema in figure 5.12, we have the following equation :

$$M_{F_{f/g}} = L_x \cdot F_{f/g}^z + L_z \cdot F_{f/g}^x \quad (5.19)$$

where  $L_x$  and  $L_z$  are the distance between the CoP and the CoR in the x and z direction respectively. We can find these parameters with the following equation :

$$L_i = |CoP^i - CoR^i| \quad (5.20)$$

where i is either x, y or z.

Finally, the last torque that must to be computed is the torque due to the forces of the tibia (and the rest of the body) on the foot. Since these forces are directly applied to the center of rotation of our model, we can directly say that this term is null. It leads us to the final relation to find the articular torque in the sagittal plane :

$$C_{t/f}^y = I_p \alpha_p^y - (L_x \cdot F_{f/g}^z + L_z \cdot F_{f/g}^x - L \cdot \cos(\beta) \cdot W_f^z) \quad (5.21)$$

where  $\vec{\alpha}_p^y$  is the angular acceleration around the y axis and  $C_{t/f}^y$  is the articular torque in the sagittal plane.

For the frontal plane, there is not the contribution of the foot weight since it only has a z component. Therefore, the torque due to the reacting forces is the only one that have to be computed (base on the schema in Figure 5.12) :

$$M_{F_{f/g}} = L_y \cdot F_{f/g}^z + L_z \cdot F_{f/g}^y \quad (5.22)$$

Hence, we obtain the final relation for the torque in the frontal plane using the equation 5.17 :

$$C_{t/f}^x = I_p \alpha_p^x - (L_y \cdot F_{f/g}^z + L_z \cdot F_{f/g}^y) \quad (5.23)$$

For the torque in the horizontal plane, we have the following relation :

$$C_{t/f}^z = I_p \alpha_p^z - (L_x \cdot F_{f/g}^y + L_y \cdot F_{f/g}^x) \quad (5.24)$$

## 5.4 Conclusion

This chapter has presented first the experimental protocol in which all the procedure followed in the laboratory tests is described. The placement of markers, the descriptions of the tests and the subjects can be found in this part.

Secondly, the data processing was described, which is the filtering and the foot contact detection. It could be interesting to know how these events are detected before use them in the results.

Finally, the dynamics model and the torques computation were described. This part present necessary

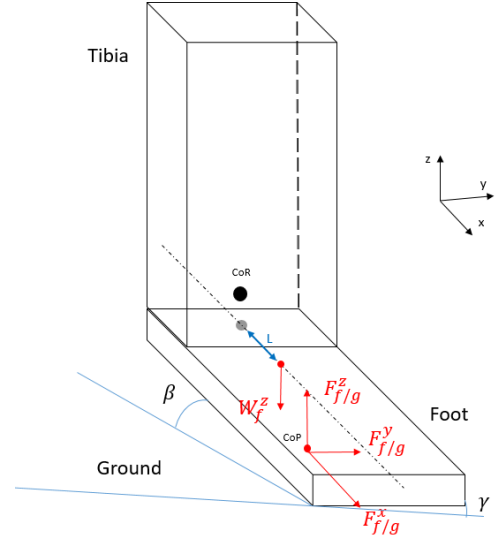


**Figure 5.12:** Schema of the foot while touching the ground with angles  $\beta$  in the sagittal plane,  $\gamma$  in the frontal plane and  $\alpha$  (not on the schema) in the transverse plane

parts for the computation of the torque in the ankle, which are the angles computations, the inertial modeling of the foot and the rotation center localization of the ankle. The next chapter will presents the results found on the basis of these parts. We will analyze the torque in the ankle computed in this part, for each tests that are described in the experimental protocol.

## CHAPTER 6

# ANKLE DYNAMICS IN-LAB EXPERIMENTATION; RESULTS & DISCUSSION

### 6.1 Introduction

In this chapter we will show the results and discuss them. The results and the discussion are in the same chapter in a readability purpose. It is easier to directly discuss the results with the graph that illustrate it just below.

The data will be analyzed in three steps. The first one will consist of a validation of the kinematics and dynamics models, i.e the validation of the angle and angular speed computation and the torque, power and energy consumption computations respectively. It is achieved by comparing our results to literature, and more precisely to the Ficanha study [7]. This part is important because it allows us to trust in our methodology to go forward in our study. The next step will be composed of the results for the different tests made. Finally, the added value of a mechanism for the inversion/eversion movement on an ankle prosthesis will be shown.

In these results, a negative angle is still an eversion angle (as for the chapter 4) while a positive one is the inversion. It is the same for the torque, when the torque tends to increase the eversion angle, it is a negative torque and when it tends to increase the inversion angle it is a positive torque. The confidence interval of each test is computed as  $M \pm 1.66 \cdot \sigma$  with M being the mean of each test of each subject and  $\sigma$  being the standard deviation computed with equation 6.1.

$$\sigma_j = \sqrt{\frac{1}{N} \sum_{i=1}^N (x_{i,j} - M_j)^2} \quad (6.1)$$

where j is the time step, i is the number of the test and N is the total number of tests (N = 24).

### 6.2 Validation of the kinematics and dynamics models

In order to validate our method, we will compare the results that are comparable to the literature. As we have seen in section 2.3 on page 17, we have access to data that have already been published [7]. A part of these experiences are the same or similar as we carried out in this work while the other experiments are different. Therefore, we will be able to compare the results they found to the results we found about the kinematics and dynamics parts.

We have done a normal walk test that is exactly the same as in the literature and a side step that is a little bit different. We will first compare the normal walk as it is exactly the same and then we will

compare the side step. We expect that the results for the normal walk are similar while the results for the side step are maybe a little bit different but close to each other.

To validate the results, the inversion/eversion kinematics and dynamic will be compared. This means that angle, the angular speed and finally the torque in the frontal plane will be analyzed. Then we will compute and compare the energy consumption<sup>1</sup>.

### 6.2.1 Normal walk comparison

At the beginning of a normal walk step, the ankle has an eversion angle until almost 25%. Then there is almost no inversion or eversion angle until 70% where there is an inversion angle.

Our results are very similar to the literature (Figure 6.1.). The standard deviation is bigger for our results than the literature, which is normal because they have more tests. But as 95% data are in the confidence interval, the literature and our results can be considered as regrouped.

It can be noticed that the standard deviation increases at the end of the stance phase for both results. This comes from the fact that when both the heel and toe are in contact with the ground, all subjects have basically the same inversion/eversion but not when the heel does not touch the ground anymore. In fact, when the heel leaves the ground, subjects could have a more different inversion/eversion angle because it depends more on the way to walk of each subject. It could also explain the difference at the beginning of the phase, before 25%, in which our results are more negative than the literature.

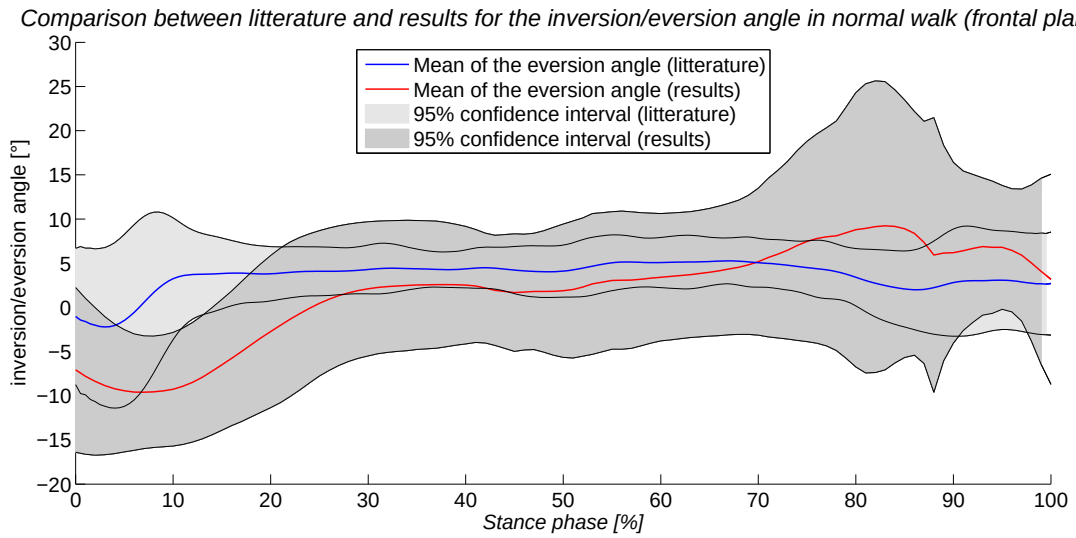


**Figure 6.1:** Evolution of the inversion/eversion angle during the stance phase of a normal walk, measured on 8 subjects in the present study (red line) and reported in the literature [7] (blue line)

Figure 6.2 shows the angular speed in the frontal plane for a normal walk for both literature and this thesis. Our results begin with a negative speed but becomes positive before 10% of the stance phase. Then it remains very slow until 85% where it becomes again negative. There is a strange step where the confidence interval becomes suddenly higher than everywhere in the test, which suggests that there is a measurement error.

The only noticeable difference with the literature is before 10%. But the confidence interval is much bigger for literature than for us. Therefore, we can say that our results are correct (except maybe the measurement error at 85%) since the confidence interval is almost everywhere inside the confidence interval of the literature results.

<sup>1</sup>The dorsiflexion/plantar flexion will also be compared in annex H because even if it is interesting, our work investigate the inversion/eversion movement first.

Comparison between literature and results for the inversion/eversion angular speed in normal walk (frontal plane)

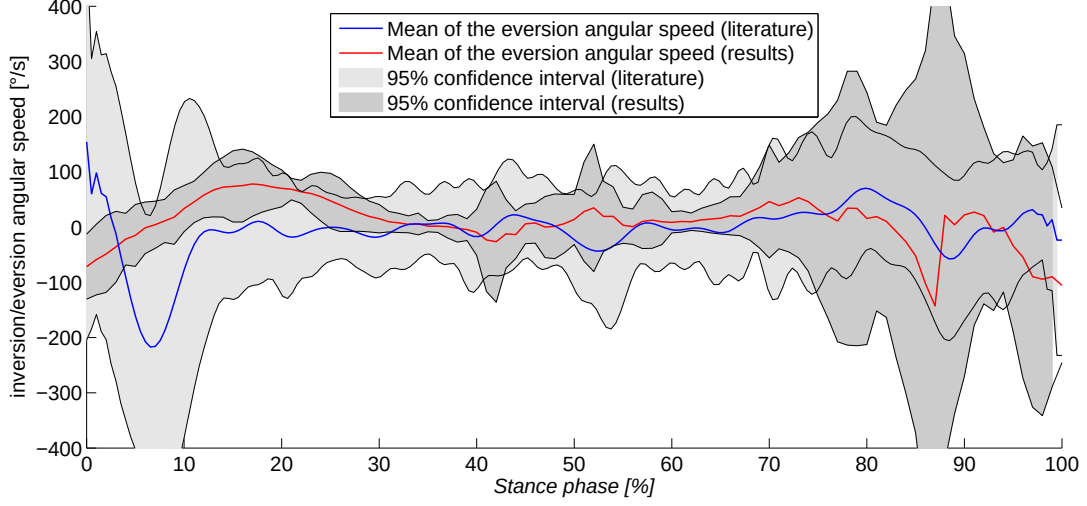


**Figure 6.2:** Evolution of the inversion/eversion angular speed during the stance phase of a normal walk, measured on 8 subjects in the present study (red line) and reported in the literature [7] (blue line)

The torque (Figure 6.3) begins by decreasing progressively until 75% of the stance phase where it reaches almost -0.2 Nm/Kg. Then it increases to reach almost 0 Nm/Kg.

Our results are very close to the results of the literature in general. It just has a little difference at 20% of the phase where the confidence intervals are not nested within each other anymore. The literature torque is also a little bit more negative for the entire phase but has a higher confidence interval than our results.

It results from the comparison between these two results that our model seems to give a correct inversion/eversion normalized torque along the stance phase, at least for the normal walk gait.

Comparison between literature and results for the inversion/eversion torque in normal walk (frontal plane)

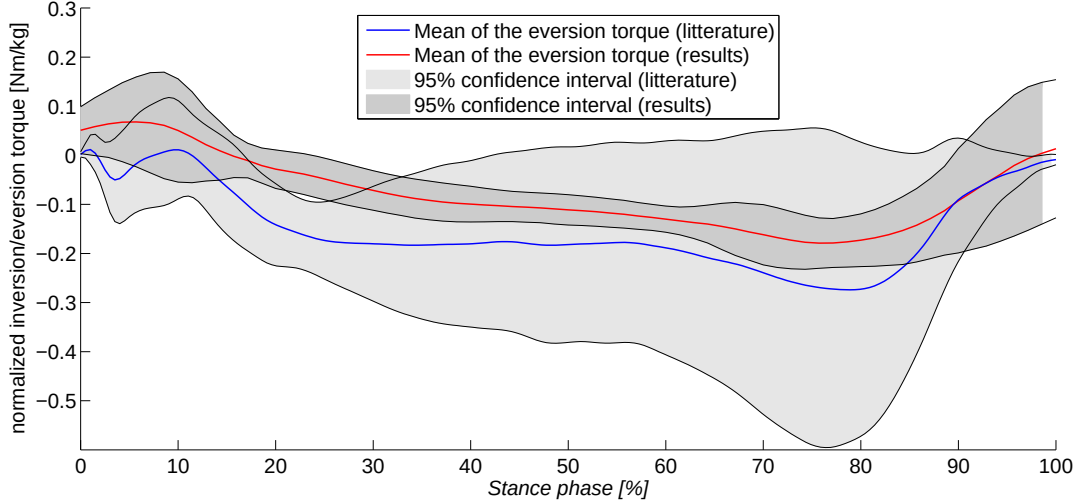


**Figure 6.3:** Evolution of the normalized torque during the stance phase of a normal walk, measured on 8 subjects in the present study (red line) and reported in the literature [7] (blue line)

Finally, the energy consumption of our results for a normal walk (considering a 75Kg person) is compared to the one of the literature. The mean of the energy consumption (not shown in the figure) is  $-0.4376$  [J] for literature and  $-0.9379$ [J] for our results.

As observed in Figure 6.4, our results are much more concentrated and are around the median of the literature results. Taking account of this and the fact that the means are very close to each other, we can say that the energy consumption for a normal walk from our data in the frontal plane corresponds entirely to literature.

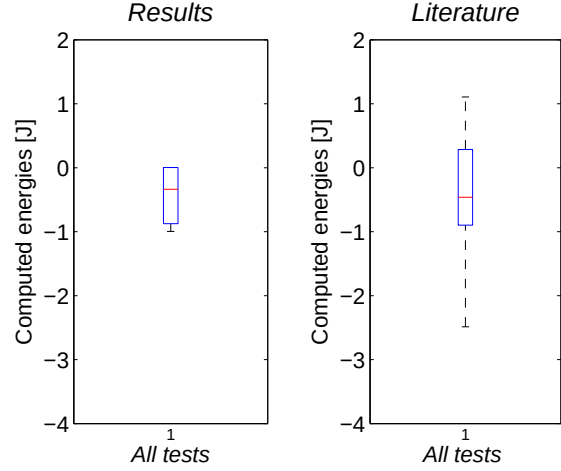


**Figure 6.4:** Energy consumption of the inversion/eversion movement during the stance phase of a normal walk for our results (left) and the literature [7] (right). The horizontal red line is the median, the edges of the box are the 25th and 75th percentiles and the black lines extend to the most extreme data point. This means that 50% of the data are inside the blue box and 100% data that are not considered as an outlier are inside the two extreme horizontal black lines.

### 6.2.2 Lateral step comparison

The lateral step of the literature is not exactly the same as the side step we used for the tests. The first one is in the continuation of the body direction while the second one interrupts the advancement of the subject. These two different sidesteps are presented in Figures 6.5 and 6.6.



**Figure 6.5:** Lateral step of literature [7]

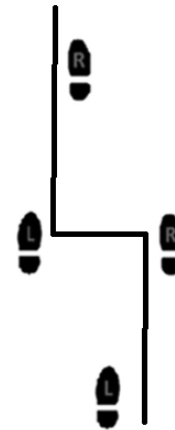


**Figure 6.6:** Our lateral step [7]

The inversion/eversion angle (Figure 6.7) of our results is decreasing on the first part of the stance phase, i.e. the eversion angle increases. At 20% of the stance phase, it reaches  $-12^\circ$  and then it increases to reach again  $0^\circ$  at 50% of the stance phase (neither inversion nor eversion angle). The eversion angle will again increase until 90% where it reaches  $15^\circ$  to decrease until the end of the stance phase.

As expected, the inversion/eversion angle of both literature and our results are really similar. The only noticeable difference is at 20% of the stance phase where the eversion angle measured in our tests is

bigger (angle more negative). It can be explained by the difference between the two kinds of lateral step. Another difference is at the end, after 90% of the stance phase. The inversion angle based on our data decreases a lot (until 4° of inversion) while the mean of the literature results just increases a little bit. This can be explained by the fact that when we reach 90% of the stance phase, the heel is not in contact with the ground anymore and the left foot is already in his own stance phase. Hence the eversion angle will not be imposed by anything else than the comfort or the natural gait of the subject. This implies that it is normal to record different eversion angles between subjects for this part of stance phase.

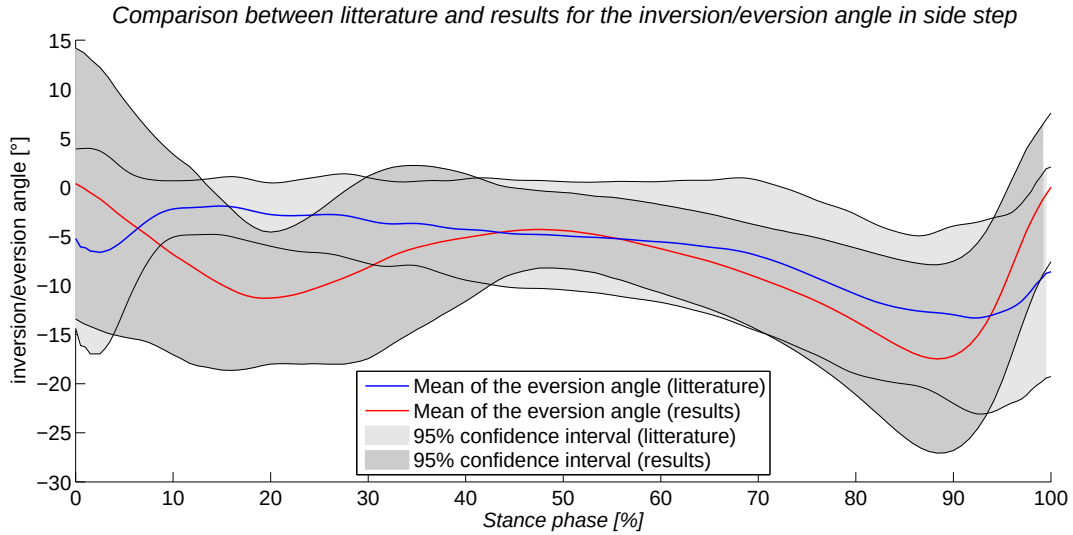


**Figure 6.7:** Evolution of the inversion/eversion angle during the stance phase of a lateral step, measured on 8 subjects in the present study (red line) and reported in the literature [7] (blue line)

The angular speed of our result (Figure 6.8) is very low until 85% of the stance phase where it reaches 200 °/s. Before 20% of the stance phase, the angular speed is increasing from -50 to almost 0°/s. Our results and the literature are quite similar concerning the angular speed of the inversion/eversion movements. The only difference is before 10% of the stance phase where the angular speed reported in the literature is higher than what we measured. However, the angular speed of the literature also has a higher confidence interval at this moment than ours, just like after 90%. It means that our data are still valid since they are within this confidence interval.

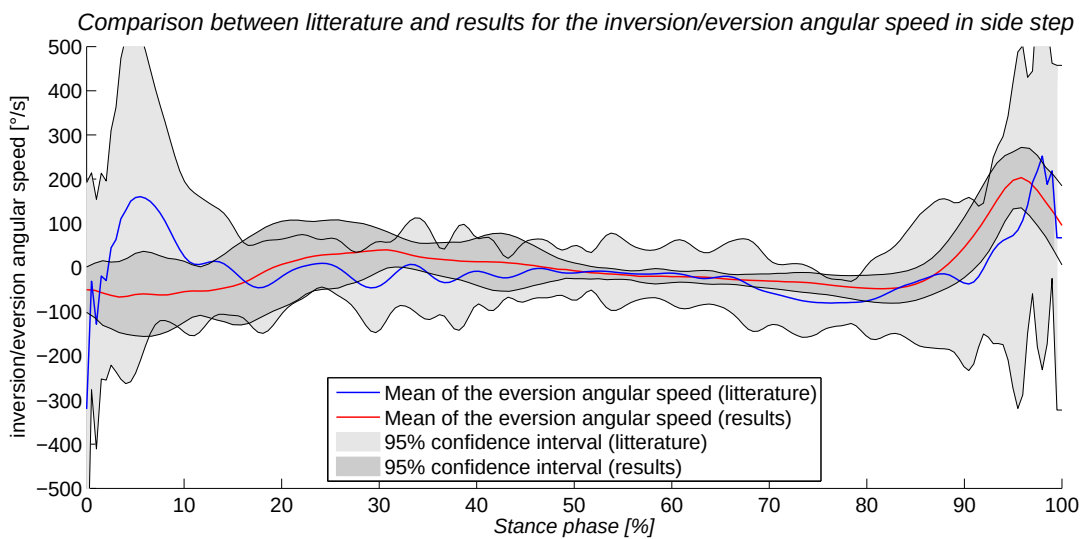


**Figure 6.8:** Evolution of the inversion/eversion angular speed during the stance phase of a lateral step, measured on 8 subjects in the present study (red line) and reported in the literature [7] (blue line)

The normalized torque measured in this study (Figure 6.9) remain at zero until 15% of the stance phase.

Then it reaches 0.3 Nm/Kg to decreases to zero again immediately after. It is only between 80 and 90% of the phase that the torque move to almost 0.2 Nm/Kg.

As it can be seen in Figure 6.9, the normalized torques are quite similar between our results and the literature even if the confidence interval of the literature is bigger than the confidence interval of our results.

But the main thing to note is that the torque we have measured before 30% of the stance phase is positive while the torque of the literature is negative. It is during the same part of the stance phase that the eversion angle we have measured was different from the one reported in the literature (Figure 6.7). Therefore it can also be due to the difference of lateral step. For the literature, the subject did not have to interrupt his speed. For our test, the subject has to slow down before initiate the lateral step. Figures 6.7 and 6.9 shows that subjects put their foot in order to slow down their speed using the eversion movement.

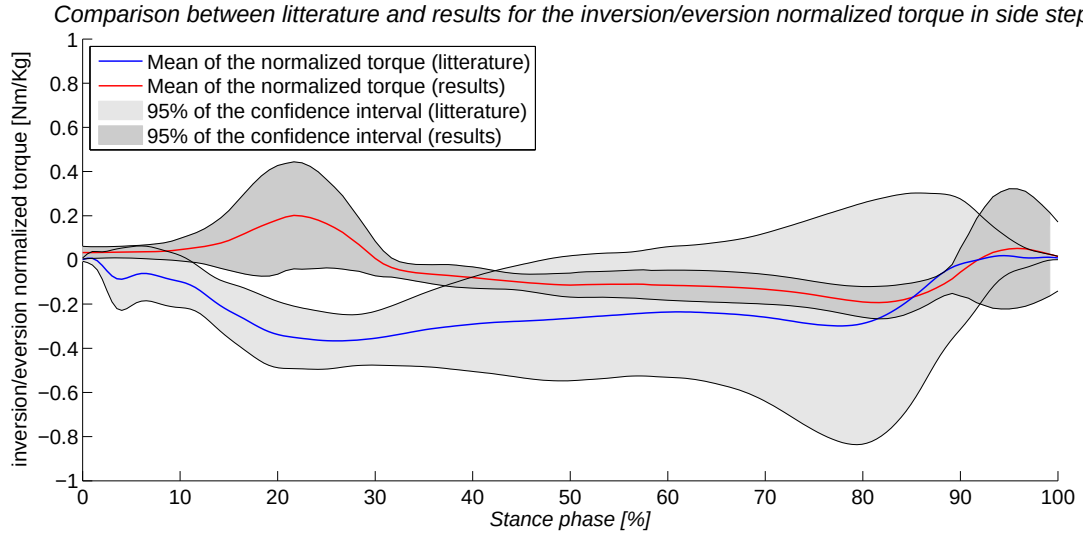
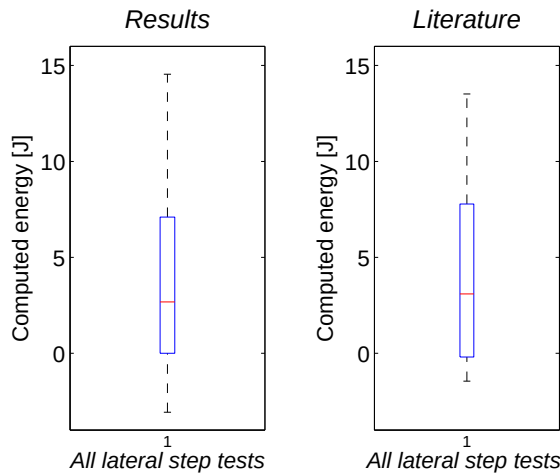


**Figure 6.9:** Evolution of the inversion/eversion normalized torque during the stance phase of a lateral step, measured on 8 subjects in the present study (red line) and reported in the literature [7] (blue line)



As for the normal walk, the energy consumption for a 75 kg person has been computed (Figure 6.10). For literature, we have a mean of 4.3856 J while we have 4.2371 J for our results (not shown in the figure). The literature results are very similar to our results since there is only a difference of 0.1485 J.

**Figure 6.10:** Energy consumption of the inversion/eversion movement during the stance phase of a lateral step measured in this study (left) and reported on the literature [7] (right). The horizontal red line is the median, the edges of the box are the 25th and 75th percentiles and the black lines extend to the most extreme data point. This means that 50% of the data are inside the blue box and 100% data that are not considered as an outlier are inside the two extreme horizontal black lines.

### 6.2.3 Conclusion

The goal of this part of our work was to validate our method. Therefore, we have measured the inversion/eversion angle, angular speed and torque for a normal walk and for a lateral step, and compared the results obtained with some results already published by Ficanha ([7]).

For the normal walk, our results were similar to those of the literature. The sidestep we used was a bit different from the one used by Ficanha ([7]). However, the difference in the results was not too high and there were logical in view of the difference in the test. It allows us to trust in our methodology to go forward in our study.

## 6.3 Analyze of the inversion/eversion movement

Our model being validated, we can trust in our methodology and go forward in the study. This is the second phase of our results which is composed of the results of all tests made in-lab. For each test, we will present the kinematics, i.e. the inversion/eversion angle and angular speed along the stance phase and the dynamics, i.e the normalized torque, the characteristic impedance, the power and the energy consumption of this movement. The purpose of this analysis is to characterize the degree of freedom that allows movements in the frontal plane for a more daily life context.

We will start by completing the normal walk and the lateral step tests. Then we will analyze the turn and the inclined plane tests. The last one will be considered as two different tests: the inversion mode and the eversion mode inclined plane tests.

### 6.3.1 Normal walk gait

The inversion/eversion angle, angular speed and normalized torque along a stance phase for a normal walk were presented in section 6.2.1. However it did not present the characteristic impedance the power and the energy consumption, which are the contains of this section.

The characteristic impedance of the normal walk gait (Figure 6.11) is very linear and both the torque and the inversion/eversion angle are very low. The angle reaches  $8^\circ$  of eversion at the very beginning of the stance phase (just after the heel impact) and  $10^\circ$  of inversion just before the heel off event. The absolute value of the normalized torque is always below 0.15 Nm.

It is the characteristic impedance of a hinge articulation with a circular spring that has  $5^\circ$  inversion as rest position. If the inversion or the eversion angle increases, the torque increases linearly.

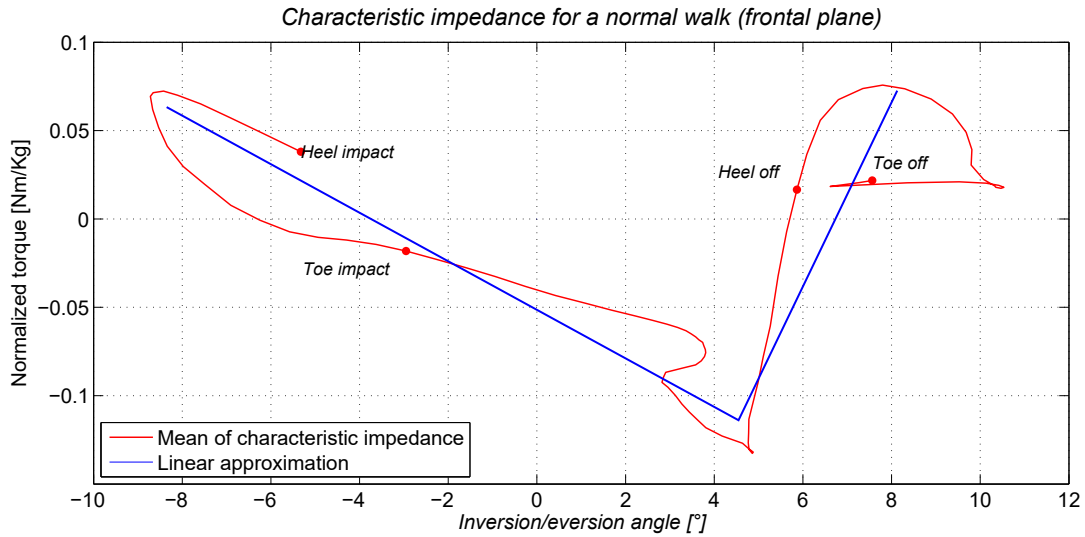


**Figure 6.11:** Characteristic impedance of the inversion/eversion motion during the stance phase of a normal walk step, measured on 8 subjects

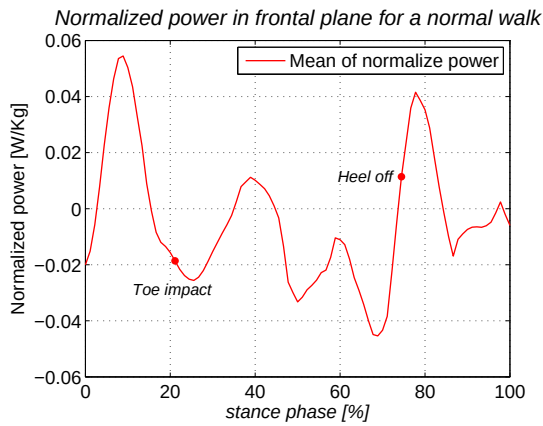


**Figure 6.12:** Power of the inversion/eversion motion during the stance phase of a normal walk step, measured on 8 subjects

The power along the stance phase in a normal walk is shown in Figure 6.12. It is logical to find a negative part bigger than a positive part when we know that we found a mean power of  $-0.9379[\text{J}]$  in section 6.2.1 (page 58). The power being so weak, it is difficult to extract a real behavior from the noise.

With these results of the normal walk gait, namely the power, the energy consumption and the characteristic impedance along the stance phase, the frontal plane degree of freedom seems to be not much used in these conditions as expected. This is why we have created more daily life conditions.

### 6.3.2 Lateral step gait

Section 6.2.2 (page 60) has presented the inversion/eversion angle, angular speed and torque along a stance phase for a lateral step. The characteristic impedance, the energy consumption and the power will now be investigated.

As a reminder, we are analyzing the right ankle for the left lateral step.

For the characteristic impedance (Figure 6.13) and the power (Figure 6.14), five phases are highlighted. They are described below and correspond to the numbers visible in Figures 6.13 and 6.14.

1. **Pose of the foot on the ground** In this phase, there is almost no torque. It is just the phase in which the foot goes from the natural position (when the foot does not touch the ground) to the

forced position (when the foot adapts itself to the ground). As a normal step, this phase has a low power consumption.

2. **Torque increase** When the subject initiates the lateral step movement, the torque in the ankle in the frontal plane increases linearly with the eversion angle until 0.15 Nm/Kg. It appears that the eversion angle decreases to reach 2° at the end of this phase. After visualization of the data thanks to *Mokka*, we found that it is the tibia that moves in the way that the inversion angle increases (and so the eversion decreases). This phase has a low power and is a little dissipative, with a minimum of -0.025 W/Kg. This means that we slow down the angular speed that increases momentarily the inversion angle.
3. **Almost constant torque** When the movement is initiated, the torque is almost constant (between 0.15 and 0.2 Nm/Kg) and the eversion angle increases rapidly, from 2° up to 11°. This is the transition phase in which the left foot is not yet on the ground but the body of the subject changes its trajectory. This phase increases the power consumption linearly from 0 to 0.12 W/Kg. This was expected since we have an almost constant torque and the inversion angular speed increases linearly (Figure 6.8 on page 61).
4. **Torque decrease** When the left foot touches the ground, the eversion torque of the right ankle will decrease from 0.2 to -0.1 Nm/Kg. Indeed, the weight of the subject and his inertia will decrease their impact on the right ankle more and more. As the inversion torque increases and the eversion angle increases as well, it is logical to find a power consumption that falls down rapidly (a minimum of -0.05 W/Kg for this phase).
5. **Push-off** When the heel of the right ankle leaves the ground, we have a brief pike of power up to 0.17 W/Kg. This is the push-off phase. But the power falls back to zero immediately after.

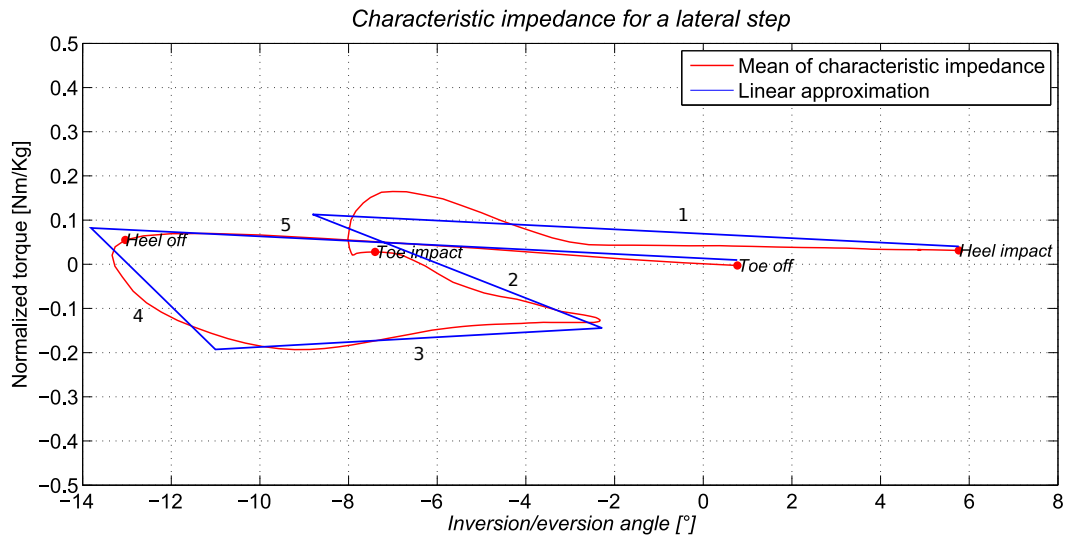


**Figure 6.13:** Characteristic impedance of the inversion/eversion motion during the stance phase of a lateral step, measured on 8 subjects

Figure 6.14 depict the power of the lateral step along the stance phase.

Two pikes of power are visible at 80 and 95% of the stance phase (up to  $0.17 \left[ \frac{W}{Kg} \right]$ )

. To explain this curve, we have added in the figure the phases detailed in the characteristic impedance.

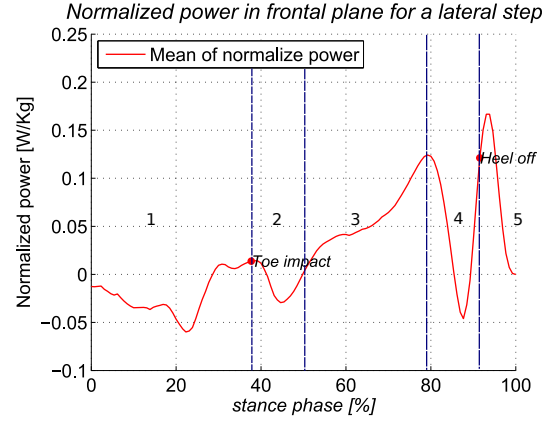


**Figure 6.14:** Power of the inversion/eversion motion during the stance phase of a lateral step, measured on 8 subjects

The graph of the characteristic impedance (Figure 6.13) shows that the inversion/eversion movement is more used than the inversion movement, which is logical as it is a left lateral step and this is the right ankle characteristic impedance. It uses 26% of the maximal range of motion compared with 21% for the normal walk. Furthermore, torques and power are higher for the lateral step than for the normal work. The power of the lateral step is between -0.05 and 0.17 W/Kg while the power of the normal step is between -0.04 and 0.055 W/Kg.

### 6.3.3 Turn gait

The turn gait has the inversion/eversion angle presented in Figure 6.15.

The eversion angle begins by increasing until almost  $10^\circ$ . This is due to the adaptation of the foot to the ground.

Then, at 20% of the stance phase (when the foot is entirely flat on the ground), the eversion angle decreases to reach zero. This is due to the first phase of the gait in which it is still a normal walk. This is why it is the same kind of eversion angle as for the normal walk test in which it begins by being as large as  $10^\circ$  and then decreases to reach  $0^\circ$  (Figure 6.1 on page 6.1).

However, instead of staying at  $0^\circ$ , the eversion angle will rapidly increase because the subject has initiated his turn movement.

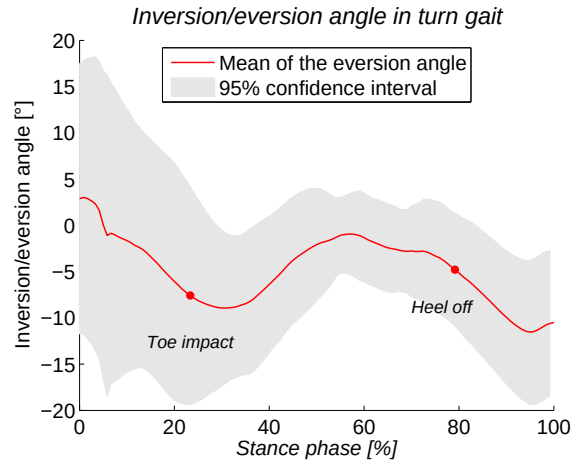


**Figure 6.15:** Evolution of the inversion/eversion angle during the stance phase of a turning step, measured on 8 subjects

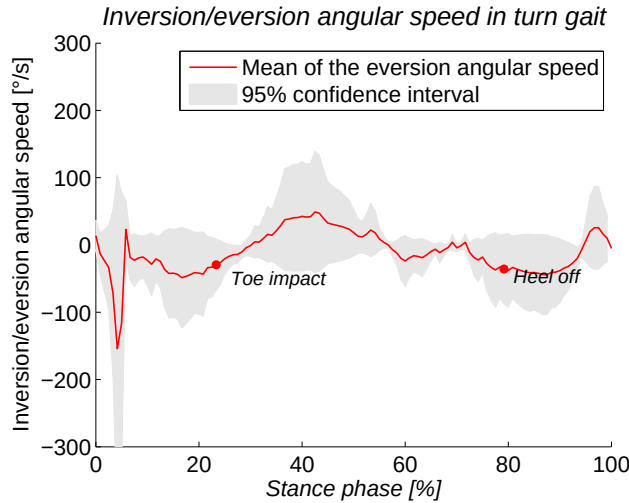


Figure 6.16 shows that the angular speed in the frontal plane is very low for this kind of gait. The biggest angular speed is before 20% of the stance phase, when the foot adapts itself to the ground and it reaches almost  $-150^{\circ}/s$ .

As the inversion/eversion angle is imposed by the position of the body and the ground after 20% of the stance phase, it is logical to find the same kind of speed magnitude as the lateral step gait (Figure 6.8 on page 61)

**Figure 6.16:** Evolution of the inversion/eversion angular speed during the stance phase of a turning step, measured on 8 subjects

The inversion/eversion torque is low (Figure 6.17), it is between  $-0.07$  and  $0.05$  Nm/Kg. Before the toe impact, the torque increases up to  $0.05$  Nm/Kg. Then it decreases linearly until  $-0.05$  Nm/Kg at the heel off moment where it increases again to zero. We can notice a small perturbation between 40 and 50% of the stance phase.

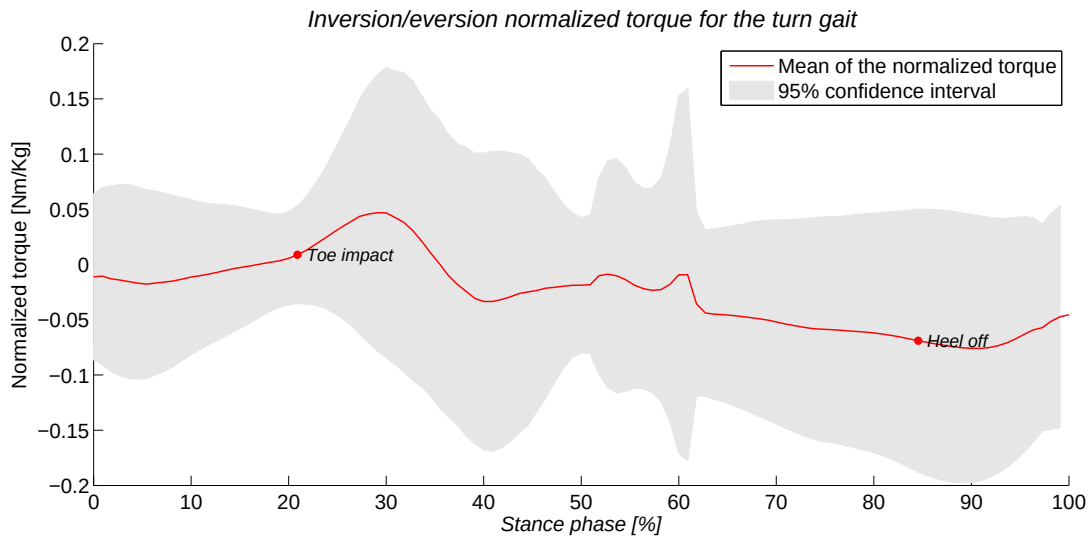


**Figure 6.17:** Evolution of the inversion/eversion normalized torque during the stance phase of a turning step, measured on 8 subjects

The characteristic impedance is depicted in Figure 6.18, in which three linear approximations can be observed. In the first part, the eversion angle increases from  $-1$  to  $9^{\circ}$  while the inversion torque increases from  $0$  to  $0.05$  Nm/Kg. This part is slowing down the eversion angular speed since a positive torque tends to increase the inversion angle and not the eversion angle. The third part is the most linear one. The eversion angle increases from  $1^{\circ}$  to almost  $12^{\circ}$  and the eversion torque increases from  $0.05$  up to almost  $0.1$  Nm/Kg. The second linear approximation is the most difficult to analyze since the linear approximation is maybe not valid. The eversion angle and the inversion torque decrease in this part but not linearly. It is interesting to observe that the motion turn counterclockwise in the graph. That means that there is a dissipation which is equal to the surface between part 1 and part 2 (hatched area).

As the turn motion is a combination of the lateral step and the normal walk, it could be interesting to compare them. Turning step gait has a much lower range of motion ( $13^\circ$ ) and torque ( $0.15 \text{ Nm/Kg}$ ) than the lateral step ( $20^\circ$  and  $0.37 \text{ Nm/Kg}$ ). Before initiating the movement (before the toe impact), the power has almost the same amplitude but after, the power in the lateral step is much larger than the turn step (up to  $0.06$  and  $0.17 \text{ W/Kg}$  for the turn step and the lateral step respectively). The difference is obviously the way in which this degree of freedom is used. The characteristic impedance of the normal walk, the lateral step and the turn step are very different. Because the lateral step has a slightly larger range of motion and most importantly a much larger torque, it is possible to analyze it phase by phase. It is more complicated for the turn step and the normal walk step.

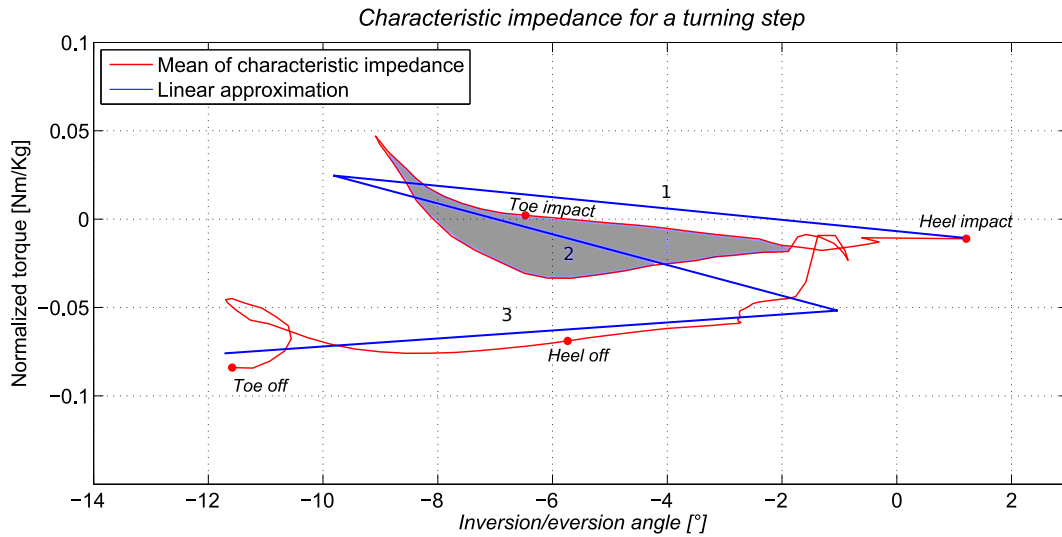


**Figure 6.18:** Characteristic impedance of the inversion/eversion normalized torque during the stance phase of a turning step, measured on 8 subjects

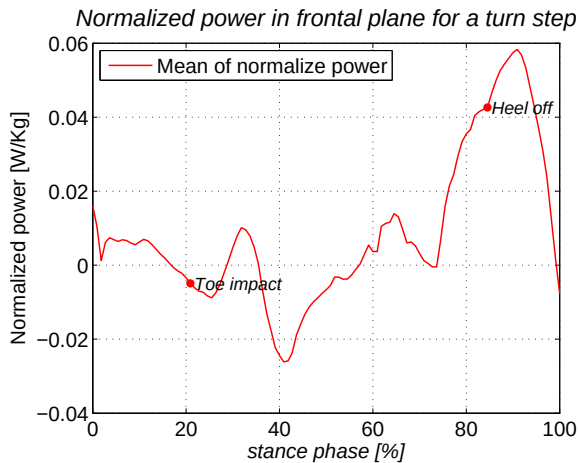
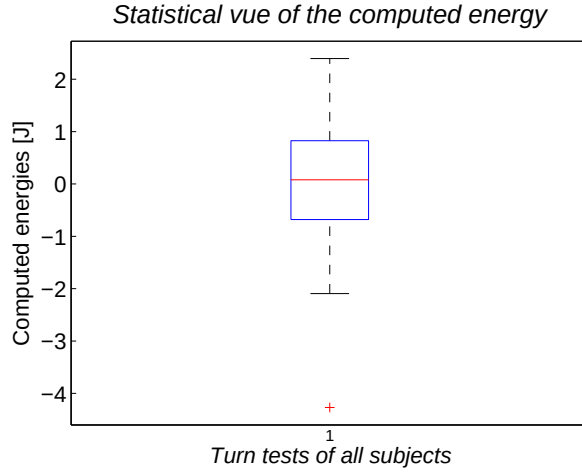


**Figure 6.19:** Evolution of the inversion/eversion power during the stance phase of a turning step, measured on 8 subjects

The power in the frontal plane for a turning step (Figure 6.19) begins with really small values. After the toe impact, there is a small increase up to  $0.01 \text{ W/Kg}$  but immediately after, the power decreases from  $0.01$  to  $-0.02 \text{ W/Kg}$ . Then there is an almost linear increase until  $0.06 \text{ W/Kg}$ , just after the heel off moment. Then it decreases rapidly.

The power between the *toe impact* moment and 50% of the stance phase is dissipative because the subject has to slow down his initial trajectory. Then we have a positive power because the subject initiates his new trajectory and there is the push off action. The power comes back to zero at the *heel off* event.



Finally, the energy developed for all tests of each subject for the turn gait has been computed (Figure 6.20). It shows that the computed energy for all turning step tests of all subjects are concentrated around zero with a median almost at zero. The mean energy (not shown in the figure) is 0.3266 [J] which is a positive energy but really weak.

**Figure 6.20:** Energy consumption of the inversion/eversion movement during the stance phase of a turning step. The central mark is the median, the edges of the box are the 25th and 75th percentiles and the black lines extend to the most extreme data point. This means that 50% of the data are inside the blue box and 100% data that are not considered as an outlier are inside the two extreme horizontal black lines. Data represented in red is data detected as outliers by the program.

In conclusion our results show that the inversion/eversion movement in a turning step is not used as much as the lateral step. Furthermore, the energy consumption is almost zero. As for the normal walk, we highlight the fact that the inversion/eversion degree of freedom seems not to be used a lot for this kind of movement.

### 6.3.4 Inclined plane gait

Section 6.3.2 has shown that the inversion/eversion movement in the normal walk and the turning step tests were not used extensively while the lateral step uses this movement a little bit more. The purpose of this section is to analyze the frontal degree of freedom in outside condition and more particularly on a non-flat ground. Thanks to the outside measurement device, we have found that the outside ground could reach 30° of roll angle at maximum (in extreme cases) but more frequently 15°.

Furthermore, for each of these two cases (15° and 30degree), there are two subparts that are the inversion mode and the eversion mode. When the direction of the inclination of the plane forces the subject to perform an inversion movement, it is said as to be the inversion mode. When the inclination of the plane force the subject to perform an eversion movement, it is the eversion mode. These two modes are presented in Figure 6.21.



**Figure 6.21:** Inversion (right) and eversion (left) modes of the inclined plane test [56]

For the purpose of clarity we will separate these two kinds of tests. We will begin with the eversion mode for both 15° and 30° slopes and then we will analyze the inversion mode for both the 15° and 30° slopes.

## Eversion mode

This part explores the eversion mode. As for the other tests, the kinematics and then the dynamics of the inversion/eversion movement will be analyzed. The 15° inclined plane will also be compared to the most extreme case which is the 30° inclined plane.

As observed in Figure 6.22, the eversion angle in both cases are very similar (except the magnitude). Three main parts can be observed:

1. In both cases the inversion angle of the foot increases linearly until the *toe impact* moment. It reaches 20° for the 15° case, and 30° for the other one.
2. Between the *toe impact* and the *heel off* events, the angle is very constant. As we expected, the inversion angle is almost the same as the inclined plane. After investigation, we have confirmed that it is the foot that is well flat on the inclined plane and the tibia is almost vertical (see appendix L for the figure made with *mokka* showing the foot and the tibia which confirms it).
3. After the *heel off* event, we have a brief spike of inversion angle. This spike reaches 20° and 25° for the 15° and 30° case respectively. This happens because when the heel leaves the ground, it will not do it as on a flat ground. The metatarsal 5 will leave before the metatarsal 1. Therefore there is an *over reaction* of the ankle.

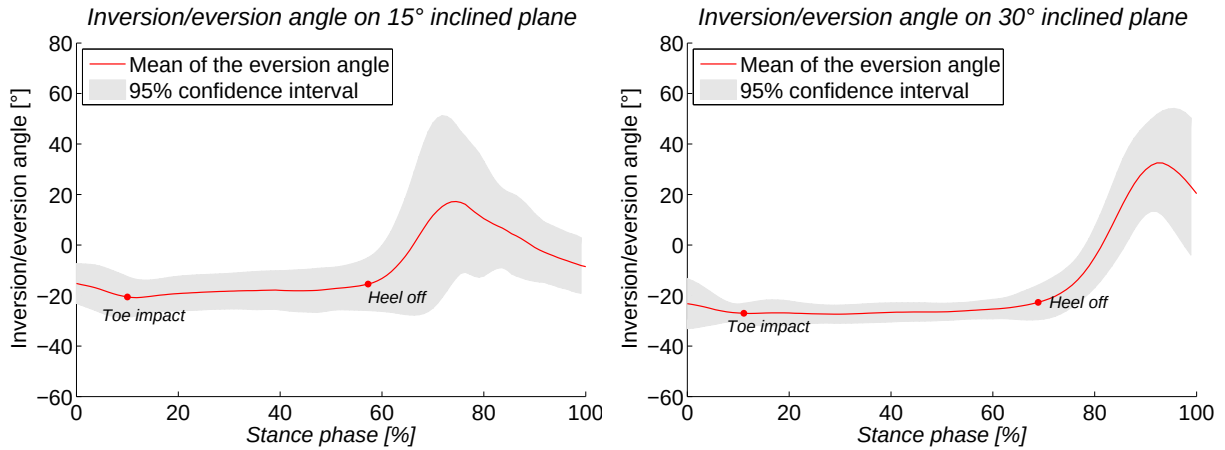


**Figure 6.22:** Evolution of the inversion/eversion angle during the stance phase of a 15° inclined plane (left) and 30° inclined plane (both in eversion mode), measured on 8 subjects

The angular speed (Figure 6.23) is again similar between the 15° and the 30° case. Two points can be noticed:

The first one is the confirmation of what was found in the angle analysis. There is a decreasing angular speed part in which the eversion angle increases. Then the second part which is the constant angle part. Finally, there is the *overreacting* part. The second point is that it reaches a high angular speed. Indeed, this magnitude of angular speed has never been reached for the normal walk, the turning step or even the lateral step. The biggest angular speed found was the end of the sidestep, with almost 200°/s (Figure 6.8 on page 61). The angular speed reaches 300°/s for the 15° inclined plane case and 400°/s for the 30° inclined plane case.

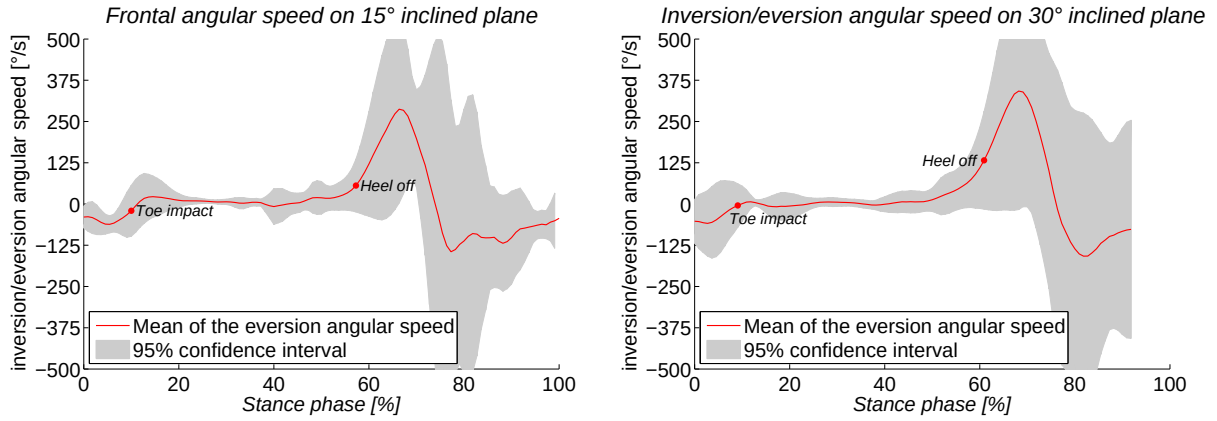


**Figure 6.23:** Evolution of the inversion/eversion angular speed during the stance phase of a 15° inclined plane (left) and 30° inclined plane (both in eversion mode), measured on 8 subjects

Regarding the torque, there is almost no difference. There is a little bit more torque for the 30° inclined plane case (almost 0.15 compared to 0.12 Nm/Kg for the 15° case) but it can reasonably be considered as negligible. There are two main parts. The first part is at the very beginning, before the *toe impact* moment. In this part, there is a negative speed, which means that the eversion angle is increasing (Figure 6.23). There is also a positive torque which is a torque that tends to increase the inversion angle (and thereby decreases the eversion angle). For this part, the subject is clearly dissipating energy to slow down this eversion angular speed.

The second part is between 70 and 85% of the stance phase. This part corresponds to the *overreaction* explained earlier. The torque and the angular speed being positive, it is a consuming part. As depicted in Figures 6.24 and 6.25, torque is null almost everywhere except on these two parts.

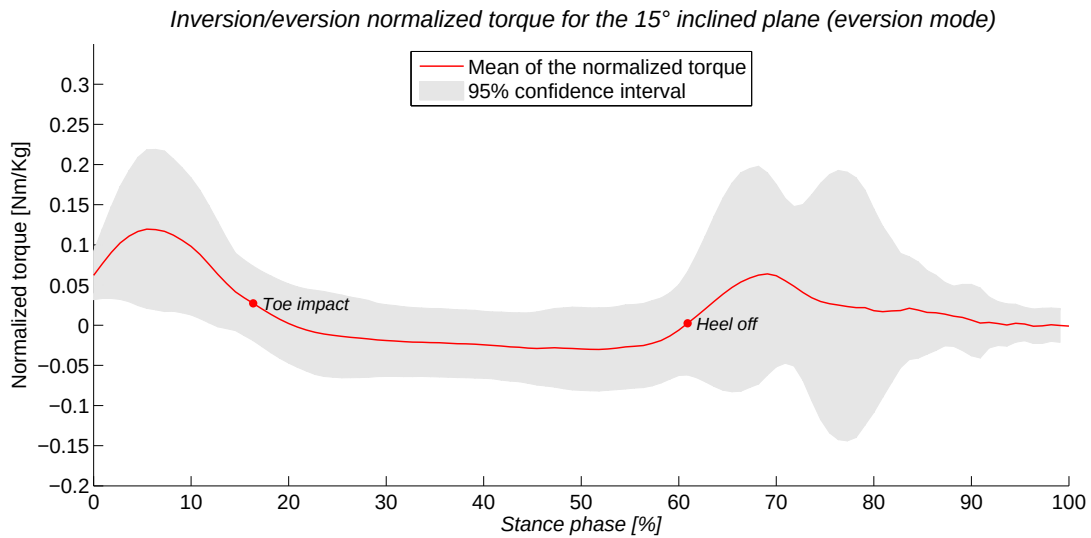


**Figure 6.24:** Evolution of the inversion/eversion normalized torque during the stance phase of a 15° inclined plane in eversion mode, measured on 8 subjects

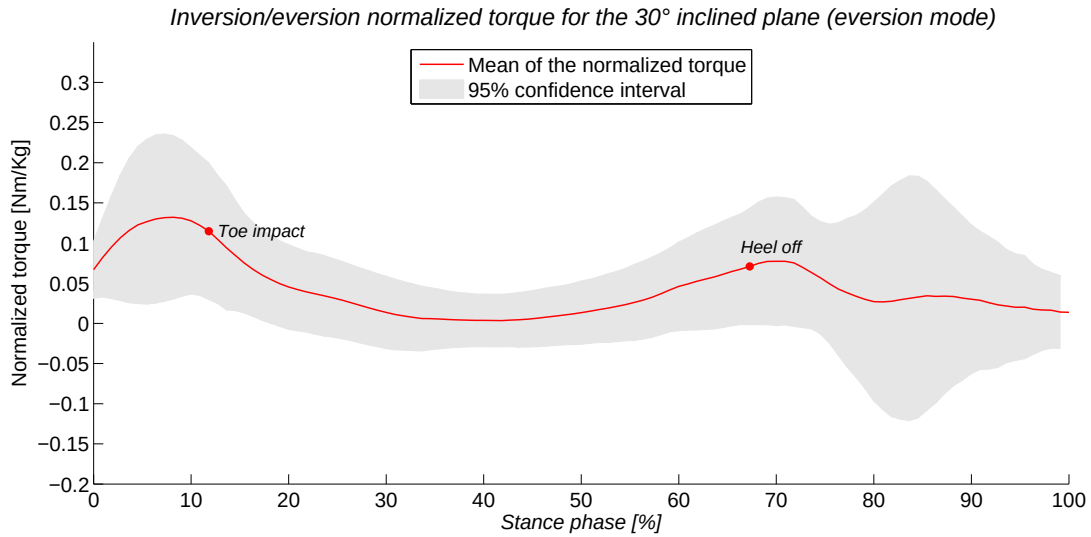


**Figure 6.25:** Evolution of the inversion/eversion angle during the stance phase of a 30° inclined plane in eversion mode, measured on 8 subjects

Figures 6.26 and 6.27 present the characteristic impedance of both inclined plane tests where the difference in the range of motion of the eversion angle can be observed. But beyond this difference, the characteristic impedance has a similar curve profile for both plane. Even the stiffness of the linear part (which is actually the slope of the line) is pretty much the same. There are three main linear approximations (namely 1, 2 and 3 in Figures 6.26 and 6.27).

1. The first linear approximation is the characteristic impedance of a spring. Indeed, the torque increases linearly as the eversion angle increases.
2. The second phase is composed of a quick dissipation part and a locked part. A torque that decreases for almost the same position is characteristic of a dissipation. It is the same if the spring was stretched and then it rest in the position without any torque. The energy of the spring is dissipated. When the torque increase in the same position, it is a locked mechanism.
3. The third linear approximation is the same as the first one except that the stiffness is lower.

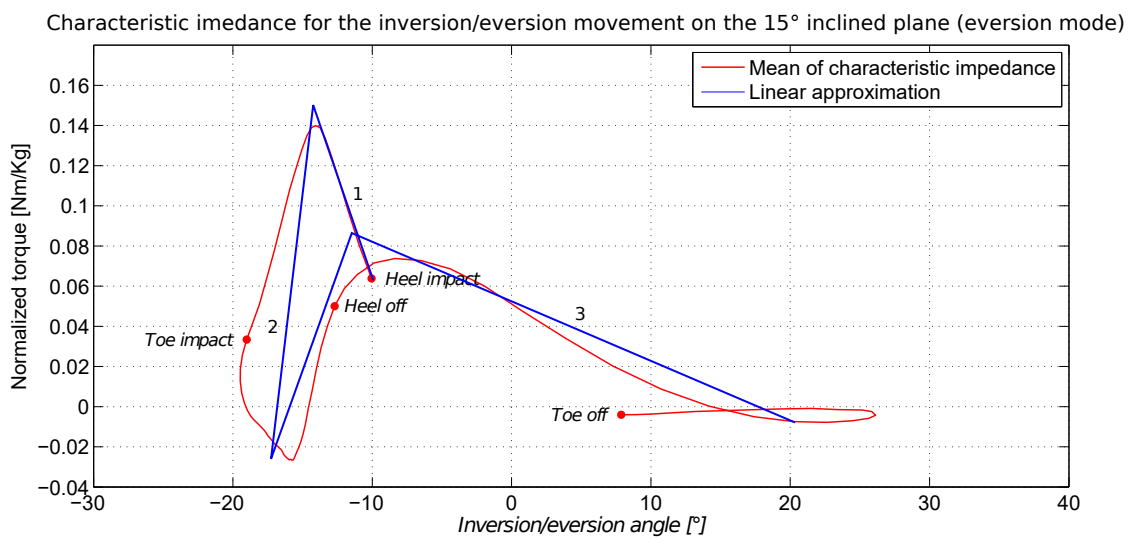


**Figure 6.26:** Characteristic impedance of the inversion/eversion degree of freedom for a 15° inclined plane ground (eversion mode), measured on 8 subjects

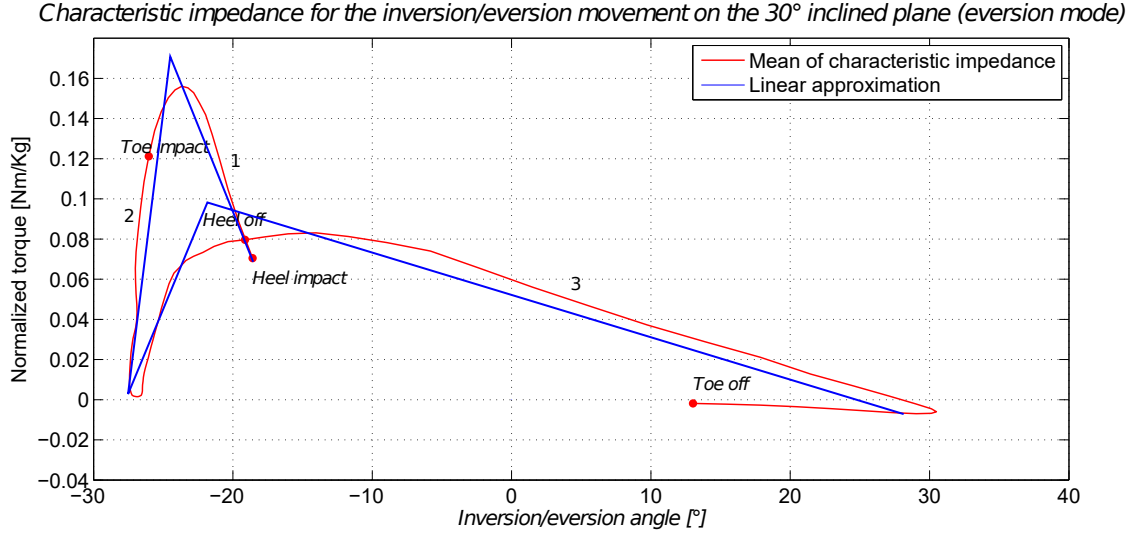


**Figure 6.27:** *Characteristic impedance of the inversion/eversion degree of freedom for a 30° inclined plane ground (eversion mode), measured on 8 subjects*

The normalized power (Figure 6.28) measured for the 15° and the 30° inclined plane show a similar curve profile. In the part before the *toe impact* event, the power is negative and has the same magnitude in both cases. This corresponds to the slow down of the eversion movement as highlighted in the torque analysis.

Secondly, we have a big growth in the *overreaction* part. This is the biggest power that was seen up to now. This time, we have a difference in magnitude between the two cases. Indeed, we reach almost 1.2 W/Kg for the 30° case while we reach 0.6 W/Kg for the 15° inclined plane case.

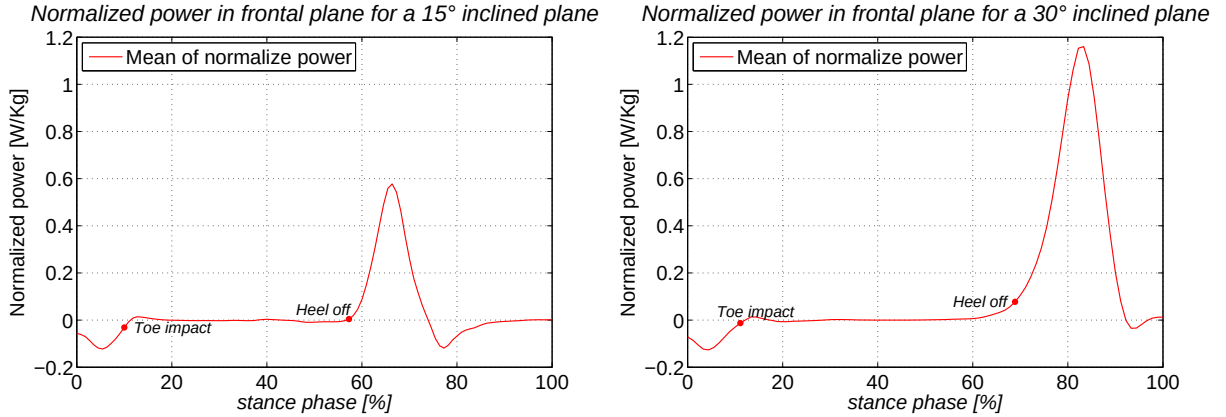


**Figure 6.28:** *Evolution of the normalized power in the inversion/eversion movement during the stance phase of a 15° inclined plane (left) and 30° inclined plane (both in eversion mode), measured on 8 subjects*

Because of this overreaction, the biggest part of the power is positive. This is the reason why we find a mean energy consumption of 0.8545 [J] for the 15° inclined plane and 3.1627 [J] for the 30° inclined plane. But there are some subjects for whom this overreaction is lower in terms of energy consumption than the slow down of the eversion angular speed for the 15° case (Figure 6.29). There are 7 negative energy consumptions for 17 positive ones (for a total of 24 tests as each subjects performed 3 times each test). For the 30°, they are all positive and the mean is higher.

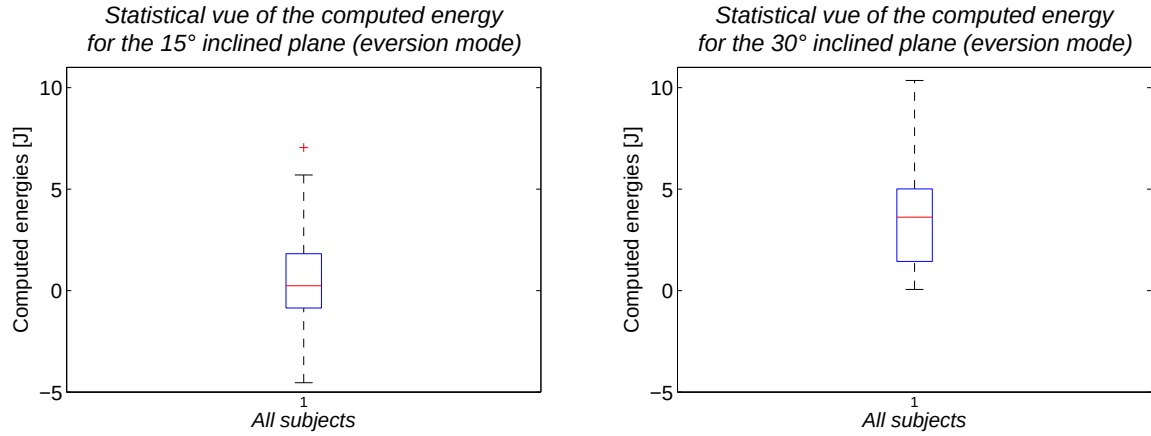


**Figure 6.29:** Energy consumption of the inversion/eversion movement during the stance phase of a 15° inclined plane (left) and a 30° inclined plane (right), both in eversion mode. The horizontal red line is the median, the edges of the box are the 25th and 75th percentiles and the black lines extend to the most extreme data point. This means that 50% of the data are inside the blue box and 100% data that are not considered as an outlier are inside the two extreme horizontal black lines.

On the basis of these results, three phases in a step were found on inclined ground in eversion mode. They are described in chronological order :

1. **Foot adaptation** The first phase is the foot adaptation to the ground. This phase is dissipative and seems to be independent of the incline of the ground in terms of power consumption.
2. **Foot flat on the ground** When the foot is entirely flat on the inclined ground, the eversion angle is adapted. It is proportional to slope. But this angle is constant (no speed) and the torque is very low. This implies almost zero power all along the phase.
3. **Push-off** From the analysis of the eversion angle to the analysis of the power, we have detected an *overreaction*. This is due to the push-off action. When the heel leaves the ground there is a peak of power in the sagittal plane.

## Inversion mode

We will now analyze the inversion mode. It is exactly the same test as before except that the inclination is in the other direction.

The inversion/eversion angle has similitudes with the eversion mode (Figure 6.30). There is an almost constant inversion angle when the foot is completely flat on the ground. But this time, it is no longer an eversion angle but it is an inversion angle, which is obviously expected since we now are in the inversion mode.

Another similitude is the *overreaction* just after the *heel off* event. However, it has a lower magnitude.

Once again, the 30° inclined plane test is very similar to the 15° inclined plane test. We have the exact same curve shape but for the inclination of 30°, we have a larger magnitude for the inversion angle and the inversion speed.

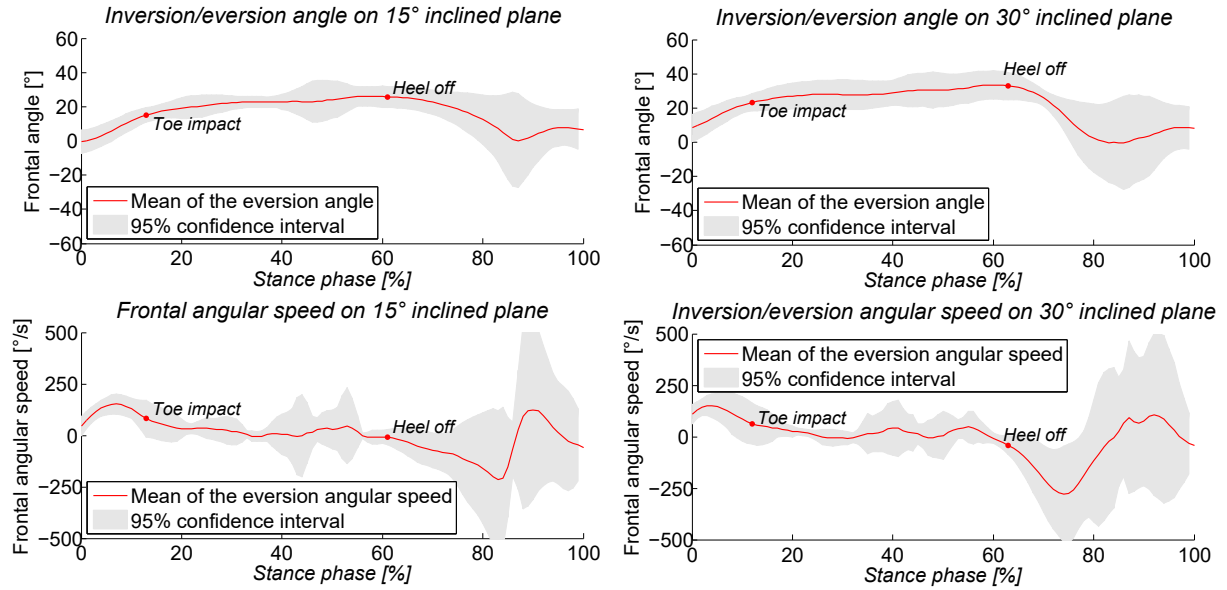


**Figure 6.30:** Evolution of the inversion/eversion angle (upper part) and angular speed (lower part) during the stance phase of a 15° inclined plane (left) and 30° inclined plane (both in inversion mode), measured on 8 subjects

The torque measured for the 15° and 30° inclined plane in inversion mode along the stance phase are more different than the eversion mode (Figures 6.31 and 6.32). Almost zero torque was observed for the eversion, except for the adaptation part (before the toe impact moment) and the overreaction part (after the heel off moment) (Figures 6.24 and 6.25 on page 71). For the inversion, no *overreaction* moment was observed. Indeed, the normalized torque is decreasing until 60% of the stance phase, which could correspond to the *heel impact* moment of the second foot, and then rises to zero.

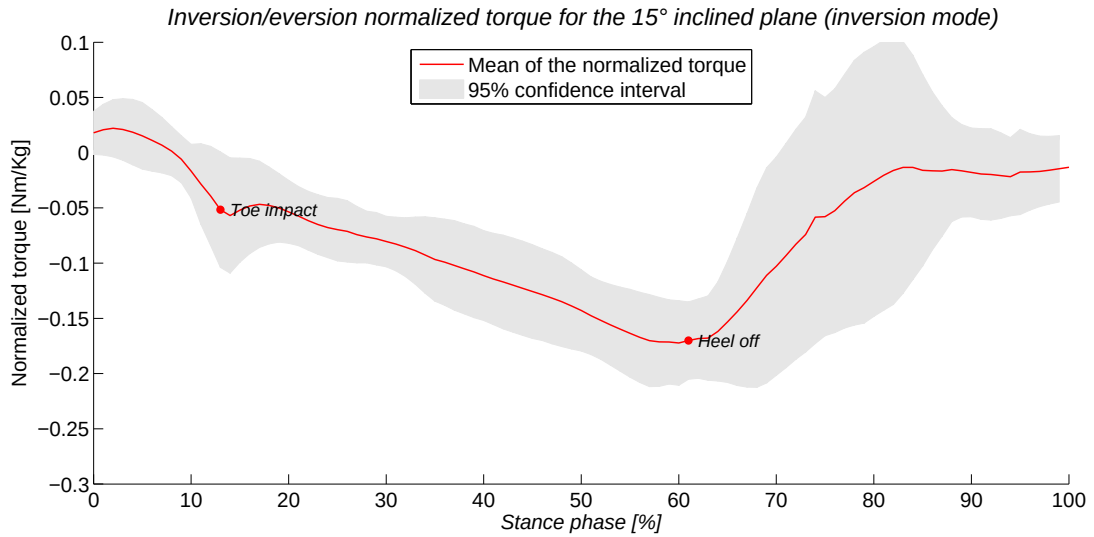


**Figure 6.31:** Evolution of the inversion/eversion normalized torque during the stance phase of a 15° inclined plane in inversion mode, measured on 8 subjects

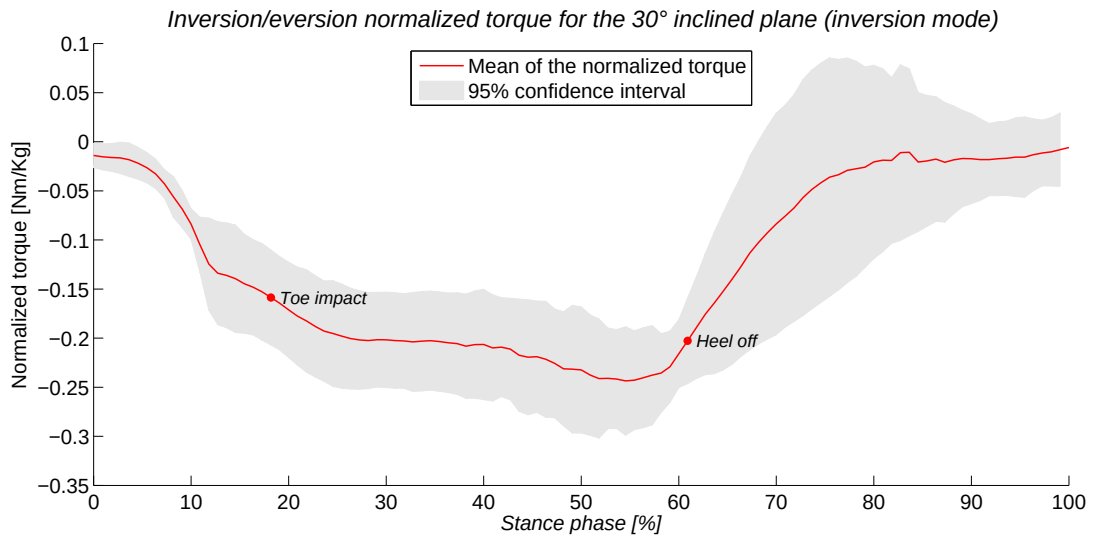


**Figure 6.32:** Evolution of the inversion/eversion normalized torque during the stance phase of a 30° inclined plane in inversion mode, measured on 8 subjects

Figures 6.33 and 6.34 represent the characteristic impedance of the 15° and 30° inclined plane in inversion mode, respectively. The linearity can easily be observed on both figures. Furthermore, both cases have a similar stiffness. The range of motion of the inversion angle is strongly correlated to the inclination of the ground. The 15° inclined plane test has 25° of inversion angle range of motion while the 30° inclined plane test has 35°.

The inversion angle seems to increase until the heel off event and then come back to the rest position. The torque decreases with the inversion angle, which means that the ankle tries to oppose the movement.

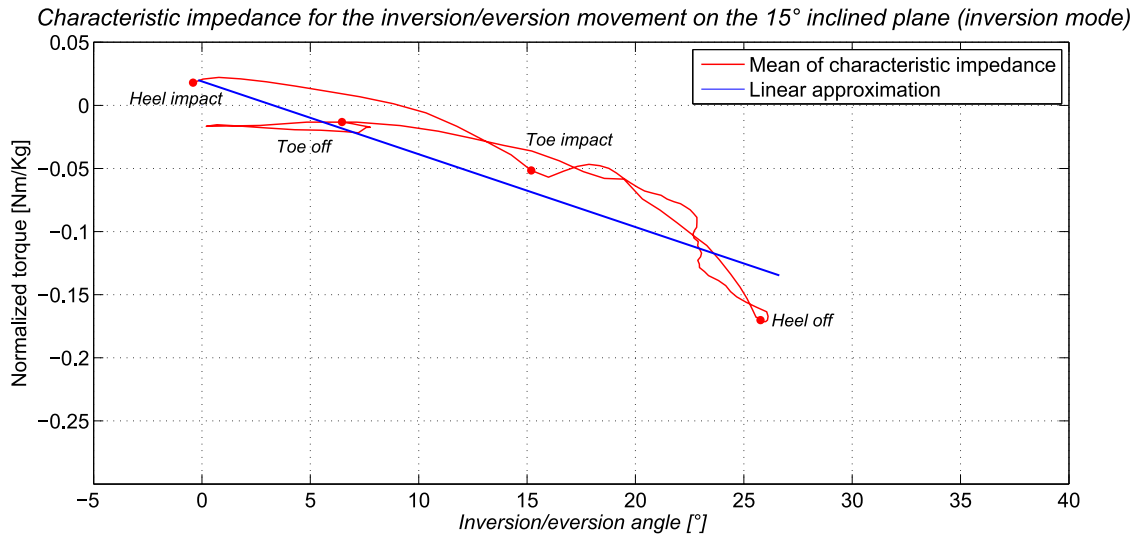


**Figure 6.33:** Characteristic impedance of the inversion/eversion degree of freedom for a 15° inclined plane ground (inversion mode)

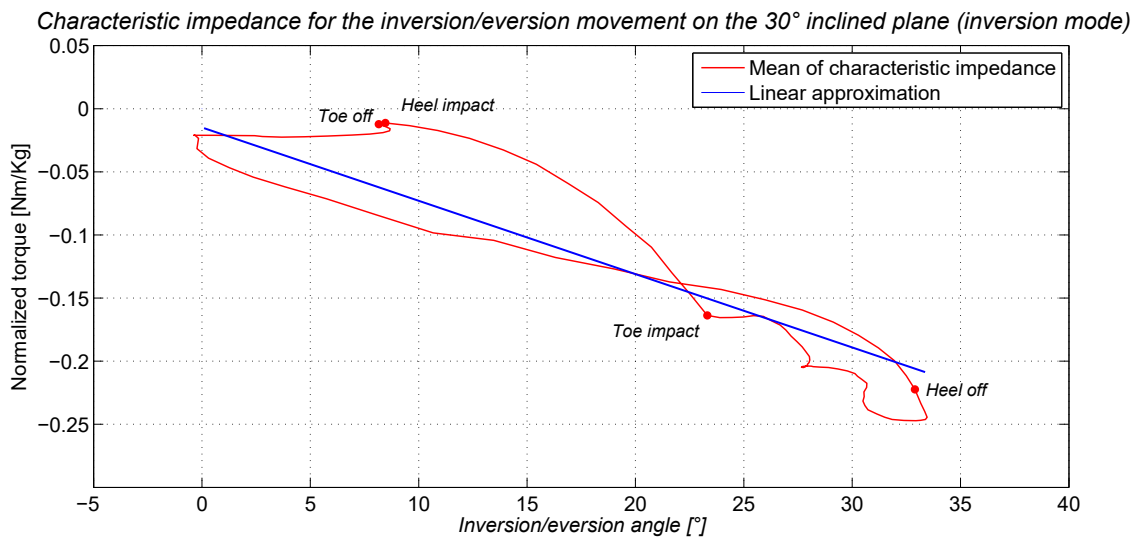


**Figure 6.34:** Characteristic impedance of the inversion/eversion degree of freedom for a 30° inclined plane ground (inversion mode)

The power observed for the inversion mode (Figure 6.35 is similar to the power measured for the eversion mode (Figure 6.28 on page 73) in the sense that there is always a negative power before the *push-off* peak of power, which is positive. This peak is again bigger for the 30° inclination case. The main difference is that there is not this almost zero torque part when the foot is flat on the ground. It can be explained by the fact that the inversion mode is a less natural gait. Subjects have more difficulty to keep their stability with this incline.

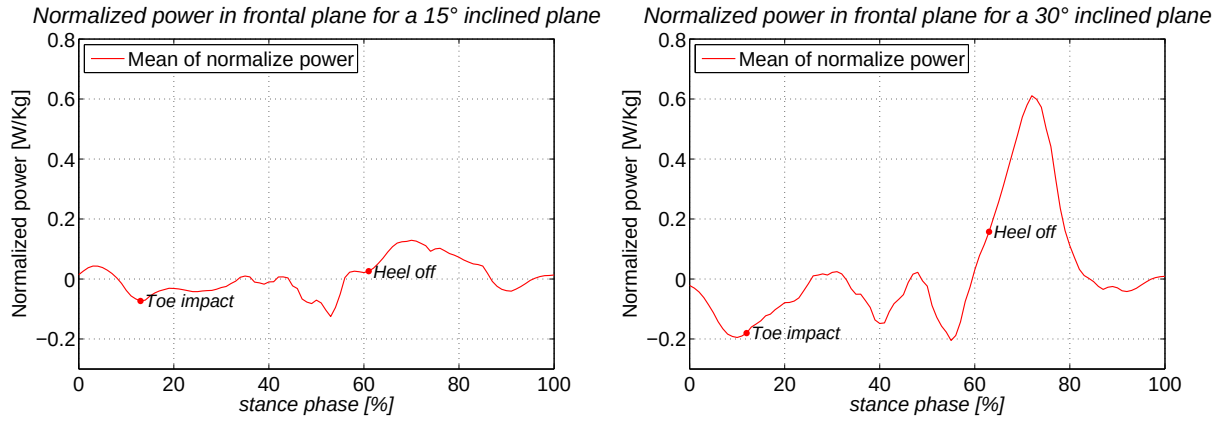


**Figure 6.35:** Evolution of the normalized power in the inversion/eversion movement during the stance phase of a 15° inclined plane (left) and 30° inclined plane (both in inversion mode), measured on 8 subjects

Both cases (15° and 30° inclined plane) have a really weak energy consumption. As observed in Figure 6.36, the median is almost zero in both cases and the mean energy consumption over all subjects is -0.3250[J] for the 15° case and 1.3654 [J] for the 30° case (for a 75Kg person, not shown in the figure).

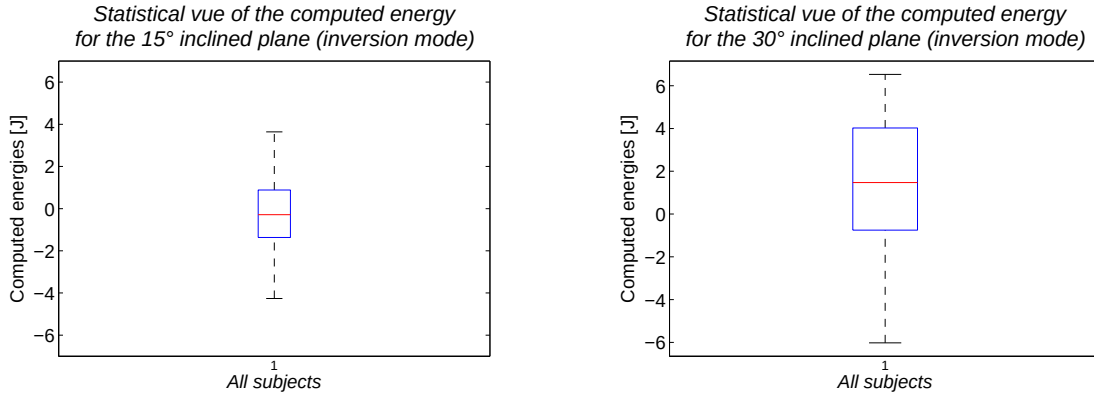


**Figure 6.36:** Energy consumption of the inversion/eversion movement during the stance phase of a 15° inclined plane (left) and a 30° inclined plane (right), both in inversion mode. The horizontal red line is the median, the edges of the box are the 25th and 75th percentiles and the black lines extend to the most extreme data point. This means that 50% of the data are inside the blue box and 100% data that are not considered as an outlier are inside the two extreme horizontal black lines.

### 6.3.5 Conclusion

From this analysis of the inversion/eversion movement, we can conclude that this degree of freedom seems to be really involved in the inclined plane gait for both the 15° and 30° cases (inversion as well as eversion mode).

The characteristic impedance observed for the inversion mode appears to be really different than for the eversion mode. Actually, they are quite similar when the anticipation is highlighted. It can be observed that the characteristic impedance of the ankle on an inclined plane in eversion mode has a different eversion angle at the heel strike moment between the 15° and 30° case and is not zero. The ankle has already 10° or 20° as eversion angle respectively for the 15° and 30° inclined plane tests in eversion mode. This means that subjects have anticipated the movement, probably unconsciously. The inversion mode has almost no anticipation, maybe because the inversion gait is more difficult to perform in the swing phase. If there was no anticipation, the characteristic impedance of the ankle on an inclined plane in eversion mode could be similar to the inversion mode (really linear and come back to the rest position) but with a dissipation phase at the maximum eversion angle reached.

## 6.4 Impact of a constrained inversion/eversion movement on the stability

We have analyzed the inversion/eversion movement in section 6.3 and we have deduced that the energy developed is very low. However, the range of motion is quite consequent and we have some pikes of power that are not negligible. Furthermore, we have a very specific characteristic impedance of the inclined plane gaits.

The aim of this section is to analyze the impact on the stability when we do not have this degree of freedom and to determine if this DoF is really necessary. As we have seen in section 5.1 on page 43, we have done that by adding an orthosis on the right ankle of the subject. Thereby, we will observe the difference in the stability when the same subject performs the tests with and without the orthosis (constrained and not constrained case).

They were still 8 subjects that participate in these tests but only 7 were taken into account because one of them had a misplaced marker which forbids us from using its results.

### 6.4.1 Impact on the body trunk angle

One of the main purposes that we expect by adding the degree of freedom allowing motion in the frontal plane is to gain in stability. To have an idea about the difference in stability, we will analyze the difference that induces a constrained ankle on the body.

We will first analyze the angle of the body thanks to the markers L5 and C7. In fact, we are able to find the angles of the body trunk in the sagittal and in the frontal planes (Figure 6.37).

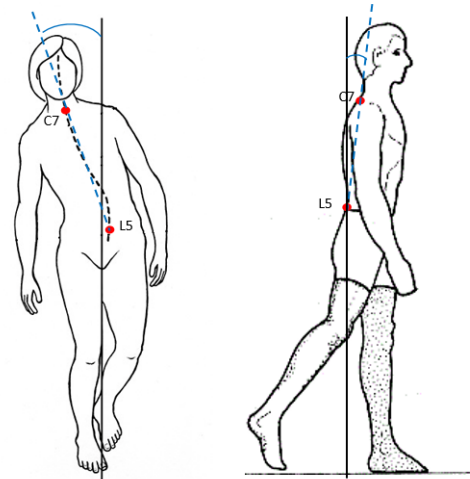


**Figure 6.37:** *Body trunk angle in the frontal (right) and sagittal (left) plane [57, 58]*

The body trunk angles in a normal walk for both constrained and not constrained conditions were compared in Figure 6.38. No changes were observed. This means that having an inversion/eversion degree of freedom may not be necessary for a walk on a completely flat ground. This confirms our analysis of section 6.3.1 on page 63.

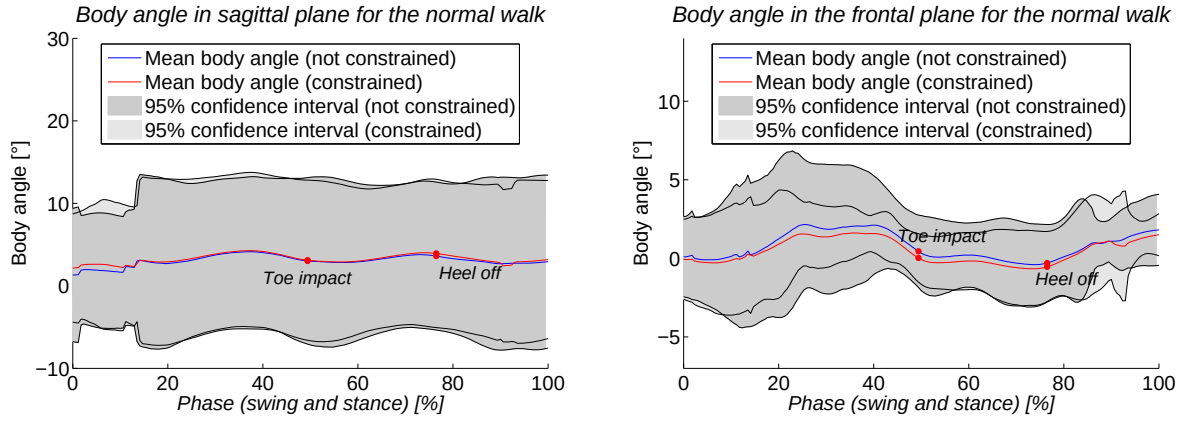


**Figure 6.38:** Evolution of the body trunk angle in the frontal (right) and sagittal (left) plane along a normal step (swing phase and stance phase), measured on 7 subjects

From the inversion/eversion movement analysis (section 6.3 on page 63), it was concluded that the turning step gait uses not very much this degree of freedom. This is confirmed by the constrained tests since no clear body trunk angle difference occurred between the constrained and not constrained cases (Figure 6.39). There is just a little higher confidence interval for the body trunk angle in the frontal plane for the constrained case.

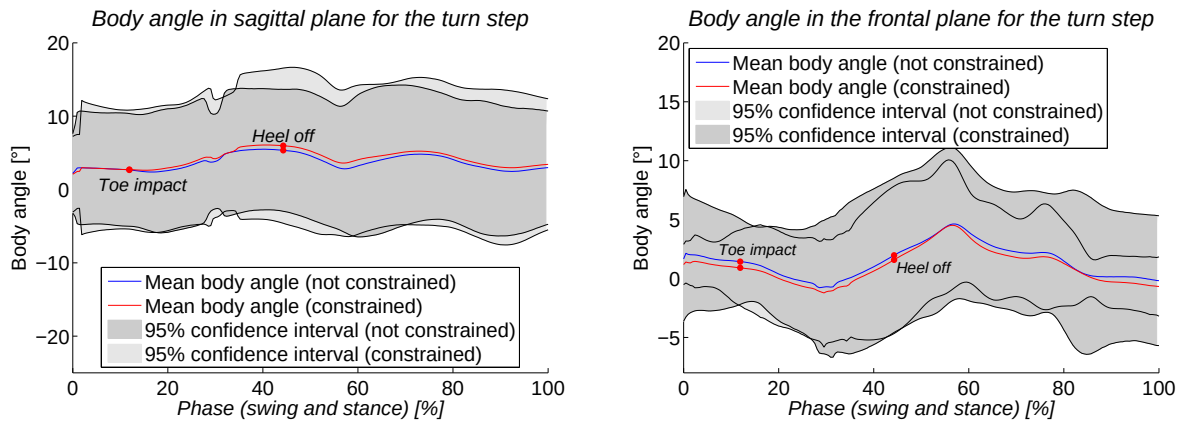


**Figure 6.39:** Evolution of the body trunk angle in the frontal (right) and sagittal (left) plane along a turning step (swing phase and stance phase), measured on 7 subjects

For the lateral step (Figure 6.40) a difference before 20% of the phase is observed. It corresponds to the adaptation of the foot to the ground. The constrained case is not really different from the not constrained case but the confidence interval is clearly higher for the first one. It means that subjects had more different behaviors with the frontal plane of the ankle locked than for the not locked case.

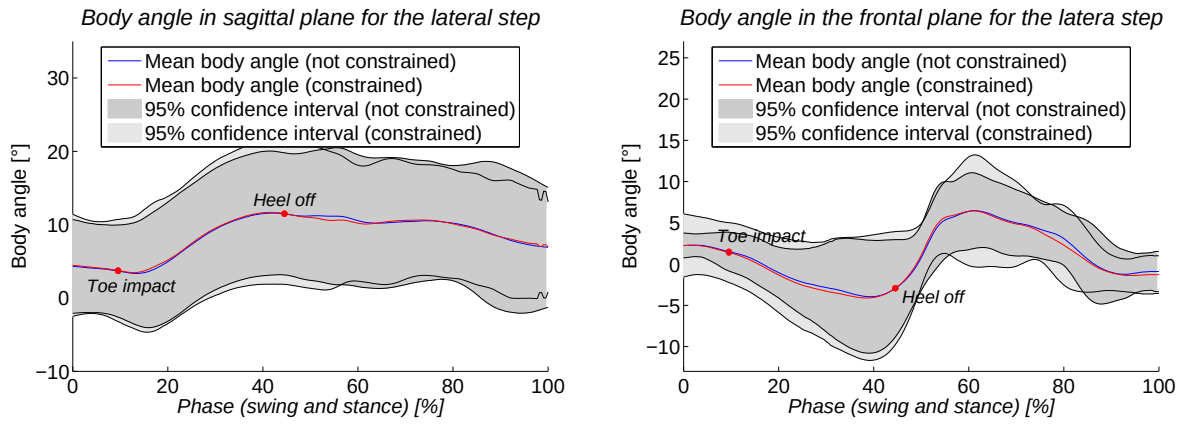


**Figure 6.40:** Evolution of the body trunk angle in the frontal (right) and sagittal (left) plane along a lateral step (swing phase and stance phase), measured on 7 subjects

Figure 6.41 shows that there is a little perturbation on the frontal angle at the end of the stance phase of the inclined plane, after the *heel off* event. Indeed, for the constrained case, the angle tends to increase. It may be considered as a small perturbation but compared to other tests, it is a not negligible perturbation (especially since it is the body trunk angle which have a standard deviation of  $1.5^\circ$  in the frontal plane [59]). Furthermore, the orthosis cannot lock entirely the inversion/eversion movement. To see a little perturbation on the body trunk angle indicates that the perturbation is in fact quite consequent.

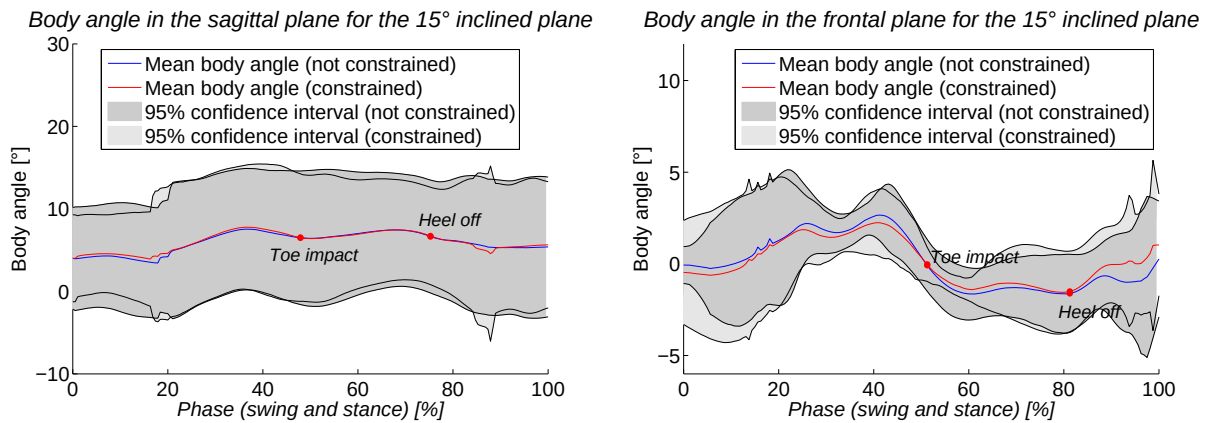


**Figure 6.41:** Evolution of the body trunk angle in the frontal (right) and sagittal (left) plane along a step on  $15^\circ$  inclined ground in eversion mode (swing phase and stance phase), measured on 7 subjects

There is the same perturbation for the 30° inclined plane than for the 15° case (Figure 6.42). But this time the confidence interval is significantly higher for the constrained case.

A similar kind of perturbation was observed for the inversion mode and for the 30° inclined plane (figures in annex I). However, it is interesting to note that the body trunk has a higher perturbation for the eversion mode than for the inversion mode. For the 15° inclined plane in inversion mode there is almost no perturbation.

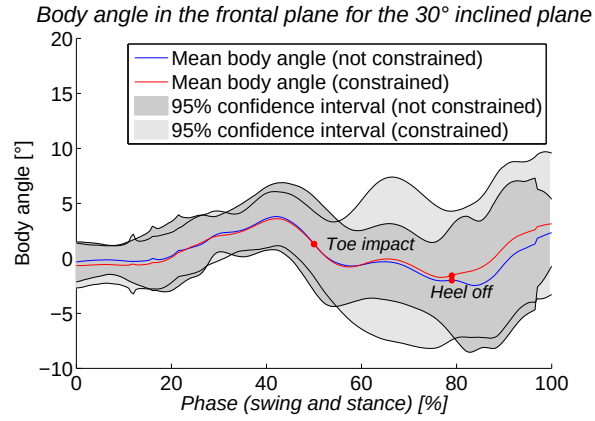


**Figure 6.42:** Evolution of the body trunk angle in the frontal plane along a step on 30° inclined ground in eversion mode (swing phase and stance phase), measured on 7 subjects

All the trunk angle perturbation generated by a constraining orthosis when walking on a 15° or 30° inclined planes are quite low. However, a few degrees of difference in the trunk angle should be seen as significant [59]. Moreover, it would have been interesting to do this angle analysis but for the arms as subjects tended to compensate a lot with their arms. Unfortunately, we did not place markers on the arms of the subjects.

### 6.4.2 Compensation of the body

When someone has an ankle prosthesis without an active degree of freedom in the sagittal plane, he has to compensate for this movement with his hip. In addition to being painful and tiresome, this compensation is not aesthetic for the person and could generate other injuries.

It is the same for the inversion/eversion degree of freedom. The subject compensates the locked movement with his body trunk (as we have seen in section 6.4.1) but mainly with his arms.

As we did not anticipate it (and it is not really the purpose of this work), we did not place markers on the arms. Consequently it will not be possible to analyze it. We did, however, observe this phenomenon experimentally.

## CHAPTER 7

# GUIDELINES FOR THE MECHANICAL DESIGN OF A TWO DEGREE OF FREEDOM ROBOTIC ANKLE PROSTHESIS

The purpose of this section is to provide some guidelines for the future design of an assistive prosthetic device targeting the eversion/inversion motion. It will not be an exhaustive list containing an advanced description of an optimal design but it will be a discussion about what design could be interesting. On the basis of our results, we will give some advice that the designer of the prosthesis should take into account in order to have the best two degrees of freedom (DoFs) robotic ankle prosthesis as possible.

## 7.1 Kinematic chain

The results show that there is a correlation between the frontal plane and the sagittal plane movements, depending on the inclination of the ground. The power needed in the frontal plane for the push-off increases when the inclination increases. For this reason, the power cannot always be in the sagittal plane of the foot if we add the inversion/eversion DOF.

It means that this degree of freedom has to be implemented below the sagittal movement in the kinematics chain (starting from the tibia).

Schema in Figure 7.1 is made in order to illustrate it. The tibia-foot articulation has been modeled as being only 2 DOFs (the sagittal plane rotation and the frontal plane rotation) which are the same as the foot prosthesis. The reader will notice the order of the rotations of these articulations in the kinematics chain.

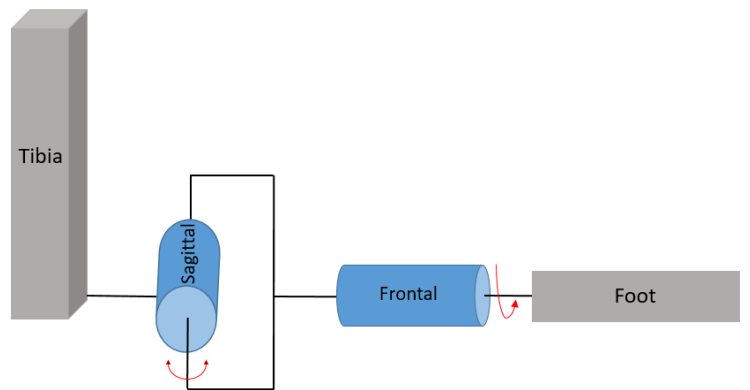


Figure 7.1: Schema to illustrate the kinematics chain

## 7.2 Mechanical design

The aim of this part is to overview some possible mechanisms (active, semi-active or passive). We will first analyze what are the benefits of an active and passive mechanism for an ankle prosthesis.

An active mechanism has the following advantages [60] :

1. Active mechanism should improve dramatically the mobility, and therefore quality of life, of people with locomotive impairments.
2. Increase self-selected gait speed while reducing metabolic expenditure.
3. Increase gait symmetry and reduce wear-and-tear on the user's unaffected joints that could otherwise arise due to compensatory movements (with the hip for instance).

A passive mechanism has the following advantages [60]:

1. No power consumption, which implies no heavy batteries. The weight is the most important factor.
2. More simple mechanism, which implies less technical incidents.
3. Low upfront costs.

We have noted in our results that the energy required by the joint in the frontal plane is much lower than the energy required in the sagittal plane for all testing cases. This observation indicates that a powerful active mechanism for the inversion/eversion movement in an ankle prosthesis is maybe not necessary, at least for users that just want to have more natural walk gaits. In fact, the increase in complexity, weight and power consumption will overcome the gain in mobility it will bring to the user.

Consequently, we will investigate the semi-active and passive mechanism. There are some advantages and drawbacks for these two options but they are both a possible solution. Section 6.3.4 (page 69) has shown that the characteristic impedance seems to be composed of three main parts for the eversion mode and is very linear for the inversion mode of the inclined plane tests. In order to behave like a healthy human ankle, the foot prosthesis have to replicate these characteristic impedances.

The three main phases for the eversion mode are:

1. **Linear part** Between the heel impact and the toe impact events, the characteristic impedance is linear.
2. **Dissipation part** Between the toe impact and the heel off events, there is a quick dissipation.
3. **Linear part** After the heel off event, the characteristic impedance is linear but it is not the same stiffness as the first part.

### 7.2.1 Passive mechanism

Thanks to the particular linearity of the characteristic impedance, we can imagine implementing the frontal plane DOF with a passive mechanism. One of the constraints being that it should be placed below the sagittal plane DOF in the kinematics chain, as explained in the section 7.1.

The mechanism should be based on spring components. The stiffness seems to be very similar between the 15° and 30° inclined ground and even for the lateral step. It encourage to think that a passive mechanism could be a good solution. Therefore, the passive mechanism would be a simple rotation controlled by spring (linear or circular). It could be a correct approximation of the real behaviors of the healthy human ankle. Unfortunately, it do not take into account the differences of stiffness between the movements and the dissipation parts. This is why a semi-active mechanism seems to be more appropriate.

### 7.2.2 Semi-active solution

If the prosthesis has already an active sagittal DOF, there is already a battery. In addition, the semi-active prosthesis does not consume a lot of energy, since it induced no power in the mechanism aside the power to control the mechanism, such as a clutch. Therefore, lighter and simpler mechanism than active system but still controlled is possible.

One of the purposes of this semi-active mechanism is to place correctly the dissipation phase of the mechanism, depending on the inclination of the ground. Our results have shown that there is a dissipation

part in the inclined plane tests in eversion mode. We can see it in figures 6.26 (page 72), 6.27 (page 73). Another advantage of a semi-active mechanism is the possibility to modify the stiffness of the frontal plane in function of the gait. In fact, the frontal DOF rarely has a non-linear characteristic impedance. However, the linear part does not always have the same slope, which means that the stiffness of the inversion/eversion movement varies between gaits. Thanks to a semi-active mechanism, we could anticipate it and adapt the stiffness to reproduce more perfectly the motion of a normal human ankle. The section 7.2.3 proposes a semi-active concept that fulfill the previous characteristics.

### 7.2.3 Concept of semi-active mechanism for frontal plane motions in two degree of freedom ankle prosthesis

This section present a semi-active principle that would be able to reproduce the healthy human ankle behavior quite accurately. It is composed of three main parts: springs, racks/worms and motors. This mechanism has two objectives: modify the impedance during the stance phase and have a quick dissipation part. The mechanism that allows the prosthesis to have a varying mechanical impedance will be described first. Then the mechanism to have a correct dissipation will be presented.

The semi-active mechanism for a varying mechanical impedance along the stance phase is depicted in Figure 7.2. The principle is to move the spring with respect to the symmetry axis with the racks shown in the schema, which means that the  $x$  variable can be modified. The following equations estimate the torque ( $\tau$ ) in the ankle with such a system.

$$\tau = xF_{Left} \cos(\beta) - xF_{Right} \cos(\beta) \quad (7.1)$$

where  $\beta$  is the inversion/eversion angle,  $F_{Left}$  is the force due to the left spring and  $F_{Right}$  is the force due to the right spring. These two forces can be given as:

$$F_{Left} = F_0 + kx \sin(\beta) \quad (7.2)$$

$$F_{Right} = F_0 - kx \sin(\beta) \quad (7.3)$$

where  $k$  is the stiffness of springs and  $F_0$  is the force due to the elongation of the spring at rest, i.e. when  $\beta = 0$ .

The torque can be rewrite as:

$$\tau = x \cos(\beta) (2kx \sin(\beta)) \quad (7.4)$$

$$= 2kx^2 \sin(\beta) \cos(\beta) \quad (7.5)$$

$$= kx^2 \sin(2\beta) \quad (7.6)$$

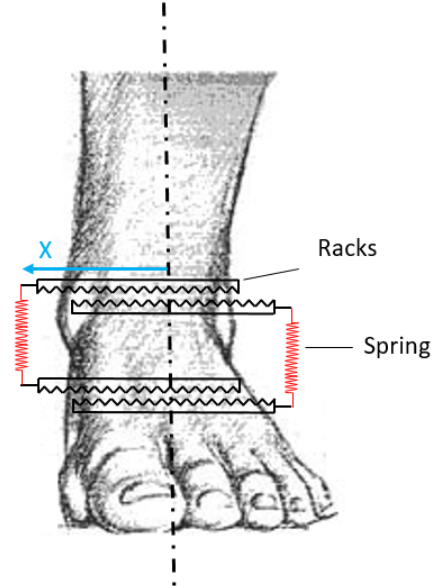


**Figure 7.2:** Semi-active mechanism for variable impedance along the stance phase [47]

Equation 7.6 shows that the torque varies with  $x^2$ . As the modification of  $x$  does not increase the elongations of the springs, the power required to modify the impedance is very low compared to the sagittal plane consumption. It is still a semi-active mechanism since no power is added in the mechanism.

An energy dissipation is similar to a spring stretched in which one of the extremity is released. All the energy contained in it is lost. The dissipative parts of the inversion/eversion movements are not entire dissipation because all the energy is not lost.

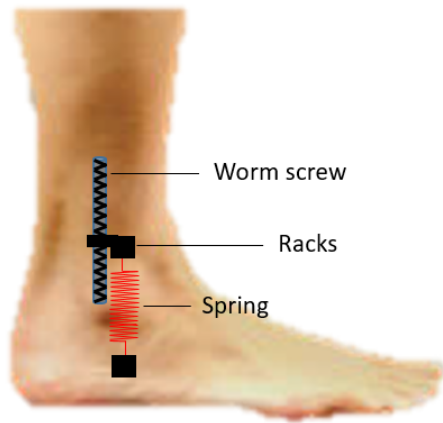


**Figure 7.3:** *Semi-active mechanism for correct dissipation part [10]*

The mechanism implementing the dissipative parts (Figure 7.3) uses the phenomenon of the released extremity spring. In order to not lost all the energy, the mechanism will just variate the extremity of the spring vertically trough a worms, which will decrease instantaneously the torque in the angle. This mechanism will be interpreted as a dissipation for the stance phase but for a consumption for the swing phase because the extremity of the spring has to reach is initial place during the swing phase. This mechanism is a kind of "series elastic actuator" and is often used in foot prosthesis [61].

The power consumption for the stance phase is very low because it is helped by the springs (the extremity of the springs goes down). However, the power consumption for the swing phase is higher because the spring has to be stretched again. This is a constraint on the choice of the spring stiffness and the  $x_0$ , i.e. the initial distance of the spring from the center (Figure 7.2).

The mechanism depicted in Figures 7.2 and 7.3 is only a principle. There are many other implementations that could achieve the characteristic of the inversion/eversion movement. However, it is a good study case to understand the behaviors that the mechanism has to fulfill. Furthermore, the semi-active mechanism for a correct dissipation part could be removed and neglected the dissipation parts of the characteristic impedance.

The previous chapter aimed to give some guidelines for the mechanical design of a two DoFs prosthesis in the light of our results. However, there are several ways to improve them or to use the data differently. Data could even be used for another purpose than the inversion/eversion movement. The following section will state a few of these possible improvements.

### 8.0.1 Limitations and perspectives

The kinematics and dynamic is composed of several assumptions that we could improve to find more reliable results. The first assumption is that the CoR of the ankle is placed constantly at the midpoint of the external and internal malleolus. To enhance this hypothesis, our model can be enhanced by using the method described in the annex K, which is based on the computation of the center of rotation (CoR) at each time step.

Even if the torque due to the inertia of the foot is really small in comparison to the torque due to the reaction forces, a better model of the foot could be used. The foot could not be considered as a box anymore but a more complicated shape to refine the inertia computations. Also, the foot could be considered as a segmented body by considering the toe joint. As the subjects had their shoes throughout the experiments, it might be difficult to do this correctly. However, we could try to enhance the model with the current data.

The data processing could be improved by the use of the solidification procedure. It consists of replacing the actual displacement of the markers triplet fixed on the same body segment by the displacement of a solid triangle associated with these same markers. These are defined by calculating the averages of the least noisy triangles determined by an iterative elimination procedure. The solidification procedure minimizes the difference between the raw trajectories of the markers and the trajectories of the markers solidified by the method of the least squares [62, 63].

It could be interesting to compute the torque in the ankle with our data but with a different method such as using *Robotran*. It could be an additional validation of our work and we could highlight some differences and hence some other limitations of our model.

The inversion/eversion motion was the main goal of this work. However, the data collected could be used in different ways, such as the transverse plane rotation.

### 8.0.2 Conclusion

Daily life activities are different than laboratory conditions. A not damaged ankle has an extraordinary compliance to the different type of environments while an ankle prosthesis is left far behind even if there is a craze in bionic foot prosthesis. There is thus a need for prosthesis improvements in order to add compliance and make life easier for amputees.

This work is focused on the inversion/eversion movement. We realized that the frontal plane is not used extensively in bionic feet prostheses which implies few available data. To respond to this lack, we decided to carry out an experimental campaign whose aim is to characterize frontal plane movements in outside conditions. This campaign was composed of two main parts: the outside characterization and the in-lab measurements. The first part allowed us to take measurements of the inversion/eversion angle of an ankle in outside condition and of the ground inclination that currently occurred in an outside normal walk. It also enables us to have a first contact with the experimental domain, to develop an electromechanical device and it provides a better understanding of the amputee problems.

During this first part, the outside conditions were characterized in order to be able to recreate similar situations inside the lab. The outside measurement device developed throughout the work was a success because the specifications needed to measure accurately the inversion/eversion angle and the ground angle, such as the precision or the frequency, were reached. This allowed us to take reliable measures of these parameters in the normal life and to perform preliminary tests which were very instructive for two reasons. The first one is that we were able to select the optimal tests to perform in-lab which were the turning step, the lateral step and the inclined plane steps. The second one, which is also very important, is that we acquired experience in the experimental domain. Thanks to this device, we were fully ready for the in-lab experiments.

The experimental campaign has been successfully completed too. Even if the data post-processing of all experiments has taken a lot of time, the validation of our model (section 6.2 on page 57) shows that our results are reliable. Some important behaviors of the ankle in the frontal plane have been highlighted. The three main results are the linear and dissipative parts of the characteristic impedance, the low power consumption of the inversion/eversion movement and the correlation between the sagittal plane and the frontal plane depending on the inclination of the ground. We have shown the added value of this DOF thanks to the body trunk angle but also with observations of subjects that perform the in-lab tests.

Because of the time spent in the data collection, we were not able to investigate the mechanical design with details. However, some important guidelines were presented for future work, i.e. the design of the 2 DoFs bionic ankle prosthesis. The need to place the frontal degree of freedom below the sagittal degree of freedom in order to have a sagittal power added in the movement in the sagittal plane of the body and not of the foot anymore was highlighted. The linear and dissipation parts found in the results is used to think about a mechanism that use spring and modify the impedance passively or semi-actively.

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## APPENDIX A

### INTERVIEW OF A TRANSFERMORAL AMPUTEE

We questioned a transfemoral amputee to have the opinion of someone that personally lives with a foot prosthesis every day. He was able to help us highlighting some problems that we would not have thought about otherwise. He have already try different kinds of prosthesis, from the Energy-Storing-and-Returning (ESR) to advanced bionic foot prosthesis. When we meet him he has an ESR one. We will summarize here the questions asked and his answers.

#### ***What is the greatest difficulty with your current prosthesis?***

The biggest difficulty is the adaptation of the foot. There are two kinds of foot adaptation : while the swing phase and while the stance phase.

The problem if the foot does not adapt itself while the swing phase is that the foot has not a great enough ground clearance. It means that the foot could touch the ground and destabilize me. This implies that I will compensate with my hip to raise my foot out of the ground.

Consequently, I could fall or develop hip pain, and I walk very slowly.

When the foot does not adapt correctly to the ground, there is two kinds of difficulties.

The first one is that I can lose my stability because the foot is not entirely in contact with the ground anymore. This problem occurs when the ground is not entirely flat (on a paved road or on a descent or ascent).

The second issue occurs when I walk in the direction of a downhill or uphill slope. As the foot does not adapt to the ground, the contact between them is limited to the heel (downhill) or to the toes (uphill). Therefore, I have to walk with really small steps in downhill slopes and I consume a lot of energy to propel my body over the standing foot in uphill slopes.

#### ***What kind of ground is the most difficult for you to walk on?***

The most difficult grounds for the stability are the trails and paved roads.

#### ***What could be the advantage of a degree of freedom allowing movements in the frontal plane of the prosthesis?***

The main advantage could be an improvement in stability. I imagine that an adaptation of the foot in the frontal plane would allow me to walk on trails or paved roads more easily. The risk of falling could be really decreased.

The drawback of this modification is the weight. If the system is too heavy, it could be more binding than anything else.

*In the daily life, do you think that the degree of freedom allowing movements in the frontal plane would be useful?*

If the ground is completely flat, it would not be useful. But you cannot imagine how the ground can be inclined. The sidewalks are often inclined toward the road. This means that walking in a town can sometimes be very difficult. The gutters are also a problem.

If the system is not too heavy or expensive, it could be a real improvement, even if it does not decrease energy consumption.

## APPENDIX B

### CONNECTION OF THE ELECTRONIC PART

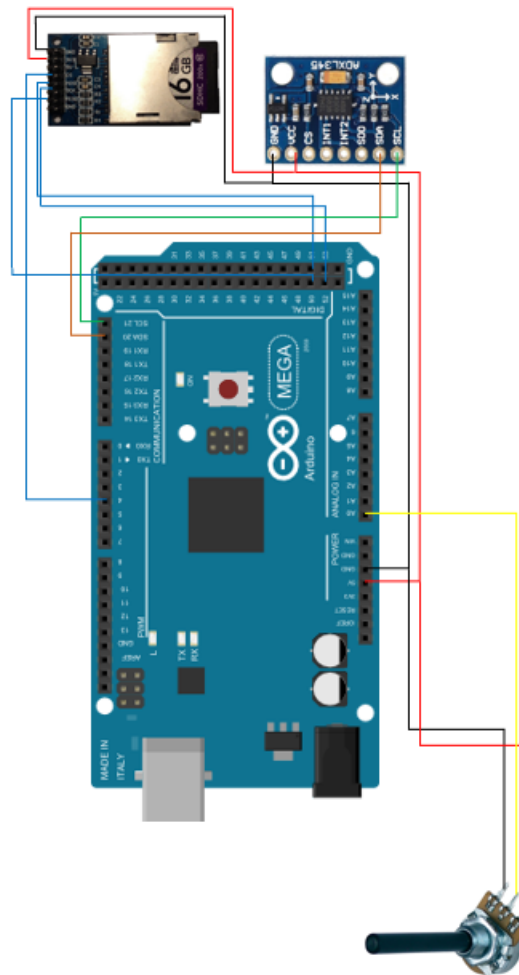


Figure B.1: *Connection of the electronic part* [64]



## APPENDIX C

### MATLAB CODE OF THE POST PROCESSING DATA COLLECTED BY THE MEASUREMENT DEVICE

```
1 function [Tn, An, Pitch, Roll] = DetectStep(T, A, ax, ay, az, error, nb, other, norme)
2 consecutive = 8;
3 Tn = zeros(length(T), nb);
4 An = zeros(length(A), nb);
5 Roll = zeros(length(A), nb);
6 Pitch = zeros(length(A), nb);
7 Tn = FillWithOthers(Tn, length(Tn), nb, other);
8 An = FillWithOthers(An, length(An), nb, other);
9 Roll = FillWithOthers(Roll, length(An), nb, other);
10 Pitch = FillWithOthers(Pitch, length(An), nb, other);
11
12 for i = 1:nb
13     count = 1;
14     for j = 1:length(T(:,i))
15         if(j < length(T(:,i))-consecutive)
16             echec = 0;
17             for k = 0:consecutive
18                 if(norme(j+k, i) > 1+error || norme(j+k, i) < 1-error)
19                     echec = 1;
20                     break;
21             end
22         end
23         if(echec == 0)
24             Tn(count,i) = T(j,i);
25             An(count,i) = A(j,i);
26             Roll(count, i) = (atan2(az(j,i), (sqrt(ax(j,i)^2+ay(j,i)^2))) * 180.0) / pi;
27             Pitch(count,i) = (atan2(ax(j,i), sqrt(ay(j,i)^2+az(j,i)^2)) * 180.0) / pi;
28             count = count + 1;
29         end
30     end
31 end
32 Tn = RemoverOthers(Tn, length(Tn(:,i)), other, i);
33 An = RemoverOthers(An, length(Tn(:,i)), other, i);
34 Roll = RemoverOthers(Roll, length(Tn(:,i)), other, i);
35 Pitch = RemoverOthers(Pitch, length(Tn(:,i)), other, i);
36 end
```



## APPENDIX D

### MATLAB CODE TO REMOVE OUTLIERS

```

1 function Outliers(postFile, fileToWrite)
2 % - postFile is the real file to convert (inclined15_1, slalom, etc)
3 % - fileToWrite is the file name and direction in which the btk file without
4 %   outliers have to be written.
5 global glob
6 s = strcat('Subjects/Subject', num2str(glob.data.sujet));
7 s = strcat(s, '/');
8 s = strcat(s, postFile);
9 s = s(1:end-4);
10 s = strcat(s, '.mdf');
11 fileName = s;
12 acq = btkReadAcquisition(fileName);
13 markers = btkGetMarkers(acq);
14 L = length(markers.R_Heel(:, 1));
15 see = 10; degrees = 1; iterations = 1;
16 T = struct2cell(markers);
17 SavedMarkers = zeros(L, 3*length(T));
18 fprintf('\n\nmarker 1/%2.0f\n', length(T));
19 for k = 1:length(T)
20     for i = 1:3
21         count = 0;
22         E = zeros(L, 1);
23         v = zeros(L, 1);
24         fct = T{k}(:, i)';
25         x = linspace(0, 100, length(fct));
26         for j = 1:length(x)
27             Variance = 0;
28             if(j > see && j < length(x)-see)
29                 f = polyval(polyfit(x(j-see:j+see), fct(j-see:j+see), degrees), ...
30                     x(j-see:j+see));
31                 v(j) = f(see+1);
32                 Variance = mean(abs(fct(j-see:j+see)-f).^2);
33             else
34                 v(j) = 0;
35             end
36             if(j > 4 && j < L-4)
37                 if(abs(fct(j)-v(j, 1)) > 3*sqrt(Variance))
38                     E(j, 1) = v(j, 1);
39                     fct(j) = v(j, 1);
40                     count = count + 1;
41                 else
42                     E(j, 1) = fct(j);
43                 end
44             end
45             if(mod(i-1, 3) == 0 && k≠1)
46                 fprintf('\n\nmarker %2.0f /%2.0f\n', k, length(T));
47             end
48             fprintf('They had %3.0f outliers\n', count);
49             T{k}(:, i) = E;
50             SavedMarkers(:, iterations) = T{k}(:, i);
51             iterations = iterations + 1;
52         end
53     end
54     btkSetMarkersValues(acq, SavedMarkers(:, 1:end));
55     btkWriteAcquisition(acq, fileToWrite);
56 end

```



## APPENDIX E

---

### MATLAB CODE TO DETECT FOOT CONTACTS

```
1 function ContactDetection()
2 global glob
3 iterMax = 10;
4 done = [0 0];
5
6 if(glob.data.sideStep == 1 || glob.data.sideStep == 3)
7     e = glob.data.sync1:glob.data.sync;
8 else
9     e = glob.data.sync:glob.data.sync2;
10 end
11 R_Heel = glob.markers.R_Heel(e, :);
12 R_Toe = glob.markers.R_Metal(e, :);
13
14 HeelMean = mean(R_Heel(:, 3));
15 ToeMean = mean(R_Toe(:, 3));
16
17 glob.data.heelOff = 1+e(1);
18 glob.data.toeIn = 1+e(1);
19 factor = 1.2;
20 for j = 1:iterMax
21     oldToe = glob.data.toeIn;
22     oldHeel = glob.data.heelOff;
23     for i = 1:length(R_Heel(:, 1))
24         if(R_Heel(i, 3) > factor*HeelMean && done(1) == 0)
25             glob.data.heelOff = i+e(1);
26             done(1) = 1;
27         end
28         if(R_Toe(i, 3) < factor*ToeMean && done(2) == 0)
29             glob.data.toeIn = i+e(1);
30             done(2) = 1;
31         end
32     end
33     HeelMean = mean(R_Heel(1:glob.data.heelOff-e(1), 3));
34     ToeMean = mean(R_Toe(glob.data.toeIn-e(1):end, 3));
35     if(glob.data.heelOff == oldHeel && glob.data.toeIn == oldToe)
36         break;
37     end
38     done = [0 0];
39 end
40 end
```



## APPENDIX F FILTER SIZING

The purpose of this part is to chose a correct size for the moving average filter. All the method described in this section comes from the internet [65]. We will find the magnitude of the frequency response as a function of the size of the filter (parameter N).

A moving average filter has a rectangular impulse response, which is stated as:

$$h[n] = \frac{1}{N} \sum_{k=0}^{N-1} \delta[n - k] \quad (\text{F.1})$$

The discrete-time Fourier transform of its impulse response can be computed as follows:

$$H(w) = \sum_{n=-\infty}^{\infty} x[n]e^{-jwn} \quad (\text{F.2})$$

$$= \frac{1}{N} \sum_{n=0}^{N-1} e^{-jwn} \quad (\text{F.3})$$

Thanks to the series relation  $\left(\sum_{k=1}^n ar^{k-1} = \frac{a(1-r^n)}{1-r}\right)$ , we get:

$$H(w) = \frac{1}{N} \frac{1 - e^{-jwN}}{1 - e^{-jw}} \quad (\text{F.4})$$

$$= \frac{1}{N} \frac{e^{-jw\frac{N}{2}}}{e^{-\frac{jw}{2}}} \frac{e^{jw\frac{N}{2}} - e^{-jw\frac{N}{2}}}{e^{\frac{jw}{2}} - e^{-\frac{jw}{2}}} \quad (\text{F.5})$$

$$= \frac{1}{N} \frac{e^{-jw\frac{N}{2}}}{e^{-\frac{jw}{2}}} \frac{\sin\left(\frac{wN}{2}\right)}{\sin\left(\frac{w}{2}\right)} \quad (\text{F.6})$$

As it is the magnitude of the frequency response ( $|H(w)|$ ) is needed and that the exponential terms does just affect the phase ( $|e^{jw}| = 1$  for all frequency), we get :

$$|H(w)| = \frac{1}{N} \left| \frac{\sin\left(\frac{wN}{2}\right)}{\sin\left(\frac{w}{2}\right)} \right| \quad (\text{F.7})$$

When this frequency response is plotted, we have the figure [F.1](#). We want -6dB for the cutting frequency and we have the relation  $20 \log(H(w)) = \text{Gain}[dB]$ . Hence, we need  $|H(w)| = 0.5$ . Therefore, we chose  $N = 5$  to have 30Hz as cutting frequency for the in-lab experiment.

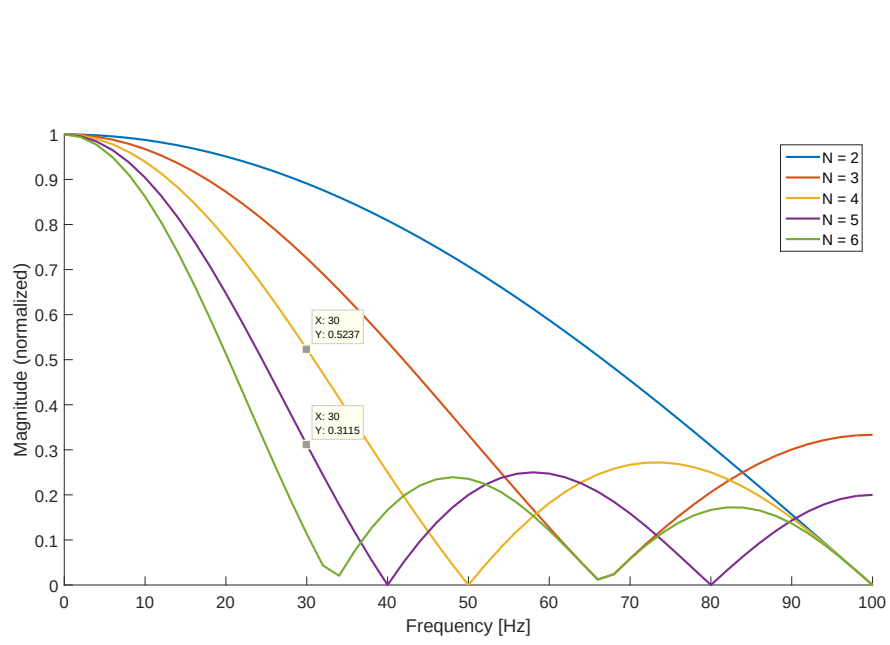
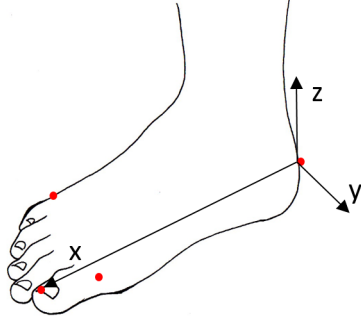


**Figure F.1:** *Magnitude of the frequency response of a moving average filter in function of the frequency*

## APPENDIX G

### FOOT FRAMEWORK COMPUTATION

The inertial frame of some markers has to be changed by the foot frame in order to compute properly the torques in the ankle. It means that we have to compute the x, y and z components of some markers on another reference frame and not on the absolute frame anymore.



This new frame is the foot frame. More precisely, the origin of the new frame is the heel of the foot, the x direction is the direction of the toe, the y direction is towards the left of the foot and finally the z direction is upward.

To change the framework, we need to take into account two types of movements: the translation and the rotation.

**Figure G.1:** Foot frame with the heel as origin, the x axis in the toe direction and the y axis toward the right of the foot [50]

Concerning the translation, we just have to subtract the x component of the heel from the x component of the markers that have to change of framework, the y component of the heel from the y component of the marker and the same for the z component.

The rotation is a little more complicated. We will have to construct the matrices at each time step. In section 5.3.2 we have explained how to compute the angle of inversion/eversion. In the same way, the  $\alpha$ ,  $\beta$  and  $\gamma$  angles are found, which are the angles around the z, y and x axis of the reference frame respectively.

With these angles, the following matrices are computed for each time step:

$$Rot_x = \begin{pmatrix} 1 & 0 & 0 \\ 0 & \cos(\gamma) & \sin(\gamma) \\ 0 & -\sin(\gamma) & \cos(\gamma) \end{pmatrix} \quad (G.1)$$

$$Rot_y = \begin{pmatrix} \cos(\beta) & 0 & \sin(\beta) \\ 0 & 1 & 0 \\ -\sin(\beta) & 0 & \cos(\beta) \end{pmatrix} \quad (G.2)$$

$$Rot_z = \begin{pmatrix} \cos(\alpha) & \sin(\alpha) & 0 \\ -\sin(\alpha) & \cos(\alpha) & 0 \\ 0 & 0 & 1 \end{pmatrix} \quad (G.3)$$

All we have to do now is to apply these matrices to the markers that have to change the framework. So we have the following relation:

$$Marker2 = Rot_x \cdot Rot_y \cdot Rot_z \cdot \begin{pmatrix} Marker_x - Heel_x \\ Marker_y - Heel_y \\ Marker_z - Heel_z \end{pmatrix} \quad (G.4)$$

where *Marker* is the marker we want to put in the frame of the foot and *Marker2* is the *Marker* in the frame of the foot.

## APPENDIX H

### COMPARISON OF OUR RESULTS AND THE LITERATURE FOR THE SAGITTAL PLANE IN A NORMAL WALK

Figures H.1, H.2 and H.3 present our result for the sagittal movement during the stance phase of a normal walk. The plantar flexion/dorsiflexion movement in the literature is very similar to our results, excepted the torque that is a little bit different between the literature and our results.

Indeed, as presented Figure H.3, the torque computed on the basis of our data is lower. This is probably due to the fact that the subject has to walk on the two force platforms, which are a little bit too close to each other. It implies that the subject has to perform a step slightly shorter than his natural step.

Intuitively, at most the step is short compared to the natural step, at most the torque in the sagittal plane is low.

Comparison between litterature and results for the plantarflexion/dorsiflexion angle in normal walk (sagittal plane)

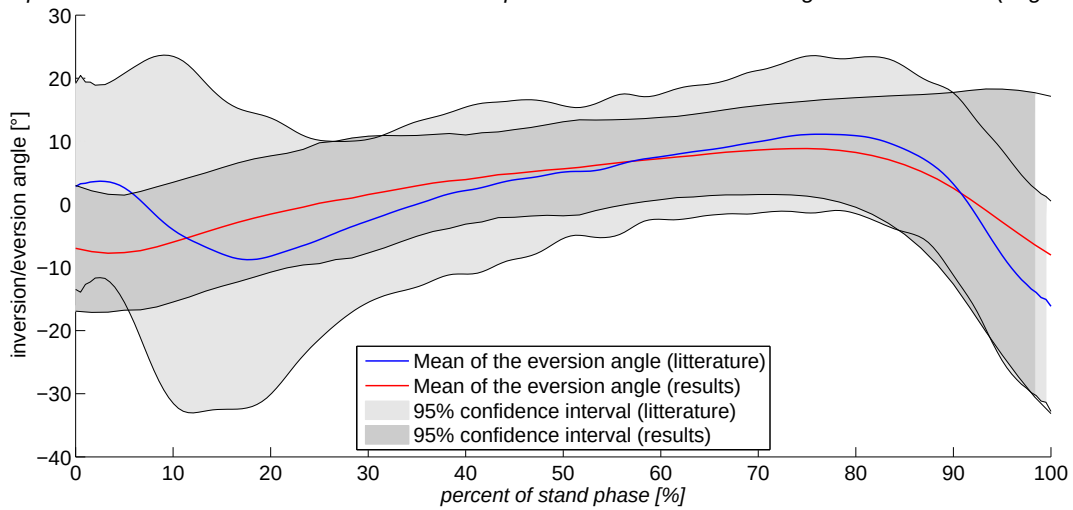


**Figure H.1:** Evolution of the plantar flexion/dorsiflexion angle during the stance phase of a normal walk, measured on 8 subjects in the present study (red line) and reported in the literature [7] (blue line)

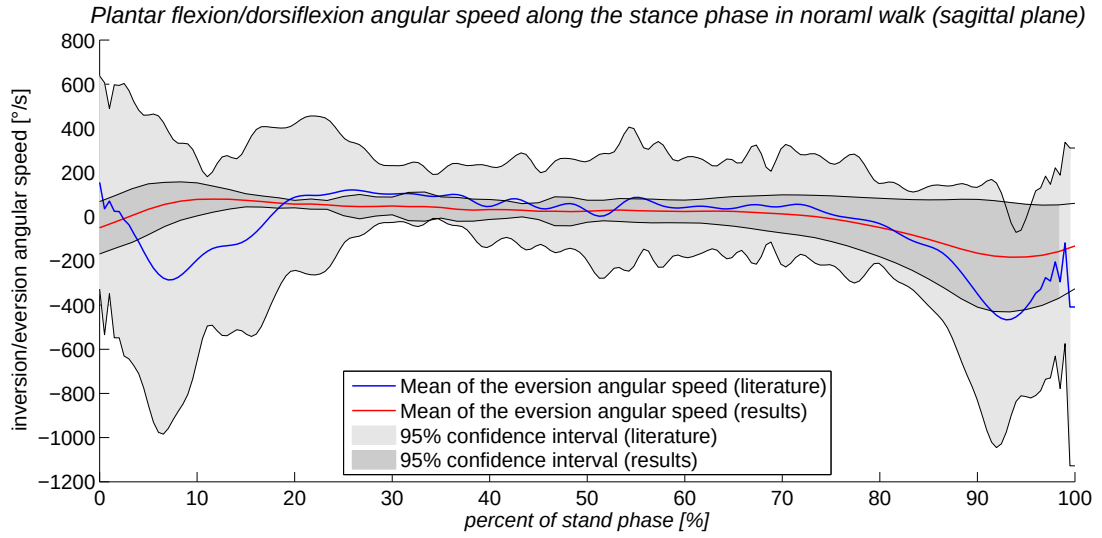


**Figure H.2:** Evolution of the plantar flexion/dorsiflexion angular speed during the stance phase of a normal walk, measured on 8 subjects in the present study (red line) and reported in the literature [7] (blue line)

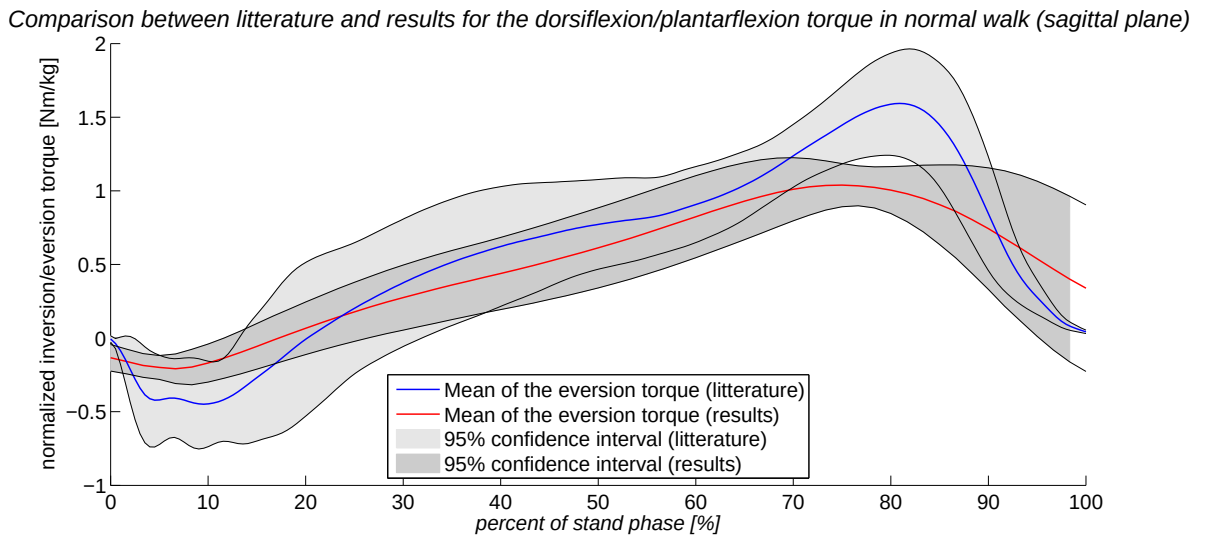


**Figure H.3:** Evolution of the plantar flexion/dorsiflexion normalized torque during the stance phase of a normal walk, measured on 8 subjects in the present study (red line) and reported in the literature [7] (blue line)

Finally, the energy consumption is compared (considering a 75Kg person). As depicted in Figure 6.4, our results are much more concentrated and are around the median of the literature results. The mean of the energy consumption (not shown in the figure) is 10.8996 [J] for literature and 8.9276 [J] for our results. Taking account of this and the fact that the means are very close each other, we can say again that the energy consumption for a normal walk from our data in the sagittal plane corresponds entirely to literature.

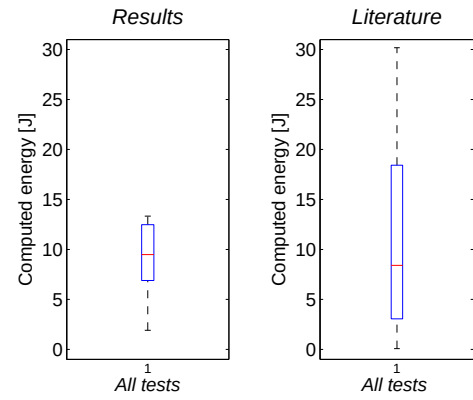


**Figure H.4:** *Energy consumption of the plantar flexion/dorsiflexion movement during the stance phase of a normal work for our results (left) and the literature [7] (right). The horizontal red line is the median, the edges of the box are the 25th and 75th percentiles and the black lines extend to the most extreme data point. This means that 50% of the data are inside the blue box and 100% data that are not considered as an outlier are inside the two extreme horizontal black lines.*



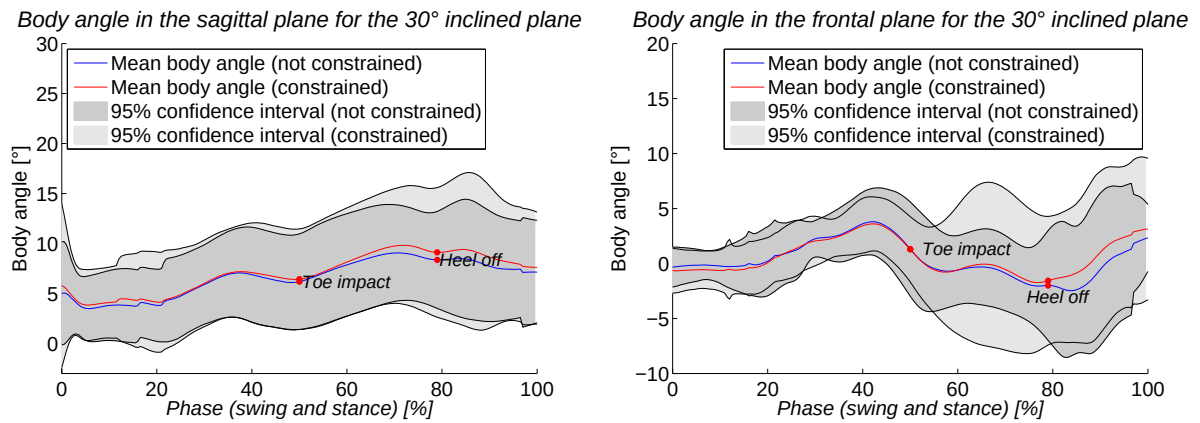


**Figure I.1:** Body trunk angles in sagittal plane (left) and frontal plane (right) for a 30° inclined plane (eversion mode)

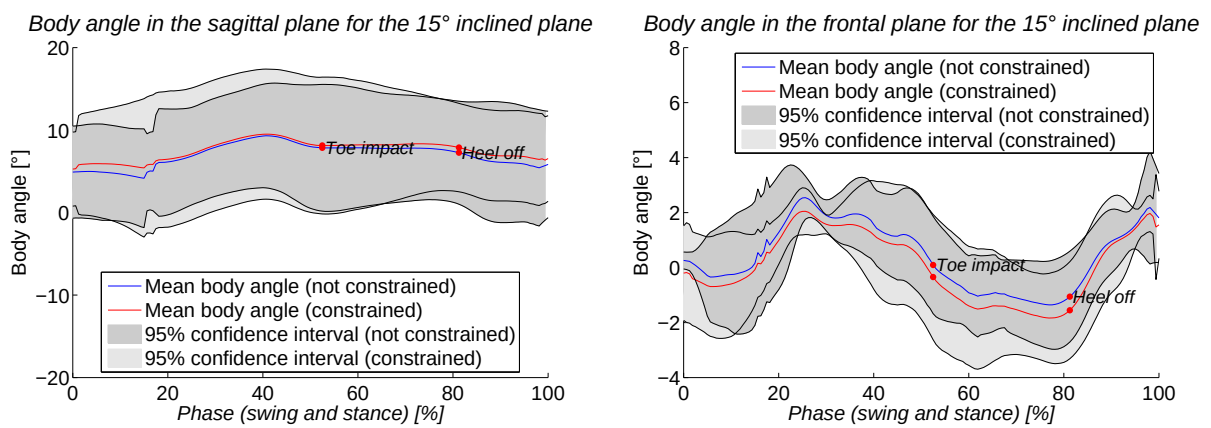


**Figure I.2:** Body trunk angles in sagittal plane (left) and frontal plane (right) for a 15° inclined plane (inversion mode)

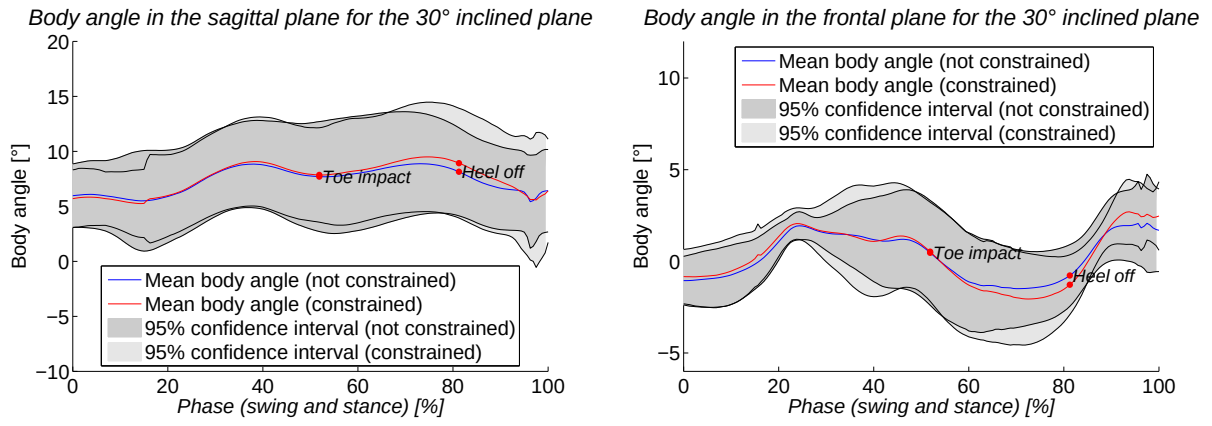


**Figure I.3:** *Body trunk angles in sagittal plane (left) and frontal plane (right) for a 30° inclined plane (inversion mode)*

## APPENDIX J

### ACQUISITION OF CONTACT FORCES

The contact forces of the foot to the ground have been collected thanks to two 3D force platform *Kistler* (size: 60 x 40 cm). The frequency of the force plate is 1000Hz and the frequency of the optoelectronic system is 100Hz. In order to switch from 1000Hz to 100Hz, we have interpolate the 1000Hz data at each time step of the 100Hz data.

The forces are computed in the center of each platform and then bring back to the point where the torque is zero in their plane. We have the torque  $\mathbf{M}_0$  [62]. So we have the following torsor for the forces and the torques :

$$\mathbf{F} = [f_x \quad f_y \quad f_z]^T$$

$$\mathbf{M} = [M_x \quad M_y \quad M_z]^T$$

where  $f_i$  are the resulting forces in the i direction and  $M_i$  are the torque around the i direction.

The third parameters computed by the force platform and that is really important for the dynamic is the forces point of application. It have been computed on the bases of the forces and torques measurement with respect to the origin of the force platform.

The origin is assumed to be  $[a \quad b \quad c]$  and is known thanks to the datasheet and the calibration of the optoelectronic system.

So, to find the application point of the ground reaction forces, we can use the origin and the measured forces and torques with the following system :

$$\begin{cases} x = -\frac{M_y + cf_x}{f_z} \\ y = \frac{M_x - cf_y}{f_z} \\ M_0 = M_z - xf_y + yf_x \end{cases}$$



## APPENDIX K

### NEW METHOD TO ESTIMATE THE AXIS OF ROTATION OF THE ANKLE

In this section we will present a way to improve our dynamic model in a further work. As this method is based on the computation of the center of rotation (CoR) at each time step, this method could be really more precise than the assumption we have done.

This method estimates the parameters of ball joints and hinge joints with a fixed axis of rotation (AoR) [55]. It is a least squares solution using all the data and without assuming rigidity but only that the markers keep the same distance from the CoR or AoR. We will here explain how to estimate the AoR.

The notation  $v_k^p$  represents the  $p$ th vector in the  $k$ th time step. As visible in figure K.1, the tips of the position vectors  $vv$  are on circles with their centers on a straight line (the rotational axis). We can have the following least squares cost function for  $P$  markers and  $N$  time step :

$$C = \sum_{p=1}^P \sum_{k=1}^N [(v_k^p - m^p) \cdot n]^2 \quad (\text{K.1})$$

where  $m^p$  is the whole points traced by the  $p$ th position vector and  $n$  the unit vector in the rotational axis direction.

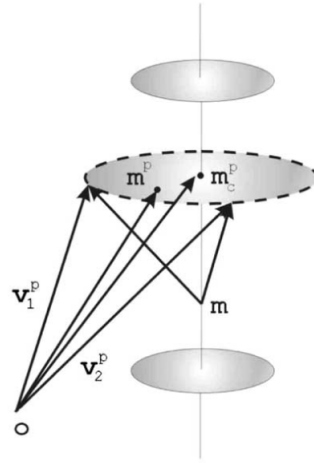


**Figure K.1:** Circular marker movement for a hinge joint [55]

The next step of the method is to differentiate the cost function (equation K.1 with respect to  $n$  and to  $m_p$  and equal the result to zero. We obtain :

$$\frac{\delta C}{\delta n} = 0 \quad (\text{K.2})$$

$$\sum_{p=1}^P \sum_{k=1}^N [(v_k^p - m^p) \cdot n] (v_k^p - m^p) = 0 \quad (\text{K.3})$$

$$\frac{\delta C}{\delta m_p} = 0 \quad (\text{K.4})$$

$$N \sum_{k=1}^N m^p n = \sum_{p=1}^P \sum_{k=1}^K v_k^p \cdot n \quad (\text{K.5})$$

$$m^p \cdot n = \left( \frac{1}{N} \sum_{k=1}^N v_k^p \right) \cdot n = \bar{v}^p \cdot n \quad (\text{K.6})$$

If we substitute  $m^p \cdot n$  from equation K.6 in equation K.3 we get the following equation :

$$\sum_{p=1}^P \sum_{k=1}^N (v_k^p \cdot n - \bar{v}^p \cdot n) v_k^p = 0 \quad (\text{K.7})$$

which can be write as<sup>1</sup> :

$$\sum_{p=1}^P \left[ \left\{ \sum_{k=1}^N v_k^p (v_k^p)^T \right\} - N \bar{v}^p (\bar{v}^p)^T \right] n = 0 \quad (\text{K.8})$$

$$\sum_{p=1}^P \left[ \left\{ \frac{1}{N} \sum_{k=1}^N v_k^p (v_k^p)^T \right\} - \bar{v}^p (\bar{v}^p)^T \right] n = 0 \quad (\text{K.9})$$

In order to define the axis of rotation, let take m a point on the axis and  $r^p$  the distance from m to the circular arc. We can form the following least squares cost function :

$$C = \sum_{p=1}^P \sum_{k=1}^N \left[ (v_k^p - m)^2 - (r^p)^2 \right]^2 \quad (\text{K.10})$$

Finally, the equation of the axis of rotation can be written as :

$$x_{AoR} = m + \tau n \quad (\text{K.11})$$

which means that it is a straight line passing trough m in the direction of h. This equation is parameterized by the scalar  $\tau$ . We are also able to find the centres of the circles of each marker.

This method, presented in [55], should be used to improve the precision of the model.

---

<sup>1</sup>Thanks to the matrix form :  $a(b \cdot c) = ab^T c$

MOKKA VIEW OF THE FOOT AND THE TIBIA DURING A 30°  
INCLINDE PLANE TEST

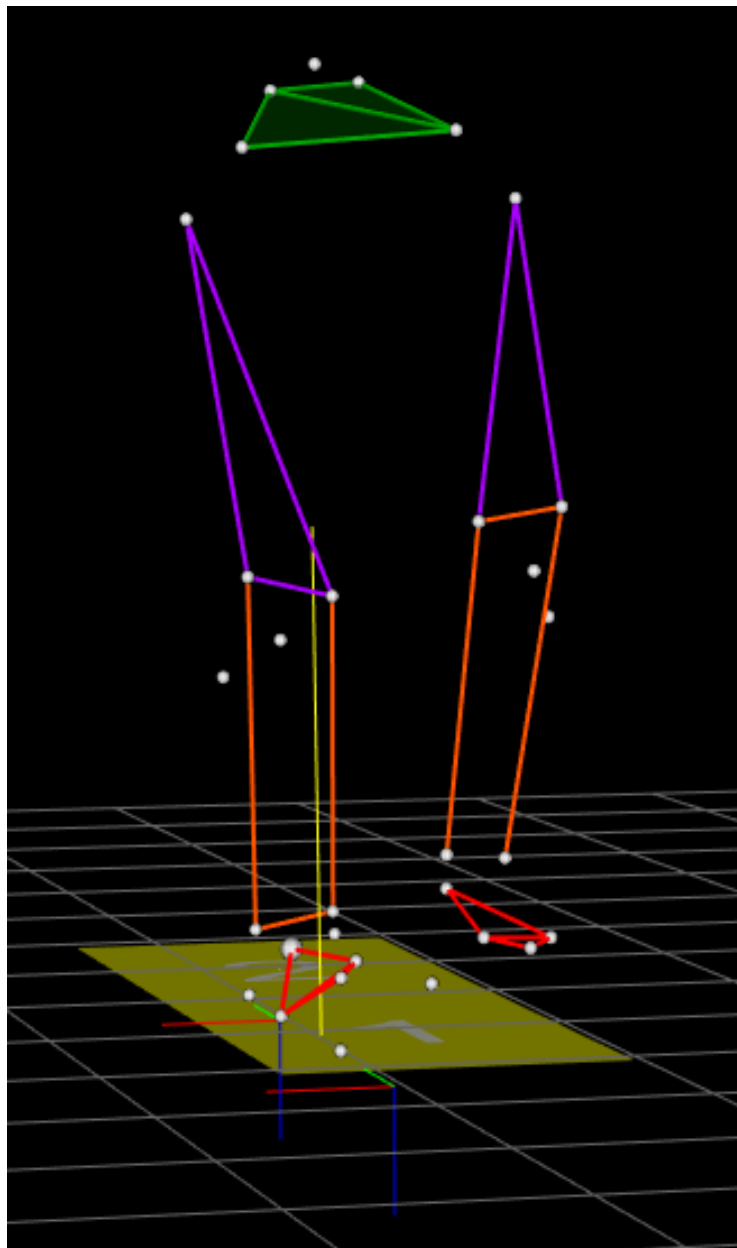


**Figure L.1:** *Mokka view of the foot and tibia during a 30° inclined plane test*



# APPENDIX M

---

## DATASHEET OF THE ADXL345 ACCELEROMETER



# Three-Axis, $\pm 2/4/8/16g$ Digital Accelerometer

Preliminary Technical Data

**ADXL345**

## FEATURES

**Ultra low power:** 25 to 130  $\mu A$  at  $V_s = 2.5 V$  (typ)  
**Power consumption scales automatically with bandwidth**  
**User selectable fixed 10-bit resolution or 4mg/LSB scale factor in all  $g$ -ranges, up to 13-bit resolution at  $\pm 16 g$**   
**32 level output data FIFO minimizes host processor load**  
**Built in motion detection functions**

- Tap/Double Tap detection
- Activity/Inactivity monitoring
- Free-Fall detection

**Supply and I/O voltage range:** 1.8 V to 3.6 V  
**SPI (3 and 4 wire) and I<sup>2</sup>C digital interfaces**  
**Flexible interrupt modes – Any interrupt mappable to either interrupt pin**  
**Measurement ranges selectable via serial command**  
**Bandwidth selectable via serial command**  
**Wide temperature range ( $-40$  to  $+85^{\circ}C$ )**  
**10,000  $g$  shock survival**  
**Pb free/RoHS compliant**  
**Small and thin:**  $3 \times 5 \times 1$  mm LGA package

## APPLICATIONS

Handsets  
Gaming and pointing devices  
Personal navigation devices  
HDD protection  
Fitness equipment  
Digital cameras

## GENERAL DESCRIPTION

The ADXL345 is a small, thin, low power, three-axis accelerometer with high resolution (13-bit) measurement up to  $\pm 16 g$ . Digital output data is formatted as 16-bit twos complement and is accessible through either a SPI (3- or 4-wire) or I<sup>2</sup>C digital interface.

The ADXL345 is well suited for mobile device applications. It measures the static acceleration of gravity in tilt-sensing applications, as well as dynamic acceleration resulting from motion or shock. Its high resolution (4mg/LSB) enables resolution of inclination changes of as little as  $0.25^{\circ}$ .

Several special sensing functions are provided. Activity and inactivity sensing detect the presence or lack of motion and if the acceleration on any axis exceeds a user-set level. Tap sensing detects single and double taps. Free-Fall sensing detects if the device is falling. These functions can be mapped to interrupt output pins. An integrated 32 level FIFO can be used to store data to minimize host processor intervention.

Low power modes enable intelligent motion-based power management with threshold sensing and active acceleration measurement at extremely low power dissipation.

The ADXL345 is supplied in a small, thin  $3 \text{ mm} \times 5 \text{ mm} \times 1 \text{ mm}$ , 14-lead, plastic package.

## FUNCTIONAL BLOCK DIAGRAM

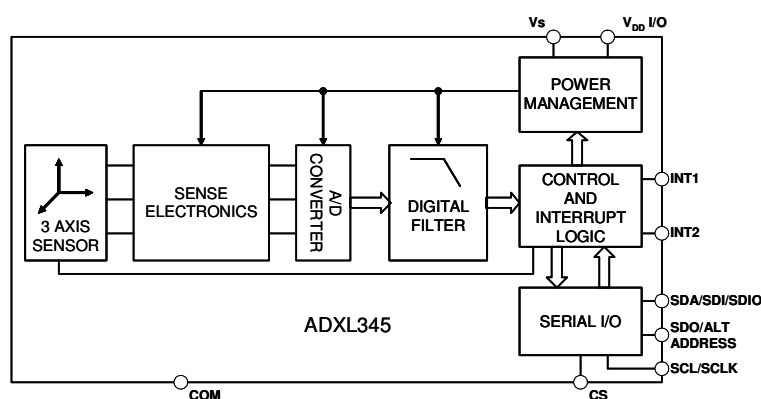


Figure 1. ADXL345 Simplified Block Diagram

## Rev. PrA

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## REVISION HISTORY

11/08—Rev. PrA - Initial Version

## SPECIFICATIONS

$T_A = 25^\circ\text{C}$ ,  $V_S = 2.5\text{ V}$ ,  $V_{DD I/O} = 1.8\text{ V}$ , Acceleration = 0 g, unless otherwise noted.

Table 1. Specifications<sup>1</sup>

| Parameter                                   | Conditions                                                     | Min   | Typ               | Max   | Unit                 |
|---------------------------------------------|----------------------------------------------------------------|-------|-------------------|-------|----------------------|
| SENSOR INPUT                                | Each axis                                                      |       |                   |       |                      |
| Measurement Range                           | User Selectable                                                |       | $\pm 2, 4, 8, 16$ |       | g                    |
| Nonlinearity                                | Percentage of full scale                                       |       | $\pm 0.5$         |       | %                    |
| Inter-Axis Alignment Error                  |                                                                |       | $\pm 0.1$         |       | Degrees              |
| Cross-Axis Sensitivity <sup>2</sup>         |                                                                |       | $\pm 1$           |       | %                    |
| OUTPUT RESOLUTION                           | Each axis                                                      |       |                   |       |                      |
| All g-ranges                                | 10-bit mode                                                    |       | 10                |       | Bits                 |
| $\pm 2\text{ g}$ range                      | Full-Resolution                                                |       | 10                |       | Bits                 |
| $\pm 4\text{ g}$ range                      | Full-Resolution                                                |       | 11                |       | Bits                 |
| $\pm 8\text{ g}$ range                      | Full-Resolution                                                |       | 12                |       | Bits                 |
| $\pm 16\text{ g}$ range                     | Full-Resolution                                                |       | 13                |       | Bits                 |
| SENSITIVITY                                 | Each axis                                                      |       |                   |       |                      |
| Sensitivity at $X_{OUT}, Y_{OUT}, Z_{OUT}$  | $V_S = 2.5\text{ V}, \pm 2\text{ g}$ 10-bit or Full-Resolution | 232   | 256               | 286   | LSB/g                |
| Scale Factor at $X_{OUT}, Y_{OUT}, Z_{OUT}$ | $V_S = 2.5\text{ V}, \pm 2\text{ g}$ 10-bit or Full-Resolution | 3.5   | 3.9               | 4.3   | mg/LSB               |
| Sensitivity at $X_{OUT}, Y_{OUT}, Z_{OUT}$  | $V_S = 2.5\text{ V}, \pm 4\text{ g}$ 10-bit mode               | 116   | 128               | 143   | LSB/g                |
| Scale Factor at $X_{OUT}, Y_{OUT}, Z_{OUT}$ | $V_S = 2.5\text{ V}, \pm 4\text{ g}$ 10-bit mode               | 7.0   | 7.8               | 8.6   | mg/LSB               |
| Sensitivity at $X_{OUT}, Y_{OUT}, Z_{OUT}$  | $V_S = 2.5\text{ V}, \pm 8\text{ g}$ 10-bit mode               | 58    | 64                | 71    | LSB/g                |
| Scale Factor at $X_{OUT}, Y_{OUT}, Z_{OUT}$ | $V_S = 2.5\text{ V}, \pm 8\text{ g}$ 10-bit mode               | 14.0  | 15.6              | 17.2  | mg/LSB               |
| Sensitivity at $X_{OUT}, Y_{OUT}, Z_{OUT}$  | $V_S = 2.5\text{ V}, \pm 16\text{ g}$ 10-bit mode              | 29    | 32                | 36    | LSB/g                |
| Scale Factor at $X_{OUT}, Y_{OUT}, Z_{OUT}$ | $V_S = 2.5\text{ V}, \pm 16\text{ g}$ 10-bit mode              | 28.1  | 31.2              | 34.3  | mg/LSB               |
| Sensitivity Change due to Temperature       |                                                                |       | $\pm 0.02$        |       | %/ $^\circ\text{C}$  |
| 0 g BIAS LEVEL                              | Each axis                                                      |       |                   |       |                      |
| 0 g Output ( $X_{OUT}, Y_{OUT}, Z_{OUT}$ )  | $V_S = 2.5\text{ V}, T_A = 25^\circ\text{C}$                   | -150  | 0                 | +150  | mg                   |
| 0 g Offset vs. Temperature                  |                                                                |       | $< \pm 1$         |       | mg/ $^\circ\text{C}$ |
| NOISE PERFORMANCE                           |                                                                |       |                   |       |                      |
| Noise (x-, y-axes)                          | Data Rate = 100 Hz, $\pm 2\text{ g}$ 10-bit or Full-Res.       |       | $< 1$             |       | LSB RMS              |
| Noise (z-axis)                              | Data Rate = 100 Hz, $\pm 2\text{ g}$ 10-bit or Full-Res.       |       | $< 1.5$           |       | LSB RMS              |
| OUTPUT DATA RATE / BANDWIDTH                | User Selectable                                                |       |                   |       |                      |
| Measurement Rate <sup>3</sup>               |                                                                | 0.1   |                   | 3200  | Hz                   |
| SELF TEST                                   |                                                                |       |                   |       |                      |
| Output Change X                             |                                                                | +0.31 |                   | +1.02 | g                    |
| Output Change Y                             |                                                                | -0.31 |                   | -1.02 | g                    |
| Output Change Z                             |                                                                | +0.46 |                   | +1.64 | g                    |
| POWER SUPPLY                                |                                                                |       |                   |       |                      |
| Operating Voltage Range ( $V_S$ )           |                                                                | 2.0   | 2.5               | 3.6   | V                    |
| Interface Voltage Range ( $V_{DD I/O}$ )    |                                                                | 1.7   | 1.8               | $V_S$ | V                    |
| Supply Current                              | Data Rate > 100 Hz                                             |       | 130               | 150   | $\mu\text{A}$        |
| Supply Current                              | Data Rate < 10 Hz                                              |       | 25                |       | $\mu\text{A}$        |
| Standby Mode Leakage Current                |                                                                |       | 0.1               | 2     | $\mu\text{A}$        |
| Turn-On Time <sup>4</sup>                   | Data Rate = 3200 Hz                                            |       | 1.4               |       | ms                   |
| TEMPERATURE                                 |                                                                |       |                   |       |                      |
| Operating Temperature Range                 |                                                                | -40   |                   | 85    | $^\circ\text{C}$     |
| WEIGHT                                      |                                                                |       |                   |       |                      |
| Device Weight                               |                                                                |       | 20                |       | mgrams               |

<sup>1</sup> All minimum and maximum specifications are guaranteed. Typical specifications are not guaranteed.

<sup>2</sup> Cross-axis sensitivity is defined as coupling between any two axes.

<sup>3</sup> Bandwidth is half the output data rate.

<sup>4</sup> Turn-on and wake-up times are determined by the user defined bandwidth. At 100 Hz data rate the turn-on/wake-up time is approximately 11.1 ms. For additional data rates the turn-on/wake-up time is approximately  $\tau + 1.1$  in milliseconds, where  $\tau$  is  $1/(\text{Data Rate})$ .

## ABSOLUTE MAXIMUM RATINGS

**Table 2. Absolute Maximum Ratings**

| Parameter                                            | Rating          |
|------------------------------------------------------|-----------------|
| Acceleration (Any Axis, Unpowered)                   | 10,000 <i>g</i> |
| Acceleration (Any Axis, Powered)                     | 10,000 <i>g</i> |
| $V_S$                                                | −0.3 V to 3.6 V |
| $V_{DD I/O}$                                         | −0.3 V to 3.6   |
| All Other Pins                                       | −0.3 V to 3.6   |
| Output Short-Circuit Duration<br>(Any Pin to Ground) | Indefinite      |
| Temperature Range (Powered)                          | −40°C to +105°C |
| Temperature Range (Storage)                          | −40°C to +105°C |

Stresses above those listed under Absolute Maximum Ratings may cause permanent damage to the device. This is a stress rating only; functional operation of the device at these or any other conditions above those indicated in the operational section of this specification is not implied. Exposure to absolute maximum rating conditions for extended periods may affect device reliability.

## ESD CAUTION

ESD (electrostatic discharge) sensitive device. Electrostatic charges as high as 4000 V readily accumulate on the human body and test equipment and can discharge without detection. Although this product features proprietary ESD protection circuitry, permanent damage may occur on devices subjected to high energy electrostatic discharges. Therefore, proper ESD precautions are recommended to avoid performance degradation or loss of functionality.



## PIN CONFIGURATION AND DESCRIPTIONS

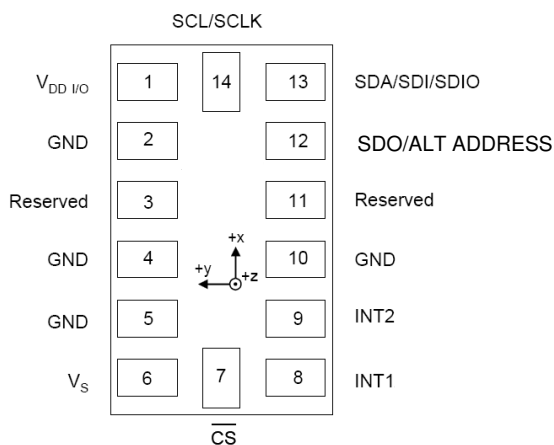


Figure 2. Pin Configuration (TOP VIEW)

Figure 2. Pin Configuration (Top View)

Table 3. Pin Descriptions

| Pin No. | Mnemonic        | Description                                                                                  |
|---------|-----------------|----------------------------------------------------------------------------------------------|
| 1       | $V_{DD\ I/O}$   | Digital Interface Supply Voltage                                                             |
| 2       | GND             | Must be connected to ground                                                                  |
| 3       | Reserved        | Reserved, must be connected to $V_S$ or left open                                            |
| 4       | GND             | Must be connected to ground                                                                  |
| 5       | GND             | Must be connected to ground                                                                  |
| 6       | $V_S$           | Supply Voltage                                                                               |
| 7       | $\overline{CS}$ | Chip Select                                                                                  |
| 8       | INT1            | Interrupt 1 Output                                                                           |
| 9       | INT2            | Interrupt 2 Output                                                                           |
| 10      | GND             | Must be connected to ground                                                                  |
| 11      | Reserved        | Reserved, must be connected to GND or left open                                              |
| 12      | SDO/ALT ADDRESS | Serial Data Out, Alternate I <sup>2</sup> C Address Select                                   |
| 13      | SDA/SDI/SDIO    | Serial Data (I <sup>2</sup> C), Serial Data In (SPI 4-Wire), Serial Data In/Out (SPI 3-Wire) |
| 14      | SCL/SCLK        | Serial Communications Clock                                                                  |

**TYPICAL PERFORMANCE CHARACTERISTICS**

## FUNCTIONAL DESCRIPTION

### DEVICE OPERATION

The ADXL345 is a complete three-axis acceleration measurement system with a selectable measurement range of either  $\pm 2\text{ g}$ ,  $\pm 4\text{ g}$ ,  $\pm 8\text{ g}$ , or  $\pm 16\text{ g}$ . It measures both dynamic acceleration resulting from motion or shock and static acceleration, such as gravity, which allows it to be used as a tilt sensor. The sensor is a polysilicon surface-micromachined structure built on top of a silicon wafer. Polysilicon springs suspend the structure over the surface of the wafer and provide a resistance against acceleration forces. Deflection of the structure is measured using differential capacitors that consist of independent fixed plates and plates attached to the moving mass. Acceleration deflects the beam and unbalances the differential capacitor, resulting in a sensor output whose amplitude is proportional to acceleration. Phase-sensitive demodulation is used to determine the magnitude and polarity of the acceleration.

### POWER SEQUENCING

Power may be applied to  $V_S$  or  $V_{DD I/O}$  in any sequence without damaging the ADXL345. All possible power on states are summarized in Table 4. The interface voltage level is set with the interface supply voltage  $V_{DD I/O}$ , which must be present to ensure that the ADXL345 does not create a conflict on the communications bus. For single-supply operation,  $V_{DD I/O}$  can be the same as the main supply,  $V_S$ . Conversely, in a dual-supply application,  $V_{DD I/O}$  can differ from  $V_S$  to accommodate the desired interface voltage. Once  $V_S$  is applied, the device enters standby state, where power consumption is minimized and the device waits for  $V_{DD I/O}$  to be applied and a command to enter measurement state (setting the MEASURE bit in the **POWER\_CTL** register). Clearing the MEASURE bit returns the device to standby state.

**Table 4. Power Sequencing**

| Condition              | $V_S$ | $V_{DD I/O}$ | Description                                                                                                                                                                                             |
|------------------------|-------|--------------|---------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------------|
| Power Off              | Off   | Off          | Completely off, potential for communications bus conflict.                                                                                                                                              |
| Bus Enabled            | Off   | On           | No functions available, but will not create conflict on communications bus.                                                                                                                             |
| Standby or Measurement | On    | On           | At power up the device is in Standby mode awaiting a command to enter measurement mode and all sensor functions are off. Once instructed to enter Measurement mode, all sensor functions are available. |

### POWER SAVING

#### Power Modes

The ADXL345 automatically modulates its power consumption proportionally with its output data rate as shown in Table 5. If

additional power savings is desired, a lower power mode is available. In this mode, the internal sampling rate is reduced allowing for power savings in the 12.5 to 400Hz data rate range at the expense of slightly greater noise. To enter lower power mode, set the LOW\_POWER bit(D4) in the **BW\_RATE** register.

**Table 5. Current Consumption versus Data Rate**

| Output Data Rate (Hz) | Bandwidth (Hz) | Rate Code | $I_{DD}$ ( $\mu\text{A}$ ) |
|-----------------------|----------------|-----------|----------------------------|
| 3200                  | 1600           | 1111      | 130                        |
| 1600                  | 800            | 1110      | 80                         |
| 800                   | 400            | 1101      | 130                        |
| 400                   | 200            | 1100      | 130                        |
| 200                   | 100            | 1011      | 130                        |
| 100                   | 50             | 1010      | 130                        |
| 50                    | 25             | 1001      | 80                         |
| 25                    | 12.5           | 1000      | 55                         |
| 12.5                  | 6.25           | 0111      | 37                         |
| 6.25                  | 3.125          | 0110      | 25                         |
| 3.125                 | 1.563          | 0101      | 25                         |
| 1.563                 | 0.782          | 0100      | 25                         |
| 0.782                 | 0.39           | 0011      | 25                         |
| 0.39                  | 0.195          | 0010      | 25                         |
| 0.195                 | 0.098          | 0001      | 25                         |
| 0.098                 | 0.048          | 0000      | 25                         |

The current consumption in Low Power Mode is shown in Table 6. Cases where there is no advantage to using Low Power Mode are shaded.

**Table 6. Current Consumption versus Data Rate in Low Power Mode**

| Output Data Rate | Bandwidth (Hz) | Rate Code | $I_{DD}$ ( $\mu\text{A}$ ) |
|------------------|----------------|-----------|----------------------------|
| 3200             | 1600           | 1111      | 130                        |
| 1600             | 800            | 1110      | 80                         |
| 800              | 400            | 1101      | 130                        |
| 400              | 200            | 1100      | 80                         |
| 200              | 100            | 1011      | 55                         |
| 100              | 50             | 1010      | 37                         |
| 50               | 25             | 1001      | 30                         |
| 25               | 12.5           | 1000      | 25                         |
| 12.5             | 6.25           | 0111      | 25                         |
| 6.25             | 3.125          | 0110      | 25                         |
| 3.125            | 1.563          | 0101      | 25                         |
| 1.563            | 0.782          | 0100      | 25                         |
| 0.782            | 0.39           | 0011      | 25                         |
| 0.39             | 0.195          | 0010      | 25                         |
| 0.195            | 0.098          | 0001      | 25                         |
| 0.098            | 0.048          | 0000      | 25                         |

**Auto Sleep Mode**

Additional power can be saved by having the ADXL345 automatically switch to sleep mode during periods of inactivity. To enable this feature set the **THRESH\_INACT** register to an acceleration value that signifies no activity (this value will depend on the application), set **TIME\_INACT** to an appropriate inactivity time period (again, this will depend on the application), and set the **AUTO\_SLEEP** bit and the **LINK** bit in the **POWER\_CTL** register. Current consumption at the sub-8Hz data rates used in this mode is typically 25  $\mu$ A.

**Standby Mode**

For even lower power operation Standby Mode can be used. In Standby Mode current consumption is reduced to 2 $\mu$ A (typical). In this mode no measurements are made and communication with the ADXL345 is limited to single-byte read or writes. Standby Mode is entered by clearing the **MEASURE** bit (D3) in the **POWER\_CTL** register. Placing the device into Standby Mode will preserve the contents of the FIFO.

## SERIAL COMMUNICATIONS

I<sup>2</sup>C and SPI digital communications are available. In both cases, the ADXL345 operates as a slave. I<sup>2</sup>C mode is enabled if the  $\overline{CS}$  pin is tied high to  $V_{DD I/O}$ . In SPI mode, the  $\overline{CS}$  pin is controlled by the bus master. In both SPI and I<sup>2</sup>C modes of operation, data transmitted from the ADXL345 to the master device should be ignored during writes to the ADXL345.

### SPI

For SPI, either 3-wire or 4-wire configuration is possible, as shown in the connection diagrams in Figure 3 and Figure 4. Clearing the SPI bit in the **DATA\_FORMAT** register selects 4-wire mode while setting the SPI bit selects 3-wire mode. The maximum SPI clock speed is 5 MHz, with 12 pF maximum loading and the timing scheme follows CPOL = 1, CPHA = 1.

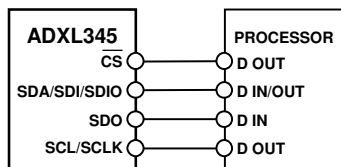


Figure 3. 4-Wire SPI connection

$\overline{CS}$  is the serial port enable line, and is controlled by the SPI master. It must go low at the start of transmissions and back

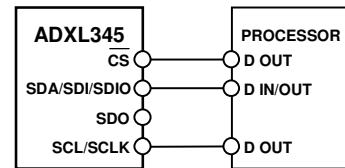


Figure 4. 3-Wire SPI connection

high at the end as shown in Figure 5. SCLK is the serial port clock and is supplied by the SPI master. It is stopped high when  $\overline{CS}$  is high, during period of no transmission. SDI and SDO are the serial data in and out respectively. Data should be sampled at the rising edge of SCLK.

To read or write multiple bytes in a single transmission, the Multi-Byte bit, located after the R/W bit in the first byte transfer, must be set. After the register addressing and the first byte of data, continued clock pulses will cause the ADXL345 to point to the next register for read or write. Continued clock pulses will continue to shift the register that is pointed to until the clock pulses are ceased and  $\overline{CS}$  is de-asserted. To perform reads or writes on different, non-sequential registers,  $\overline{CS}$  must be de-asserted between transmissions and the new register must be addressed separately.

The timing diagram for 3-wire SPI reads or writes is shown in Figure 5. The 4-wire equivalents for SPI reads and writes are shown in Figure 6 and Figure 7 respectively.

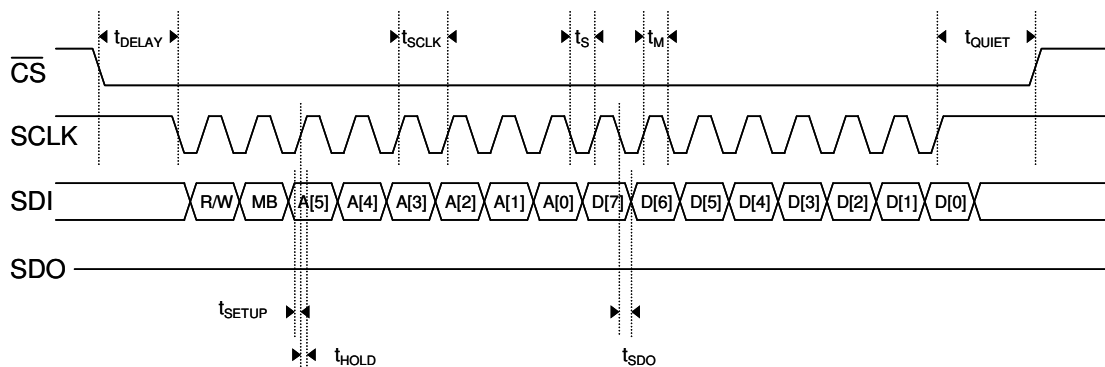


Figure 5. SPI 3-wire Timing Diagram

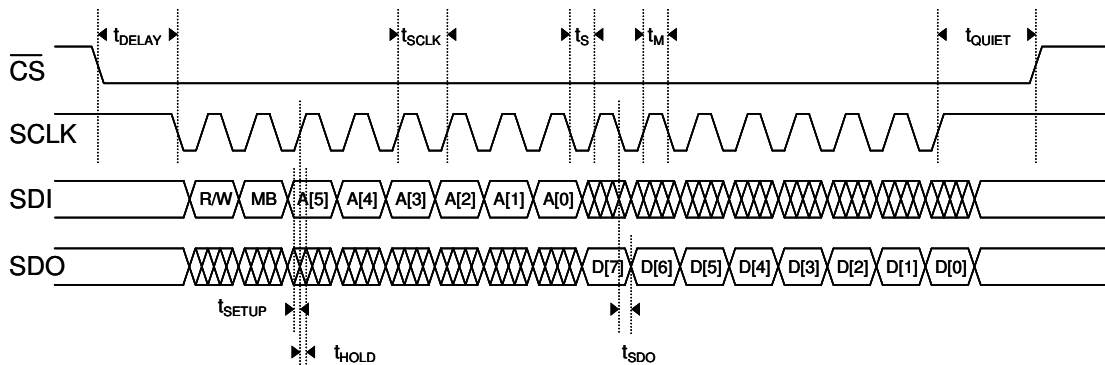


Figure 6. SPI 4-wire Read Timing Diagram

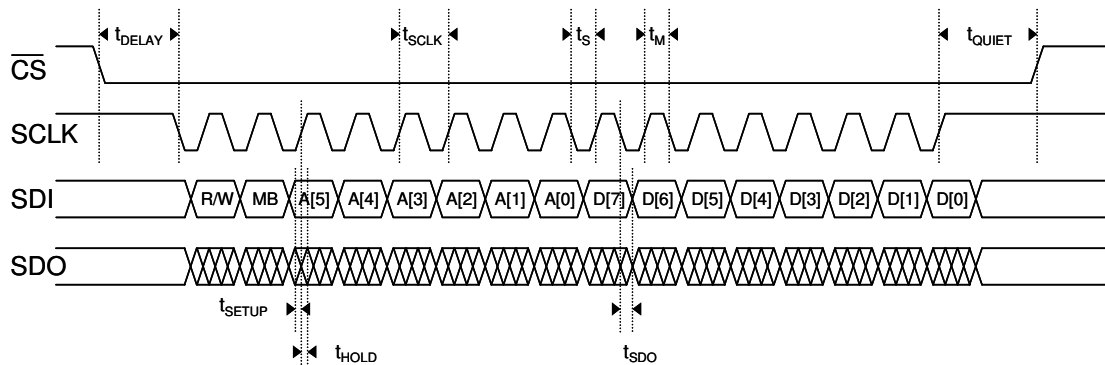


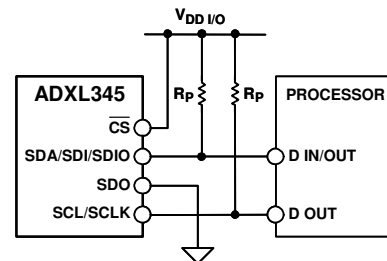
Figure 7. SPI 4-wire Write Timing Diagram

Table 7. SPI Timing Specifications

(TA = 25°C, VS = 2.5V, VDD I/O = 1.8V)

| Parameter   | Limit                 | Unit   | Description                                                                   |
|-------------|-----------------------|--------|-------------------------------------------------------------------------------|
| $f_{SCLK}$  | 5 MHz max             |        | SPI clock frequency                                                           |
| $t_{SCLK}$  | 200 ns min            |        | 1/(SPI clock frequency) Mark/space ratio for the SCLK input is 40/60 to 60/40 |
| $t_{DELAY}$ | 200 ns min            |        | falling edge to SCLK falling edge                                             |
| $t_{QUIET}$ | 200 ns min            |        | SCLK rising edge to rising edge                                               |
| $t_S$       | $0.4 \times t_{SCLK}$ | ns min | SCLK low pulse width (space)                                                  |
| $t_M$       | $0.4 \times t_{SCLK}$ | ns min | SCLK high pulse width (mark)                                                  |
| $t_{SDO}$   | 8 ns max              |        | SCLK falling edge to SDO transition                                           |
| $t_{SETUP}$ | 10 ns min             |        | SDI valid before SCLK rising edge                                             |
| $t_{HOLD}$  | 10 ns min             |        | SDI valid after SCLK rising edge                                              |

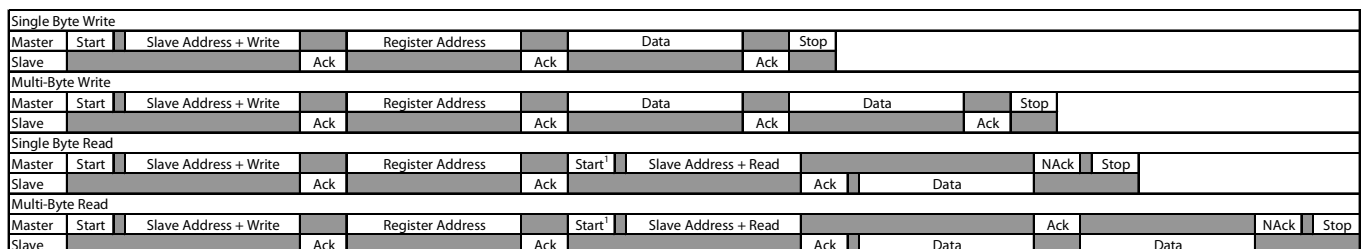
for the device is 0x1D, followed by the read/write bit. This translates to 0x3A for write, 0x3B for read. An alternate I<sup>2</sup>C address of 0x53 (followed by the read/write bit) may be chosen by grounding the SDO pin (pin 12). This translates to 0xA6 for write, 0xA7 for read.

Figure 8. I<sup>2</sup>C Connection Diagram (Address = 0x53)

## I<sup>2</sup>C

With  $\overline{CS}$  tied high to VDD I/O, the ADXL345 is in I<sup>2</sup>C mode, requiring a simple 2-wire connection as shown in Figure 8. The ADXL345 conforms to *The I<sup>2</sup>C Bus Specification*, Version 2.1, January 2000, available from Phillips Semiconductor. It supports standard (100 kHz) and fast (400 kHz) data transfer modes. Single or multiple byte read/writes are supported as shown in Figure 9. With the SDO pin high the 7 bit I<sup>2</sup>C address

If other devices are connected to the same I<sup>2</sup>C bus, the nominal operating voltage level of these other devices cannot exceed VDD I/O by more than 0.3 V. Pull-up resistors, R<sub>P</sub>, should be in the range of 1k to 20kΩ.



This Start is either a restart or a Stop followed by a Start

Figure 9. I<sup>2</sup>C Timing Diagram

## INTERRUPTS

The ADXL345 provides two output pins for driving interrupts: INT1 and INT2. Each of the interrupt functions are described in detail below. All functions can be used simultaneously, with the only limiting feature being that some functions may need to share interrupt pins. Interrupts are enabled by setting the appropriate bit in the **INT\_ENABLE** register and are mapped to either the INT1 or INT2 pins based on the contents of the **INT\_MAP** register. It is recommended that interrupts be configured with the interrupts disabled, preventing interrupts from being accidentally triggered during configuration. This can be done by writing a value of 0x00 to the **INT\_ENABLE** register.

### DATA\_READY

**DATA\_READY** is set when new data is available and cleared when no new data is available.

### SINGLE\_TAP

**SINGLE\_TAP** is set when single acceleration event that is greater than the value in the **THRESH\_TAP** register occurs for a time shorter than specified in the **DUR** register.

### DOUBLE\_TAP

**DOUBLE\_TAP** is set when two acceleration events that are greater than the value in the **THRESH\_TAP** register occur that are shorter than the time specified in the **DUR** register, with the second tap starting after the time specified by the **LATENT** register and within the time specified in the **WINDOW** register. See the Tap Detection description in the Application section for more details.

### ACTIVITY

**ACTIVITY** is set when acceleration greater than the value stored in **THRESH\_ACT** is experienced.

### INACTIVITY

**INACTIVITY** is set when acceleration of less than the value stored in the **THRESH\_INACT** register is experienced for longer than the time specified in the **TIME\_INACT** register. The maximum value for **TIME\_INACT** is 255 s.

### FREE\_FALL

**FREE\_FALL** is set when acceleration of less than the value stored in the **THRESH\_FF** register is experienced for longer than the time specified in the **TIME\_FF** register. The **FREE\_FALL** interrupt differs from **INACTIVITY** interrupt in that all axes always participate, the timer period is much smaller (1.28 s maximum) and it is always DC coupled.

### WATERMARK

**WATERMARK** is set when the FIFO has filled up to the value stored in **SAMPLES**. It is cleared automatically when the FIFO is read and its content emptied below the value stored in **SAMPLES**.

## OVERRUN

**OVERRUN** is set when new data has replaced unread data. The precise operation of **OVERRUN** depends on the FIFO mode. In Bypass Mode, **OVERRUN** is set when new data replaces unread data in the **DATA\_X**, **DATA\_Y**, and **DATA\_Z** registers. In all other modes, **OVERRUN** is set when the FIFO is filled. **OVERRUN** is cleared by reading the FIFO contents, and is automatically cleared when the data is read.

## FIFO

The ADXL345 contains a 32 level FIFO that can be used to minimize host processor intervention. The FIFO has four modes as described in Table 15 in the Register Definitions section. Mode selection is made by setting the appropriate **MODE** bits in the **FIFO\_CTL** register. Each FIFO mode is described below.

### Bypass Mode

In Bypass Mode the FIFO is not operational and remains empty.

### FIFO Mode

In FIFO Mode data from X, Y, and Z measurements go into the FIFO. When the FIFO is filled to the level specified in **SAMPLES** (in the **FIFO\_CTL** register), the **WATERMARK** interrupt is set. The FIFO will continue filling until it is full (32 X, Y, and Z samples) and then stop collecting data. After the FIFO has stopped collecting data the device still continues to operate, so features like Tap detection, for example, may still be used once the FIFO is full. The **WATERMARK** interrupt will continue to occur until the number of samples in the FIFO is less than the value of **SAMPLES** in the **FIFO\_CTL** register.

### Stream Mode

In Stream Mode data from X, Y, and Z measurements go into the FIFO. When the FIFO is filled to the level specified in **SAMPLES** (in the **FIFO\_CTL** register), the **WATERMARK** interrupt is set. The FIFO will continue filling, and will hold the latest 32 X, Y, and Z samples, discarding older data as new data arrives. The **WATERMARK** interrupt will continue to occur until the number of samples in the FIFO is less than the value of **SAMPLES** in the **FIFO\_CTL** register.

### Trigger Mode

In Trigger Mode, the FIFO fills and holds the latest 32 X, Y, and Z samples. Once a trigger event occurs (as described by the **TRIG\_SOURCE** bit in the **FIFO\_CTL** register), the FIFO will keep the last n samples (where n is the value specified by **SAMPLES** in the **FIFO\_CTL** register) and then operate in FIFO mode, collecting new samples only when the FIFO is not full. Additional trigger events will not be recognized until Trigger Mode is reset. This can be done by setting the device in Bypass Mode, reading the **FIFO\_STATUS** register and then setting the device back into Trigger Mode. The FIFO data should be read first, as placing the device into Bypass Mode will clear the FIFO.

**Retrieving Data from the FIFO**

FIFO data is read through the **DATA\_X**, **DATA\_Y** and **DATA\_Z** registers. When the FIFO is in FIFO, Stream, or Trigger Modes, reads to the **DATA\_X**, **DATA\_Y**, and **DATA\_Z** registers read data stored in the FIFO. Each time any data is read from the FIFO the oldest X, Y, and Z data is placed into the **DATA\_X**, **DATA\_Y** and **DATA\_Z** registers. If a single byte read operation is performed, the remaining bytes worth of data will be lost. Therefore, all axes of interest should be read in a burst (or multi-byte) read operation. To ensure that the FIFO has completely popped, there must be at least 5  $\mu$ s between the end of reading the data registers, signified by the transition to register 0x38 from 0x37 or the **CS** pin going high, and the start of new reads of the FIFO or reading the **FIFO\_STATUS** register. For SPI operation at 1.5 MHz or lower, the register addressing portion of the transmission is sufficient delay to ensure the FIFO has completely popped. It is necessary for SPI operation greater than 1.5 MHz to de-assert the **CS** pin to ensure a total of 5  $\mu$ s, which is at most 3.4  $\mu$ s at 5 MHz operation. This is not a concern when using I<sup>2</sup>C, as the communication rate is low enough to ensure a sufficient delay between FIFO reads.

**SELF TEST**

The ADXL345 incorporates a Self Test feature that effectively tests both its mechanical and electronic systems. When the Self Test function is enabled (via the **SELF\_TEST** bit in the **DATA\_FORMAT** register), an electrostatic force is exerted on the mechanical sensor. This electrostatic force moves the mechanical sensing element in the same manner as acceleration, and it is additive to the acceleration experienced by the device. This added electrostatic force results in an output change in the X, Y, and Z-axes. Because the electrostatic force is proportional to  $V_s^2$ , the output change varies with  $V_s$ .

The Self Test feature on the ADXL345 exhibits a bi-modal behavior which depends upon which phase of the clock Self Test is enabled. Due to this, a typical value for Self Test is not reported; however, the limits shown in Table 1 and below are valid for both potential values.

**Table 8. Self Test output in LSB for 2 g and Full-Resolution**

|        | $V_s = 2.5\text{ V}$ |      |      | $V_s = 3.3\text{ V}$ |      |      |
|--------|----------------------|------|------|----------------------|------|------|
|        | Min.                 | Typ. | Max. | Min.                 | Typ. | Max. |
| X-Axis | +80                  |      | +260 | +140                 |      | +455 |
| Y-Axis | -80                  |      | -260 | -140                 |      | -455 |
| Z-Axis | +120                 |      | +420 | +210                 |      | +730 |

**Table 9. Self Test output in LSB for 4 g 10-bit**

|        | $V_s = 2.5\text{ V}$ |      |      | $V_s = 3.3\text{ V}$ |      |      |
|--------|----------------------|------|------|----------------------|------|------|
|        | Min.                 | Typ. | Max. | Min.                 | Typ. | Max. |
| X-Axis | +40                  |      | +130 | +70                  |      | +225 |
| Y-Axis | -40                  |      | -130 | -70                  |      | -225 |
| Z-Axis | +60                  |      | +210 | +105                 |      | +365 |

**Table 10. Self Test output in LSB for 8 g 10-bit**

|        | $V_s = 2.5\text{ V}$ |      |      | $V_s = 2.5\text{ V}$ |      |      |
|--------|----------------------|------|------|----------------------|------|------|
|        | Min.                 | Typ. | Max. | Min.                 | Typ. | Max. |
| X-Axis | +20                  |      | +65  | +35                  |      | +113 |
| Y-Axis | -20                  |      | -65  | -35                  |      | -113 |
| Z-Axis | +30                  |      | +105 | +52                  |      | +183 |

**Table 11. Self Test output in LSB for 16 g 10-bit**

|        | $V_s = 2.5\text{ V}$ |      |      | $V_s = 2.5\text{ V}$ |      |      |
|--------|----------------------|------|------|----------------------|------|------|
|        | Min.                 | Typ. | Max. | Min.                 | Typ. | Max. |
| X-Axis | +10                  |      | +33  | +17                  |      | +57  |
| Y-Axis | -10                  |      | -33  | -17                  |      | -57  |
| Z-Axis | +15                  |      | +53  | +26                  |      | +92  |

## REGISTER MAP

Table 12. Register Map

| Hex | Dec | Name           | Type | Reset Value | Description                       |
|-----|-----|----------------|------|-------------|-----------------------------------|
| 0   | 0   | DEVID          | R    | 11100101    | Device ID.                        |
| 1   | 1   | Reserved       |      |             | Reserved. Do not access.          |
| to  |     |                |      |             | Reserved. Do not access.          |
| 1C  | 28  | Reserved       |      |             | Reserved. Do not access.          |
| 1D  | 29  | THRESH_TAP     | R/W  | 00000000    | Tap threshold                     |
| 1E  | 30  | OFSX           | R/W  | 00000000    | X axis offset                     |
| 1F  | 31  | OFSY           | R/W  | 00000000    | Y axis offset                     |
| 20  | 32  | OFSZ           | R/W  | 00000000    | Z axis offset                     |
| 21  | 33  | DUR            | R/W  | 00000000    | Tap duration                      |
| 22  | 34  | LATENT         | R/W  | 00000000    | Tap latency                       |
| 23  | 35  | WINDOW         | R/W  | 00000000    | Tap window                        |
| 24  | 36  | THRESH_ACT     | R/W  | 00000000    | Activity threshold                |
| 25  | 37  | THRESH_INACT   | R/W  | 00000000    | Inactivity threshold              |
| 26  | 38  | TIME_INACT     | R/W  | 00000000    | Inactivity time                   |
| 27  | 39  | ACT_INACT_CTL  | R/W  | 00000000    | Axis enable control for ACT/INACT |
| 28  | 40  | THRESH_FF      | R/W  | 00000000    | Free-fall threshold               |
| 29  | 41  | TIME_FF        | R/W  | 00000000    | Free-fall time                    |
| 2A  | 42  | TAP_AXES       | R/W  | 00000000    | Axis control for Tap/Double Tap   |
| 2B  | 43  | ACT_TAP_STATUS | R    | 00000000    | Source of Tap/Double Tap          |
| 2C  | 44  | BW_RATE        | R/W  | 00001010    | Data Rate and Power Mode control  |
| 2D  | 45  | POWER_CTL      | R/W  | 00000000    | Power Save features control       |
| 2E  | 46  | INT_ENABLE     | R/W  | 00000000    | Interrupt enable control          |
| 2F  | 47  | INT_MAP        | R/W  | 00000000    | Interrupt mapping control         |
| 30  | 48  | INT_SOURCE     | R    | 00000000    | Source of interrupts              |
| 31  | 49  | DATA_FORMAT    | R/W  | 00000000    | Data format control               |
| 32  | 50  | DATAX0         | R    | 00000000    | X axis data                       |
| 33  | 51  | DATAX1         | R    | 00000000    |                                   |
| 34  | 52  | DATAY0         | R    | 00000000    | Y axis data                       |
| 35  | 53  | DATAY1         | R    | 00000000    |                                   |
| 36  | 54  | DATAZ0         | R    | 00000000    | Z axis data                       |
| 37  | 55  | DATAZ1         | R    | 00000000    |                                   |
| 38  | 56  | FIFO_CTL       | R    | 00000000    | FIFO control                      |
| 39  | 57  | FIFO_STATUS    | R/W  | 00000000    | FIFO status                       |

**REGISTER DEFINITIONS****0x00 DEVID (read-only)**

| D7 | D6 | D5 | D4 | D3 | D2 | D1 | D0 |
|----|----|----|----|----|----|----|----|
| 1  | 1  | 1  | 0  | 0  | 1  | 0  | 1  |

**DEVID** holds a fixed device ID code of 0xE5 (345 octal).

**0x1D THRESH\_TAP (read/write)**

| D7  | D6 | D5 | D4 | D3 | D2 | D1 | D0  |
|-----|----|----|----|----|----|----|-----|
| MSB | D6 | D5 | D4 | D3 | D2 | D1 | LSB |

**THRESH\_TAP** holds the threshold value for tap interrupts. The data format is unsigned, so the magnitude of the tap event is compared to **THRESH\_TAP**. The scale factor is 62.5 mg/LSB (i.e. 0xFF = +16 g). A zero value may result in undesirable behavior if Tap/Double Tap interrupts are enabled.

**0x1E, 0x1F, 0x20 OFSX, OFSY, OFSZ (read/write)**

| D7  | D6 | D5 | D4 | D3 | D2 | D1 | D0  |
|-----|----|----|----|----|----|----|-----|
| MSB | D6 | D5 | D4 | D3 | D2 | D1 | LSB |

**OFSX/OFSY/OFSZ** offer user offset adjustments in two's-complement form with a scale factor of 15.6 mg/LSB (i.e. 0x7F = +2 g).

**0x21 DUR (read/write)**

| D7  | D6 | D5 | D4 | D3 | D2 | D1 | D0  |
|-----|----|----|----|----|----|----|-----|
| MSB | D6 | D5 | D4 | D3 | D2 | D1 | LSB |

**DUR** is an unsigned time value representing the maximum time that an event must be above the **THRESH\_TAP** threshold to qualify as a tap event. The scale factor is 625  $\mu$ s/LSB. A zero value will prevent Tap/Double Tap functions from working.

**0x22 LATENT (read/write)**

| D7  | D6 | D5 | D4 | D3 | D2 | D1 | D0  |
|-----|----|----|----|----|----|----|-----|
| MSB | D6 | D5 | D4 | D3 | D2 | D1 | LSB |

**LATENT** is an unsigned time value representing the wait time from the detection of a tap event to the opening of the time window **WINDOW** for a possible second tap event. The scale factor is 1.25 ms/LSB. A zero value will disable the Double Tap function.

**0x23 WINDOW (read/write)**

| D7  | D6 | D5 | D4 | D3 | D2 | D1 | D0  |
|-----|----|----|----|----|----|----|-----|
| MSB | D6 | D5 | D4 | D3 | D2 | D1 | LSB |

**WINDOW** is an unsigned time value representing the amount of time after the expiration of **LATENT** during which a second tap can begin. The scale factor is 1.25 ms/LSB. A zero value will disable the Double Tap function.

**0x24 THRESH\_ACT (read/write)**

| D7  | D6 | D5 | D4 | D3 | D2 | D1 | D0  |
|-----|----|----|----|----|----|----|-----|
| MSB | D6 | D5 | D4 | D3 | D2 | D1 | LSB |

**THRESH\_ACT** holds the threshold value for activity detection. The data format is unsigned, so the magnitude of the activity event is compared to **THRESH\_ACT**. The scale factor is 62.5 mg/LSB. A zero value may result in undesirable behavior if Activity interrupt is enabled.

**0x25 THRESH\_INACT (read/write)**

| D7  | D6 | D5 | D4 | D3 | D2 | D1 | D0  |
|-----|----|----|----|----|----|----|-----|
| MSB | D6 | D5 | D4 | D3 | D2 | D1 | LSB |

**THRESH\_INACT** holds the threshold value for inactivity detection. The data format is unsigned, so the magnitude of the inactivity event is compared to **THRESH\_INACT**. The scale factor is 62.5 mg/LSB. A zero value may result in undesirable behavior if Inactivity interrupt is enabled.

**0x26 TIME\_INACT (read/write)**

| D7  | D6 | D5 | D4 | D3 | D2 | D1 | D0  |
|-----|----|----|----|----|----|----|-----|
| MSB | D6 | D5 | D4 | D3 | D2 | D1 | LSB |

**TIME\_INACT** is an unsigned time value representing the amount of time that acceleration must be below the value in **THRESH\_INACT** for inactivity to be declared. The scale factor is 1 second/LSB. Unlike the other interrupt functions, which operate on unfiltered data (See Threshold description in Application section), the inactivity function operates on the filtered output data. At least one output sample must be generated for the inactivity interrupt to be triggered. This will result in the function appearing un-responsive if the **TIME\_INACT** register is set with a value less than the time constant of the Output Data Rate. A zero value will result in an interrupt when the output data is below **THRESH\_INACT**.

**0x27 ACT\_INACT\_CONTROL (read/write)**

| D7           | D6              | D5              | D4              | D3             | D2                | D1                | D0                |
|--------------|-----------------|-----------------|-----------------|----------------|-------------------|-------------------|-------------------|
| ACT<br>AC/DC | ACT_X<br>Enable | ACT_Y<br>Enable | ACT_Z<br>Enable | INACT<br>AC/DC | INACT_X<br>Enable | INACT_Y<br>Enable | INACT_Z<br>Enable |

X/Y/Z Enable: A '1' enables X, Y, or Z participation in activity or inactivity detection. A '0' excludes the selected axis from participation. If all of the axes are excluded, the function is disabled.

AC/DC: A '0' = DC coupled operation and a '1' = AC coupled operation. In DC coupled operation, the current acceleration is compared with **THRESH\_ACT** and **THRESH\_INACT** directly to determine whether activity or inactivity is detected. In AC coupled operation for activity detection, the acceleration value at the start of activity detection is taken as a reference value. New samples of acceleration are then compared to this reference value and if the magnitude of the difference exceeds **THRESH\_ACT** the device will trigger an activity interrupt. In AC coupled operation for inactivity detection, a reference value is used again for comparison and is updated whenever the device exceeds the inactivity threshold. Once the reference value is selected, the device compares the magnitude of the difference between the reference value and the current acceleration with **THRESH\_INACT**. If the difference is below **THRESH\_INACT** for a total of **TIME\_INACT**, the device is considered inactive and the inactivity interrupt is triggered.

**0x28 THRESH\_FF (read/write)**

| D7  | D6 | D5 | D4 | D3 | D2 | D1 | D0  |
|-----|----|----|----|----|----|----|-----|
| MSB | D6 | D5 | D4 | D3 | D2 | D1 | LSB |

**THRESH\_FF** holds the threshold value for Free-Fall detection. The data format is unsigned. The root-sum-square(RSS) value of all axes is calculated and compared to the value in **THRESH\_FF** to determine if a free fall event may be occurring. The scale factor is 62.5 mg/LSB. A zero value may result in undesirable behavior if Free-Fall interrupt is enabled. Values between 300 and 600 mg (0x05 to 0x09) are recommended.

**0x29 TIME\_FF (read/write)**

| D7  | D6 | D5 | D4 | D3 | D2 | D1 | D0  |
|-----|----|----|----|----|----|----|-----|
| MSB | D6 | D5 | D4 | D3 | D2 | D1 | LSB |

**TIME\_FF** is an unsigned time value representing the minimum time that the RSS value of all axes must be less than **THRESH\_FF** to generate a Free-Fall interrupt. The scale factor is 5 ms/LSB. A zero value may result in undesirable behavior if Free-Fall interrupt is enabled. Values between 100 to 350 ms (0x14 to 0x46) are recommended.

**0x2A TAP\_AXES (read/write)**

| D7 | D6 | D5 | D4 | D3       | D2              | D1              | D0              |
|----|----|----|----|----------|-----------------|-----------------|-----------------|
| 0  | 0  | 0  | 0  | SUPPRESS | TAP_X<br>Enable | TAP_Y<br>Enable | TAP_Z<br>Enable |

TAP\_X/Y/Z Enable: A '1' in TAP\_X, Y, or Z Enable enables X, Y, or Z participation in Tap detection. A '0' excludes the selected axis from participation in Tap detection.

Setting the SUPPRESS bit suppresses Double Tap detection if acceleration greater than **THRESH\_TAP** is present between taps. See Tap Detection in the Application Section for more details.

**0x2B ACT\_TAP\_STATUS (read)**

| D7 | D6              | D5              | D4              | D3     | D2              | D1              | D0              |
|----|-----------------|-----------------|-----------------|--------|-----------------|-----------------|-----------------|
| X  | ACT_X<br>Source | ACT_Y<br>Source | ACT_Z<br>Source | ASLEEP | TAP_X<br>Source | TAP_Y<br>Source | TAP_Z<br>Source |

X/Y/Z Source: Indicate the first axis involved in a Tap or Activity event. A '1' corresponds to involvement in the event and a '0' corresponds to no involvement. They are not cleared, but overwritten by new data. **ACT\_TAP\_STATUS** should be read before clearing the interrupt. Disabling an axis from participation will clear the corresponding Source bit when the next Activity or Tap/Double Tap event occurs.

ASLEEP: A '1' indicates that the part is in the Auto Sleep Mode. A '0' indicates that the part is not using Auto Sleep Mode. See the **POWER\_CTL** description for more information on Auto Sleep Mode.

**0x2C BW\_RATE (read/write)**

| D7 | D6 | D5 | D4        | D3   | D2 | D1 | D0 |
|----|----|----|-----------|------|----|----|----|
| X  | X  | X  | LOW_POWER | RATE |    |    |    |

LOW\_POWER: A '0' = Normal operation and a '1' = Reduced power operation with somewhat higher noise. (See Power Modes section for details).

RATE: Selects device bandwidth and output data rate. See Table 5 and Table 6 for details. Default value is 0x0A, or 100 Hz Output Data Rate. An Output Data Rate should be selected that is appropriate for the communication protocol and frequency selected. Selecting too high of an Output Data Rate with a low communication speed will result in samples being discarded.

**0x2D POWER\_CTL (read/write)**

| D7 | D6 | D5   | D4         | D3      | D2    | D1     | D0 |
|----|----|------|------------|---------|-------|--------|----|
| X  | X  | LINK | AUTO_SLEEP | MEASURE | SLEEP | WAKEUP |    |

**LINK:** A '1' with both the activity and inactivity functions enabled will delay the start of the activity function until inactivity is detected. Once activity is detected, inactivity detection will begin and prevent the detection of activity. This bit serially links the activity and inactivity functions. When '0' the inactivity and activity functions are concurrent. Additional information can be found in the Application section under Link Mode.

**AUTO\_SLEEP:** A '1' sets the ADXL345 to switch to Sleep Mode when inactivity (acceleration has been below THRESH\_INACT for at least TIME\_INACT) is detected and the LINK bit is set. A '0' disables automatic switching to Sleep Mode. See SLEEP for further description.

**MEASURE:** A '0' places the part into standby mode and a '1' places the part into measurement mode. The ADXL345 powers up in standby mode with minimum power consumption.

**SLEEP:** A '0' puts the part into a normal mode of operation. A '1' places the part into Sleep Mode. This suppresses DATA\_READY, stops sending data to the FIFO, and switches the sampling rate to one specified by the WAKEUP bits. In Sleep Mode, only the Activity function can be used.

When clearing the LINK, AUTO\_SLEEP, or SLEEP bits, it is recommended that the part be placed into Standby when clearing the bits and then re-enabling Measurement mode during a following write. This is done to ensure the device is properly biased if Sleep mode is manually disabled. Not toggling Measurement mode may result in the first few after LINK, AUTO\_SLEEP, or SLEEP is cleared having additional noise, especially if the device was asleep when the bits were cleared.

**WAKEUP:** Controls the frequency of readings in Sleep Mode as shown in

Table 13 below:

**Table 13. WAKEUP Rates**

| D1 | D0 | Frequency (Hz) |
|----|----|----------------|
| 0  | 0  | 8              |
| 0  | 1  | 4              |
| 1  | 0  | 2              |
| 1  | 1  | 1              |

**0x2E INT\_ENABLE (read/write)**

| D7         | D6         | D5         | D4       | D3         | D2        | D1        | D0      |
|------------|------------|------------|----------|------------|-----------|-----------|---------|
| DATA_READY | SINGLE_TAP | DOUBLE_TAP | ACTIVITY | INACTIVITY | FREE_FALL | WATERMARK | OVERRUN |

Setting bits with a value of '1' in this register to enable their respective functions and generate interrupts. A value of '0' will prevent the functions from generating an interrupt. DATA\_READY, WATERMARK, and OVERRUN bits only enable the interrupt output; the functions are always enabled. It is recommended that interrupts be configured before enabling their outputs.

**0x2F INT\_MAP (read/Write)**

| D7         | D6         | D5         | D4       | D3         | D2        | D1        | D0      |
|------------|------------|------------|----------|------------|-----------|-----------|---------|
| DATA_READY | SINGLE_TAP | DOUBLE_TAP | ACTIVITY | INACTIVITY | FREE_FALL | WATERMARK | OVERRUN |

Any '0' bits in this register send their respective interrupts to the INT1 pin. Bits set with a '1' send their respective interrupts to the INT2 pin. All selected interrupts for a given pin are ORed.

**0x30 INT\_SOURCE (read)**

| D7         | D6         | D5         | D4       | D3         | D2        | D1        | D0      |
|------------|------------|------------|----------|------------|-----------|-----------|---------|
| DATA_READY | SINGLE_TAP | DOUBLE_TAP | ACTIVITY | INACTIVITY | FREE_FALL | WATERMARK | OVERRUN |

Bits set with a '1' in this register indicate that their respective functions have triggered. A value of '0' indicates that the corresponding event has not occurred. DATA\_READY, WATERMARK and OVERRUN bits will always be set if corresponding event occurs, regardless of INT\_ENABLE, and are cleared by reading data from the DATA\_X/Y/Z registers. DATA\_READY and WATERMARK may require multiple reads, as per the FIFO Mode descriptions in the FIFO section. Other bits are cleared by reading INT\_SOURCE.

**0x31 DATA\_FORMAT (read/write)**

| D7        | D6  | D5         | D4 | D3       | D2      | D1    | D0 |
|-----------|-----|------------|----|----------|---------|-------|----|
| SELF_TEST | SPI | INT_INVERT | X  | FULL_RES | JUSTIFY | RANGE |    |

**DATA\_FORMAT** controls the presentation of data at registers 0x32 to 0x37. All data, except  $\pm 16$  g range, must be clipped to avoid rollover.

**SELF\_TEST:** A '1' applies a Self Test force to the sensor causing a shift in the output data. A value of '0' disable Self Test.

SPI: A value of '1' sets the device to 3-wire SPI and a value of '0' sets the device to 4-wire SPI.

INT\_INVERT: A value of '0' sets the interrupts to Active High while a value of '1' sets the interrupts to Active Low.

FULL\_RES: When this bit is set with a value of '1' the device is in Full-Resolution Mode, where the output resolution increases with RANGE to maintain a 4 mg/LSB scale factor. When this bit is '0' the device is in 10-bit Mode and RANGE determine the maximum g-Range and scale factor.

JUSTIFY: A '1' = Left (MSB) justified and a '0' = Right justified with sign extension.

RANGE: Sets the g-Range based on Table 14 below.

**Table 14. g-Range Setting**

| D1 | D0 | g-Range |
|----|----|---------|
| 0  | 0  | ±2 g    |
| 0  | 1  | ±4 g    |
| 1  | 0  | ±8 g    |
| 1  | 1  | ±16 g   |

#### 0x32, 0x33 DATA0, DATA1 (read only)

| D7 | D6 | D5 | D4 | D3 | D2 | D1 | D0 |
|----|----|----|----|----|----|----|----|
|    |    |    |    |    |    |    |    |
|    |    |    |    |    |    |    |    |

#### 0x34, 0x35 DATAY0, DATAY1 (read only)

| D7 | D6 | D5 | D4 | D3 | D2 | D1 | D0 |
|----|----|----|----|----|----|----|----|
|    |    |    |    |    |    |    |    |
|    |    |    |    |    |    |    |    |

#### 0x36, 0x37 DATAZ0, DATAZ1 (read only)

| D7 | D6 | D5 | D4 | D3 | D2 | D1 | D0 |
|----|----|----|----|----|----|----|----|
|    |    |    |    |    |    |    |    |
|    |    |    |    |    |    |    |    |

These six bytes hold the output data for each axis. The output data is two's complement with **DATAx0** as the LSByte and **DATAx1** as the MSByte. The **DATA\_FORMAT** register (0x31) controls the format of the data. It is recommended that a burst read of all of the registers is performed to prevent the change of data between reads of sequential registers.

#### 0x38 FIFO\_CTL (read/write)

| D7        | D6      | D5      | D4 | D3 | D2 | D1 | D0 |
|-----------|---------|---------|----|----|----|----|----|
| FIFO_MODE | TRIGGER | SAMPLES |    |    |    |    |    |

FIFO\_MODE: Corresponds to the FIFO Mode as shown in Table 15 below.

TRIGGER: A value of '0' sets the trigger event of Trigger Mode to INT1 and a value of '1' sets the trigger event to INT2.

SAMPLES: Function depends on the FIFO Mode as shown in Table 16 below. Entering a value of zero in SAMPLES will immediately set the WATERMARK status bit in **INT\_SOURCE**, regardless of FIFO mode. Undesirable operation may occur if a value of zero is used for SAMPLES when Trigger Mode is used.

**Table 15. FIFO Modes**

| D7 | D6 | MODE    | Function                                                                                                        |
|----|----|---------|-----------------------------------------------------------------------------------------------------------------|
| 0  | 0  | Bypass  | The FIFO is bypassed                                                                                            |
| 0  | 1  | FIFO    | FIFO collects up to 32 values then stops collecting data                                                        |
| 1  | 0  | Stream  | FIFO holds the last 32 data values. Once full, the FIFO's oldest data is lost as it is replaced with newer data |
| 1  | 1  | Trigger | When triggered by the TRIGGER the FIFO holds the last 32 data values and stops when full.                       |

**Table 16. SAMPLES Functions**

| FIFO Mode | Samples Function                                                                                |
|-----------|-------------------------------------------------------------------------------------------------|
| Bypass    | None                                                                                            |
| FIFO      | Specifies how many FIFO entries are need to trigger a Watermark interrupt                       |
| Stream    | Specifies how many FIFO entries are need to trigger a Watermark interrupt                       |
| Trigger   | Specifies how many FIFO samples <i>before</i> the trigger event are retained in the FIFO buffer |

#### 0x39 FIFO\_STATUS (read)

| D7        | D6 | D5      | D4 | D3 | D2 | D1 | D0 |
|-----------|----|---------|----|----|----|----|----|
| FIFO_TRIG | X  | ENTRIES |    |    |    |    |    |

FIFO\_TRIG: A '1' corresponds to a trigger event occurring while a '0' means that a FIFO trigger event has not yet occurred.

ENTRIES: Reports how many data values are stored in the FIFO. To collect the data from the FIFO, access is through the standard **DATAx**, **DATAY**, and **DATAZ** registers. FIFO reads must be done in burst, or multi-byte, mode as each FIFO level is cleared after any read, single- or multi-byte, of the FIFO. The FIFO stores a maximum of 32 entries, which equates to a maximum of 33 entries available at any given time, due to the fact that an additional entry is available at the output filter of device.

## APPLICATION

### POWER SUPPLY DECOUPLING

In many applications, a 0.1  $\mu\text{F}$  capacitor at  $V_S$  and  $V_{DD\ I/O}$  placed close to the ADXL345 supply pins adequately decouples the accelerometer from noise on the power supply. However, in applications where noise is present at the 50 kHz internal clock frequency, or any harmonic thereof, additional care in power supply bypassing is required because this noise may cause errors in acceleration measurement. If additional decoupling is necessary, a 10  $\Omega$  resistor or ferrite in series with  $V_S$  and an additional larger bypass capacitor (2.2  $\mu\text{F}$  or greater) at  $V_S$  may be helpful.

Care should be taken that the connection from the ADXL345 ground to the power supply ground be low impedance because noise transmitted through ground has an effect similar to noise transmitted through  $V_S$ .

### MECHANICAL CONSIDERATIONS FOR MOUNTING

The ADXL345 should be mounted on the PCB in a location close to a hard mounting point of the PCB to the case. Mounting the ADXL345 at an unsupported PCB location (that is, at the end of a “lever,” or in the middle of a “trampoline”), as shown in Figure 10, may result in large apparent measurement errors because the accelerometer will see the resonant vibration of the PCB. Locating the accelerometer near a hard mounting point ensures that any PCB resonances at the accelerometer are above the accelerometer’s mechanical sensor resonant frequency and, therefore, effectively invisible to the accelerometer.

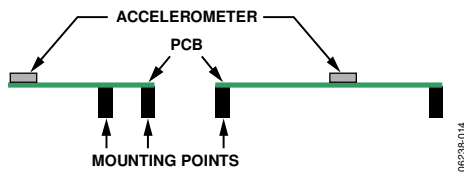


Figure 10. Where Not to Mount an Accelerometer

### TAP DETECTION

The tap interrupt function is capable of detecting either single or double taps. The following parameters are shown graphically in Figure 11 for a valid single and valid double tap event:

- Tap detection threshold is defined by the **THRESH\_TAP** register.
- Maximum tap duration time is defined by the **DUR** register.
- Tap latency time is defined by the **LATENT** register and is the waiting period from the end of the first tap

until the opening of the time window, whose value is contained in the **WINDOW** register, for a possible second tap.

- The interval time after the expiration of **LATENT** is defined by the **WINDOW** register and is the period of time during which a second tap must begin. The second tap need not finish before the end of **WINDOW**.

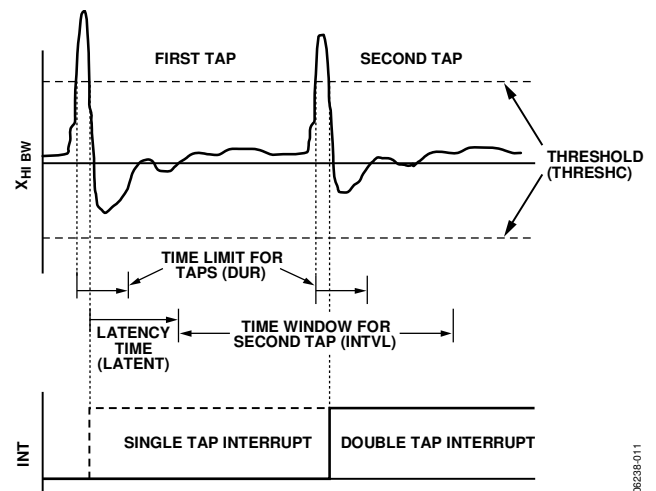


Figure 11. Tap Interrupt Function with Valid Single and Double Taps

If only the single tap function is in use, the single tap interrupt will trigger at the point that the acceleration goes below the threshold as long as **DUR** is not exceeded. If both single and double tap functions are in use the single tap interrupt will trigger once the double tap event has been either validated or invalidated.

Several events can occur to invalidate the second tap of a double tap event. First, if the **SUPPRESS** bit in the **TAP\_AXES** register is set, any acceleration spikes above the threshold during the **LATENT** time window will invalidate the double tap as seen in Figure 12.

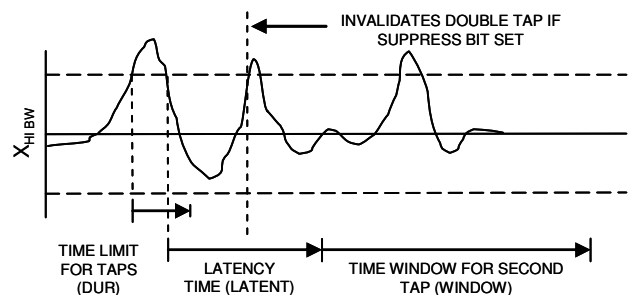


Figure 12. Double Tap event invalid due to high-g event with **SUPPRESS** set

A double tap event can also be invalidated if an acceleration above the threshold is detected at the start of **WINDOW**, resulting in an invalid double tap at the start of **WINDOW**, shown in Figure 13. Additionally, a double tap event can be invalidated by having an acceleration exceed **DUR**, resulting in an invalid double tap at the end of **DUR** for the second tap event, also seen in Figure 13.

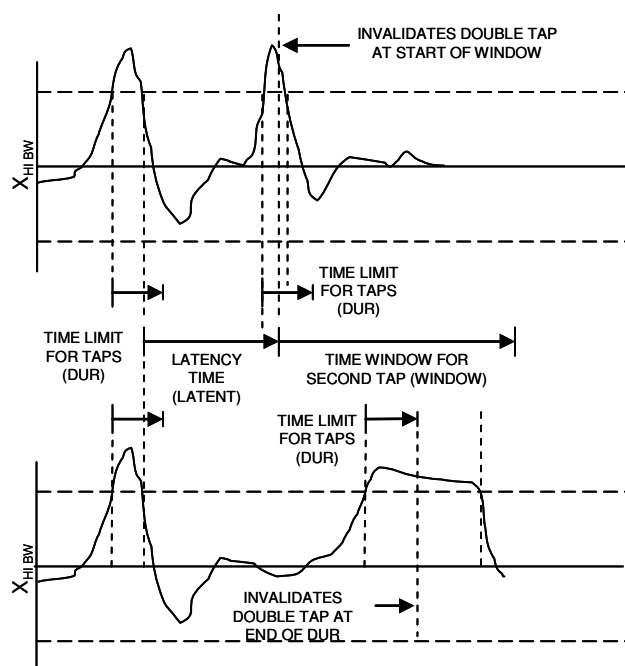


Figure 13. Tap Interrupt Function with Invalid Double Taps

Single taps, double taps, or both may be detected by setting their respective bits in the **INT\_ENABLE** register. Control over participation of each of the three axes in tap/double tap detection is exerted by setting the appropriate bits in the **TAP\_AXES** register. For the double tap function to operate, both **LATENT** and **WINDOW** must be non-zero.

Every mechanical system will have somewhat different tap/double tap response based on the system's mechanical characteristics, so some experimentation with values for the **LATENT**, **WINDOW**, and **THRESH\_TAP** registers will be required. In general a good starting point is **LATENT**>0x10, **WINDOW**>0x10, and **THRESH\_TAP**>3g. Setting very low values in the **LATENT**, **WINDOW**, and/or **THRESH\_TAP** registers may result in unpredictable response due to the accelerometer picking up "echoes" of the tap inputs.

After a tap interrupt is received, the first axis to exceed the **THRESH\_TAP** level is reported in the **ACT\_TAP\_STATUS** register. This register is never cleared, but overwritten with new data.

## THRESHOLD

The lower Output Data Rates are achieved by decimation of a common sampling frequency inside the device. The activity, free-fall and tap/double tap detection functions are performed using the un-filtered data. Since the output data is filtered, the high frequency and high-g data that is used to determine activity, free-fall and tap/double tap events may not be present if the output of the accelerometer is examined. This may result in trigger events appearing to occur when acceleration does not appear to trigger an event, such as exceeding a threshold or remaining below a threshold for a certain period of time.

## LINK MODE

The LINK function can be used to reduce the number of activity interrupts the processor must service by only looking for activity after inactivity. For proper operation of the link feature, the processor must still respond to the activity and inactivity interrupts by reading the **INT\_SOURCE** register to clear them. If the activity interrupt is not cleared, the part will not go into Auto Sleep Mode. The **ASLEEP** bit in the **ACT\_TAP\_STATUS** register indicates if the part is in Auto Sleep Mode.

RECOMMENDED PWB LAND PATTERN

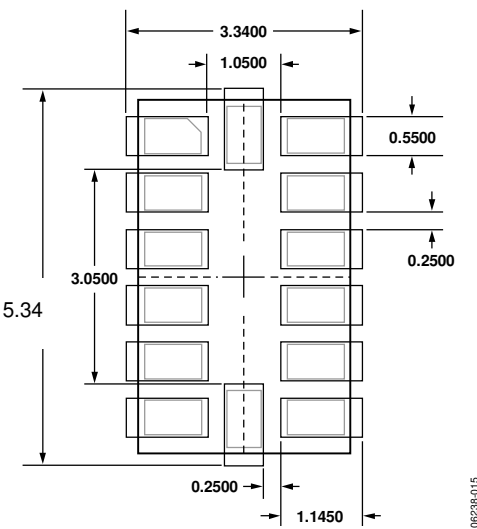


Figure 14. Recommended Printed Wiring Board Land Pattern  
(Dimensions Shown in Millimeters)

## RECOMMENDED SOLDERING PROFILE

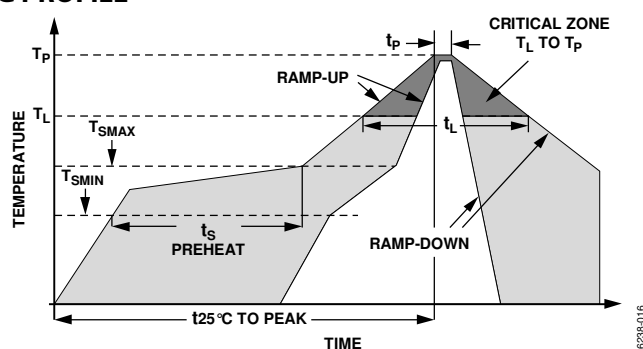
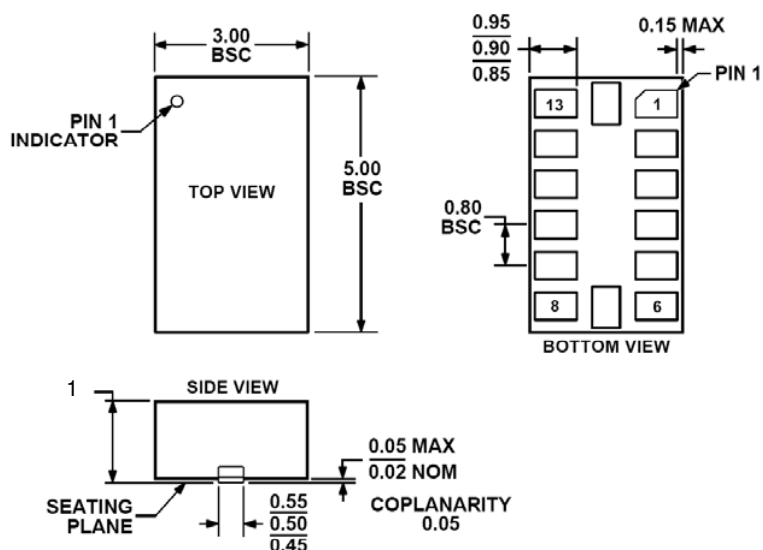


Figure 15. Recommended Soldering Profile

Table 17. Recommended Soldering Profile<sup>1</sup>

| Profile Feature                                      | Condition         |                   |
|------------------------------------------------------|-------------------|-------------------|
|                                                      | Sn63/Pb37         | Pb-Free           |
| Average Ramp Rate ( $T_L$ to $T_P$ )                 | 3°C/sec max       | 3°C/sec max       |
| Preheat                                              |                   |                   |
| Minimum Temperature ( $T_{SMIN}$ )                   | 100°C             | 150°C             |
| Maximum Temperature ( $T_{SMAX}$ )                   | 150°C             | 200°C             |
| Time ( $T_{SMIN}$ to $T_{SMAX}$ ) ( $t_s$ )          | 60 sec to 120 sec | 60 sec to 180 sec |
| $T_{SMAX}$ to $T_L$                                  |                   |                   |
| Ramp-Up Rate                                         | 3°C/sec max       | 3°C/sec max       |
| Time Maintained Above Liquidous ( $T_L$ )            |                   |                   |
| Liquidous Temperature ( $T_L$ )                      | 183°C             | 217°C             |
| Time ( $t_L$ )                                       | 60 sec to 150 sec | 60 sec to 150 sec |
| Peak Temperature ( $T_P$ )                           | 240 + 0/-5°C      | 260 + 0/-5°C      |
| Time Within 5°C of Actual Peak Temperature ( $t_p$ ) | 10 sec to 30 sec  | 20 sec to 40 sec  |
| Ramp-Down Rate                                       | 6°C/sec max       | 6°C/sec max       |
| Time 25°C to Peak Temperature                        | 6 minutes max     | 8 minutes max     |

## OUTLINE DIMENSIONS



## ORDERING GUIDE

| Model                        | Measurement Range | Specified Voltage (V) | Temperature Range | Package Description                   | Package Option |
|------------------------------|-------------------|-----------------------|-------------------|---------------------------------------|----------------|
| ADXL345BCCZ <sup>1</sup>     | ±2, 4, 8, 16g     | 2.5                   | −40°C to +85°C    | 14-Lead Land Grid Array Package [LGA] | TBD            |
| ADXL345BCCZ−RL <sup>1</sup>  | ±2, 4, 8, 16g     | 2.5                   | −40°C to +85°C    | 14-Lead Land Grid Array Package [LGA] | TBD            |
| ADXL345BCCZ−RL7 <sup>1</sup> | ±2, 4, 8, 16g     | 2.5                   | −40°C to +85°C    | 14-Lead Land Grid Array Package [LGA] | TBD            |
| EVAL-ADXL345Z <sup>1</sup>   |                   |                       |                   | Evaluation Board                      |                |

<sup>1</sup> Z = Pb-free part.

## NOTES

**NOTES**

