Exploratory Study on Lower Limb Amputee Patients: Use of IMUs to Monitor the Gait Quality During the Rehabilitation Period

Förberedande studie på patienter med amputerad nedre extremitet: Användning av IMUs för att övervaka gångkvaliteten under rehabiliteringsperioden

AUDE BARTHÉLEMY
Exploratory study on lower limb amputee patients: 
Use of IMUs to monitor the gait quality during the rehabilitation period

Aude Barthélemy  
940926-4109

Supervised by: Emeline Simonetti & Coralie Villa

January 3rd 2019
Acknowledgements

First of all, I would like to thank wholeheartedly Emeline SIMONETTI, PhD student at École Nationale Supérieure des Arts et Métiers for the daily support and the opportunity to work on such an interesting and challenging subject related to her thesis, that she provided me. By sharing her codes, guiding me through the development of new algorithms and associating me to her research, she allowed me to be involved in a clinical study in the heart of the rehabilitation of lower-limb amputee patients.

I would also like to thank warmly Coralie VILLA, research project officer at the Centre d’Etude et de Recherche sur l’Appareillage des Handicapés (CERAH) who accepted the charge of supervising my master thesis despite her tight schedule. Always ready to bring support, she gave me a lot of advice that will technically help me in my work, but will also stay in my mind to guide me in the future.

Then, I would like to thank Joseph BASCOU, head of the research department of the CERAH, for welcoming me in his team and being always happy to give advice and help whatever his workload.

I also thank a lot our patient, who accepted to wear all our sensors so many times during his rehabilitation sessions. Always ready to give us time for the instrumentation and highly motivated for giving us the best data to record, this study wouldn’t have been possible without his participation.

This study could also be made only with the involvement of the physiotherapist team from the Institution Nationale des Invalides (INI). I would like to thank all of them for letting us being immersed in their daily work and for allowing us to intervene during their sessions. Their interest in our study made us feel even more confident in the tremendous interest of our research.

This master thesis wouldn’t have taken place without the agreement of the INI and the CERAH. I thus thank their respective heads for agreeing on welcoming me for the last months of my studies.

I also thank very much Rodrigo MORENO, responsible of the supervision group at KTH, for all the relevant reviews he made all along the master thesis. His comments and advice were of great help to provide a work fitting at best the requirements from KTH.

Finally, I would like to warmly thank the whole team from the CERAH, Julie, Galo, Christelle and Alvaro for their daily good mood. The discussions always full of humour were very enriching and this joyful atmosphere contributed to make this master thesis even more enjoyable.
Abstract

Specific rehabilitation is a key period for a lower-limb amputee patient. While learning how to walk with a prosthesis, the patient needs to avoid any gait compensations that may lead to future comorbidities. To reach a gait pattern close to the one of a healthy person, objective data may be of great help to complement the experience of the clinician team. By using 6 IMUs located on the feet, shanks and thighs accompanied by 3 accelerometers on the pelvis, sternum and head, data could be recorded during walking exercises of 7 rehabilitation sessions of a patient. To compute the absolute symmetry index of the stance phase duration and the stride duration all over the instrumented sessions, the gait events defining the transitions between gait phases were determined thanks to several algorithms. By first comparing the error obtained in the calculation of the stance phase duration with all tested algorithms as compared to the data from pressure insoles considered as a reference system, the algorithm developed by Trojaniello and collaborators [1] was found to be the most adapted to this situation. Using this algorithm on the data from all sessions highlighted the possibility to detect changes in the symmetry of stance phase duration and stride duration, that are relative to the gait quality. This means that IMUs seem to be able to monitor the progress of a patient during his rehabilitation. Hence, IMUs have proven themselves to be a system of great interest in the analysis of the gait pattern of a lower-limb amputee patient in rehabilitation, by allowing for an embedded measurement of much more parameters than the pressure insoles, whose calibration constituted a real limitation.
Contents

List of Figures 5
List of Tables 6
List of Abbreviations 7

1 Introduction 8

2 Materials & Methods 10
  2.1 Subjects ........................................... 10
  2.2 Equipment ....................................... 10
  2.3 Measurement Protocol ........................... 10
  2.4 Data processing .................................. 12
    2.4.1 Synchronisation ............................. 12
    2.4.2 Signal preparation .......................... 12
    2.4.3 Gait events detection ....................... 12
    2.4.4 Parameters calculation ................. 13
    2.4.5 Presentation of the results .......... 14

3 Results 14
  3.1 Determination of the most adapted algorithm . 14
  3.2 Relevance of a monitoring of stance phase and stride durations 15

4 Discussion 17
  4.1 Gait event detection algorithms ................ 17
  4.2 Relevance of the monitoring of stance phase and stride durations 17
  4.3 Sources of error and limitations of the study 18

5 Conclusion 19

Appendices  i

A Background Chapter ii
  A.1 Introduction ................................... i
  A.2 Rehabilitation of lower-limb amputees ........... i
  A.3 Tools for a quantitative analysis of human gait .. ii
  A.4 Analysis of a walking gait cycle ............... v
  A.5 Detection of gait events from IMUs signals .... vii
  A.6 Adequate frequency of measurement in lower-limb amputee rehabilitation  ix

B Some detailed results xi
  B.1 Numerical values corresponding to Figure 2 .......... xi
  B.2 Absolute errors of the SPD on prosthetic side for all trials .......... xi
  B.3 ASI of the SPD and stride duration numerical values .......... xii

C Personal contribution to the code xii

References xiv
## List of Figures

<table>
<thead>
<tr>
<th>Figure</th>
<th>Description</th>
<th>Page</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>Sensors positions with corresponding axes</td>
<td>11</td>
</tr>
<tr>
<td>2</td>
<td>Prosthetic stance phase duration errors in seconds as compared to insoles data</td>
<td>15</td>
</tr>
<tr>
<td>3</td>
<td>Evolution of the ASI of the mean stance phase duration during the rehabilitation</td>
<td>16</td>
</tr>
<tr>
<td>4</td>
<td>Evolution of the mean stride duration during the rehabilitation</td>
<td>16</td>
</tr>
<tr>
<td>5</td>
<td>Lower extremity amputation levels ©2016 Data Trace Internet Publishing, LLC. All rights reserved</td>
<td>i</td>
</tr>
<tr>
<td>6</td>
<td>Reconstructed subject with Vicon data Source: Figure 2.16 from [21]</td>
<td>iii</td>
</tr>
<tr>
<td>7</td>
<td>Forceplates with associated ground reaction forces Source: <a href="http://www.centredesante-pep06.fr">www.centredesante-pep06.fr</a></td>
<td>iii</td>
</tr>
<tr>
<td>8</td>
<td>Pressure Insoles</td>
<td>iv</td>
</tr>
<tr>
<td>9</td>
<td>Instrumented mat Source: shoreorthotics.co.nz</td>
<td>iv</td>
</tr>
<tr>
<td>10</td>
<td>IMUs</td>
<td>v</td>
</tr>
<tr>
<td>11</td>
<td>Right gait cycle and main gait events Source: Modified from Figure 15.6 from [25]</td>
<td>vi</td>
</tr>
<tr>
<td>12</td>
<td>TUG test Source: <a href="http://www.frailtytoolkit.org">www.frailtytoolkit.org</a></td>
<td>x</td>
</tr>
<tr>
<td>13</td>
<td>Prosthetic stance phase duration errors in seconds as compared to insoles data</td>
<td>xi</td>
</tr>
</tbody>
</table>

All the figures are reproduced with permission from the corresponding editors.
List of Tables

1 Characteristics of the algorithms ........................................ 13
2 Results of extra-detection and missed-steps for each algorithm during the first trial . 15
3 Summary table of literature review ........................................ ix
4 Errors of the stance phase duration on the prosthetic limb during the first trial of the first session as compared to insoles data (errors in seconds) ......................... xi
5 Errors of the stance phase duration on the prosthetic limb for all trials as compared to insoles data (errors in seconds) ...................................................... xi
6 Values of the ASI of the SPD and the stride duration during rehabilitation ........ xii
List of Abbreviations

AP  Antero-posterior
ASI  Absolute Symmetry Index
HO  Heel Off
HS  Heel Strike
IMU  Inertial Measurement Unit
LHS  Left Heel Strike
LTO  Left Toe Off
ML  Medio-lateral
RHS  Right Heel Strike
RMS  Root Mean Square
RTO  Right Toe Off
SPD  Stance Phase Duration
Std  Stride Duration
TO  Toe Off
TS  Toe Strike
TUG  Timed Up and Go test
V-acceleration  Vertical acceleration
1 Introduction

Following lower-limb amputation, patients are taken into care by a multidisciplinary team whose principal aim is the return home of the patient with an autonomy in daily-life activities. However, while learning to walk with a prosthesis, the patient often has recourse to gait compensations such as asymmetries or limping. As this may lead to severe comorbidities, the rehabilitation period also aims at reducing these compensations [2]. The visual analysis of the clinician, some basic functional tests and the patient feelings are for now the main ways to detect gait defaults. Gait quality is thus assessed thanks to the experience of clinicians applied on subjective data.

Today, some tools can be used to obtain quantitative and objective data that may help guiding the rehabilitation. Using optoelectronic cameras with 3D markers, forceplates or instrumented mats may indeed be used, but in addition to being very expensive, they also limit the analysis to a few steps within the lab and are often not accessible to the clinicians. Even if optoelectronic cameras in parallel with forceplates are considered as a reference, they are thus not always adapted to a clinical routine [3].

Using inertial measurement units (IMUs) thus seem to represent an interesting alternative for clinical use as they have a reduced cost and are light and small enough to be worn outside the lab. Consisting into 3-axis accelerometers, 3-axis gyroimeters and a magnetometer, they permit to get access to accelerations and angular velocities of the segments and thus to calculate some parameters describing the motion. A previous bibliographic study has permitted to determine the clinically-relevant parameters the most representative of motion and found them to be: spatio-temporal parameters (stance phase duration, step length and their symmetry, stride duration), variability parameters (root mean square (RMS) of pelvic accelerations), flexion angles at the knee and hip and the quantification of the prosthetic use [4].

The calculation of these parameters is however not immediate and algorithms must be used, first to detect gait events such as heel-strikes (HS) or toe-offs (TO), which respectively mark the beginning and end of the stance phase, and then to calculate the parameters themselves. Several studies have already been conducted about the detection of those events from IMUs signals thanks to algorithms, however, to our knowledge, none of them have been tested on a population of lower-limb amputee patients in rehabilitation. Most of them have been developed for healthy subjects [5, 6], or people suffering from another disease affecting gait [1, 7, 8]. Only a few research teams developed algorithms for amputee patients [9–11] but only amputee patients already rehabilitated were studied.

Pilot study

The general objective of the study, in the very long term, is the development of a tool, based on the use of IMUs, to monitor the evolution of a lower-limb amputee patient in rehabilitation. Just like pressure sensors have been used to develop pressure insoles to be used with a smartphone application, a tool based on IMUs could also be developed. Providing direct results relative to the gait quality of a patient to the clinician team would allow for a complementarity between these quantitative objective data and the clinicians experience.
Exploratory study

In the present study, most of the previously mentioned algorithms have been used, shared by the original authors or re-coded on the basis of the corresponding articles and validated under the scope of a doctoral thesis. The aim of the study was double. First, to determine which algorithm is the most successful in detecting gait events in people with lower-limb amputation currently undergoing rehabilitation as compared to the data from pressure insoles. Then, to determine if it is relevant to monitor the stance phase duration and its symmetry and the stride duration with IMUs all along the rehabilitation period of a lower-limb amputee patient. This means determining if an evolution can be detected while instrumenting a walking exercise all along the rehabilitation. A protocol was designed and validated on a healthy subject before applying the algorithms to gait data recorded on a lower-limb amputee patient during the walking exercises of 7 instrumented rehabilitation sessions and comparing it to data obtained from pressure insoles for validation. The best algorithm to use could be determined thanks to the first session. Then, only this algorithm was used to determine the relevance of monitoring the symmetry of stance phase duration and the stride duration along the rehabilitation.

This study was part of a general protocol approved by the French ethical committee for protection of persons (CPP).
2 Materials & Methods

2.1 Subjects

The study took place in the hospital of the institution hosting the research centre: hospital of the Invalides in Paris, France.

The population of interest was defined as lower-limb amputee patients (transtibial or transfemoral) in specific rehabilitation period after a first prosthesis fitting. The patients considered into the study had to be able to walk with their prosthesis.

Due to the number and physical condition of lower-limb amputee patients corresponding to the population of interest present in the rehabilitation service during the period of the study, only one 63-year old male patient, transfemoral amputee, could be part of the protocol.

The patient provided informed written consent before the beginning of the study.

2.2 Equipment

The patient was instrumented at the beginning of the sessions with different sensors.

6 IMUs (Mtw, developed by Xsens) containing a 3-axis accelerometer, a 3-axis gyrometer and a magnetometer with a sampling frequency of 100 Hz were positioned on the feet, shanks and thighs, and held by strips. The x-axis was oriented along the anatomical axis of the corresponding limb as shown on Figure 1(a).

4 accelerometers (Actigraph) were also placed on the patient, recording at 100Hz. Three of them were positioned on the upper body, on the pelvis, sternum and head (on the occiput) and the last one was placed on the prosthetic shank (Figure 1(b)). A headband was used for the sensor on the head and a harness for the one on the sternum. The pelvic sensor, located on the lower-back, was covered by a foam and held by a strip. In each position, one of the axis was corresponding to the craniocaudal axis.

A pair of pressure insoles (Loadsol, developed by Novel) was also inserted into the shoes of the subject. These pressure insoles are integrating one single pressure sensor covering the whole surface of the insole, permitting to measure the normal component of the ground reaction force. Their sampling frequency was also 100 Hz, with a resolution of 10 Newtons.

2.3 Measurement Protocol

Some of the patient rehabilitation sessions were instrumented, when the patient was in a physical condition good enough to walk and the respective plannings could allow it, with the initial aim of instrumenting at least one session a week. In order to record data right from the beginning of the walking rehabilitation sessions, data were recorded even if the patient could walk only between parallel bars. Thus, 7 sessions were instrumented, respectively 15 (one in the morning and one in the afternoon), 20, 28 and 49 (morning and afternoon) days after the first instrumentation.

The patient was first equipped with all sensors before the beginning of the session. The pressure insoles were calibrated according to the manufacturer’s instructions in the application: each leg raised to set the zero and then the whole weight on the same leg to scale the sensors, knowing the weight of the patient. The exercise of standing on one leg being to difficult for a transfemoral patient, the prosthetic side insole was calibrated on the sound side before being inserted in the shoe of the prosthetic side.

The session was led by the physiotherapist without intervention of the experimenter regarding the choice of the exercises. Recordings were made when a walking exercise was executed by the
Figure 1: Sensors positions with corresponding axes
patient, in which case the patient was asked to execute a sound foot stomp on the floor to ease the synchronisation of the different sensors during post-processing, thanks to the generated spike.

2.4 Data processing

To obtain clinical parameters that can be understood and useful for the clinicians from raw data recorded, several steps of data processing must be made. The signals of the different sensors must be first synchronised and then the successive gait events can be determined. From these gait events detection, the parameters of interest can finally be calculated and their evolution studied.

2.4.1 Synchronisation

In order to allow for a comparison between data obtained from IMUs and pressure insoles, which recordings do not start exactly at the same time, the respective signals must be synchronised. An algorithm is thus in charge of asking the user to select a point in the 50 frames following the peak corresponding to the foot striking the floor at the beginning of the exercise in data sets from all sensors. The algorithm then detects the peak as the maximum in the 50 frames and defines a starting and ending frame for each sequence.

2.4.2 Signal preparation

The raw data obtained from the sensors may need to be prepared in order for the algorithms to run correctly.

The lower-limb amputee patient often loads far more weight on his sound limb than a healthy person. If the patient also has trouble in loading completely one insole for calibration, the output signal is saturated. A step of signal reconstruction must thus be done before running the gait event detection algorithms to remove this saturation.

On the other hand, the patient was walking back and forth between parallel bars. As the algorithms have not been validated on u-turns and since the motion data are less clear during these periods, the data corresponding to the u-turn periods have been cut-off for this study.

2.4.3 Gait events detection

Several algorithms have been used to detect gait events from raw data recorded with IMUs. They had been selected from existing literature and implemented on Matlab\textsuperscript{®} thanks to a share from the original authors or thanks to the elements provided in the corresponding articles [4]. Their characteristics are presented in Table 1.
<table>
<thead>
<tr>
<th>Algorithm</th>
<th>Description</th>
</tr>
</thead>
<tbody>
<tr>
<td>Trojaniello et al. [1]</td>
<td>Algorithm based on the identification of peaks in the medio-lateral (ML) angular velocity of the shank that enables the definition of intervals in which gait events are susceptible to be found. HS are identified as minima in the ML angular velocity preceding a maximum in the antero-posterior acceleration (AP-acceleration) and TO as minima in the AP-acceleration preceding the last maximum in AP-acceleration.</td>
</tr>
<tr>
<td>Selles et al. [9]</td>
<td>Algorithm based on the identification of peaks in the vertical acceleration (V-acceleration) of the shank that enables the definition of intervals in which gait events are susceptible to be found. HS are finally determined as maxima in V-acceleration and TO as minima in the AP-acceleration.</td>
</tr>
<tr>
<td>Maqbool et al. [11]</td>
<td>Algorithm based on the identification of peaks in the ML angular velocity of the shank that enables the detection of mid-swing events. HS are identified as the first negative local minima following mid-swing, combined with a negative slope of the signal, or as a minimum in a $\pm 10^\circ/s$ range occurring in the next 80ms. TO are identified as local minima occurring at least 300ms after a HS with a speed lower than -20$^\circ/s$.</td>
</tr>
</tbody>
</table>

These three algorithms thus permit to determine the instants of HS and TO that define the transitions between swing and stance phases and allow for the calculation of several parameters.

### 2.4.4 Parameters calculation

From the HS and TO identification, the stance phase duration (SPD) is calculated as the temporal difference between TO and HS:

$$SPD = t(\text{TO}) - t(\text{HS})$$

In order to assess the quality of a gait pattern, the Absolute Symmetry Index (ASI) for the stance phase duration can be determined. The ASI was first described for healthy subjects [12] and then adapted to amputee people with the corresponding expression [13]:

$$ASI = 100 \times \frac{I - P}{0.5(I + P)}$$

where I and P refer to the parameters of the intact limb and prosthetic limb respectively. The mean values of the SPDs obtained during an instrumented session for each leg will be hence compared thanks to this index. This parameter will thus allow for an evaluation of the symmetry of stance phase duration between the two legs, with the aim for the clinical team to reach values close to the ones of healthy people (between 2.5 and 3.1 according to Nolan and collaborators [13]) meaning that the time spent on each leg while walking would be equally distributed.

The other parameter to be calculated is the stride duration (StD) which refers to the temporal difference between two consecutive HS. This parameter, that will be represented through the temporal evolution of its mean value across sessions, can provide an insight on the walking velocity of the patient as it is decreasing when walking speed increases, which is expected to happen when the patient is progressing and thus permits to monitor the evolution of the patient.
2.4.5 Presentation of the results

Determination of the most adapted algorithm

In order to determine the best algorithm to be used to study the gait of a lower-limb amputee patient in rehabilitation, the first result to be analysed is the error (in seconds) of the SPD. It is calculated on the prosthetic limb thanks to the three algorithms, by comparison with the data obtained with the reference system, the pressure insoles during the first trial recorded at the first session. The median value is thus represented, with the 25th and 75th percentiles. Outlier values, that correspond to values outside the interval \([q_1-2.7\sigma(q_3-q_1); q_3+2.7\sigma(q_3-q_1)]\) where \(\sigma\) refers to the standard deviation and \(q_1\) and \(q_3\) to the 25th and 75th percentiles respectively, are also represented thanks to circles.

These results being calculated thanks to the detection of the HS and TO at each right (R) and left (L) step by the algorithms, having a look at the extra-detection and missed-steps for each algorithm may also be relevant to confirm the choice of the algorithm. A table showing the number of extra-detection and missed-steps of the LHS, LTO, RHS and RTO is thus also drawn.

Relevance of a monitoring of stance phase and stride durations

The evolution of the ASI of the stance phase duration and the stride duration are represented with the mean values of the parameters hereby calculated from the IMUs data taken over all the trials made during each instrumented session. For the stride duration, the data used are the ones recorded on the sound limb.

3 Results

3.1 Determination of the most adapted algorithm

An internal study made by the team but not yet published has shown that the algorithm from Trojaniello and collaborators [1] was the most adapted to lower-limb amputee people once rehabilitated. The first part of this study was thus to determine if this algorithm was still the most adapted for a lower-limb amputee patient within his period of rehabilitation.

First, as can be seen in Table 5, Trojaniello algorithm does not make many extra-detection (false positive) and misses a low number of steps (false negative) as compared to the two other algorithms. As a total, over the 134 detected events, Trojaniello algorithm provides 5 misinformation when Maqbool and Selles provide 23 and 28 respectively. This hence seems to indicate that the algorithm developed by Trojaniello and collaborators is the most adapted to calculate the parameters of interest on amputee patients in rehabilitation, as is already made for amputee people already rehabilitated.

The results on Figure 2 show a representation of the distribution of the absolute error (in seconds) of the SPD between the three algorithms.

As can be seen on the figure, the lowest median value is obtained with Maqbool algorithm (-0.43s, corresponding to -27%) followed by Trojaniello (0.525s). The median value for the absolute error obtained with Selles algorithm is a bit further (1.45s). However, the error values obtained with Maqbool and Selles algorithms are much more spread than the ones from Trojaniello algorithm that remain closer to the median value.

Moreover, Maqbool algorithm provides one outlier value. This representation of the error results thus confirms the fact that Trojaniello algorithm seems to be the most adapted to the analysis of
the gait of a lower-limb amputee patient in rehabilitation. This have been confirmed by representing the same errors on all the trials, as presented in appendix.

Table 2: Results of extra-detection and missed-steps for each algorithm during the first trial

<table>
<thead>
<tr>
<th></th>
<th>Maqbool</th>
<th>Selles</th>
<th>Trojaniello</th>
</tr>
</thead>
<tbody>
<tr>
<td>ExtraEvent LHS</td>
<td>2</td>
<td>0</td>
<td>0</td>
</tr>
<tr>
<td>ExtraEvent LTO</td>
<td>5</td>
<td>4</td>
<td>0</td>
</tr>
<tr>
<td>ExtraEvent RHS</td>
<td>1</td>
<td>3</td>
<td>1</td>
</tr>
<tr>
<td>ExtraEvent RTO</td>
<td>4</td>
<td>5</td>
<td>0</td>
</tr>
<tr>
<td>Total ExtraEvent</td>
<td>12</td>
<td>12</td>
<td>1</td>
</tr>
<tr>
<td>MissedEvent LHS</td>
<td>2</td>
<td>0</td>
<td>0</td>
</tr>
<tr>
<td>MissedEvent LTO</td>
<td>5</td>
<td>6</td>
<td>1</td>
</tr>
<tr>
<td>MissedEvent RHS</td>
<td>1</td>
<td>5</td>
<td>2</td>
</tr>
<tr>
<td>MissedEvent RTO</td>
<td>3</td>
<td>5</td>
<td>1</td>
</tr>
<tr>
<td>Total MissedEvent</td>
<td>11</td>
<td>16</td>
<td>4</td>
</tr>
</tbody>
</table>

Figure 2: Prosthetic stance phase duration errors in seconds as compared to insoles data

3.2 Relevance of a monitoring of stance phase and stride durations

The best algorithm being determined, only this algorithm has been used to calculate the parameters of interest. The results hence show the temporal progression of the parameters of interest calculated thanks to Trojaniello and collaborators algorithm.

Day 0 refers to the first session instrumented. The second instrumented session occurred 15 days after the first one and the third one was made on the afternoon of the same day. Two other sessions
were then instrumented at 20 and 28 days after the first instrumentation respectively. The two last sessions took place in the morning and afternoon of a session following the Christmas break.

Thus, Figure 3 is representing the evolution of the ASI of the stance phase duration over the sessions instrumented. The symmetry index is first increasing between the two first days of instrumentation (thus between session 1 and sessions 2 & 3). Then, for the two next sessions, the ASI decreases and reaches a mean value of 7.6 during session 5. Finally, the ASI comes back to a higher value during the two last sessions.

![Figure 3: Evolution of the ASI of the mean stance phase duration during the rehabilitation](image)

Figure 3: Evolution of the ASI of the mean stance phase duration during the rehabilitation

Figure 4 is representing the evolution of the mean stride duration calculated on the sound limb of the patient. The stride duration increases between the 2 first instrumented sessions but then decreases and remains almost constant around 3.1s during the 3 following instrumented sessions. Finally, the stride duration increases during the two last sessions.

![Figure 4: Evolution of the mean stride duration during the rehabilitation](image)

Figure 4: Evolution of the mean stride duration during the rehabilitation
4 Discussion

4.1 Gait event detection algorithms

While comparing the performance of the different algorithms that could be used in this study, the one developed by Trojaniello and collaborators was found to be the most adapted to amputee patients in rehabilitation. The trustworthiness of this result is increased by the fact that this algorithm was already proven to be the most adapted to a population of lower-limb amputee people of larger scale, already rehabilitated, in a study intern to the research team, not yet published.

This study reveals that among the three algorithms used, that were not initially developed for amputee patients in rehabilitation, only the one from Trojaniello and collaborators is really able to detect the gait events right from the beginning of the rehabilitation period. The two others are able to detect many gait events, but remain far from the reference.

However the three algorithms are using thresholds that are defined by their authors or in the literature, but that may not be adapted for amputee patients in rehabilitation. Indeed, most of the time intervals that guide the algorithms in the detection of the gait events are calculated on a gait pattern close to the one of a healthy person, that is much faster than the one of a patient in rehabilitation. For example, Selles and collaborators designed their algorithm so that it differentiates the stride durations under 1.5m/s from those higher than 1.5m/s in order to adapt the thresholds, but this differentiation does not seem adapted to our population whose stride duration is more than twice the one used to differentiate [9]. Hence, the performance of the algorithms may be increased for a population of lower-limb amputee patients in rehabilitation thanks to an adaptation of the thresholds to be used for this very slow walking pace. Thus, these algorithms would be able to provide even more accurate results while running on data from an amputee patient in rehabilitation, which confirms the large range of possibilities that they bring with IMUs.

4.2 Relevance of the monitoring of stance phase and stride durations

With the two parameters (ASI of the SPD and StD), different changes may reflect a progress of the patient. If the ASI of the SPD is decreasing without an increase of the StD, it would indicate that the patient is improving his symmetry of walk without slowing his walking pace. On the other hand, if the ASI of the SPD remains constant while the StD decreases, this would indicate that the patient is able to walk faster without losing his gait symmetry.

The results showed that the ASI of the SPD was first increasing and then decreasing, which means that there was a progression of the patient over the period of study preceding Christmas break. Then a regression can be seen following the break with an increase in the ASI of the SPD.

Here, a large deterioration can be seen both in the ASI of the SPD and in the StD between the first and second sessions instrumented. This may be due to the fact that the patient had not donned his prosthesis correctly and was thus losing it while walking. This means that monitoring the progression of the patient with IMUs allows for the detection of a bad performance that can be due to a fatigue of the patient or a problem with the prosthesis.

The following 3 sessions then show a real decrease in the ASI of the SPD in parallel with a StD that remains almost constant. This reveals that the patient is progressing as he becomes able to improve his symmetry in the stance phase duration between his two legs, without slowing the walking pace. Thus, IMUs allowed for the detection of a patient progress through an improvement of his symmetry without a decrease in the StD. Finally, the two last sessions revealing an increase in both the ASI of the SPD and the StD, a detection of a regression of the patient due to a decrease in his activity is allowed by IMUs. Hence, using IMUs to monitor the rehabilitation of lower-limb patients
seems to be relevant as it actually detects the progression in the ASI of the SPD and the StD, but also provides the possibility for the calculation of other parameters reflecting the gait quality, such as other spatio-temporal parameters, variability parameters or flexion angles, that may also reflect the progression of the patient.

Pressure insoles have already been validated for the measurement of gait asymmetries on amputee patients [14] and also permit to get access to the stride duration and the ASI of the stance phase duration. However, this system requires a calibration that involves an exercise to be made by the patient, during which he is asked to raise one foot and then to stand only on the corresponding leg before repeating the exercise on the other side. This exercise seems simple but actually, a transfemoral amputee that is still in rehabilitation period is not able to stand on his prosthetic leg. Thus, the insoles cannot be calibrated as expected and a "trick" had to be found, consisting in calibrating the insole of the prosthetic side on the sound side before putting it into the right shoe and continuing the calibration. The calibration of the insoles can thus still be made, but only approximately, which may lead to problems of saturation of the signal if the whole weight is not loaded on the insole during this calibration. Moreover, for an amputee patient in rehabilitation, the gait pattern is not very clear, with a stance phase signal that may not be as smooth as for healthy people. Thus gait events may sometimes be badly detected by the insoles. IMUs hence permit to get rid of these limitations, without a need for a calibration and thus provide much more possibilities for studying the gait of amputee patients in rehabilitation, still allowing for more objectivity and precision than the visual analysis or the functional tests made with a basic chronometer in clinical routine, but without losing the advantage of being an embedded system usable outside a gait laboratory [3].

4.3 Sources of error and limitations of the study

The main source of error of the study stands in the calibration difficulties that were mentioned earlier. Indeed, pressure insoles are supposed to represent a reference but the difficulties of calibration led sometimes to some problems of saturation and it happens that bad detection are made by the system when the gait pattern is less clear. This source of error was not expected as the insoles were tested on transfemoral amputees, but already rehabilitated. The corresponding algorithm has been corrected to avoid these problems, but the load measured by the system remains erroneous while the calibration is approximate. If the symmetry of load was to be studied (thus with another system than IMUs), pressure insoles would not be able to stand as a reference when the patient is unable to do the calibration exercise as initially required. The difference in the dispersion obtained with Trojaniello algorithm as compared to the two others being rather large, this source of error should nevertheless not invalidate the results.

On the other hand, a few limitations are also inherent to the study. Indeed, the way to turn around of our patient being rather unconventional, the algorithms couldn’t detect the different steps during these periods and we chose to cut the corresponding frames, hence loosing some information that we hope to be able to analyse later on. Having only one amputee subject can also constitute a limitation, as it’s not possible to know to what extent those results would have also been obtained on another subject, especially on a transtibial amputee. However, this study aimed at determining the relevance of using IMUs to monitor the progress of a lower-limb amputee patient in rehabilitation and not at determining the general pattern of progression, which thus reduces the impact of this low number of subjects.
5 Conclusion

IMUs have been found to be able to provide an estimation of the asymmetry of the stance phase duration of both legs and the stride duration all along the rehabilitation period of a lower-limb amputee patient. Thus, using IMUs to monitor the evolution of a lower-limb amputee patient seems to be a promising alternative to the use of pressure insoles. By permitting to detect the progress of a patient over only a few sessions, using IMUs enables to avoid the difficulties of calibration of the insoles, in addition to bringing the opportunity to calculate plenty of other parameters that may help clinicians in their patients follow-up.

These promising results let think about a bright future in the use of IMUs. In the very short-term, thanks to the data recorded, other parameters can be calculated: the step length from the AP-acceleration data and its ASI, the RMS of the accelerations of the upper body and their attenuation coefficients to get access to the gait stability, and finally, the flexion angles at the hip and knee. Moreover, this study constituting a pre-study under the scope of a doctoral thesis, a longer study will take place, including more patients over a longer period with the aim of determining if a general pattern of progression can be found for all the patients, that could then be used by the clinicians to determine the level of progress of their patients. In the very long-term, this study around IMUs may lead to the development of a tool, using IMUs, in order to provide direct results on the patient progress to the clinician team during the rehabilitation sessions.
Appendices

A  Background Chapter

A.1  Introduction

After a lower-limb amputation, the rehabilitation constitutes a key period for the patient to recover as much as possible of his functional abilities. In order to enhance the rehabilitation methods, a lot of elements must be taken into account and thus understood. The purpose of this background chapter is therefore, at first, to provide some elements of context about lower-limb amputation and then to detail the current tools used to increase the quality of the rehabilitation. Afterwards, following a presentation of the human gait cycle, a definition of the relevant parameters to monitor and the way to detect them will be presented. Finally, some elements about the optimal frequency of biomechanical testing on an amputee will be given.

A.2  Rehabilitation of lower-limb amputees

A lower-limb amputation significantly alters the locomotive abilities of a person, thus reducing his/her autonomy in daily life activities [15] hereby lowering his quality of life. In Sweden, around 3500 amputations occur each year, mainly of the lower-limb [16] and in France, lower-limb loss affects about 8300 people per year [17]. These lower-limb amputations are mainly transtibial (under the knee) and transfemoral (above the knee) (Figure 5) and caused by different kinds of etiologies: vascular (∼82%), traumatic (∼16%), congenital (∼1%) or tumour-related (∼1%) [2]. Population ageing coupled with a sedentary lifestyle increasing the incidence of diabetes, the rate of vascular amputation may keep rising in the years to come.

Figure 5: Lower extremity amputation levels
©2016 Data Trace Internet Publishing, LLC. All rights reserved
Rehabilitation and prosthetic fitting aim at reducing and supplementing the functional loss induced by amputation. In most cases, the patient is able to return home (87.9%), but only a small proportion of patients fitted with a prosthesis is able to walk again (56 to 97%) and even less of them are able to ambulate outdoors (26 to 62%) after their stay in a rehabilitation centre [18].

This rehabilitation period is of great importance since it determines the level of mobility to be reached by the patient. Many aspects are taken into account during this period that starts as soon as residual limb healing occurs, in order to focus initially on residual limb acceptance and muscle strengthening. After this first stage, the patient is instructed on how to take care of his residual limb and fitted with a temporary prosthesis. He can thus tame the prosthesis by learning to don and doff it and starting gaining confidence on loading or balance [4, 14]. A third and last phase finally takes place: the “specific rehabilitation” that is prosthesis- and patient-specific. Focus is made on restoring balance, increasing joints range of motion during gait, but also reducing limping and gait asymmetries. Indeed, patients often have recourse to gait compensations such as hip hiking (elevation of residual limb) or vaulting (early flexion of the contralateral ankle) to ease foot clearance and reduce tripping risks during prosthetic swing phase. But not only non-aesthetic, these gait compensations introduce long-term disabilities and comorbidities like arthritis, low-back pain, ...

To restore a gait pattern as close as possible to the asymptomatic one, the rehabilitation is conducted by a multidisciplinary team comprising doctors, physiotherapists, occupational therapists, ortho-prosthetists... During different kinds of exercises, the team analyses the walking gait of the patient aiming at describing his movement, detecting his potential gait defects, but also evaluating his performance and progress. To this end, clinical evaluations are based upon visual observations from practitioners, investigations of the patient self-perception of the prosthesis and assessment of overall performance metrics during specific motor tasks (walking speed during a ten-meter walk for instance) [19, 20]. These evaluations are easy to set-up, without the need for lots of equipment nor complicated protocols. However, they are mainly built on subjective impressions and depend on the experience of the clinical team.

In order to improve the rehabilitation methods and bypass this subjectivity, there is a need for quantitative and objective data that can provide clinically-relevant indicators to the clinicians.

A.3 Tools for a quantitative analysis of human gait

To provide useful information to the clinicians, several kinds of tools are used to analyse the walking gait patterns. These tools do not replace the experience of the clinicians but give them the possibility to refine their analysis thanks to precise and objective data.

An **optoelectronic system**, such as Vicon, is a set of infrared cameras that can detect the positions of a set of reflective markers placed at specific locations on the subject body (articulations, extremities...). It is then possible to reconstruct 3D trajectories of all markers and thus the movement of the subject. Starting from these data, accelerations and angular velocities at the joints can be calculated permitting an analysis of the whole movement (Figure 6).
Forceplates enable the measurement of the ground reaction force for each step made on the surface of the platform (Figure 7). It can provide useful elements about balance of the walking pattern (is the same load applied on each leg?) or contact times of each foot (how long does it stay on the ground?). The experimenter can use only the trials during which the patient puts only one foot per platform, what reduces the workable data, but the patient is free from any body-worn sensors and more easily keeps his natural gait pattern without discomfort.

Pressure insoles are a kind of embedded system as they are worn by the patient in his shoes. Thanks to pressure sensors spread on the whole insole, they provide an access to pressure data all along the step and on the whole surface of the foot (Figure 8). Hence, they enable the evaluation
of balance of the patient and the determination of contact times at each step, using a threshold on pressure values that are much higher when the foot is on the ground. Furthermore, they can be used for an undetermined number of steps and are not limited to the room of the lab.

Figure 8: Pressure Insoles

**Instrumented mats** are also used by some gait analysis labs. Like pressure insoles, the mat is equipped by pressure sensors and thus permits to follow the evolution of the centre of pressure during the whole gait (Figure 9). All steps made on the mat can thus provide data about symmetry and balance of gait but also about gait metrics.

Figure 9: Instrumented mat

Source: shoreorthotics.co.nz

Except for pressure insoles, all these tools are commonly used in gait analysis laboratories. However they are rather expensive and thus rarely used during rehabilitation. The combination of the optoelectronic system with forceplates has become a gold standard in gait analysis as it provides all the necessary data for a complete analysis of motion. However the time needed to set up the equipment is long, with the need to attach all markers to the patient. Furthermore, they are limited to a few steps within the lab and provide data that may not be highly representative of the daily-life gait, as made on a flat ground, completely free from obstacles.

Thanks to the evolution of technologies in the last decades, it has now become possible to use embedded sensors to analyse gait patterns and get rid of these laboratory-limited systems [22–24]. Pressure insoles already allow for data acquisitions over a long time and distance with a reduced set-up time and provide instantaneous results. They are involved in more and more research studies and only start being used in gait analysis labs. On the other hand, accelerometers and gyroscopes are now small and light enough to be fixed to the patient, combined into a single sensor: Inertial Measurement Units (IMUs)(Figure 10).

iv
**Inertial measurement units (IMUs)** are indeed a combination of one to three linear acceleration sensors (accelerometers), one to three angular velocity sensors (gyroscopes) and sometimes a magnetometer. These sensors measure the acceleration and angular velocity of their own movement (and by transitivity, the movement of the rigid body to which they are fixed) by using the inertial principle applied to a free mass contained in the sensor. Hence, the acceleration and angular velocity can be calculated from the reluctance to move (inertia) of the free mass, when accelerated by an external force or torque.

These accelerations can provide many useful information, thanks to numerical integration, allowing for the determination of linear velocity and displacement or angular displacement for instance. However, the integration may be affected by a time-increasing drift that invalidates the measurement and gravity also has to be removed from the accelerometer data prior to integration. To this last end, the orientation of the sensor in space has to be known and is thus calculated by integration of angular velocities. Each time an event or a quasi-static position is identified, the drift affecting the angular displacement can be reset, otherwise, algorithms often use gravity to get rid of this numerical integration drift. To determine the vertical orientation of the sensor, a magnetometer is usually added into the sensor [3]. Once this feature is considered, data obtained can be treated correctly.

These sensors are thus very practical as they are placed on the subject directly, enabling the measurement in all kinds of conditions and for a long time. As explained by Iosa, once the placement of IMUs is made on the subject, with the right orientation, they enable gait analysis, stabilometry (postural balance assessment), upper body mobility assessment, monitoring of daily-life activity, tremor assessment and instrumented clinical tests [3]. These lightweight sensors thus permit the instrumentation of any patient and could be used also on amputee patients during their rehabilitation in order to quantify the quality of their walking gait.

### A.4 Analysis of a walking gait cycle

The analysis of walking gait often rely on cyclic properties of the human gait cycle. As can be seen on Figure 11, the cycle can be divided in different phases according to the events taking place in this pattern.
The gait cycle is firstly described as the cyclic succession of strides. A stride starts when the heel of one leg touches the ground (event called heel strike (HS) or initial contact) and ends when this same heel touches the ground again. A right (respectively left) stride happens when this heel belongs to the right leg (respectively left). In one stride, two steps are gathered, a step being defined as starting at HS and ending at the opposite HS. Finally each step is also divided into two periods: a stance and a swing phase. The stance phase corresponds to the period during which the foot is in contact with the ground (between HS and toe off (TO)) and in contrast, the swing phase takes place when the foot does not touch the ground. During the stance phase, intermediate events can be defined, namely heel-off (HO) when the heel leaves the ground and toe-strike (TS) when the toes start touching the floor.

All these characteristics allow for the definition of many spatio-temporal parameters that can then be used to assess the quality of a walking gait. Indeed, by detecting different gait events (more especially HS and TO as they determine the initiation and termination of stance phase), the symmetry of gait can be analysed. Gait symmetry refers to the similarity of contralateral limbs cycles. Three different aspects can be characterised: step length (spatial symmetry), stance duration (temporal symmetry) and limb loading (loading symmetry). However, only the two first ones can be assessed thanks to IMUs, the last one being determined mainly thanks to pressure insoles or forceplates.

Spatial symmetry refers to the similarity of distances covered by each step (right and left). Once body metrics are known, step lengths can be determined either by trigonometric calculations if the gait is assumed symmetric, by modelling the gait with pendulums [26–28], or by integration of the antero-posterior acceleration [1, 8, 29]. Thus IMUs enable the detection of spatial asymmetry that is a common gait defect in population with lower-limb amputation [17]. Lower-limb amputees indeed tend to have longer prosthetic steps than contralateral steps. Using IMUs would thus permit to quantify this asymmetry and help in adapting the rehabilitation to correct this default. On the other hand, temporal symmetry refers to the similarity of time spent in each phase (swing and stance) for each limb. By detecting gait events, the contact time, corresponding to the stance phase duration, can be determined for each foot, together with step or stride durations. Thus, it is possible to detect if less time is spent on one leg, meaning that the gait is not balanced between the two legs. For example, lower-limb amputees often spend more time in stance phase on their intact limb than prosthetic limb [30].
IMUs also allow for the measurement of other parameters assessing gait quality, without using the detection of gait events. These are for example the postural and dynamic stability, actimetry corresponding to a measure of activity, or joint kinematics.

Stability can mainly be studied thanks to calculations of the root mean square (RMS) of accelerations measured with a lower-trunk-worn IMU [3]. The RMS of accelerations is a measure of dispersion of accelerations that corresponds to the standard deviation of acceleration signals. Postural stability can be studied in statics thanks to these RMS of accelerations as they reflect the movement of the body centre of pressure in the horizontal plane. In dynamics, a higher RMS indicates a higher range of motion of body segments and thus a higher risk of falls. RMS can also be used to evaluate the transmission of accelerations from lower-limbs to the upper body and especially to the head. Indeed, head stability is essential to ensure a steady optical flow and a trustworthy processing of vestibular signals which contribute to the control of equilibrium [3].

An evaluation of the level of activity (actimetry) is also possible thanks to IMUs. Indeed, these light-weight sensors can be worn by the patient for a long time in real-life conditions, enabling an evaluation of the use of the prosthesis and even of the number of steps made with it.

Finally, placing IMUs on two consecutive body segments permits the calculation of joint kinematics. The temporal evolution of the angle of flexion of the knee or hip for example may give the possibility to detect an uncoordinated gait pattern, a diminished joint range of motion or to quantify a gait defect such as hip hiking.

A.5 Detection of gait events from IMUs signals

As explained in the previous section, the detection of gait events after a measurement made with IMUs is one of the core steps if one is to assess a gait quality. Thus, a lot of studies have already been made in order to define the best way to distribute the sensors over the body and the best algorithm to detect gait events from these data (cf Table 3). The variety of studies illustrates the variety of possibilities to conduct such a measurement and reveals the level of challenge that it represents.

The detection of gait events from gait data is obviously easier on people who have a traditional gait pattern. Many studies have therefore been conducted on asymptomatic subjects in order to define an algorithm working on a classic pattern before testing it on subjects with gait defects. For instance, in 2012, McCamley and collaborators published a study aiming at detecting gait events from the vertical acceleration measured on asymptomatic subjects [5]. To that end, they used only one waist-worn IMU and applied a continuous wavelet transform to the data. Their method was shown to be very accurate in identifying initial and final contacts as compared to results taken from an instrumented mat, but one can suggest that another positioning of sensors may provide even more accurate results. The influence of the sensors position was precisely one of the elements under investigation in Storm et al.’s study [6], in which they compared the results obtained with one waist-worn and 2 shank-worn IMUs. The study revealed that the results, obtained thanks to a previously made algorithm, were more accurate with the shank-worn IMUs than the waist-worn one. This seems indeed logical since shanks are more distal than the waist and can thus provide less attenuated signals and without any risk of mixing the left and right events. Moreover, one of the specificities of this study is that the tests were made on an outdoor walk, rather long (15 min), thus permitting to validate the method on a real-life walk with all walking velocity changes that it induces. These two teams hence demonstrated that a high accuracy could be obtained in the detection of gait events with IMUs thanks to different kinds of algorithms, with a slight preference for distal-worn sensors. However, as McCamley and collaborators mention themselves as a limit of their study [5], these results were only obtained on young healthy subjects without any gait deficiencies and thus the accuracy of methods still have to be assessed on people with gait pathologies.
The necessity of defining detection methods to be used on people with gait defaults or pathologies was hopefully considered by some researchers. Indeed, Trojaniello and collaborators for example developed a detection method that was tested on people with pathologies affecting gait pattern (hemiparetic, choreic and Parkinsonian subjects) to be compared with healthy elderly subjects [1]. An algorithm based on the identification of time intervals within which gait events were susceptible to happen during a straight walk was thus developed. They demonstrated the robustness, precision and validity of their algorithm that used 2 IMUs mounted on the shanks, just above the ankles and thus proved that using IMUs to study pathological gait patterns was also possible without losing accuracy. Bertoli and collaborators recently extended the validation of Trojaniello’s algorithm, that was renamed TEADRIP (Trusted Events and Acceleration Direct and Reverse Integration along the direction of Progression) [7]. They extended the measurements to a very large cohort comprising healthy elderly, Parkinsonian and mildly cognitively impaired people and still demonstrated the validity of the algorithm. Finally, another example of study made on people whose gait is affected is the one from Mariani and collaborators [8]. They created an algorithm that they demonstrated to be precise and accurate both on healthy people and on subjects before and after surgical treatment for ankle osteoarthritis. The aim was to detect HS and TO, but also the intermediate events of gait (HO and TS) in order to determine foot-flat phases thanks to two foot-worn IMUs. The algorithm was a bit less efficient in detecting intermediate events, but it remained accurate and precise even on a population with an affected gait pattern. These examples of studies show that using IMUs to detect gait events is also possible on people with a gait pattern that can be affected by a disease.

However, none of these studies has yet validated the method with lower-limb amputees.

By 2005, Selles and collaborators tried to use two piezoresistive accelerometers mounted just under the knees to detect gait events on asymptomatic people and transtibial amputees [9]. They created a complex algorithm but which is robust and fast, adapted to all subjects whatever their walking speed. They thus proved that acceleration data could permit an analysis of gait, even for transtibial amputees. However, transtibial amputees still having their own knee, their gait remains close to the traditional one. The main challenge in gait analysis of amputees thus lies in the analysis of transfemoral amputees, who have to deal with both artificial knee and ankle. Ledoux has thus tried to use one single IMU with only a two-axis accelerometer and one-axis gyrometer placed on the prosthetic shank of transfemoral amputees [10]. She compared the results from 3 different kinds of algorithms to treat the data from the IMU, one based on thresholds, one on linear discriminant analysis and one on quadratic discriminant analysis. In the end, all algorithms provided reliable results, but even if the algorithm based on the definition of thresholds was thought to be the most subject-dependant, due to the definition of thresholds that could depend on the subject and his velocity, it was the one that provided most accurate results as compared with data from forceplates. Hence, an algorithm using thresholds to detect gait events can also be used on transfemoral amputees. Finally, Maqbool and collaborators, in 2017, used only a one-axis gyrometer mounted at the shank to detect gait events from healthy subjects, one transtibial amputee and one transfemoral amputee making the test several times with 5 different prosthesis [11]. By comparing the results with data from pressure insoles, the authors validated the method both on a flat ground and on a slope. The precision was a bit lower for the transfemoral amputee, but the algorithm remained valid in all cases.
In the end, many studies have already been done on asymptomatic people, patients with a gait pattern affected by a disease and even on lower-limb amputees. However, many strategies concerning the positioning of sensors (at the waist, on the shanks or feet) and the kind of data extracted from these (use of accelerometers or gyroscopes and the number of axis considered) were used. Each study was made on different kinds of subjects and compared to data taken from different gold standard references (pressure insoles, instrumented mats or forceplates). Hence, there is no consensus about the type and placement of sensors, nor about the kind of algorithm to use. Moreover, none of these authors considered the study of an amputee during a rehabilitation period as the difficulty increases due to the irregularity of gait pattern or the organisation specificities of this period. Instrumenting the rehabilitation of a lower-limb amputee in order to analyse his gait pattern thanks to the detection of gait events thus represents a new challenge.

A.6 Adequate frequency of measurement in lower-limb amputee rehabilitation

In order to provide useful information to the clinician team conducting the rehabilitation of lower-limb amputees, the adequate frequency of instrumentation with IMUs should be defined. Indeed, providing them parameters permitting to determine the quality of the gait with measurements too spaced apart in time may not be as helpful as it could be, but on the other hand, too close measurements may risk to be too heavy to process or may overload the clinicians with too many data. Finding the right frequency of instrumentation during the rehabilitation process is thus an important part of the protocol definition, but turns out to be not that easy.

During a rehabilitation, lots of tests can be made to evaluate progresses made by a patient, all providing different indicators. According to the advancement of rehabilitation and to the clinician conducting it, the tests used will not be the same and the frequency of measurement may have to be adapted to it. Deathe revealed that no consensus was made on the tests to use [31], however, among the classical tests that can be made, the “timed up and go” test (TUG), 2-minute walk, 10-meter walk and AMPpro may be the most widely used. During the 3 first tests, the patient is asked to

<table>
<thead>
<tr>
<th>Study</th>
<th>Sensors</th>
<th>Position of sensors</th>
<th>Reference (Gold Standard)</th>
<th>Subjects</th>
<th>Conclusions</th>
</tr>
</thead>
<tbody>
<tr>
<td>McCamley et al., 2012[3]</td>
<td>IMU</td>
<td>Waist</td>
<td>Instrumented mat</td>
<td>18 AS</td>
<td>Validation of a method using the vertical acceleration</td>
</tr>
<tr>
<td>Storm et al., 2016[4]</td>
<td>3 IMU (3 axis acc. + gyro + magnet)</td>
<td>Shanks + Waist</td>
<td>Pressure insoles</td>
<td>10 AS</td>
<td>2 shank-worn IMUs provide better results than the waist-worn IMU</td>
</tr>
<tr>
<td>Trojaniello et al., 2014[5]</td>
<td>2 IMU (3 axis acc. + gyro + magnet)</td>
<td>Ankles</td>
<td>Instrumented mat</td>
<td>10 hemiparetic +10 choreic + 10 PD +10 SA</td>
<td>Interval identification-based algorithm during straight-walk Robust, precise, valid</td>
</tr>
<tr>
<td>Bertoli et al., 2018[6]</td>
<td>2 IMU (3 axis acc. + gyro + magnet)</td>
<td>Ankles</td>
<td>Instrumented mat</td>
<td>80 ELD + 125 PD + 31 MCI</td>
<td>Validation of TEADRIP algorithm</td>
</tr>
<tr>
<td>Mariani et al., 2013[7]</td>
<td>2 IMU (3 axis acc. + gyro)</td>
<td>Feet</td>
<td>Pressure insoles</td>
<td>10 SA + 32 with ankle troubles</td>
<td>Algorithm precise and accurate Highly efficient for HS and TO</td>
</tr>
<tr>
<td>Selles et al., 2005[8]</td>
<td>2 acc piezoresistive (1 axis)</td>
<td>Under the knees</td>
<td>Forceplates</td>
<td>15 AS + 10T</td>
<td>Complex algorithm, robust, fast but adapted to all subjects</td>
</tr>
<tr>
<td>Ledoux, 2018[9]</td>
<td>1 IMU (2 axis-acc. + 1 axis-gyro)</td>
<td>Shank</td>
<td>Forceplates</td>
<td>10 AS + 5 TF</td>
<td>Reliability of 3 tested-algorithms THR&gt;QDA&gt;LDA</td>
</tr>
<tr>
<td>Maqbool et al., 2017[10]</td>
<td>1 IMU (1-axis-gyro)</td>
<td>Shank</td>
<td>Pressure insoles</td>
<td>8 AS + 17T + 1 TF (5 prosthesis)</td>
<td>Validation on flat ground and slope Lower precision for TF</td>
</tr>
</tbody>
</table>

AS: Asymptomatic Subject TT: Transtitual amputee TP: Transfermoral amputee ELD: Elderly PD: Parkinson disease (M)IMU: (Magno) - Inertial Measurement Unit MCI: Middly Cognitively Impaired Q/LDA: Quadratic/Linear Discriminant Analysis THR: Threshold
perform an exercise: perform a sit-to-stand motion, walk along a 3m path, turn back and perform a stand-to-sit motion for the TUG (Figure 12), walk as far as possible within 2min for the 2-minute walk or walk along a 10-meter pathway in the 10-meter walk test. All these tests are easy to set up and provide direct information through the time needed to perform the exercise or the distance covered. The AMPpro test is a series of small exercises that give more or less points according to the ability of the patient to perform them autonomously and safely. All these tests have already been studied many times to assess their intra-rater or inter-rater reliability [32–35] but the studies always took place after the rehabilitation or at its very end. Only Brooks mentions the fact that 2 rehabilitation sessions were needed to detect real changes between two 2-minute walk tests [36]. No frequency thus seems to be yet recommended to follow the progress of an amputee in rehabilitation.

Figure 12: TUG test
Source: www.frailtytoolkit.org

Holden and Baker both studied the rehabilitation period of amputees and obtained curves representing the evolution of the number of steps and the walking speed respectively [37, 38]. This does not indicate the frequency to follow but provides insights about progress rate in rehabilitation and could thus guide the reflection about the frequency of recording that will be chosen for the study.

In the end, literature about biomechanical tests does not give enough elements to define the adequate frequency of measurement. Reflection about this frequency will thus be mainly based upon direct discussions with clinical teams.
B Some detailed results

B.1 Numerical values corresponding to Figure 2

Table 4: Errors of the stance phase duration on the prosthetic limb during the first trial of the first session as compared to insoles data (errors in seconds)

<table>
<thead>
<tr>
<th>Absolute error of the SPD on prosthetic side (s)</th>
<th>Maqbool</th>
<th>Selles</th>
<th>Trojaniello</th>
</tr>
</thead>
<tbody>
<tr>
<td>Median</td>
<td>-0.81</td>
<td>-0.88</td>
<td>0.43</td>
</tr>
<tr>
<td>25th percentile</td>
<td>-1.18</td>
<td>-1.397</td>
<td>0.12</td>
</tr>
<tr>
<td>75th percentile</td>
<td>-0.44</td>
<td>-0.675</td>
<td>0.8775</td>
</tr>
<tr>
<td>Maximum</td>
<td>1.3</td>
<td>0.32</td>
<td>1.39</td>
</tr>
<tr>
<td>Minimum</td>
<td>-1.62</td>
<td>-2.07</td>
<td>-0.32</td>
</tr>
</tbody>
</table>

B.2 Absolute errors of the SPD on prosthetic side for all trials

Table 5: Errors of the stance phase duration on the prosthetic limb for all trials as compared to insoles data (errors in seconds)

<table>
<thead>
<tr>
<th>Absolute error of the SPD on prosthetic side (s)</th>
<th>Maqbool</th>
<th>Selles</th>
<th>Trojaniello</th>
</tr>
</thead>
<tbody>
<tr>
<td>Median</td>
<td>-0.85</td>
<td>-1.04</td>
<td>0.006</td>
</tr>
<tr>
<td>25th percentile</td>
<td>-1.57</td>
<td>-1.99</td>
<td>0</td>
</tr>
<tr>
<td>75th percentile</td>
<td>0</td>
<td>0</td>
<td>0.88</td>
</tr>
<tr>
<td>Maximum</td>
<td>4.61</td>
<td>1.91</td>
<td>5.16</td>
</tr>
<tr>
<td>Minimum</td>
<td>-5.6</td>
<td>-6.75</td>
<td>-6.31</td>
</tr>
</tbody>
</table>

Figure 13: Prosthetic stance phase duration errors in seconds as compared to insoles data

Table 5: Errors of the stance phase duration on the prosthetic limb for all trials as compared to insoles data (errors in seconds)
In the light of the results obtained for all trials over all sessions, the fact that Trojaniello algorithm is more adapted to this situation is confirmed.

B.3 ASI of the SPD and stride duration numerical values

Table 6: Values of the ASI of the SPD and the stride duration during rehabilitation

<table>
<thead>
<tr>
<th></th>
<th>Session 1</th>
<th>Session 2</th>
<th>Session 3</th>
<th>Session 4</th>
<th>Session 5</th>
</tr>
</thead>
<tbody>
<tr>
<td>Stride duration (s)</td>
<td>2.7674</td>
<td>3.4785</td>
<td>3.1459</td>
<td>3.0969</td>
<td>3.1846</td>
</tr>
</tbody>
</table>

C Personal contribution to the code

This study taking place under the larger scope of a doctoral thesis, the fundamental algorithms were already developed. These algorithms had already been used on lower-limb amputee people data but never on data from lower-limb amputee patients in rehabilitation. Thus, a personal contribution to the codes could be made, by modifications of the previously implemented algorithms, or by the development of new algorithms to enable the analysis on the new data recorded.

Saturation deletion
Due to the difficulties of calibration of the pressure insoles for a transfemoral patient, a phenomenon of saturation of the signal sometimes occurred. Thus an algorithm aiming at restoring the original signal was developed.

The algorithm is run with 3 inputs: the signal, the saturation threshold and the threshold above which to look for a saturation beginning or end. After referencing all the frames above this last threshold, the algorithm looked for the presence of an abrupt change of value right before or right after each referenced frame, meaning that it corresponds to the beginning or the end of a saturation period. Thus, in case of saturation, the value of the saturation threshold was added to all the frames considered as saturated.

As a saturation sometimes occurred into the saturation itself, the value corresponding to the frame right before the saturation end was finally checked to determine if the following period is a normal signal or a double saturation in order to adapt the signal if necessary.

U-turn cut
Our patient not yet being able to turn around in a proper way, the algorithms had trouble in detecting the steps during these periods. The algorithms had even never been validated on u-turns for healthy patients. An algorithm was thus implemented to cut these periods and allow for the analysis of one-ways only in a first time.

The algorithm only takes the signal as input (the angular velocity around the medio-lateral axis) and aims at providing a matrix indicating the starting and ending frames of the u-turns to cut. The signal is first filtered with a Butterworth filter of second order with a sampling frequency of 100Hz and a cut-off frequency of 1 Hz. Then the RMS envelope of the signal over a sliding window of 1000 frames is used to determine the u-turn periods that correspond to the lowest values. All the values under half of the envelope were considered as to be cut. The beginning and ending frame of these periods were returned in the output matrix.

Instead of having one signal for a full trial, the signal is thus separated in several shorter signals corresponding to each one-way. All the algorithms thus had to be adapted to take into account these
separations and work with structure arrays instead of structures.

**Trojaniello and collaborators algorithm**

The patient under study being still learning how to walk again, he sometimes stumbled. Trojaniello and collaborators algorithm tended to detect two separate steps at these moments, and then ran an error as the number of steps between the two sides was different (an alternation of right and left steps should provide an equal number of steps for both sides, plus or minus one). Thus, the algorithm was adapted to check if the alternation between right and left steps is correct and if not, to show a graphical representation of the corresponding signal and let the user choose between keeping only one of the two peaks or merge them in order to keep the right number of steps. Thus even the trials during which the patient stumbled could be analysed.
References


