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Quantitative mobility assessment in patients with knee osteoarthritis

From lab to an ecological approach by means of wearable technology

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"Measurement is the first step that leads to control and eventually to improvement. If you can't measure something, you can't understand it. If you can't understand it, you can't control it. If you can't control it, you can't improve it."

James Harrington

"Life is suddenly reduced to a one dimension picture, known as physical function, and continually referred to as 'outcome.' The typical outcome process ignores emotional and interpersonal needs and skills. Within those parameters lies the answer to true recovery. I refuse to have an 'outcome.' I do have a life!"

Cannon, 1994

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Préambule

Venant de la clinique où le qualitatif prend une place importante, je fus porté par le désir **d'apporter un outil de mesure quantitatif pertinent et accessible aux kinésithérapeutes**. La pertinence d'une mesure en clinique se réfère souvent aux besoins/demandes du patient ou aux contraintes pratiques, tandis que l'établissement de la pertinence par une méthodologie scientifique requiert de prendre en compte de nombreux aspects tels que la précision, la fiabilité, le coût, etc. Je persiste à croire que c'est ma sensibilité pour le besoin du patient d'être compris dans les difficultés de son quotidien qui m'a poussé à choisir **l'approche en situation écologique** qui sera développée comme premier axe de recherche dans cette thèse.

Ma curiosité m'a poussé à prendre un angle transversal sur la problématique des **technologies de mesure** en kinésithérapie, le deuxième axe de la thèse. On pourrait me reprocher un **manque de profondeur**. Il s'avère que j'ai dû accepter dans le parcours que mon chemin serait une répétition d'essais/erreurs qui me pousseraient à rediriger mon chemin en conséquent. Les limites d'une **étude exploratoire** sont nombreuses mais permettent de discerner les options les plus pertinentes et porteuses des options moins avantageuses. La largeur de la thèse peut être expliqué par le nombre d'options technologiques envisagées (plus de treize), de logiciels de traitement de données ou langages de programmation (plus de huit), ainsi que par les défis techniques rencontrés avec les centrales inertielles. Ces défis m'ont guidé vers la recherche de **pluridisciplinarité** sans laquelle ce travail n'aurait pu voir le jour : informaticiens, ingénieurs, entrepreneurs, cliniciens de terrain, chercheurs spécialistes de mon approche, ou de l'intelligence artificielle. La

thèse ne serait pas la même si je pouvais la commencer aujourd'hui avec toute cette équipe.

Les discussions au sujet de ma thèse avec les patients et les cliniciens de terrain m'ont permis de garder le sens humain de la clinique sans tomber dans le technologisme. Ils m'ont aussi fait prendre conscience du **manque de formation à l'utilisation des technologies** ou à la prise de conscience de leur apport potentiel pour les soins. J'ai donc développé un fameux tropisme pour l'enseignement au fil des années de contacts avec les étudiants en kinésithérapie, mais également en ingénieurie. Je suis persuadé que la formation initale et continue de ces professionnels devra s'adapter pour correspondre aux besoins de terrain et aux changements liés aux technologies. Si certaines sociétés technologiques sont parvenues à nous simplifier la vie par des outils clés en mains, le travail pour parvenir au même résultat dans les soins de santé est semé d'embûches.

Ces apprentissages m'ont confirmé mon intérêt pour la **recherche appliquée** plutôt que pour la recherche fondamentale dont j'admire les fervents adeptes. La mise en pratique de connaissances dans le but de résoudre un problème concret agite mes neurones et provoque des débats animés avec mes confrères. Ces deux types de recherche se nourrissent l'une de l'autre. J'ai encore beaucoup à apprendre pour combler le fossé entre ces approches. Cela fera probablement partie intégrante de mes projets futurs. J'espère que ma thèse pourra avoir de nombreuses **retombées** positives avec l'émergence de nouvelles possibilités de mesure en milieux non contrôlés.

Que la lecture de ce travail puisse vous inspirer !

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Summary

Mobility assessment is an essential task of physiotherapists, who are faced with patients suffering from severe knee **osteoarthritis** and activities limitations. These patients walk slower, display less range of motion at their joints and are often more sedentary than healthy individuals. However, the mobility assessment tools accessible in clinics are limited and the laboratory technologies are costly and cumbersome. As the goal of rehabilitation is to improve patients' functioning in daily life, it is necessary to be able to assess them in their **daily environment**. The technological advances have allowed the assessment of several mobility variables by **wearables sensors**, at low cost and not limited by spatial constraints. However, their performance has not yet been demonstrated in daily life conditions. Moreover, the sensor-based mobility variables must meet certain **psychometric qualities** to be considered useful by physiotherapists: reliability, validity, responsiveness, no modification of movement, etc.

For these reasons, the goal of this thesis was to test wearable technologies in daily life conditions with patients suffering from severe knee osteoarthritis. Therefore, we assessed the sensors' quality by a step-by-step approach, starting from the lab, towards uncontrolled daily life conditions.

First, the assessment of gait variables by means of an instrumented treadmill has not been conclusive **in a hospital environment**: the majority of patients with knee arthroplasty refused or abandoned the study because of the long commuting between their home and the hospital.

In order to measure mobility variables in **semi-controlled conditions**, we developed a measurement system of joint angles, based on open source inertial sensors. Next, this system was validated against an eight-camera motion analysis system, which is the current gold standard. Afterwards, the wearable technologies were tested outside-laboratory conditions as a means to assess the changes in lower limb mobility associated with an innovative injection in the genicular nerves of the knee(s) of **Cameroonian patients with osteoarthritis**. The improvement in range of motion was observed at the hip and pelvis during walking on a flat surface and in stairs. To our knowledge, this was the first study using inertial sensors in Cameroon.

In the third and last part of this thesis, we used activity trackers in **daily life settings**. One tracker was used continuously **for three month**s on patients with a recent hip or knee arthroplasty, undergoing a telerehabilitation treatment. This study revealed the return to the preoperative physical activity level around the 7th postoperative week.

Mobility assessment in outdoor conditions is challenging as it renders certain variables, such as speed, **sensitive to various environmental perturbations** that are often non measurable and hardly standardizable. Future research should study the **large variability in mobility** throughout a day, in particular by considering the optimum state (the best 10% of an individual's distribution) and the lower range of an individual's mobility in various outdoor contexts.

This thesis opens perspectives concerning the use of wearable sensors in various contexts: in the practice of physiotherapists (e.g. at the home of the patients or in low-funded hospitals). **Artificial intelligence**, combined with wearable sensors, could at some point automate mobility assessment, consider more variables and provide a more comprehensive picture to clinicians. Those technologies could **push forward the practice** and the **clinical decision process of physiotherapists**. We underline that this could add value, as we preliminarily observed in a survey that physical activity is hardly being assessed by Belgian physiotherapists, while joint angles are preferentially assessed with the bare eye.

This thesis made a unique contribution to out-of-laboratories studies. It explored new horizons for the ambulatory measurement of joints angles, spatio-temporal parameters and physical activity. It will allow physiotherapists to move forward in a holistic understanding of the mobility limitations in real-life settings.

Key words: osteoarthritis, ambulatory assessment, wearable devices, gait

Résumé

L'évaluation de la mobilité est une tâche essentielle des kinésithérapeutes qui se voient confrontés à des patients soufrant d'**arthrose** sévère du genou dont les activités sont limitées. Ces patients marchent plus lentement, génèrent moins d'amplitude articulaire et sont souvent plus sédentaires que la moyenne de la population. Les moyens d'évaluation disponibles en clinique sont limités et les technologies de laboratoire sont couteuses et encombrantes. Comme l'objectif de la rééducation est d'améliorer le fonctionnement du patient dans la vie de tous les jours, il est nécessaire de pouvoir également l'évaluer dans son **environnement de vie quotidien**. Les avancées technologiques rendent maintenant possible l'évaluation de divers paramètres de mobilité par des **capteurs portables**, peu couteux et qui ne sont pas limités à un certain périmètre d'analyse. Cependant, leur fonctionnement n'a pas été démontré dans des situations de la vie quotidienne. De plus, les paramètres de la mobilité mesurés avec ces capteurs doivent répondre à certaines **qualités psychométriques** pour être utiles aux kinésithérapeutes : fiabilité, validité, sensibilité au changement, non perturbation du mouvement, etc.

Pour ces raisons, ce travail de thèse a eu pour but de tester des technologies portables dans des contextes de la vie quotidienne de patients atteints d'arthrose sévère du genou. Les qualités des capteurs ont été évaluées dans une approche pas à pas en partant du laboratoire, en allant vers des conditions non contrôlées de la vie quotidienne.

Premièrement, l'évaluation de paramètres de la marche au moyen d'un tapis roulant instrumenté n'a pas été concluante en **contexte hospitalier**: la majorité des patients participants opérés d'une prothèse de genou ont refusé ou abandonné l'étude à cause du trajet trop important entre leur domicile et l'hopital.

Afin de mesurer les paramètres de mobilité en **conditions semi-contrôlées**, un système de mesure des angles articulaires basé sur des centrales inertielles « open source » a été développé. Ensuite, il a été validé par rapport à un system de 8 caméras optoélectroniques, qui est le système de référence. Ces technologies portables ont par la suite été testées comme moyen d'évaluer les changements de cinématique articulaire après une thérapie innovante d'injection des nerfs géniculés du genou chez des patients soufrant d'arthrose au **Cameroun**. Les améliorations ont été observées au niveau de la hanche et du bassin lors de la marche et de la descente et montée d'escaliers. A notre connaissance, il s'agit de la première étude ayant utilisé des capteurs inertiels au Cameroun.

Dans la dernière partie de cette thèse, nous avons utilisé des traqueurs d'activité dans des **conditions de vie quotidienne**. Un traqueur a été utilisé continuellement pendant **trois mois** par des patients ayant bénéficié d'une prothèse de hanche ou de genou et d'un traitement de télérééducation. Cette étude a permis de révéler le retour à un niveau d'activité physique préopératoire aux alentours de la 7ème semaine postopératoire.

L'évaluation de la mobilité en conditions extérieures est un défi car elle rend les variables comme la vitesse **sensible à divers perturbateurs environnementaux** non mesurables et non standardisables. Des futures études

devront aider à interpreter la haute **variabilité de la mobilité** au cours d'une journée, notamment en considérant les cas extrèmes d'optimum de mobilité, ou de mobilité non optimales dans divers contextes extérieurs.

Cette thèse ouvre des perspectives pour le développement de l'utilisation de capteurs portables dans divers contextes pratique des kinésithérapeutes (i.e. à domicile du patient ou dans des structures hospitalières à moyens limités). **L'intelligence artificielle,** combinée aux capteurs portables, pourrait un jour permettre d'automatiser l'évaluation de la mobilité, prendre en compte plus de variables et de fournir aux cliniciens une représentation plus complète. Ces technologies seront amenées à **faire évoluer la pratique** et le **processus de décision des kinésithérapeutes**. En effet, nous avons observé dans une enquête préliminaire que l'activité physique est peu évaluée à ce jour par les kinésithérapeutes belges, tandis que les angles articulaires sont préférentiellement évalués à l'œil nu.

Cette thèse apporte une contribution unique aux études hors-laboratoires. Elle développe de nouveaux horizons pour la mesure ambulatoire d'angles articulaires, de paramètres spatio-temporels et de l'activité physique. Elle permettra au kinésithérapeute de progresser dans sa compréhension holistique des limitations de mobilité dans des conditions de vie réelle.

Mots-clés: arthrose, capteurs portables, locomotion, évaluation ambulatoire

General introduction

Mobility assessment

The ability to move is a core element in the way we live our lives. For most of us, it is something normal, whether we walk inside our home, commute to work, or hike into nature. Limitations in the ability to walk may have important consequences on the mobility of people in several daily life situations. The assessment of mobility is therefore of major importance for our health system. The International Classification of Functioning disability and health (ICF) describes the concept of mobility as '*moving by changing body position or location or -by transferring from one place to another, by walking, running or climbing'* (activities and participation section - d4) [1]. In the ICF domains, the relationship between mobility limitation, and participation restrictions of people with disabilities is influenced by contextual factors that consist of environmental factors, such as the physical, social and attitudinal environment in which people live and conduct their life [1]. Thus, mobility is not a fixed state but the result of a dynamic interaction between the health condition, the environment and the person [2]. The difference between ICF domains (*i.e.* body functions, activities and participation) depends on the level observed. The body function component focuses on mechanisms, while the activity component focuses on the execution of tasks and the participation component reflects the involvement in a real-life situation. In order to assess the execution of a task or an action performed by an individual, a distinction between two concepts in the activity component should be made: capacity and performance (Figure 1). Capacity reflects the person's ability to execute a task or an action in a given domain at a given moment, measured in a standardized

environment. Performance, on the other hand, reflects the person's performance of a task or action in his usual environment. Capacity is much more often assessed in research and clinical practice than performance [3]. This specification is of major importance as we will see later on. Moreover, the performance could also be subdivided into perceived and objective performance. Indeed, patient perception is a different construct that often fails to capture functional decline or overstates functional improvement after some treatments [4].

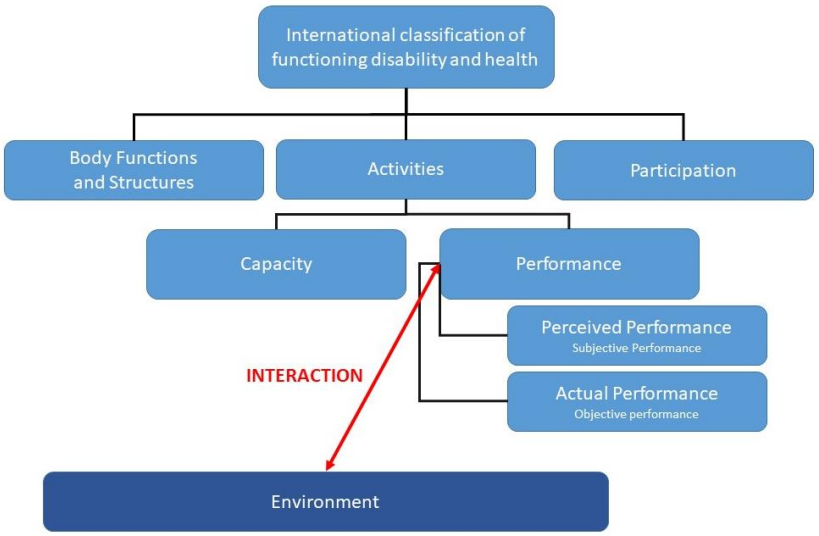

Figure 1: International classification of functioning disability and health (ICF) model explained by 3 components (body functions, activity and participation) and environmental factors. Mobility (d4) belong to the activity component. The Activity component can be divided into capacity and performance. Capacity is the person's ability to execute a task at a given moment, measured in a standardized environment. Performance is the person's performance of a task in his current environment: Performance can be subdivided into perceived or objective. Adapted from De Baets 2017[3]

Rehabilitation specialists, including physiotherapists, are particularly interested in mobility. In the rehabilitation process, the assessment is the first major step [5] (Figure 2). Assessing the patient's mobility is critical to draw a prognosis, assign the best possible treatment and monitor the patient's evolution. Usual mobility assessment consists of different questionnaires (Patient reported outcome measure [PROM]) and motor performance tests (Performance based outcome measure [PBOM]) e.g. 6 minutes walking test (6MWT), 10 meters fast walking test, timed get up and go (TUG)[6] or dynamic gait index (DGI) [7]. These motor tests provide quantitative information about the capacity of the patient to perform a specific activity, such as walking, in a standardized clinical environment. Patient-reported outcome measures (PROM) are self-reported questionnaires that assess symptoms, activities and quality of life. Examples are the knee injury and osteoarthritis outcome score (KOOS) [8], or the International Physical Activity Questionnaire (IPAQ) for physical activity measurement [9]. While these tests and questionnaires are recommended for osteoarthrosis management [6], they provide little information about specific movement characteristics and the quality of movement. In the clinical practice, physiotherapists appreciate movement quality without devices [10]. However, in a research perspective, we believe that we cannot understand a phenomenon and improve it without measurement analysis. Therefore, the understanding of human mobility limitations requires a more in-depth analysis, that is, quantitative, comprehensive and multifactorial.

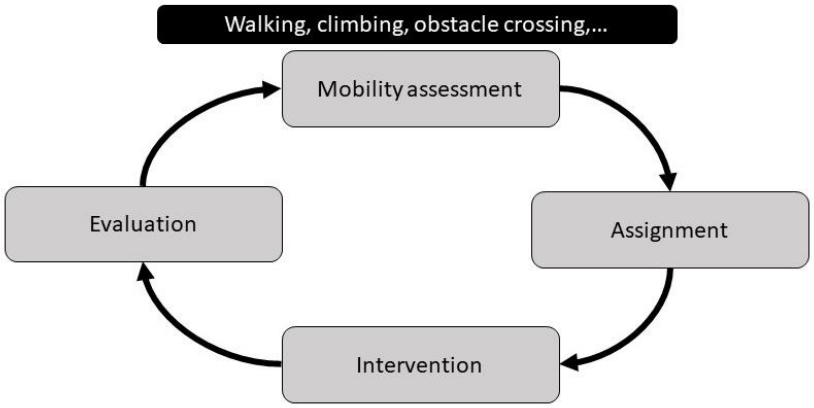

Figure 2. The Rehabilitation cycle and the measurement modalities used in this thesis. The cycle includes four major stages: (1) assessment, (2) assignment, (3) intervention, (4) evaluation. Adapted from Stucki et al [5].

In the last decades, the quantitative assessment of mobility has been mainly focused on gait. The ultimate goal is to provide the best treatment to improve gait performance and participation. Quantitative gait analysis is considered to be any objective means that can be used to measure walking performance. It is most frequently used as a quantitative aid in surgical decision making for children with cerebral palsy, a step that complements the clinical information obtained through careful physical examination [11, 12]. In contrast, instrumented quantitative gait analysis has not been systematically adopted in the clinical routine to evaluate gait in other patient populations [13].

Gait assessment is only a part of mobility assessment, and the motricity needed in daily life functioning. Other dimensions of mobility such as obstacle crossing or stair climbing, are also important activities to perform to allow social participation. For example, the ability to walk long distance or climb stairs are the main treatment goals for patients who are candidates for a total knee

arthroplasty [14]. Such challenging activities must therefore also be assessed in order to display a comprehensive overview of mobility. Moreover, the parameters measured should not be limited to a single variable, such as the time to perform a task. The comprehensive assessment of the whole kinetic chain, instead of focusing on one joint, is justified to provide an overview of mobility limitations. Although measures of mobility performed in laboratory could deliver accurate results, another limitation is that they are often restricted to laboratory settings. Indeed, the time and spatial constraints, as well as the cost limit the translation to clinical practice. How can we then understand mobility in daily life?

Ecological validity

Mobility behavior can be understood by examining the action constraints which are divided into three main categories [15]. First, organismic constraints are all the inherent properties and characteristics of the individual, such as height, age or flexibility. Second, task constraints are external to the individual and specific to the situation the performer is faced with. For example, the task could be to move from the house to the market store. Third, environmental constraints are external constraints related to the environment in which the task is performed [15] (Figure 3). Environmental constraints could be the support surface of gait, obstacles, slope or the type of footwear. They have been classified in eight dimensions : minimum walking distance, time constraints, ambient condition, terrain characteristic, external physical load, attentional demands, postural transition and traffic level [16].

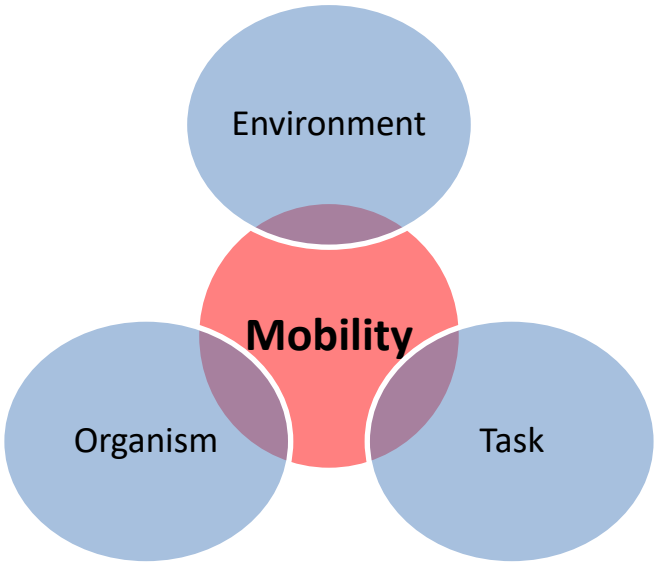

Figure 3: *Newell Theory of constraints: Interrelationships of organism, task and environment. Mobility behavior is an emergent pattern arising from the interplay of constraints. Reproduced from Newell [15].*

Regarding the environmental constraints, mobility is an adaptive performance to the environment and its perturbations, mobility limitations arise when a person cannot meet the demand of the environment. For example, regarding gait, most experiments are conducted in laboratories with artificial settings, where gait is assessed in steady-state conditions, without perturbations. These conditions have little in common with the natural everyday environments of subjects. It is therefore unlikely, that we will be able to generalize the findings from laboratory studies to real-life situations. Such research lacks 'ecological validity', or representativeness [17]. Brunswick introduced the term ‘ecological validity’ in 1943 to justify the opposition of representative design (i.e., ecological approach) to systematic design (i.e., laboratory approach) [18]. The term reflects the tension between the need for

control and the need to preserve the essence of the phenomenon under investigation. A recent study [19] among patients with multiple sclerosis shows that the 10 meters walking test assessed in standardized conditions was only moderately correlated with daily-life walking speeds [19]. Another example is that conventional measures of motor symptom severity in Parkinson's disease do not reflect daily-life activity [20]. One of the assumptions is that an increased attention to task performance has an influence on symptoms such as freezing of gait [21, 22]. Therefore daily-life measures provide information that is not captured in a conventional, laboratory-based assessment [23].

Limited external and ecological validity account for a significant gap between research evidence and research implementation. External validity relates to how applicable the findings are in the real world and internal validity relates to how the findings could be attributed to a few independent variables. In gait research, internal validity is often prioritized over external validity. The interval validity optimization is performed by setting strict inclusion criteria, and reproducible methods that limit bias. Although some environmental perturbations could be reproduced in laboratory, the randomness of the real world is hardly reproduced. For example, in a complex environment, gait control could become vision-dependent[24] for safe foot placement, and decision making in trajectory [25, 26]. Moreover, assessments in rehabilitation settings are often perceived as test situations which may cause patients to perform differently than they would otherwise [21, 27]. For example, patients suffering from Parkinson's disease have attenuated gait performance in free-living conditions compared to laboratory-based data [20].

From an interventional perspective, some argue that the ecological approach could promote improvement across levels of human functioning [28] and that rehabilitation should focus on meaningful concrete goals. The main goal of functional therapeutics is to improve engagement in functional tasks. Such aim cannot be achieved by considering the individual in isolation from its context. The paradigm shift from the laboratory to ecological settings contribute to the shift from the medical model to the biopsychosocial approach [29], as this latter health approach integrates the impact of more contributory factors to disease than biological processes alone.

A shift is therefore needed from laboratory-based to real-life centered research, and a smooth transition between clinical performance and daily life activities. However, laboratory and real-life measures do not deliver comparable results [17]. The sources of data should complement each other to result in a comprehensive outlook for the therapist. While controlled conditions are needed to test hypotheses on mechanisms of human functioning, the understanding of the individual-environment system requires an ecological approach [2].

However, some challenges need to be addressed in an ecological approach. In ecological settings, task requirements are not systematically manipulated and the use of the traditional measurement system is impractical. As a challenge we could thus mention recognition of behavioral abnormalities and normal changes in behavior, context awareness, unobstruviness, and minimal standardization of protocols [30].The ultimate goal is to reach interpretable measurements in uncontrolled, free-living conditions. To do so, a smooth transition between approaches could be executed by going from

controlled condition to semi-controlled condition in a circuit [31] (Figure 4). Researchers could then pragmatically provoke ecological perturbations to understand behavior by using a methodology called the ecological task analysis [32]. However, there is a lack of consensus of the type and scope of measures, the most appropriate approach for data capture, and the extraction of timely information that meets the needs of patients, clinicians, caregivers, and health care regulators [33].



Figure 4: *Step-by-step approach from controlled condition in lab to uncontrolled condition in free-living. Semi-controlled condition means that efforts have been made to standardize the task, i.e. on a circuit in a relevant environment, but that that subjects kept a certain freedom in determining their movements.*

Room for innovation: wearable sensors for data-based healthcare

An ecological approach can only be executed by technology that allows ambulatory recording of movement A wide array of approaches have been investigated using various technologies [34]. Figure 5 displays an overview of the measurement tools, protocols and quantitative outcome measures that are currently being used.

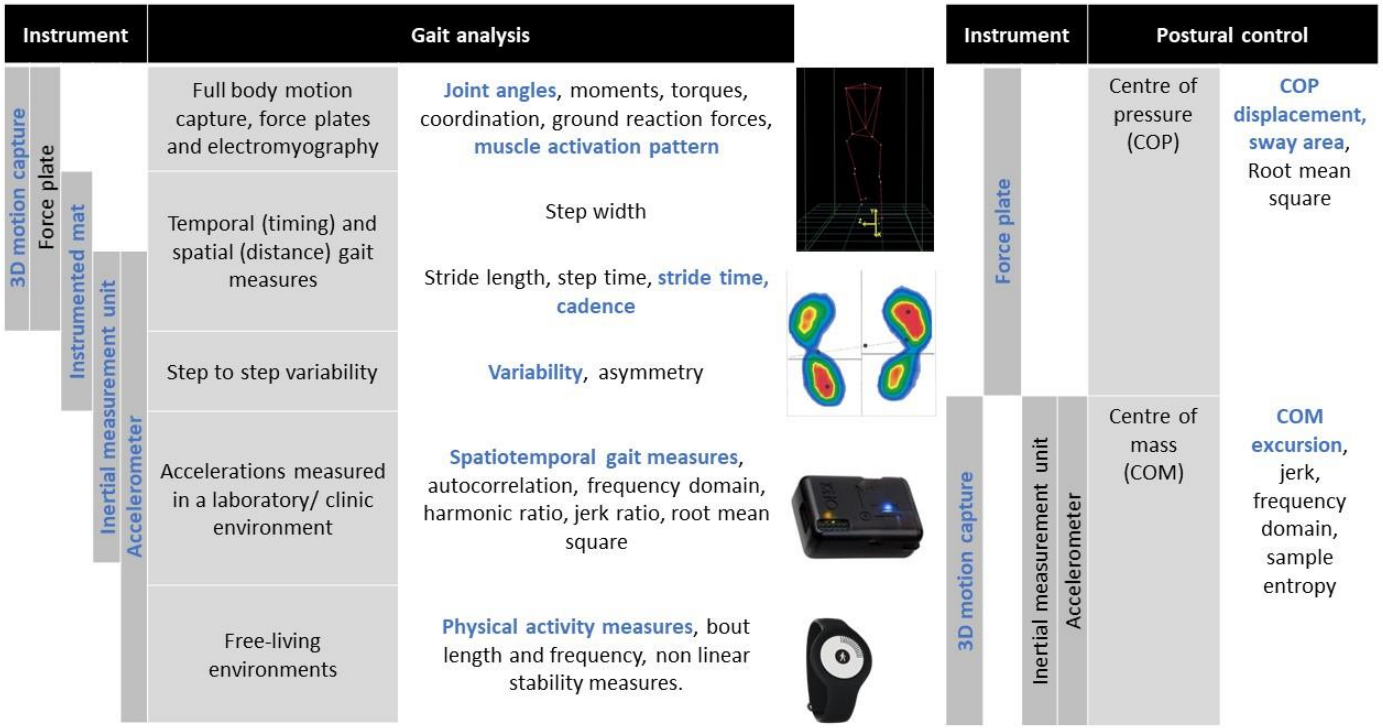

Figure 5: Summary of outcome measures that may be obtained from a quantitative assessment of gait and postural control. Items in blue were used in the current thesis. Adapted from *The Role of Movement Analysis in Diagnosing and Monitoring Neurodegenerative Conditions: Insights from Gait and Postural Control, Buckley C et al. Brain sciences 2019[35].*

When choosing a technology, criteria such as commercial availability, battery life, data communication and storage, range, sensitivity, robustness, usability, price, invasiveness or portability should be considered. 3D Motion capture is highly precise and accurate, but requires experienced technical expertise and a dedicated space [35]. Wearable sensors monitor individuals unobtrusively for extended periods of time in their home and community settings [36]. Portable sensors do not suffer from space constraints, and allow to measure subjects outdoor, at work places, sport centers, patient's home, etc. They are also cheaper and portable. But it often requires complex algorithms to extract features, and data collection is limited by battery power or data storage and lack context of measurement [36].

Various physiological outcomes such as heart rate, muscle activity, stress, sleep patterns, temperature, level of mental attention, etc. can be obtained using noninvasive sensors attached to the body. Multiple measures can then be incorporated into miniature sensors to monitor a range of medical outcomes [37]. Multiple stakeholders, such as healthcare regulators or healthcare companies have invested a lot in order to come to the objective of data-based healthcare [38, 39]. In 2018, 172 million wearable trackers were sold worldwide, often combined with smartphone applications [40].

Mobility is one dimension of activities [1], which is affected by joint or muscle function, as well as the context studies, whereas physical activity is one aspect of health behavior affected by mobility itself [41]. Each of these aspects can be recorded by different types of sensors [34]. The different parameters underlying daily life activities can be recorded by the different components of inertial sensors. Physical activity can be measured by accelerometers or GPS in

terms of energy expenditure, the number of steps per day, or the time of active/sedentary behavior. Figure 6 shows the mobility outcomes studied in this thesis: spatio-temporal paremeters, joint angles and physical activity.

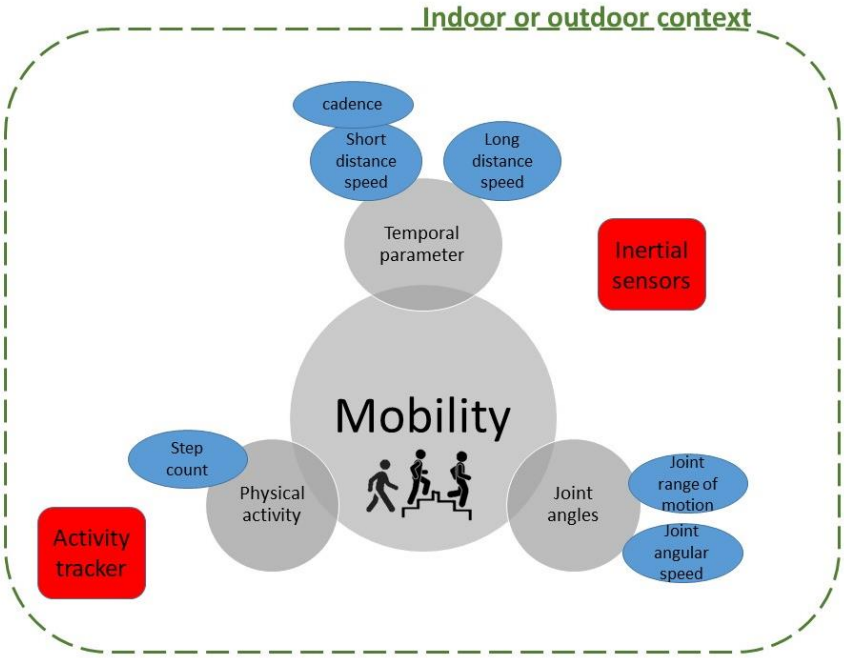

Figure 6: Mobility outcomes used in this thesis. Red rectangles are the sensors used to record the outcomes in the blue circles. Adapted from Lowe [34].

For decades, physical activity was assessed mainly by self-reported questionnaires. Nowadays, however, the same outcome is more easily accessible through smart watches or smartphones which are used worldwide for sport applications, as well as for health purposes. However, their validity is limited by the accuracy of the sensor and algorithms, and by real-world movement complexity [42]. Inertial sensors are a more recent technology and could be considered as immature, as they are not widely used so far. The

maturity of technology can be described using the technology readiness level (TRL), a scale developed by NASA during the 1970s [43]. TRLs are based on a scale from 1 to 9 with 9 being the most mature technology. This scale was adopted in particular by the European Commission in 2010 [44], and used in the Horizon 2020 program. For example, a TRL 4 means that the technology is validated in lab, while a TRL 5 means that the technology is demonstrated in the relevant environment. The complexity of inertial sensor use depends on the number of sensors, as well as the number of components used in the sensor fusion algorithms (accelerometer and gyroscope, combined or not with a magnetometer) [45]. In this work, we explored TRL 4 and 5 for joints angles measurement with inertial sensors.

Spatio-temporal parameters can be measured with only one sensor in the back [46], but require accurate event detection algorithms [47] as will be shown later in this thesis (Part II - Chapter 1). 3D joint angles measurement require multiple sensors attached to body segments [48], as well as sensor fusion algorithms to compensate for drift and accurate calibration procedures [49]. Inertial sensor technology is promising but most of the research papers focus on technology development, demonstration in laboratories and psychometric properties assessment for kinematic analysis [50]. Very few studies measure subjects in daily life situations. Demonstrations in relevant environment (TRL 6) are needed to test the potential of the inertial sensors for measuring valuable mobility outcomes for rehabilitation.

To achieve reliable measurements of daily-life mobility, we need to (1) further validate sensors that unobtrusively record movement relevant to rehabilitation, (2) design a system that integrates multiple sensors, with user-

friendly recording procedures and (3) develop a methodology to extract info from data in a clinically relevant manner and measure mobility outcomes in patients with conditions that limit their level of independence. A step towards this goal can be achieved by following an innovation process, starting from the actual knowledge of inertial sensors, to explore the potential of their use for a specific patient population with mobility problems.

Awareness of technology characteristics for clinical use

Measuring mobility, with or without technology, only makes sense in a clinical context. It is a means for clinicians to make more informed decisions. Several applications of mobility measurement exist, in both research and clinics. Large studies comparing disabled populations to normative values allow to define targets for rehabilitation in clinics. Then, those measures could help to identify the disability level and to choose appropriate treatments. Afterwards, clinical studies are needed to assess the efficacy of those interventions on the parameters measured. Finally those measures could also be used in fundamental research to enhance performance or to undertstand the mechanisms of gait deviation [51]. Regarding these applications, mobility measurements can provide important information for decision making in healthcare. However, there is a gap between the three different mobility outcomes, being subjective patient input regarding their mobility (PROM), the performance-based measures used in clinics and quantitative description of gait in a standardized laboratory. Moreover, performance measures need to fulfil some criteria to be useful in clinics for managing mobility limitations.

First, these measurements should have good psychometric properties [52]. Among these, validity refers to how accuractely a method measures what it is intended to measure. If a measure has high validity, that means it produces resultats that correspond to real variations. Reliability refers to how consistently a method measures something. If the same result can be consistently achieved by using the same methods under the same circumstances, the measurement is considered reliable. Granularity refers to the extent to which a measure is decomposed of different levels. Responsiveness denotes the ability of a tool to detect a change and interpretability refers to whether the user can confer clinical meaning to a score or a change in that score. Furthermore, the measurement tools have to be easy-to-use and non-expensive.

Second, several other criteria were proposed to attest the clinical utility of measures [53]: (1) the measure should not modify the movement as for example in laboratory, with the patient concentrating on what they are doing, (2) the measures are identifiable as a physical or physiological quantity, (3) the measures have to be correlated to the human potentials, (4) the measures should not be observable and quantifiable by the user of the results (e.g. the physiotherapist), and (5) the parameters should be clearly different for people suffering from a pathology compared to healthy subjects in order to be able to perform a diagnostic function. Apart from those qualities, the acceptability of measurement technology can be driven by therapist perceptions such as the effort expectancy (time/cost) or the performance expectancy (perceived value) [54]. It means that the measurement must be worth the cost in temporal or financial terms. Fast and low cost measurements, providing valuable outcomes are therefore needed to achieve translation from the laboratory to clinical practice. We used a step by step approach for the assessment of the

characteristics of the mobility outcomes are illustrated in Figure 7. Finally, we need a specific patient population with mobility problems to test the sensors and explore the clinical utility of mobility measures.

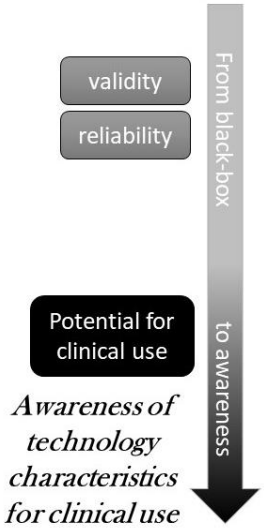

Figure 7: Step-by-step approach towards awareness of technology characteristics for clinical use. First, we assessed the validity and reliability of measurement systems. Second, we assessed the clinical differences of patients before and after treatments to determine the ability of systems to detect improvement or deteriotation of a patient's conditon

Knee osteoarthritis, a challenging context

Knee osteoarthritis (OA) is a growing chronic condition which is a major cause of difficulty in walking and climbing stairs for the elderly in Europe [55]. Treatment of OA is directed towards reducing joint pain and stiffness, maintaining and improving joint mobility, reducing physical disability and limitation of activities, improving health-related quality of life, limiting the progression of joint damage and educating patients about the nature of the disorder and its management. Non-pharmacological, pharmacological and surgical therapy are proposed depending on OA severity. Total knee arthroplasty is recommended for end-stage symptomatic osteoarthritis. In Belgium, a total of 23035 knee arthroplasties (TKA) were performed in 2016 [56]. In the US, the number is expected to increase significantly until 2030 [57]. Treatment choice depends on many factors, such as OA severity, health culture, patients' preferences, access to the health system, expected improvement [58, 59], etc. Nowadays, the main data used by clinicians to perform decisions are x-rays, clinical symptoms and known risk factors. Until now, few clinics have taken quantitative movement assessment into account.

The optimization of the health system regarding knee OA is of major importance, because of its increasing cost. Belgium has adopted the triple aim model which is a framework specifying that new designs must be developed to simultaneously pursue three dimensions: improving the patient experience of care (including quality and satisfaction); improving the health of populations; and reducing the cost of health care per capita [60, 61].

The assessment of patient experience and the impact of care on mobility in daily life is therefore needed to guide clinicians in shared-decision

making with the patients [62]. In knee OA, a substantial amount of research focuses on patient satisfaction. Seventeen percent of patients appears to be unsatisfied after receiving a TKA. However, the concept of satisfaction is not well defined and the patient's perception interacts with multiple dimensions. Satisfaction is modulated by expectancy [63], which are patients' beliefs about treatment results. Improvement of mobility in the activities of daily life is one of the most important expectancies of patients after treatment (*unpublished data – Thesis Marichal-Barrere*). The ability to climb stairs and to walk long distances are some of the highest expectancies encountered in consultation. However, their assessment remains limited in clinical practice.

Studies in laboratories show that patients suffering from knee OA present reduced walking speed, shorter stride length and a more prolonged stance phase of the gait cycle compared to healthy subjects [64]. They also have less range of motion at the hip, knee and ankle, joints and generate less muscular moments and powers at the ankle and more moments at the knee [64]. There are fewer studies in more demanding tasks [65]. It is recommended to record kinematics of joints adjacent to the knee as they can be affected as well [50]. Moreover, patients suffering from knee OA are often sedentary and treatments such as TKA do not change their physical activity level [66].

After a total knee arthroplasty, physical performance measures decrease initially to then increase in the long term; however, the perceived function measured with PROM does not follow the same trend. PROM display small responsiveness early, and excellent long-term responsiveness that is twice as large as performance measures. Patient perception fails to capture the acute

functional declines after TKA and may overstate the long-term functional improvement with surgery [4].

In clinics, mobility assessment in this population is limited to PROM and time-based performance outcomes. Developing tools for this population that provide a quantitative description of mobility and bridge the gap between localized evaluations (e.g. joint level) and subjective patient input regarding their mobility is therefore warranted. The achievement of data-based healthcare can only be performed if data on those outcomes are recorded [67].

It is necessary to measure and understand the impact of biomechanical adaptations on movement strategies if treatment goals are not to be limited to treating pain symptoms and focusing on mobility dimensions. As seen before, adequate measurement technologies for clinical use should be developed and tested.

Aims of this thesis

Given the lack of quantitative tools to measure mobility outcomes in ecological settings, and the specific functional limitations of patients suffering from knee OA, the implementation of ambulatory tools is the main topic of this thesis. The characteristics of the mobility outcomes recorded with different technologies are assessed in an exploratory research process from controlled conditions in laboratory settings to uncontrolled conditions in ecological settings (Figure 8).

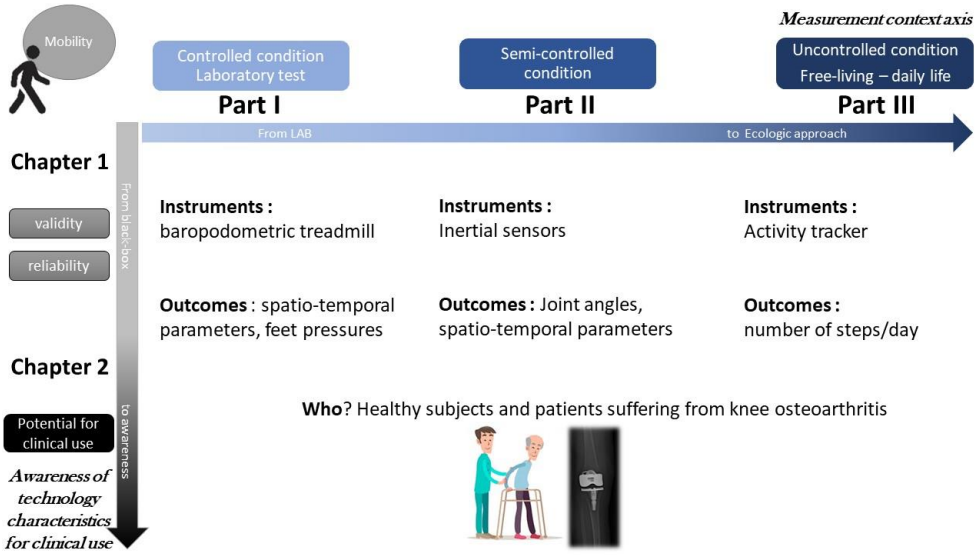

Figure 8: Road Map of this thesis. The x-axis represents the research process from laboratory settings to uncontrolled conditions. The y-axis represents the increase of awareness of technology characteristics for clinical use. The instruments and outcomes used in each context to measure the mobility of healthy patients and those suffering from knee OA are described as well.

This thesis is therefore subdivided into **three parts**, illustrating a research process from the lab to ecological settings. **Each part illustrates** the increase in awareness of technology characteristics for clinical use. We focus on instrument validation, and assessment of psychometric properties of mobility outcomes; after, we explore the potential of using locomotor outcomes for clinical use of healthy subjects and patients suffering from knee OA (Figure 9).

Part I focuses on the use of technology to measure the mobility of patients in a clinical condition **Chapter 1** We investigated the responsiveness

of spatio-temporal and posturography characteristics of gait on a treadmill after total knee arthroplasty.

Part II focuses on the use of technology to assess movement outside the laboratory, on a circuit, in semi-controlled settings. **Chapter 1** addresses the question of the validity of a commercial trunk accelerometer for measuring spatio-temporal parameters of gait. A second paper investigates the validity and reproducibility of different functional calibrations for an inertial measurement system to record 3D joint angles of the lower limbs. **Chapter 2** aims to assess the effect of pain intervention on 3D joint angles during locomotion in patients suffering from knee osteoarthritis in Cameroun.

Part III explores the use of sensors in free-living conditions. **Chapter 1** addresses the question of the validity of a commercial activity tracker to measure the number of steps. **Chapter 2** explores the continuous use of this activity tracker after arthroplasty during a telerehabilitation treatment.

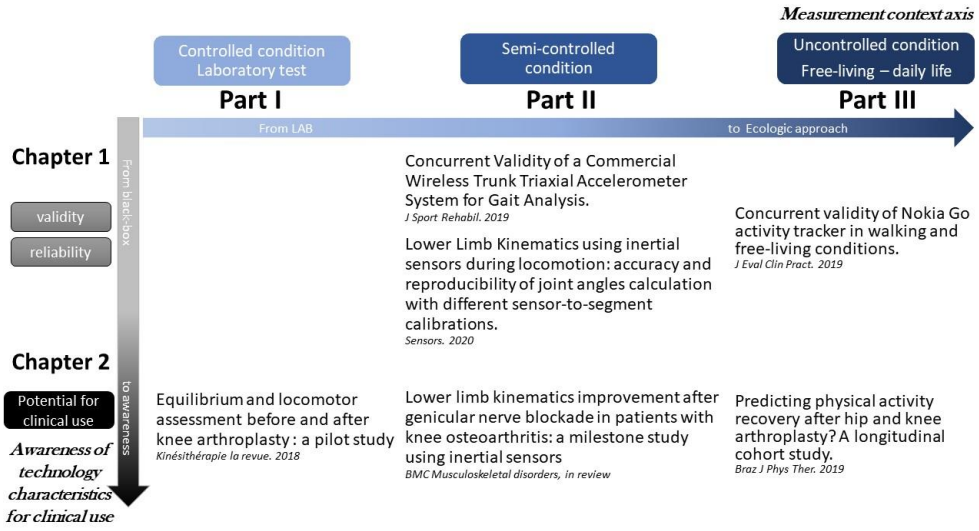

Figure 9: Thesis overview

Overview of the studies included in this thesis

Part	Chapter - article	Subject	Design	Data collection	n screened	n included	KOOS (mean (SD)/median [IQR])	Age (mean (SD))	Context
I	2	**Posturography** responsiveness after arthroplasty	Cohort pilot study – Case series	Prospective	OA:164	OA:9	OA: 35.5 (10.5)	65.9 (7.4)	Hospital
II	1-1	**Inertial sensor** validation for gait analysis	Cross-sectional	Prospective	HS: 30	HS: 30	NA	37.8 (19)	Laboratory
	1-2				HS: 7	HS: 7	NA	22.6 (1.5)	Laboratory
	2-1	Responsiveness after injection	Therapeutic pilot study	Prospective	OA: 14 HS: 10	OA: 14 HS: 10	OA: 50.25 [37.0-68.5]	OA:64.5 (11.3) HS: 50.6 (11.9)	Cameroon center
III	1	**Activity tracker** validation	Cross-sectional	Prospective	HS: 24 HS2: 60	HS1: 24 HS2: 60	NA	HS: 23.4 (1.3) HS2: 39.4 (12)	1:Outdoor on a circuit 2: 24h of daily life
	2	Responsiveness after artrhoplasty	Cohort study – Case series	Retrospective	OA: 146	OA: 132	KOOS: 49.6 (16.3) HOOS: 47.5 (17.3)	OA: 62 (10)	3 months of daily life

KOOS: Knee osteoarthritis outcome score; HOOS: Hip osteoarthritis outcome score; IQR: interquartile range; SD: standard deviation; NA: not applicable; OA: Osteoarthritis subjects HS: healthy subjects;

Part I. Controlled conditions, Laboratory test

Potential of using locomotor outcomes for clinical use

Equilibrium and locomotor parameters assessments in patients before and after total knee arthroplasty: A pilot study

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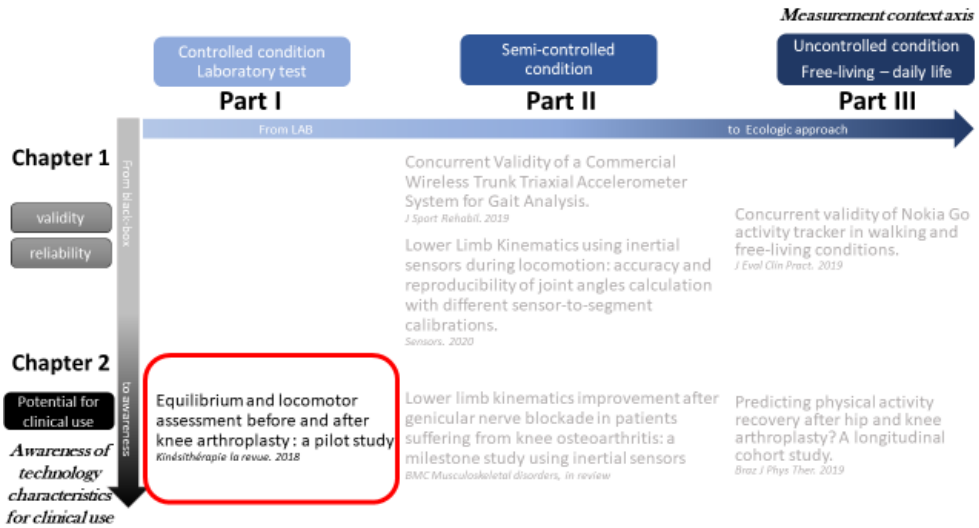

Equilibrium and locomotor parameters assessments in patients before and after total knee arthroplasty: a pilot study

Abstract

Objective. – Our objective was to analyze the feasibility and the need for the assessment of postural and locomotor capacities of patients before and after total knee arthroplasty, as well as the relationship between functionality and postural change.

Methods. – The assessment was divided in three times: one in pre-operative (T1), two in postoperative (T2, T3). Data were collected on 9 subjects by means of a baropodometric treadmill and self-administered questionnaires.

Results. – Refusal/exclusion rate amount to 95%. The main reason for refusal was the distance between home and the rehabilitation center. Only the walking cadence increased significantly between T1 and T3. The medial load on the operated side during walking increased in relationship to the functional score. Simultaneously, those improved between T1 and T3.

Discussion and conclusion. – The low participation rate is a barrier to other studies. The use of wearable technologies could compensate the main reason for refusal. None posturographic parameter evolved in the two months following arthroplasty. There were few relationships between functional score change and posturographic parameter change.

Keywords

Postural control; total knee arthroplasty; static balance; locomotor parameters;

Introduction

L'arthrose se caractérise par une dégénérescence progressive de l'articulation affectant le cartilage, les os et les tissus mous péri-articulaires [68]. L'arthrose des membres inférieurs, affectant principalement la hanche ou le genou, représente actuellement le trouble articulaire dégénératif le plus fréquent chez les personnes âgées. En Europe, plus de 20% de la population âgée de plus de 60 ans présentent une arthrose modérée à sévère[69], ceci devenant un problème majeur de santé publique.

L'arthrose sévère de genou génère des douleurs musculo-squelettiques et une gêne fonctionnelle importante réduisant la mobilité de l'articulation. Cette réduction est généralement associée à une faiblesse musculaire péri-articulaire et à une altération des fonctions motrices, diminuant progressivement l'indépendance fonctionnelle et la qualité de vie de la personne. Les capacités posturo-locomotrices [70] sont progressivement altérées augmentant le risque de chutes chez ces personnes [71]. En effet, les sujets atteints d'arthrose du genou marchent plus lentement, avec une longueur de pas diminuée et une cadence plus faible [72]. Ils présentent également des oscillations posturales plus importantes en statique les yeux ouverts et fermés que des sujets asymptomatiques quantifiées par une augmentation de la surface de l'ellipse de confiance et de la vitesse de déplacement du centre de pression (CdP) [73].

Un des traitements proposés afin de réduire au mieux la douleur et de rendre à l'articulation sa mobilité est l'arthroplastie totale de genou, plus communément appelée prothèse totale de genou (PTG). En Belgique, plus de 25000 PTG sont posées annuellement, classant la Belgique au 3ème rang

mondial en termes d'arthroplasties de genou par 100000 habitants. Par conséquent, une bonne prise en charge de ces personnes, avant opération mais également avec un suivi adéquat après cette opération, revêt une importance considérable. Après une PTG, les patients augmentent leur vitesse de marche par une augmentation de la longueur du pas et de la cadence [74, 75]. Cependant la plupart conservent malgré tout un schéma de marche asymétrique avec un temps d'appui unipodal plus court du côté opéré que du côté non-opéré [76].

Les capacités posturo-locomotrices telles que la station debout et la marche doivent donc être étudiées afin d'améliorer au mieux la prise en charge des patients bénéficiant d'une PTG. La vitesse de marche est un paramètre dont la mesure est facile et recommandée [6] mais cependant insuffisant pour décrire la fonction ambulatoire [16]. De nouveaux outils d'évaluation apparaissent en clinique tels que les tapis baropodométriques permettant de quantifier plus précisément la posture et les paramètres spatio-temporels de la marche.

À notre connaissance, seule l'étude de Quagliarella et al. [77] a conjointement évalué la posture à partir d'une plate-forme de forces et les capacités fonctionnelles au moyen d'un seul questionnaire lié à la pathologie à 6 mois et 1 an après PTG. Cette étude a montré peu de congruence entre l'amélioration fonctionnelle et les paramètres posturographiques dans le suivi de patients avec une PTG. Cependant, une évaluation intégrant la mesure objective non seulement de la posture mais également de la marche ainsi que l'ensemble des capacités fonctionnelles au moyen de plusieurs questionnaires

portant sur les domaines de la Classification Internationale du fonctionnement, du handicap et de la santé (CIF) pourrait donner des résultats différents.

C'est pourquoi nous nous sommes proposés de réaliser une étude pilote afin de déterminer i) la faisabilité de ce type d'évaluation en milieu clinique [78], ii) l'intérêt d'évaluer l'évolution des paramètres reflétant les capacités posturo-locomotrices de patients avant et après la pose d'une PTG, ainsi que iii) les relations entre les capacités posturo-locomotrices. Nous formulons l'hypothèse selon laquelle les paramètres posturographiques et spatio-temporels de la marche seront modifiés en post-opératoires et que ces modifications seront reliées à l'amélioration des capacités fonctionnelles.

Matériel et Méthodes

1. Patients

Le recrutement des patients a été réalisé entre août 2016 et avril 2017, soit par contact direct au Centre Reine Fabiola (CRF) à Montignies-sur-Sambre en Belgique soit par téléphone selon la liste proposée par le chirurgien de l'établissement. Au total, 185 patients ont été contactés afin de leur proposer de participer à l'étude se déroulant exclusivement au CRF (Figure 1).

Les participants devaient présenter un stade sévère d'arthrose de genou quantifié par radiographie, être prêts à recevoir une intervention de type PTG bénéficiant de la voie d'abord chirurgicale de type mini-invasive subvastus, avoir entre 60 et 85 ans et présenter une indication pour une PTG primaire.

Étaient non-inclus dans notre étude, tous les patients n'ayant pas suivi la rééducation dans sa totalité au CRF, présentant une PTG controlatérale, ou une prothèse totale de hanche, ou un traumatisme du genou et/ou ayant reçu

une chirurgie récente de moins de deux ans, ou nécessitant des semelles orthopédiques, ou un dispositif permanent d'assistance à la marche ou encore présentant une pathologie pouvant altérer la compréhension du protocole expérimental.

Tous les patients inclus (n=21) ont été informés oralement et par écrit du protocole expérimental à suivre et ont signé un consentement éclairé. Seuls 9 patients ont complété les 3sessions d'évaluation (Figure1).

Cette étude expérimentale interventionnelle a été approuvée par l'Université catholique de Louvain et par le Comité d'Éthique du Grand Hôpital de Charleroi.

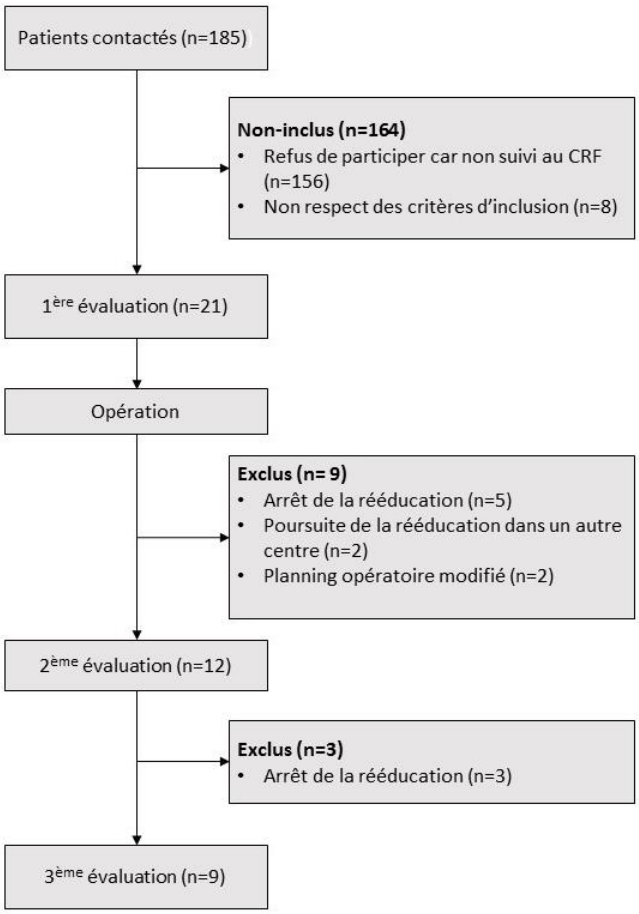

Figure 1 : Diagramme de recrutement des patients

2. Protocole expérimental

Tous les participants ont suivi un protocole de réadaptation standardisé au CRF. Ce protocole se composait de neuf séances de kinésithérapie en préopératoire et de trente séances au minimum en postopératoire.

Trois séances d'évaluation ont été réalisées à trois temps précis de la prise en charge. La première séance (T1) a été effectuée au début de la période

préopératoire, à savoir trois semaines avant l'opération, la deuxième séance (T2), dite à court-terme, ce qui correspond à une période entre deux et quatre semaines postopératoires selon la rééducation et la disponibilité du patient et enfin une troisième séance (T3), dite à moyen-terme, correspondant à une période entre deux et trois mois postopératoires (Figure 2).

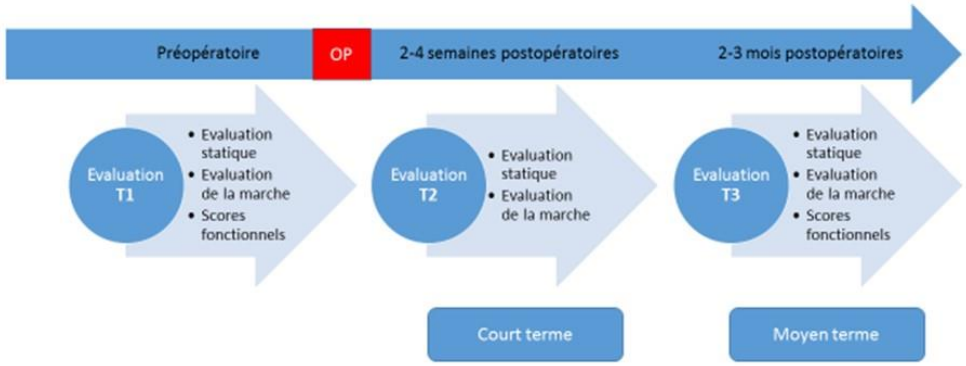

Figure 2: Déroulement temporel des trois séances d'évaluation de l'étude

3. Outil de mesure

Les évaluations posturographiques en statique ainsi que la marche ont été effectuées à l'aide d'un tapis baropodométrique de marche RunTime®, conçu par la société française CAPRON Podologie. Ce tapis roulant présente une surface de 120x40 cm munie de 24000 capteurs de pression permettant une analyse des déplacements du CdP et du pourcentage de mise en charge en statique et à la marche. Une évaluation fonctionnelle a également été réalisée à l'aide de questionnaires auto-administrés validés en français. Le fonctionnement physique a été évalué au moyen du « Knee Injury and Osteoarthritis Outcome Score » (KOOS), la qualité de vie générale au moyen du « Medical Outcome Study Short Form 12 » (SF-12), la peur du mouvement

au moyen du « Tampa Scale of Kinesiophobia » (TSK-CF) et les représentations cognitives et émotionnelles de la maladie au moyen du « Brief illness perception questionnaire » (B-IPQ).

4. Collecte des données

Posturographie en statique

L'analyse posturographique [79] permet, via une analyse de la trajectoire du CdP de recueillir un certain nombre d'indices rendant compte de la répartition des forces verticales appliquées au sol et de la stabilité du sujet. Elle consistait en un examen d'une durée de 51,2 secondes à une fréquence d'échantillonnage de 25 Hz afin d'étudier la trajectoire du CdP et la répartition des pressions plantaires.

Après avoir recueilli les données anthropométriques de chaque sujet comme la taille, le poids et la pointure de chaussure, il a été demandé au patient de se placer debout sur le tapis, pieds nus, le regard à l'horizontale, les bras le long du corps et les pieds disposés à sa convenance, à savoir de la manière la plus confortable pour lui [80]. Les perturbations externes (bruits, passages, échanges) ont été limitées au maximum par la demande du silence sur le lieu des recueils de mesures.

Les paramètres calculés à partir de la trajectoire du CdP ont été la longueur totale du déplacement du CdP (mm), la position moyenne du CdP en antéro-postérieure (Xmoy) et en médio-latérale du CdP (Ymoy), la vitesse de déplacement du CdP (mm/s), la surface de l'ellipse de confiance (mm2) correspondant à la dispersion de 90% des positions successives du CdP (exluant 10% des points extrêmes quel que soit leur instance d'occurrence

durant l'enregistrement) [79, 81] et le ratio longueur/surface des déplacements du CdP.

Les paramètres suivants ont été calculés à partir de la répartition de la charge du corps sur chaque pied (exprimées en % du poids corporel): la charge totale du corps répartie à la fois sur les pieds du côté prothésé et sain et sur l'avant et l'arrière de chaque pied (Figure 3).

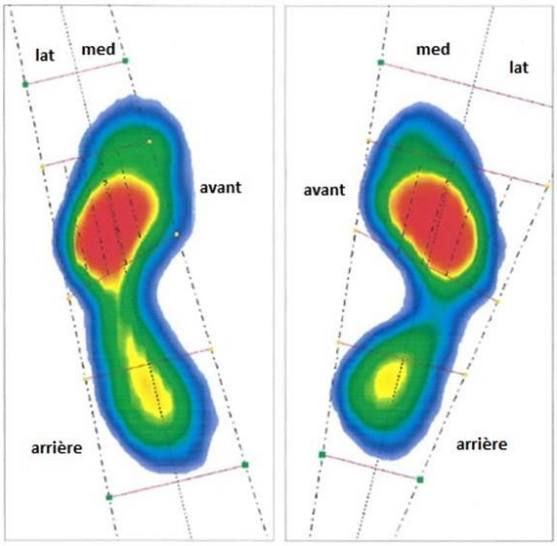

Figure 3 : Présentation schématique de la répartition des pressions sous les deux pieds utilisée pour mesurer la mise en charge

Paramètres spatio-temporels de la marche

Les différents paramètres ont été mesurés lors d'une marche sur le tapis baropodométrique à vitesse spontanée calculée précédemment lors d'un test de 4x10 mètres (10mWT). Une minute de marche sans enregistrement a été laissée au patient afin qu'il puisse s'habituer au tapis roulant. La durée d'enregistrement était de 20 s à une fréquence d'échantillonnage de 200 Hz.

Les paramètres obtenus ont été les suivants : vitesse de marche (km/h), cadence (pas/minute), longueur du pas gauche et droit (cm), temps d'appui et oscillant gauche et droit (ms), temps de double appui, charges (% du poids corporel) sous l'avant, l'arrière, la partie médiale et la partie latérale de chaque pied.

Evaluation fonctionnelle

Une évaluation fonctionnelle a également été réalisée à T1 et T3

5. Analyses statistiques

Le logiciel SPSS (version IBM SPSS Statistics 24.0) a été utilisé pour réaliser les analyses statistiques. Une ANOVA à mesures répétées à un facteur (effet temps) a été réalisée. En présence d'un effet principal, un test Post-Hoc de Bonferroni ($p<0.017$) a été effectué afin de déterminer le sens de la différence des modifications et entre quelle(s) période(s) s'effectuai(en)t ces changements. Dans le cas de données se distribuant selon la loi normale (34 variables), les résultats ont été présentés en terme de moyenne et écart-type. Pour les données ne se distribuant pas selon la loi normale (5 variables), une analyse de variance sur les rangs a été effectuée et les résultats ont été présentés en terme de médiane et quartiles (Tableau II).

L'évolution entre T1 et T3 des tests fonctionnels a été déterminé par un test de t-pairé.

De plus, une matrice de corrélation de Spearman a été réalisée afin d'évaluer la relation entre l'évolution des scores fonctionnels (Δ T3-T1) et des paramètres calculés lors de l'évaluation sur tapis (Δ T3-T1).

Résultats

Parmi les 185 patients contactés, seuls 21 ont accepté de participer à l'étude soit 11% de la population initiale. Le principal motif de refus était dû à la distance entre le domicile et le centre ainsi que la fréquence de visites nécessaires, à savoir trois fois. Douze patients (/21) ont par la suite abandonné l'étude donnant un taux d'abandon de 57% (Figure 1). Aucun incident (e.g. malaise, chute, etc.) n'a été observé lors des évaluations.

Tableau I : Caractéristiques des patients

	(n=9) Moyenne (SD)
Age (années)	65,9 (7,4)
Taille (cm)	165,4 (5,6)
Poids (kg)	83,2 (8,7)
BMI (kg/cm^2)	30,3 (3,2)
Satisfaction personnelle du traitement (/10)	8,6 (1,2)
Activités physiques (heures/semaine)	6,4 (5,8)
Genre (homme/femme)	1/8
Statut (retraité/actif)	8/1
Fumeur (oui/non)	1/8
Comorbidités (oui/non)	8/1
Traumatisme (oui/non)	1/8
Antécédents d'opérations du genou (oui/non)	3/6
Douleurs pluriarticulaires (oui/non)	9/0
Rééducation préalable pour le genou (oui/non)	3/6
Antécédents infiltrations (oui/non)	9/0

SD : déviation standard, BMI : Body mass index (indice de masse corporelle)

1. Posturographie en statique

Les paramètres enregistrés en position statique ne présentent aucun changement significatif au cours du temps. La surface de l'ellipse de confiance (mm2) présente une tendance à l'augmentation à T2 (100.8 (78.9)) par rapport à T1 (50.1 (55.8)) et T3 (56.8 (57.9)) mais reste non significative. Consécutivement le ratio longueur/surface montre une diminution à T2 (5.6 [3.2-24.9]) par rapport à T1 (12.5 [6.0-53.2]) et T3 (26.7 [6.3-28.6]) sans présenter cependant de différences statistiquement significatives (Tableau II).

Tableau II : Evolution de l'équilibre posturographique en statique

	T1	T2	T3
	Moyenne (EC) ou Médiane [25-75%]		
Analyse sur 51,2 secondes			
Longueur du tracé (mm)	454.9 (94.6)	501.7 (113.1)	514.1 (126.8)
Surface de l'ellipse de confiance (mm2)	50.1 (55.8)	100.8 (78.9)	56.8 (57.9)
Ratio longueur/surface	12.5 [6.0-53.2]	5.6 [3.2-24.9]	26.7 [6.3-28.6]
Vitesse moy (mm/s)	9.0 (1.9)	10.1 (2.8)	10.3 (2.6)
Xmoy (mm)	13.9 (13.4)	2.5 (15.4)	9.4 (13.5)
Ymoy (mm)	-4.7 (5.9)	-4.9 (5.1)	-5.1 (3.4)
Charge globale jambe opérée (%)	48.0 (9.8)	49.2 (6.2)	49.8 (7.1)
Charge globale jambe non opérée (%)	52 (9.8)	50.1(6.2)	50.2 (7.1)
Charge avant-pied jambe opérée (%)	50 [46.5-52]	46 [43-50.5]	46 [42-49.5]
Charge avant-pied jambe non opérée (%)	48.3 (8.9)	47.6 (6.4)	48.7 (4.4)
Charge arrière-pied jambe opérée (%)	50.7 (4.5)	53.2 (4.3)	54.0 (3.8)
Charge arrière-pied jambe non opérée (%)	51.7 (8.9)	52.8 (6.4)	51.7 (4.5)

EC : ecart-type, [25-75%] : intervalle interquartile 25-75%

2. Paramètres spatio-temporels de marche

L'étude des paramètres spatio-temporels de la marche a montré une augmentation significative de la cadence (pas/minute) entre T1(96.8 (24.2)) et T3 (112.6 (5.2)) (Tableau III).

Par ailleurs, la vitesse spontanée (km/h) montre une tendance à l'augmentation entre T1 (2.9 (0.9)) et T3 (3.3 (0.5)) mais cela reste non significatif.

Quant au temps d'appui (ms) des deux côtés, ils montrent une tendance à la diminution entre T1 (côté opéré (OP) : 719 [647-1082], côté non opéré (NOP) : 812 (233)) et T3 (OP : 670[632-751], NOP : 681 (53)) sans présenter de différences statistiquement significatives. Alors que ceux-ci diffèrent d'environ 100 ms à T1, ils sont quasiment identiques à T3.

Tableau III : Evolution des paramètres spatio-temporels de marche

	T1	T2	T3
	Moyenne (EC) ou Médiane [25-75%]		
Vitesse (km/h)	2.9 (0.9)	2.8 (0.8)	3.3 (0.5)
Cadence (pas/minute)	**96.8 (24.2)***	99.4 (19.3)	**112.6 (5.2)***
Longueur du pas jambe opérée (cm)	52.02 (5.9)	48.5 (7.5)	52.7 (6.2)
Longueur du pas jambe non opérée (cm)	52.3 (6.5)	48.8 (7.7)	51.2 (6.1)
Temps appui jambe opérée (ms)	719 [647-1082]	721 [628-870]	670 [632-751]
Temps appui jambe non opérée (ms)	812 (233)	786 (159)	681 (53)
Temps de double appui (ms)	183 (73)	163 (48)	146 (25)
Temps oscillant jambe opérée (ms)	410 [387-612]	415 [400-518]	410 [387-435]
Temps oscillant jambe non opérée (ms)	421 [403-591]	454 [411-481]	415 [393-423]
Charge avant-pied jambe opérée (%)	75.4 (7.1)	74 (6.9)	74 (4.3)
Charge avant pied jambe non opérée (%)	75.3 (4.9)	75.1 (6.4)	74.7 (3.8)
Charge arrière-pied jambe opérée (%)	24.9 (7.6)	25.9 (6.7)	25.7 (4.3)
Charge arrière-pied jambe non opérée (%)	24.4 (4.6)	25.2 (6.8)	25.4 (4.1)
Charge médiale jambe opérée (%)	48.8 (5.2)	49.6 (4.9)	51.2 (4.4)
Charge médiale jambe non opérée (%)	50.3 (2.7)	52.3 (5.6)	52 (3.2)
Charge latérale jambe opérée (%)	50.3 (4.4)	50.4 (4.9)	48.8 (4.4)
Charge latérale jambe non opérée (%)	49.6 (2.7)	47.7 (5.6)	48.7 (3.2)

Données extraites sur une moyenne de 35 empreintes; EC : ecart-type; [25-75%] : intervalle

Scores d'évaluation fonctionnelle

Tous les scores s'améliorent significativement entre T1 et T3 sauf la sous-catégorie mentale du SF-12 (Tableau IV).

L'étude des corrélations entre les évolutions des scores fonctionnels (Δ T3-T1) et des paramètres calculés sur tapis (Δ T3-T1) montre trois corrélations significatives entre l'évolution du pourcentage de charge sur le côté médial du pied opéré à la marche et l'évolution du KOOS (r=0.80) (Figure 4), le TSK (r=-0.76) et le B-IPQ (r=-0.739). Cela signifie que l'augmentation de la charge du côté médial évolue avec une amélioration du fonctionnement physique (KOOS), une diminution de la peur du mouvement (TSK) et une meilleure représentation cognitive et émotionnelle de la maladie (B-IPQ).

Tableau IV : Evolution des scores fonctionnels

	T1 Moyenne (EC)	T3 Moyenne (EC)	T3-T1 Différence
SF-12 phys. (points /100)	30,3 (4,2)	39,9 (8,1)	9,62**
SF-12 mental (points /100)	38,4 (9,6)	43,7 (12,2)	5,25 NS
KOOS symptômes (%)	42,1 (11,4)	62,7 (16,1)	20,6*
KOOS douleur (%)	46,3 (14,2)	71,3 (19,5)	24,9*
KOOS fonction (%)	39,7 (14,6)	75,2 (20,9)	35,5**
KOOS activités (%)	5,6 (9,8)	32,8 (27,1)	27,2*
KOOS QoL (%)	16,3 (17,9)	52,8 (25,6)	36,5*
KOOS total (%)	35,5 (10,5)	65,5 (17,5)	29,9*
TSK-CF (/68)	47,7 (6,1)	39,9 (9,6)	-7.8*
BIPQ (/80)	45,7 (5,3)	35,2 (12,0)	-10,5*

ecart-type, [25-75%], T1 : évaluation pré-opératoire, T3 : évaluation post-opératoire (entre 2 et 3 mois), SD : déviation standard, NS : non significatif, * : $p<0.05$, ** : $p<0.01$

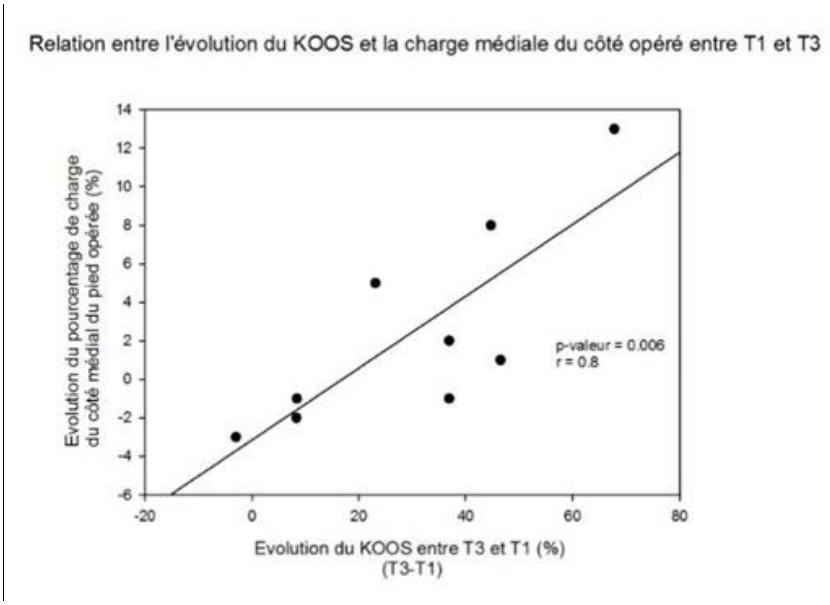

Figure 4 : Relation entre l'évolution du KOOS et la charge médiale du côté opéré entre T1 et T3

Discussion

Notre étude a montré i) un faible taux de participation rendant une étude de plus grande ampleur peu faisable dans les conditions actuelles, ii) aucun changement postural en statique et une augmentation de la cadence à 3 mois postopératoire, (iii) ainsi qu'une relation entre un changement postural et les capacités fonctionnelles.

Faisabilité

Le faible taux de participation (11%) à l'étude par rapport aux personnes contactées est un obstacle à la réalisation d'une étude sur un nombre important de patients de l'analyse des capacités posturo-locomotrices de patients recevant une PTG à court et moyen termes. Cela s'explique notamment par des refus dû à la distance entre le domicile et le centre ainsi que la fréquence de visites nécessaires, à savoir trois fois. En effet, la population fréquentant le centre provient de zones géographiques parfois très éloignées et le transport s'est avéré difficile pour beaucoup d'entre eux. Le taux d'abandon élevé s'explique notamment par un arrêt du traitement ou un changement de thérapie.

Cependant, la proximité géographique du lieu d'évaluation par rapport au lieu de rééducation du patient (CRF) a permis aux évaluateurs de limiter au maximum le temps consacré au recueil des données (déplacement et passation d'une durée d'environ 45 minutes). En dernier lieu, il faut noter que l'évaluation s'est passée sans complication pour les 9 patients ayant suivi l'étude jusqu'à son terme. En conclusion, ces résultats semblent donc être un frein à la mise en place d'une étude de plus grande ampleur dans ces conditions.

Evolution des paramètres posturo-locomoteurs en postopératoire

Aucun paramètre postural évaluant l'équilibre statique n'a changé significativement au cours du temps.

En ce qui concerne les paramètres spatio-temporels de la marche, seule la cadence a augmenté significativement entre T1 et T3. En revanche, d'autres auteurs n'ont pas retrouvé cette augmentation à 2 mois postopératoires [74] mais l'ont observé seulement à un an ou deux ans postopératoires [74, 82]. Dans ces études, ces patients adoptaient une vitesse de marche réduite associée à une diminution de la longueur du pas et une augmentation du temps de double appui tandis que la cadence augmentait avec un pas plus court mais seulement à partir d'un an postopératoire. Ces différences observées entre nos résultats et ces deux études [74, 82]. pourraient s'expliquer notamment par des différences entre les protocoles de réadaptation en terme de précocité de mise en place d'exercices actifs, de rééducation à la marche ou d'intensité des exercices [83].

Quant au temps d'appui, on note un retour à un équilibre du ratio entre le temps d'appui du côté opéré et du côté non opéré, pouvant illustrer une marche symétrique à 3 mois postopératoire [84]. Ces résultats ne concordent pas avec ceux d'autres études telle que celle conduite par Casartelli et al. [85]. Cette étude réalisée sur 23 patients à 6 mois en postopératoire a montré que les patients opérés d'une PTG marchaient avec une phase d'appui plus courte comparativement au côté non opéré. Cependant, ils n'avaient pas évalué les patients en préopératoire ne permettant pas de comparer les patients au fil du temps, ni évalué l'intensité de la réadaptation suivie ne permettant pas de comparer un potentiel effet de la thérapie.

Relations entre capacité fonctionnelle, posturographie statique et paramètres spatio-temporels de la marche

Les scores fonctionnels se sont nettement améliorés entre l'évaluation préopératoire et l'évaluation à 3 mois, caractérisant une meilleure capacité fonctionnelle de nos 9 patients dans leur vie quotidienne. Cette évolution contribue probablement de manière importante à la satisfaction globale du patient, facteur qui est de plus en plus recherché en orthopédie [86].

Parmi tous les paramètres calculés, seule la charge médiale sous le pied du côté opéré a été corrélée à l'évolution des scores fonctionnels. Malgré notre petit échantillon, cela signifie que les sujets ayant le mieux récupéré ont également modifié la répartition de leur poids lors de la marche, mettant plus de poids du côté interne sur le pied opéré. Cependant, cette observation est difficile à interpréter sans données cinématiques des membres inférieurs, notamment dans le plan frontal [87].

Excepté cette corrélation, nos résultats ne sont donc pas différents de l'étude de Quagliarella et al [77] montrant peu de congruence entre la douleur et les paramètres posturographiques. Celui-ci avait également remarqué que les résultats posturaux de 71% des patients opérés d'une PTG se trouvaient dans la normalité. Cela signifie que la posturographie ne permet pas à elle seule de mettre en évidence les améliorations de patients opérés d'une PTG.

En clinique, la fonctionnalité du patient est actuellement mesurée par des variables comme la vitesse de marche, la distance maximale de marche ainsi que l'observation clinique de la boiterie [88]. Ces évaluations sont largement recommandées en complément des scores fonctionnels pour

caractériser des aspects du fonctionnement physique non capturés par des échelles subjectives [4]. Le tapis baropodométrique a le potentiel d'apporter des informations complémentaires permettant d'adapter la prise en charge. Cependant, l'intérêt des variables apportées par ce dispositif doit pouvoir être perçu par les cliniciens de terrain pour envisager une implémentation de ce dispositif. Masui et coll. [73] ont pu objectiver des différences d'oscillations posturales entre patients arthrosiques et sujets asymptomatiques. Cependant, le lien entre le contrôle postural, l'arthrose du genou, la diminution de force musculaire, de proprioception ou la présence d'instabilité est complexe et nécessite d'autres études plus spécifiques.

Perspectives

Les paramètres spatio-temporels de la marche peuvent être objectivés avec d'autres outils, notamment au moyen de centrales inertielles qui sont des technologies innovantes, portables et peu coûteuses [89]. De plus, ces technologies présentent l'avantage de pouvoir être utilisées au domicile du patient et de pallier le premier motif de refus de participation à l'étude, à savoir l'éloignement du centre d'évaluation. C'est donc peut-être au clinicien-chercheur d'effectuer le déplacement vers le patient pour envisager un plus fort taux de participation.

Les performances posturales sont souvent étudiées en examinant le maintien de l'équilibre en station debout, sans mouvement. L'étude des corrections posturales dans des tâches plus complexes comme le maintien de l'équilibre unipodal ou les réactions suite à une perturbation soudaine de l'équilibre pourraitent permettre de dégager des variables plus sensibles représentant l'évolution des patients. D'ailleurs, des études antérieures

semblent démontrer l'efficacité des exercices d'équilibre par des perturbations chez les patients atteints d'arthrose de genou ainsi que chez les patients opérés d'une PTG [90, 91]. Cependant, les paramètres utilisés tels que la vitesse de marche, le temps d'appui sur une jambe ou le temps pris pour se lever d'une chaise ne reflètent qu'une partie des mécanismes d'équilibration mis en jeu dans les activités quotidiennes des individus [92]. D'autres études sont nécessaires pour étudier l'effet d'une rééducation spécifique et adaptée sur les mécanismes d'équilibration dans des conditions dynamiques tels que les parcours de type Posture-Equilibre-Mouvement [93] qui sont de plus en plus proposés aux personnes âgées [94].

Limites

Le choix des paramètres de posturographie a été limité par le matériel utilisé. D'autres variables, plus sensibles, comme la variabilité de la vitesse de déplacement du CdP [95] auraient pu mettre en évidence des changements dans notre échantillon. Nous nous sommes également limités à des tâches simples afin de garder une évaluation sans danger et réalisable. La répétition des évaluations posturographiques est recommandée pour assurer une bonne fiabilité [96]. Cependant, il nous a fallu rester réaliste avec la disponibilité des patients et des évaluateurs en routine clinique.

Par ailleurs, le déséquilibre de la représentation des genres (l'échantillon est composé de 8 femmes pour 1 homme) peut constituer un biais. À la suite d'une PTG, les femmes présentent généralement un niveau de douleur plus élevé que les hommes [97]. Ceci peut donc avoir des conséquences sur la biomécanique de marche. Cette répartition est cependant représentative de la population atteinte d'arthrose de genou, étant

donné que les femmes sont les plus touchées par cette pathologie dégénérative [98]. L'âge élevé de la population arthrosique peut également avoir eu une influence sur l'hétérogénéité des paramètres stabilométriques. En effet, l'âge joue un rôle très important sur les mécanismes de l'équilibre et plus particulièrement sur de nombreux changements physiologiques des différents systèmes sensoriels comme par exemple les afférences visuelles ou encore vestibulaires qui participent énormément au contrôle de l'équilibre [99].

Conclusion

En conclusion, le faible taux de participation semble un frein à la réalisation d'une étude de ce type à plus grande échelle. Une perspective pour améliorer le taux de participation pourrait être de réaliser les évaluations à domicile avec des technologies portables. Aucun paramètre posturographique n'a évolué tandis que la cadence s'est améliorée dans les trois mois postopératoire d'une PTG. Il semble également y avoir peu de liens entre les scores fonctionnels et les paramètres objectifs issus de l'évaluation sur tapis hormis la charge médiale du côté opéré.

Part II. Semi-controlled settings - Circuit

Chapter 1. Instrument validation and psychometrics properties

Article 1

Concurrent validity of a commercial wireless trunk tri-axial accelerometer system for gait analysis

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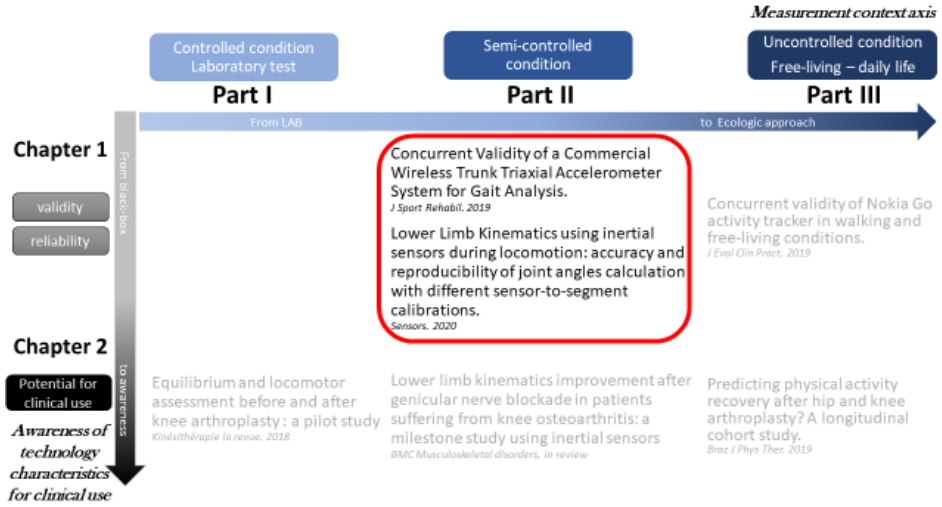

Concurrent validity of a commercial wireless trunk tri-axial accelerometer system for gait analysis

Abstract

Context: Wearable sensor devices have notable advantages, such as cost effectiveness, easy-to-use and real-time feedback. Wirelessness ensures full body motion which is required during movement in a challenging environment such as during sports. Research on the reliability and validity of commercially available systems, however, is indispensable.

Objective: To confirm the test-retest reliability and concurrent validity of a commercially available body worn sensor - BTS G-WALK® sensor system - for spatio-temporal gait parameters with the GAITRite® walkway system as golden standard.

Design: Reliability and concurrent validity study.
Setting: Laboratory setting.
Participants: Thirty healthy subjects.
Main outcome measures: Spatiotemporal parameters: speed, cadence, stride length, stride duration, stance duration, swing duration, double support, and single support.

Results: In terms of test-retest reliability of the BTS G-WALK® sensor system, ICC values for both the spatial and temporal parameters were excellent between consecutive measurements on the same day with ICC-values ranging from 0.85 – 0.99. In terms of validity, ICC values between measurement systems showed excellent levels of agreement for speed, cadence, stride length, and stride duration (range 0.88 – 0.97). Poor to moderate levels of agreement (range 0.12 – 0.52),

however, were found for single/double support and swing/stance duration. Bland Altman plots showed overall % bias values equal to or smaller than 3% with limits of agreement ≤ 15% (speed, cadence, stride length, stride duration, swing duration and stride duration). Only for single and double support, the limits of agreement were higher with respectively -15,4 to 19.5% and -48.0 to 51.4 %.

Conclusion: The BTS G-WALK® sensor system is reliable for all measured spatio-temporal parameters. In terms of validity, excellent concurrent validity was shown for speed, cadence, stride length and stride duration. Cautious interpretation is necessary for temporal parameters based on final foot contact (stance, swing and single/double support time).

Introduction

Wearable sensor devices have notable advantages, such as cost effectiveness, easy-touse and real-time feedback. Wirelessness ensures full body motion which is required during movement in a challenging environment such as during sports. These technology systems allow clinicians to benefit from data gathered during the performance of everyday activities or sports and data recorded under controlled conditions in clinical settings[36]. For gait analysis, such wearables are potentially useful for assessing abnormal gait and evaluating the effectiveness of rehabilitation approaches and therapeutic interventions.

It has been demonstrated that spatio-temporal gait parameters can be determined during overground walking using only one tri-axial trunk accelerometer [100]. However, to overcome some associated critical issues as the need for gravity compensation and the presence of drift error in the position data, data fusion of linear acceleration (accelerometer) and angular velocity (gyroscope) combined in an inertial measurement unit (IMU) permits compensation [47]. Notwithstanding the closer the IMU is positioned to the point of contact (e.g. on the shank) the better gait events can be correctly detected[101], lower trunk acceleration patterns have been consistently associated with initial and final foot contact[102]. Recently, industry transfers technology development made in research[103] to implement it in clinical routine. For example, the G-WALK is a user-friendly device, without post-processing handling needed to obtain the gait parameters. This aspect is really important to facilitate technology adoption[104]. However, before using them for clinical interpretation, we need to define their reliability and validity. The latter can be done by comparing the spatio-temporal parameters of gait

obtained from the IMU to a golden standard, such as the GAITRite® system with proven reliability and validity of spatio-temporal gait parameters[105].

The purpose of this study was to confirm the test-retest reliability and concurrent validity of a commercially available body worn sensor - BTS G-WALK® sensor system – for spatio-temporal gait parameters in a healthy population with the GAITRite® walkway system as golden standard.

Methods

Subjects

Thirty healthy subjects (15 males, 15 females, mean age of 37.8 years with 20-56 range) volunteered to participate in this study. The average weight and height of the subjects is 73.8 ± 16.84 kg and 173.6 ± 9.42 cm, respectively. Subjects had to be healthy and were excluded if they reported to have any musculoskeletal, neurological or systemic pathology potentially affecting their gait. All participants were informed about study procedures prior to signing an institutionally approved informed consent. This study was approved by the ethical committee of the Ghent University (n°: BE670201526917).

Experimental approach and Materials

The BTS G-WALK® (BTS Bioengineering S.p.A. Italy) inertial sensor was placed on the lower back with the centre of the device at the fifth lumbar vertebrae (L5) (Fig 1). The inertial platform is equipped with 4-Sensor Fusion technology and consists out of a tri-axial accelerometer 16bit/axes (8g), a tri-axial magnetometer 13 bit (1200uT), a tri-axial gyroscope 16 bit/axes (250°/s) and a GPS Receiver. The BTS G-WALK® sensor system determines spatio-temporal parameters as well as pelvic rotations during gait and other

movements realtime through a Bluetooth® 3.0 connection (G-Studio® software). For the current research question only spatio-temporal parameters were used for further analysis.

The GAITRite® walkway system used for this study measured 7.03 m x 0.89 m with an active sensor area of 6.13 m x 0.56 m. The sensors are triggered when mechanical pressure is applied at a used sampling rate of 100 Hz. Data is collected by on-board processors for a connected computer to receive through a serial port (GAITRite® Software).

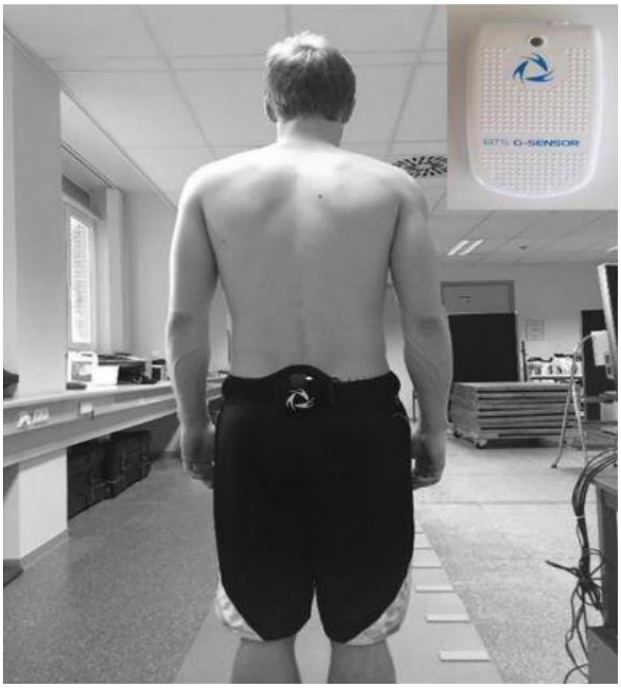

Figure 1

Procedures

Each subject performed 5 walking trials on the GAITRite® sensor system. All subjects wore their own comfortable and non-restricting clothing. Every individual was instructed to walk towards the end of the walkway mat at their own everyday pace. During each trial, both the GAITRite® sensor system and the BTS G-WALK® sensor system (G-Studio® software) recorded equivalent parameters to be analyzed and compared. The parameters that were registered by both systems were speed (m/s), cadence (steps/min), stride length (m), stride duration (sec), stance duration (% of gait cycle (GC)), swing duration (%GC), double support (%GC), single support (%GC).

Data analysis and statistics

Based on comparison with the GAITRite® sensor system, the following methods were used to determine the reliability and validity of the BTS G-WALK® sensor system. For unilateral (left and right) outcome measures only the data from the right side was included for analysis. Normality of the data was verified using the Shapiro-Wilk test. The reliability of the BTS G-WALK® sensor system outcome measures across the five walking trials were examined using Intra Class Correlations (two-way mixed, absolute agreement). For establishing validity, means and standard deviations were calculated for each parameter for both sensor systems over the 5 walking trials. Then, Intra Class Correlations between the two methods were calculated. A paired sample T-test was performed to determine systematic differences between the two systems. Subsequently, Bland-Altman plots with 95% limits of agreement (LoA) (mean difference ± 1.96 SD) were generated to visualize the degree of agreement between the BTS G-WALK® sensor system and the

GAITRite® sensor system measurements. Based on the Bland Altman plots the percentage bias was calculated with the GAITRite system as reference standard (((mean G-Walk – mean GAITRite)/ mean GAITRite)*100). Statistical significance for all tests was determined at the 5% level.

Results

Inter trial reliability

Reliability of all spatio-temporal gait variables recorded by means of the BTS GWALK® sensor system across the 5 trials is presented in Table 1. Overall, excellent inter trial reliability (ICC values between 0.84 and 0.99) is shown.

Concurrent validity

Table 1 also displays comparative data for both gait analysis systems. ICC values showed excellent levels of agreement for speed, cadence, stride length, and stride duration (range 0.88 – 0.97). Poor to moderate levels of agreement (range 0.12 – 0.47), however, were found for the relative temporal parameters that divide the gait cycle in phases (single/double support and swing/stance duration). Paired t-tests revealed that there were no significant systematic differences for speed single and double support ($p>0.05$). Comparison for cadence ($p<0.001$), stride length ($p=0.031$) and stride ($p=0.0012$)/swing ($p=0.009$)/stance ($p=0.009$) duration did show a significant difference between devices (table 1). However, Bland Altman plots showed overall % bias values equal to or smaller than 3% with limits of agreement ≤ 15% (speed, cadence, stride length, stride duration, swing duration and stride duration) (Table1; Fig 2). Only for single and double support, the limits of

agreement were higher with respectively 15,4 to 19.5% and -48.0 to 51.4 % (Table1; Fig 2).

Table 1: Reliability of G-walk and Validity outcome measures between G-walk and Gaitrite measuring systems.

	Reliability	Validity						
				Paired sample T-test				
	ICC G-walk (95%CI)	G-walk Mean (SD)	GAITRite Mean (SD)	Mean diff (95%CI)	p-value	Bias %	LoA%	ICC (95%) G-walk vs GAITRite
Speed (m/s)	0.99 (0.98-0.99)	1.42 (0.21)	1.44 (0.17)	-0.02(-0.06:0.2)	0.277	-1.6	-15.1; 0.4;	0.92 (0.83-0.96)
Cadence (st/min)	0.98 (0.97-0.99)	115.3 (1.2)	113.0 (6.6)	2.3 (2.0;2.7)	<0.001	2.1	3.8	0.96 (0.02-0.99)
Stride length (m)	0.99 (0.98-0.99)	1.49 (0.19)	1.53 (0.14)	-0.04 (-0.08)	0.031	-2.9	-15.7; 9.8	0.88 (0.73-0.94)
Stride duration (sec)	0.84 (0.73-0.92)	1.05 (0.06)	1.06 (0.06)	-0.02 (-0.03)	0.012	-1.3	-6.7; 4.1	0.93 (0.83-0.97)
Single support (%GC)	0.91 (0.85-0.95)	38.7 (3.2)	37.9 (1.3)	0.7 (-0.5; 2.0)	0.231	2	-15.4;	0.18 (-0.70-0.60)
Double support (%GC)	0.89 (0.80-0.94)	11.0 (2.5)	11.0 (1.5)	0.02 (-1.0;1.1)	0.972	1.7	-48.0;	0.12 (-0.92-0.59)
Swing duration	0.85 (0.75-0.92)	39.3 (2.3)	38.1 (1.3)	1.1 (0.3; 1.9)	0.009	3	-8.5; 14.4	0.47 (-0.04-0.74)
Stance duration	0.85 (0.75-0.92)	60.8 (2.3)	61.8 (1.3)	-1.1 (-1.9;-0.3)	0.009	-1.8	-8.6; 5.0	0.47 (-0.36-0.74)

SD=standard deviation; CI=confidence interval; ICC=intraclass correlation coefficient; GC=gait cycle

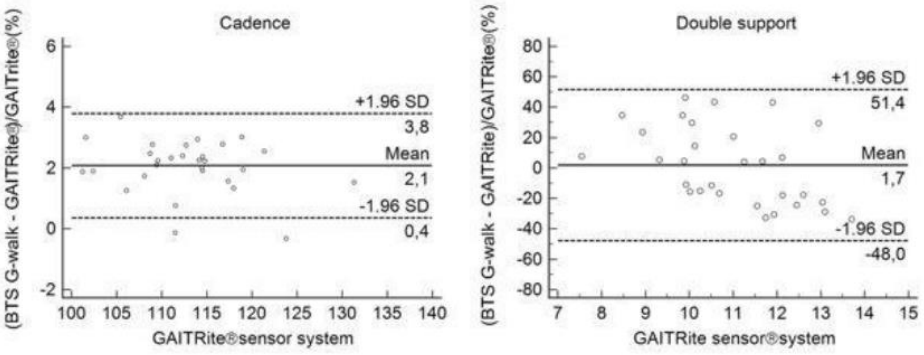

Figure 2

Discussion

In terms of test-retest reliability of the BTS G-WALK® sensor system, ICC values for both the spatial and temporal parameters are excellent between consecutive measurements on the same day with ICC-values ranging from 0.85 – 0.99. This signifies that BTS G-WALK® sensor system is reliable for all measured parameters.

Concerning the validity, the results require some careful considerations. First of all, paired sample T-test did show significant differences between the two measurement devices for cadence, stride length and stride/swing/stance duration; with no differences for speed, single and double support. However, when taken the % bias into consideration, the overall % bias was equal to or lower than 3%, which is within clinically acceptable limits. Speed, cadence, stride length and stride duration showed excellent ICC values between test devices with Limits of Agreement for % bias ≤ 15%. For swing and stance duration termination moderate ICC values between devices were moderate to good, also with Limits of Agreement for % bias ≤ 15%. Poor ICC values with Limits of Agreement for % bias ≥15%, however, were observed for single and double support. This corresponds with Trojaniello et al. (2014) who evaluated several methods of single IMU's mounted on the lower trunk and showed an acceptable accuracy, sensitivity and robustness for temporal parameters based on the identification of the initial foot contact, i.e. step and stance duration [106]. The accuracy was lower for parameters based on final foot contact (stance, swing and double support time) as larger errors in event determination of final foot contact were observed. To improve validity, future research should focus on ameliorating IMU algorithms for identifying this final foot contact.

Limitations of the current study are that we are unaware of the precise algorithm used by this commercial IMU, which prohibits us from presenting a potential underlying explanation for the reported results. Furthermore, we measured spatio-temporal parameters at a rather homogenous speed in our sample and speed might have an impact on accuracy [107]. Spatiotemporal parameters were registered only over a short distance in a controlled setting and we only included healthy subjects. Reliability and validity (specific parameters) results suggest results that the BTS G-WALK® sensor system might be used in community and athletic settings or clinical studies evaluating treatment effects, however, research in these specific populations is warranted.

In conclusion, the BTS G-WALK® sensor system is a reliable IMU for all measured spatio-temporal parameters. In terms of validity, cautious interpretation is necessary for temporal parameters based on final foot contact (stance, swing and single/double support time). Excellent concurrent validity was shown for speed, cadence, stride length and stride duration.

Concurrent validity of a commercial wireless trunk tri-axial accelerometer system for gait analysis

Article 2

Lower Limb Kinematics using inertial sensors during locomotion: accuracy and reproducibility of joint angles calculation with different sensor-to-segment calibrations.

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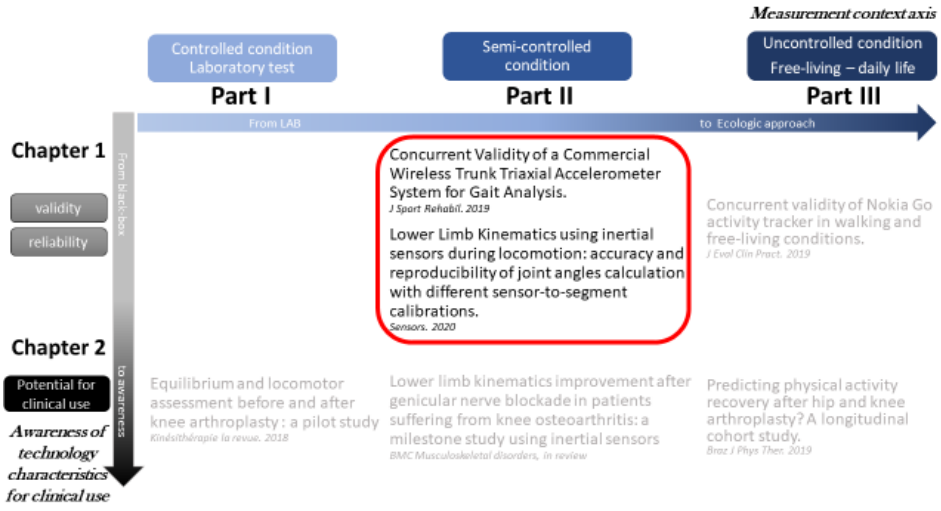


Lower Limb Kinematics using inertial sensors during locomotion: accuracy and reproducibility of joint angles calculation with different sensor-to-segment calibrations.

Abstract:

Inertial measurement unit (IMU) records of human movement can be converted into joint angles using a sensor-to-segment calibration, also called functional calibration. This study aims to compare the accuracy and reproducibility of four functional calibration procedures for the 3D tracking of the lower limb joint angles of young healthy individuals in gait. Three methods based on segment rotations and one on segment accelerations were used to compare IMU records with an optical system for their accuracy and reproducibility. The squat functional calibration movement, offering a low range of motion of the shank, provided the least accurate measurements. A comparable accuracy was obtained in other methods with a root mean square error below 3.6° and an absolute difference in amplitude below 3.4°. The reproducibility was excellent in the sagittal plane (intra-class correlation coefficient (ICC) > 0.91, standard error of measurement (SEM) < 1.1°), good to excellent in the transverse plane (ICC > 0.87, SEM < 1.1°), and good in the frontal plane (ICC > 0.63, SEM < 1.2°). The better accuracy for proximal joints in calibration movements using segment rotations was traded to distal joints in calibration movements using segment accelerations. These results encourage further applications of IMU systems in unconstrained rehabilitative contexts.

Keywords: inertial sensor; gait; validity; functional calibration; accuracy; wearable electronic devices

Introduction

Quantitative assessment of lower limb kinematics is required in various applications, such as motion analysis, sports science, and rehabilitation. Although opto-electronic motion capture systems are considered the gold standard for this assessment, their widespread use is limited by their restricted area of measurement, their optical limitations due to marker occlusion or reflection, and their cost. Moreover, opto-electronic trackers are generally used in a restricted lab environment, which further limits the exploration of real-life movements and exercises. Wearable sensors, such as inertial measurement units (IMUs), have been developed to overcome these limitations [108], allowing for human motion analysis in unconstrained real-life conditions [20, 109].

Although they generally contain a 3D accelerometer, a 3D gyroscope, and an optional 3D magnetometer, an IMU does not measure joint angles perfectly. Joint angles obtained via signal integration typically drift over time [45, 110] and their accuracy varies with the joint assessed and the movement complexity [50, 111]. While an error under 5° is generally accepted for most clinical gait applications [112], the measurement error typically ranges from 5° to 18.8° depending on the joint and the plane of motion [111]. Another major challenge in IMU-based human motion analysis that the IMUs' local coordinate systems are not aligned with physiologically meaningful axes [113]. Such alignment, required to compute joint angles, can be performed via a "sensor-to-segment" calibration procedure [114].

The first approach to ensure this alignment consists of a rigorous positioning of the sensor in relation to the anatomy [115]. This method

assumes that the segment axes are parallel to the IMU axes, is approximate, and requires user expertise to locate the sensor axes relative to the joint axes for both segments around each joint. The second approach consists of placing an IMU on each segment and aligning the IMU and joint axes via a set of calibration postures [116] and/or movements around physiological motion axes. The latter functional method [117, 118] consists of making the subject stand upright with straight legs for a few seconds to define the vertical axis for each IMU or segment, while the other axes are defined via active or passive movements [117, 118] . Since the movements are generally human-controlled, the accuracy of the axes definition essentially relies on the subjects' ability to precisely hold a given posture and on the execution of a given movement [113]. The third approach consists of exploiting the kinematic constraints of the joints and use almost arbitrary movements to perform the sensor-to-segment calibration [113, 119]. This method is particularly adapted to single axis joints that can be satisfactorily modelled as a hinge joint like the knee; however, the modeling of spherical joints requires the execution of movements mostly around one axis to identify the joint axes [113], which resemble the functional method.

The IMU-based tracking of lower limb spherical joints, using one of the two aforementioned methods, therefore requires the execution of a functional calibration movement. While the accuracy of the sensor-to-segment calibration is determined by the quality of the functional calibration movement, to our knowledge, it has only been investigated in one study for upper limb motion tracking [120] and no study has compared functional calibration movements for the tracking of lower limb joint angles.

This study aimed to (1) assess the accuracy of different functional calibration methods in order to compute the lower limb joint angles during walking, (2) assess the reproducibility of different functional calibration movements, and (3) compare the accuracy provided by functional calibration movements in different gait movements.

Material and Methods

1. Participants

Seven healthy young adults participated in this study (6 females, 1 male, mean (SD) age = 22.6 (1.5) years, height = 1.67 (0.08) m, body mass = 65.4 (11.6) kg). Participants were included in the study if they were between 20 and 25 years old and free of any injury at the time of participation. The study protocol was approved by the ethics committee of our university (agreement number: B403201523492) and each patient provided written informed consent to the use of their anonymized data.

2. Experimental Setup and Recordings

To assess the lower limb joint kinematics, seven wearable IMUs; (x-IMU, x-io Techologies, Bristol, UK) were fixed in matched 3D-printed ABS (acrylonitrile butadiene styrene) enclosures and attached by means of a semi-elastic belt to seven lower body segments, as shown in Figure 1: the waistline at the level of the fifth lumbar vertebra (L5), the middle of the thighs, the middle of the shanks, and at the dorsal side of the feet. The IMUs were firmly strapped on the skin or clothes. Although this could lead to undesirable artifacts, it is more representative of records in an unconstrained context, such as outdoor conditions. Each IMU included a tri-axial accelerometer (full scale ±6 g), a

gyroscope (±2000 °/s), and a magnetometer (±8.1 G) that were sampled at a frequency of 128 Hz. The IMUs were connected to a computer by means of a Bluetooth connection using a custom application based on open source software [121] (C# program, github.com/xioTechnologies). Each movement was recorded independently. The synchronization between the IMUs was ensured by a custom-built magnetic coil that sent a magnetic impulse at the beginning of each recording.

Four reflective markers were fixed at each corner of the ABS enclosures to define clusters for each segment (Figure 1). Motion capture data were collected at a rate of 200 Hz using an eight-camera motion analysis system (Vicon V5 Motion Systems, Oxford Metrics Ltd,. Oxford, UK)) and processed using Nexus 2.5 software. The position of each marker on the cluster allowed for the orientation of each segment to be computed in the lab reference frame.

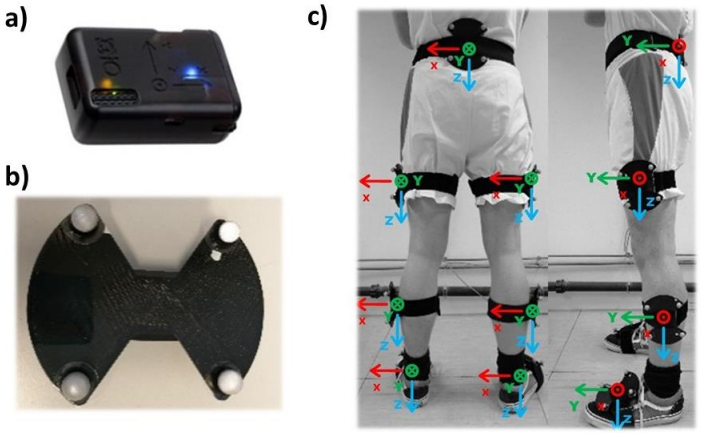


Figure 1. Sensor locations: (a) x-IMU from xi-o Technologies, (b) 3D-printed acrylonitrile butadiene styrene (ABS) enclosures (4 markers of 14 mm diameter) with the inertial measurement unit (IMU) reference frame, and (c) segment reference frame of the 7 IMUs on the subject.

3. Functional Calibration and Test Movements

The experimental protocol is illustrated in Figure 2. Four functional calibration movements were performed to assess their reproducibility and accuracy regarding lower limb joint angle measurement with the IMUs. These movements were designed to include a rotation in the sagittal plane of each lower body segment, including the pelvis, while being easy to explain and reproduce. Each functional calibration movement included (1) an upright static posture with the arms alongside the body and the feet parallel beside each other that was used to define the segment vertical axis and (2) a functional movement spanning a range of orientations for each segment in the sagittal plane that was used to define a second segment axis. In the static posture, the segment was supposed to have a zero angle in all three planes such that the segment reference frames were aligned with the lab reference frame. The X axis was defined as the medio-lateral axis, pointing to the left of the subject; the Y axis as the anterio-posterior axis, pointing in front of the subject; and the Z axis as the vertical axis, pointing downward (Figure 1c).

The instructions were as follows:

- Calibration movement 1: "Tilted to stand": Start in a leaned-back position with extended legs on the chair, bend the knees, bend the trunk forward, get up from the chair, and stop moving.
- Calibration movement 2: "Extension stand up": Sit on the chair, extend the knees in front of you, bend the knees, bend the trunk forward, get up from the chair, and stop moving.
- Calibration movement 3: "Squat": Stand in front of the chair, rise on your heels, squat deeply, get up, and stop moving.

- Calibration movement 4: "Walking" 5 m:
- 4a. Walk at your pace to the red line on the floor, then stop moving.
- 4b. Fast: walk five meters as fast as you can without running to the red line on the floor as if you were late to a meeting.
- 4c. Slow: walk five meters slowly to the red line on the floor but keep moving.

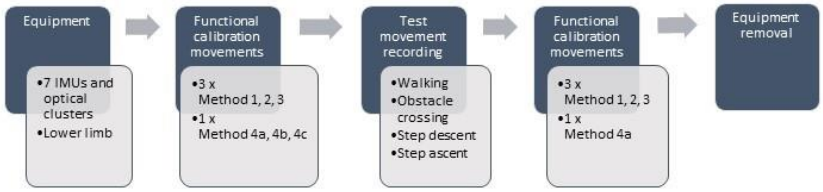
The functional calibration movements were demonstrated by the operator and each participant received practice trials to get used to each movement. Each functional calibration movement was recorded three times before and three times after the execution of the test movements. The walking movement at self-selected speed was only performed two times, before and after the test movements.

Four test movements were performed in the same order:

- walking five meters at a self-selected speed;
- stepping over an obstacle 28 cm in height while walking 5 m at a self-selected speed;
- ascend a step 20 cm in height;
- descend a step 20 cm in height.

The mean recorded times for test movements were 11 s for walking, 11 s for stepping over an obstacle, 14 s for the step ascent, 16 s for the step descent.

a) Experimental protocol



b) Functional calibration movements

Segment-rotation-based



Calibration movement 1 : Tilted to stand



Calibration movement 2 : Extension



Calibration movement 3: Squat

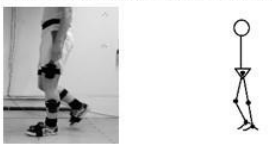
Segment-acceleration-based



Calibration movement 4a : Walking



Calibration movement 4b : Walking **fast**

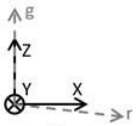


Calibration movement 4c : Walking **slow**

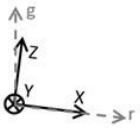
c) Options to determine segment reference frame

Frontal view of the segment axis

Option 1 : Z axis is aligned with gravity (g) and X axis is the closest to the rotation axis (r)



Option 2 : X axis is aligned with the rotation axis (r) and Z axis is the closest to gravity (g)



d) Joint angle and accuracy indicators calculation

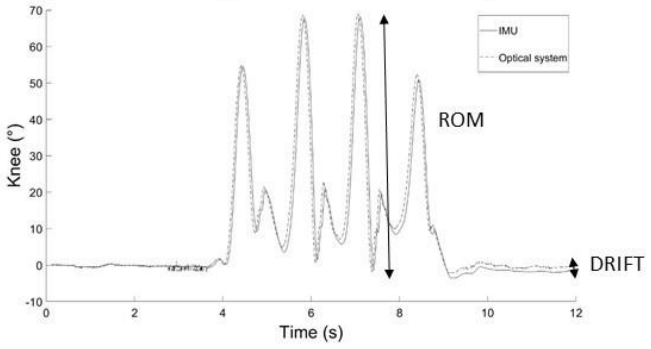


Figure 2. (a) Experimental protocol. (b) Functional calibration movements: for movements 1, 2, and 3, where the second axis was defined as the principal rotational axis as determined by a principal component analysis (PCA) on gyroscope signals; for movements 4a, 4b, and 4c, the second axis was defined as the principal acceleration axis through a PCA on accelerometer signals. (c) Two options to determine the segment reference frames, as shown in segment frontal views. (d) The accuracy was assessed using the root mean square error (RMSE), the absolute difference in the range of motion (ΔROM) between both systems, and the absolute drift accumulated during the movement (DRIFT).

4. Signal Processing

An open source attitude and heading reference system (AHRS) algorithm was used for sensor fusion between the accelerometer and gyroscope sensor data of the IMU (Mahony's AHRS algorithm) [122]; the magnetometer signals were omitted. The four calibration movements were used to compute the orientation of each segment relative to the lab reference frame in different ways. The gravity vector during the static upright posture was used to define the vertical axis for each segment. A second segment axis was defined in one of two ways depending on the method of functional calibration. For functional calibration movements 1, 2, and 3, it was defined as the principal rotational axis as determined by a principal component analysis (PCA) on gyroscope signals. Two options were used to determine the segment reference frame (see frontal views of the segment reference frame in Figure 2c): either the gravity vector (g) was defined as the vertical axis and the lateral axis was forced to be the orthogonal axis closest to the rotation axis of the functional calibration movement (r), or the lateral axis was defined as the functional calibration movement rotation axis and the vertical axis was the orthogonal axis closest to the gravity vector (and thus the transversal plane was not perfectly horizontal in

the static upright posture). For functional calibration movements 4a, 4b, and 4c, the second axis was defined as the principal acceleration axis through a PCA on accelerometer signals transformed in a lab-fixed reference frame. The 3D orientation of the pelvis and joint angles for the hips, knees, and ankles were calculated from the segment orientations based on the recommendations of the International Society of Biomechanics [123] for the different functional calibration movements. Flexion-extension were rotations around the X axis, abduction-adduction was around the Y axis, and internal-external rotations were around the Z axis.

The lower body 3D kinematics derived from the optical system were computed in two different ways. They were either computed in the lab frame or computed through the same functional calibration procedures described above, using the principal axis of rotation or acceleration determined from optical records.

For each participant, a static period (about 5 s) in a standing position was captured at the beginning of each test to define the segment's initial orientation for the IMU AHRS algorithm.

5. Data Analysis

Joint angles of the walking test movement were calculated for all functional calibration procedures. The accuracy of the IMU kinematics was computed for each calibration procedure as the difference in joint angle between the IMU and optical measurements. The accuracy was assessed using the root mean square error (RMSE) during the movement period, the absolute difference in the range of motion (ROM) between both systems (ΔROM), and

the absolute drift accumulated during the movement due to the error in the angular rate integration (DRIFT). The RMSE, ΔROM, and DRIFT parameters were computed using Matlab 2018 (Mathworks Inc, Natick, MA, USA) and are expressed in degrees.

A generalized linear model was used to assess the effect of (1) the functional calibration movement, (2) the option used to determine the segment's reference frame to compute the IMU orientation, and (3) the functional calibration method for the optical system on the amplitude of the RMSE, ΔROM, and DRIFT parameters for each joint angle and plane of motion. This analysis was performed with SPSS (version 25, IMB Corporation, Amonk, NY, USA) and the significance level was set to $\alpha = 0.05$.

The reproducibility of each functional calibration movement was assessed as the difference in joint angle computed from each repetition of the functional calibration movement. The reproducibility of the ROM parameter in all movement planes and joints was determined based on the intra-class correlation coefficient (ICC) [124] and standard error of measurement (SEM) [124]. Values of ICC ≥ 0.90 were considered as excellent, 0.70–0.89 as good, 0.40–0.69 as acceptable, and <0.40 as low [125]. The SEM estimates the non-systematic variance and reflects the within-subject variability among repeated calibrations. A proportional SEM (SEM%) was calculated by expressing the SEM relative to the mean ROM (SEM% = (SEM/mean) × 100%)) [126]. An SEM% above 10% was considered as high.

Once the most accurate and reproducible functional calibration method was selected for a walking test, the accuracy was determined for the other test movements, namely the step ascent, descent, and stepping over an

obstacle, using the RMSE and ΔROM parameters. The parameters were calculated for the front leg (i.e., the first leg to touch the step in the step ascent, the first leg to touch the floor leg in the step descent, and the first leg to touch the ground in obstacle stepping) and for the back leg in the different test movements. Differences in the RMSE and ΔROM parameters between test movements were assessed with a one-way ANOVA. Tukey's post hoc test was used to reveal which groups differed in the case of significant p-values. The significance level was set to $\alpha = 0.05$.

Results

The pelvis orientation and the joint angles during a time-normalized typical gait stride at a self-selected speed of 1 m·s−1 are illustrated in Figure 3 for measurements with the IMUs and with the optical system using the walking functional calibration (Calibration movement 4a). The traces displayed a classical movement pattern as indicated by the similarity between both set of measurements, with a mean RMSE value lower than 5° for all joints in all planes.

1. Assessment of the IMU Accuracy

The mean differences between measurements from the IMUs and from the optical system, as well as the factors affecting this difference, are presented in Table 1. The highest mean error was observed at the knees and at the ankles, bilaterally, as shown by the RMSE between 3.0° and 4.1° and by the ΔROM between 1.9° and 5.1° in any plane of motion. The generalized linear model also indicated that these differences were significantly linked to the functional calibration movement; see the details in Figure 4. These errors

were also significantly associated with the methods of determination of the IMUs reference frame, although to a lesser extent: the mean (SD) RMSE was 2.3° (1.7°) when the vertical axis was aligned to the gravity vector and 3.0° (3.1°) when the lateral axis was aligned to the segment rotation axis. The errors were not significantly linked to the two different methods used to compute the 3D kinematics from the optical system (lab frame or computed through the functional calibration). Table 1 also shows that the DRIFT was largely independent of the factors considered in the generalized linear model.

The variation of the RMSE across different functional calibration movements is illustrated in Figure 4. While the RMSE varied between approximately 1° and 6° for most joints, planes of motion, and functional calibration movements, it was larger for the "squat" calibration movement, especially at the knee and ankle where the upper confidence limit of the RMSE reached up to 8° and the upper confidence limit ΔROM reached up to 10° or more. Tilted, extension, and walking functional calibration movements tended to provide less accurate measurements at the knee. Walking functional calibration movements tended to report more accurate angles for distal lower limb joints, although without a clear visible impact on the ROM accuracy. After excluding the "squat" movement, the mean accuracy for all other calibration movements is summarized in Table 2, indicating a mean RMSE of less than 2° at the pelvis, less than 3° at the hip and ankle, and less than 4° at the knee with a trend for larger errors in the frontal plane. The mean errors were smaller in ΔROM. DRIFT values at the hips and at other lower limb joints in the sagittal plane were on average under 2.7°, while higher mean values up to 4.9° were observed at the knee and at the ankle in the frontal and transverse planes.

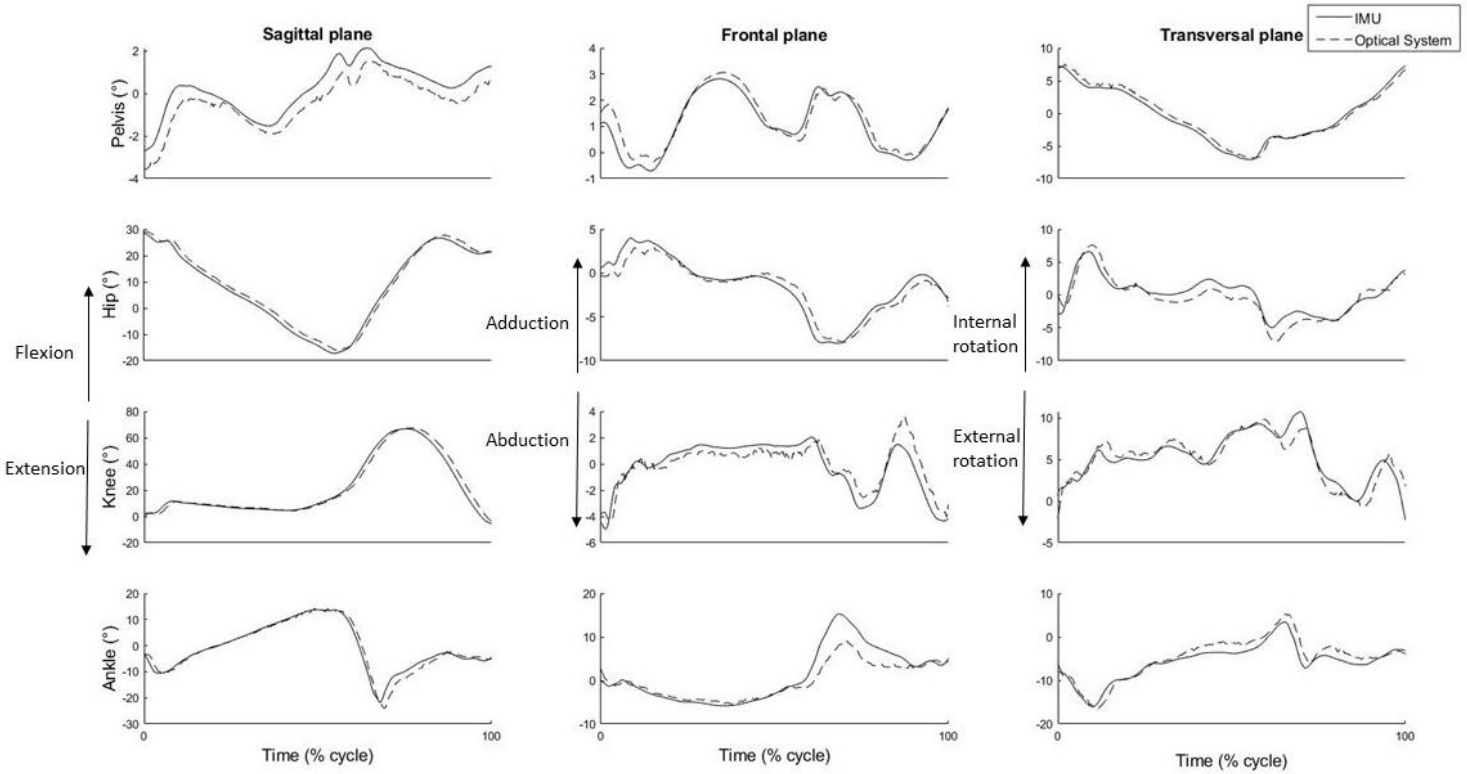


Figure 3. Typical trace of the 3D pelvis orientation and joint angles during a gait stride for measurements with the IMUs (plain lines) and with the optical system (short dash lines) using the walking functional calibration.

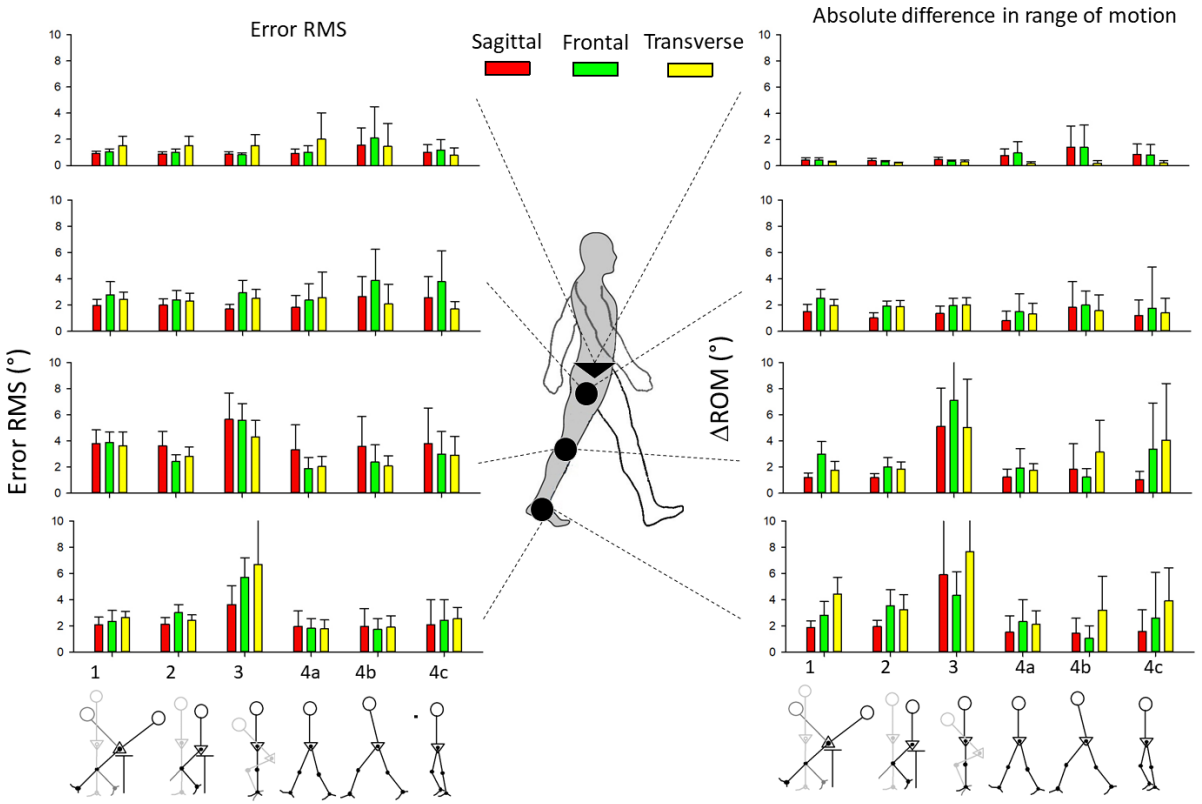


Figure 4. Accuracy across functional calibration movements (1. Tilted to stand, 2. Extension, 3. Squat, 4a. Walking (1 ± 0.1 ms−1), 4b. Walking fast (1.5 ± 0.1 ms−1), 4c. Walking slow (0.6 ± 0.1 ms−1)). Error bars are the confidence interval means at 95%.

Table 1. Generalized linear model: effect of functional calibration movement, segment reference frame, and optical functional calibration on the accuracy.

Accuracy Indicator		**RMSE (°)**				**ΔROM (°)**				**DRIFT (°)**			
Calibration Factor			**Functional Calibration Movement**	**Segment Reference Frame**	**Optical Functional Calibration**		**Functional Calibration Movement**	**Segment Reference Frame**	**Optical Functional Calibration**		**Functional Calibration Movement**	**Segment Reference Frame**	**Optical Functional Calibration**
		Mean (SD)	Mean difference between methods			Mean (SD)	Mean difference between methods			Mean (SD)	Mean difference between methods		
Pelvis	Sagittal	0.9 (0.5)	-	-	-	0.4 (0.6)	1.0	-	-	0.4 (0.6)	-	-	-
	Frontal	1.1 (0.9)	1.3	-	-	0.5 (0.7)	1.1	-	-	0.5 (0.7)	-	-	-
	Transverse	1.5 (1.8)	-	-	-	0.2 (0.2)	-	-	-	0.2 (0.2)	-	-	-
Hip right	Sagittal	2.0 (1.2)	-	-	-	1.6 (1.6)	-	1.0	-	1.6 (1.6)	-	-	-
	Frontal	2.7 (2.1)	-	-	-	2.3 (1.5)	-	-	-	2.3 (1.5)	-	-	-
	Transverse	2.4 (1.5)	-	-	-	2.2 (1.3)	-	-	-	2.2 (1.3)	-	-	-
Knee right	Sagittal	**4.1 (3.1)**	-	1.5	-	2.4 (3.8)	**3.9**	-	-	2.4 (3.8)	1.2	-	-
	Frontal	**3.6 (2.3)**	**4.0**	-	-	**4.7 (6.4)**	**9.2**	-	-	**4.7 (6.4)**	-	-	-
	Transverse	**3.3 (2.1)**	-	2.4	-	**3.2 (5.2)**	**4.3**	**3.1**	-	**3.2 (5.2)**	-	-	-
Ankle right	Sagittal	2.5 (1.7)	-	0.9	-	2.7 (5.1)	-	-	-	2.7 (5.1)	-	-	-
	Frontal	**3.3 (2.5)**	**3.7**		-	**4.2 (5.6)**	**7.2**	-	-	**4.2 (5.6)**	-	-	0.5
	Transverse	**2.4 (4.3)**	**4.3**	2.6	-	**5.1 (6.6)**	**5.6**	2.8	-	**5.1 (6.6)**	2.0	1.0	-
Hip left	Sagittal	2.5 (0.8)	-	-	-	0.8 (0.7)	-	-	-	0.8 (0.7)	-	-	-
	Frontal	2.1 (1.6)	-	-	-	1.9 (1.7)	-	-	-	1.9 (1.7)	-	-	-
	Transverse	2.1 (1.6)	-	-	-	2.0 (1.8)	1.9	0.6	-	2.0 (1.8)	-	-	-
Knee left	Sagittal	**3.7 (3.5)**	-	1.9	-	1.9 (3.6)	-	-	-	1.9 (3.6)	-	-	-
	Frontal	**3.5 (1.7)**	1.9	-	-	**4.3 (4.0)**	**4.9**	-	-	**4.3 (4.0)**	-	-	-
	Transverse	**3.0 (1.6)**	-	1.4	-	2.7 (3.9)	3.0	2.6	-	2.7 (4.0)	-	-	-
Ankle left	Sagittal	**3.2 (1.5)**	-	-	-	2.3 (4.0)	**4.6**	2.2	-	2.3 (4.0)	-	-	-
	Frontal	2.9 (1.6)	2.8	-	-	**3.6 (3.9)**	**3.8**	**-**	-	**3.6 (3.9)**	-	-	-
	Transverse	**3.2 (4.3)**	**4.8**	2.4	-	**3.6 (5.1)**	**6.7**	2.0	-	**3.6 (5.1)**	-	-	-

Numerical values indicate significant effects, dashes indicate non-significant effects. Mean differences between calibration factors are presented in degrees for significant effects ($p < 0.05$) and in bold for angles >3°. For the functional calibration movement factor, the highest mean difference between the six movements is presented. RMSE: root mean square error, ΔROM: absolute difference in range of motion, DRIFT: absolute difference between IMU and optical system at end of recording, SD: standard deviation.

Table 2. Summary of the accuracy without the squat calibration movement and with reference frames aligned with gravity.

		RMSE (°)	ΔROM (°)	DRIFT (°)
			Mean (SD)	
Pelvis	Sagittal	1.0 (0.7)	0.7 (0.8)	0.4 (0.6)
	Frontal	1.2 (1.1)	0.7 (0.9)	0.5 (0.4)
	Transverse	1.5 (1.8)	0.2 (0.2)	0.2 (0.2)
Hip right	Sagittal	2.1 (1.3)	1.0 (1.0)	1.6 (1.6)
	Frontal	2.9 (2.2)	2.0 (1.7)	2.3 (1.6)
	Transverse	2.2 (1.5)	1.7 (1.1)	2.1 (1.3)
Knee right	Sagittal	**3.6 (2.5)**	1.1 (0.6)	1.9 (1.8)
	Frontal	2.8 (1.6)	2.5 (2.3)	**4.0 (5.3)**
	Transverse	2.2 (1.2)	1.8 (1.9)	2.9 (4.8)
Ankle right	Sagittal	2.0 (1.3)	1.6 (1.1)	2.7 (5.4)
	Frontal	2.4 (1.6)	2.6 (2.7)	**3.5 (4.6)**
	Transverse	2.2 (1.0)	**3.4 (2.9)**	**4.9 (6.6)**
Hip left	Sagittal	2.5 (1.7)	1.1 (0.7)	0.8 (0.7)
	Frontal	2.4 (1.9)	1.8 (1.8)	2.0 (1.7)
	Transverse	2.0 (1.5)	1.5 (1.3)	2.0 (1.8)
Knee left	Sagittal	**3.2 (2.6)**	1.0 (0.7)	1.4 (1.1)
	Frontal	**3.2 (1.7)**	**3.1 (3.0)**	**4.0 (3.9)**
	Transverse	2.3 (0.9)	1.5 (1.1)	2.5 (3.9)
Ankle left	Sagittal	2.9 (1.4)	0.9 (0.8)	2.1 (4.1)
	Frontal	2.3 (1.2)	2.3 (1.9)	**3.1 (3.3)**
	Transverse	1.9 (0.8)	1.7 (1.7)	**3.1 (5.1)**

Bold values indicate angles >3°.

2. Assessment of the IMU Reproducibility

The mean ROM recorded for each joint in each plane of motion during one walking test movement are presented in Table 3, together with the reproducibility indices computed across repetitions of each calibration movement. The mean ROM displayed symmetrical values for both limbs and classical movement amplitudes for walking at 1 m·s−1. Overall, the mean reproducibility was excellent for all calibration movements in the sagittal plane (ICC: 0.96–0.99) and it was good to excellent in the transverse plane (ICC: 0.87–0.93). In the frontal plane, the mean reproducibility was good for all calibration movements (ICC: 0.79–0.86), except for walking, which had an acceptable reproducibility on average (ICC: 0.63) due to the low ICC observed for the hip and knee. The reproducibility was uniformly good to excellent across the calibration movements, except for the walking movement, which reported slightly lower reproducible movement amplitudes for proximal joints (as low as ICC = 0.76 for the hip compared to ICC = 0.94 for any other movement) and slightly higher reproducibility in distal joints (as high as ICC = 0.91 for the ankle compared to ICC = 0.82 for other movements). For all functional calibration movements, whatever the joint, the mean SEM was within 1.2° in all planes, although the mean SEM remained generally higher at the knee (0.9°) and ankle (1.8°) compared to the hip (0.6°) and pelvis (0.1°). This resulted in acceptable variations between movements, as shown by a mean SEM% of 1.8% in the sagittal plane, 4.8% in the transverse plane, and 7.9% in the frontal plane, where articular amplitudes were smaller for all lower limb joints.

Table 3. Joint angle measurement reproducibility for one walk task across four functional calibration movements.

		ROM (°)	Tilted (Movement 1)			Extension (Movement 2)			Squat (Movement 3)			Walking (Movement 4a)		
		Mean (SD)	ICC	SEM (°)	SEM %	ICC	SEM (°)	SEM%	ICC	SEM (°)	SEM %	ICC	SEM (°)	SEM%
Pelvis	Sagittal	*12.8 (2.1)*	1.00	0.1	0.7%	1.00	0.1	0.5%	1.00	0.1	0.5%	0.98	0.3	2.5%
	Frontal	*6.8 (0.8)*	0.98	0.1	1.7%	0.98	0.1	1.6%	1.00	0.1	0.8%	0.76	0.4	6.1%
	Transverse	*17.9 (4.1)*	1.00	0.0	0.0%	1.00	0.0	0.0%	1.00	0.0	0.0%	1.00	0.0	0.0%
Hip right	Sagittal	*50.4 (2.4)*	0.98	0.3	0.7%	1.00	0.2	0.4%	1.00	0.3	0.6%	0.90	0.9	1.8%
	Frontal	*14.0 (1.9)*	0.81	0.8	5.8%	0.90	0.7	5.1%	0.90	0.6	4.2%	0.50	1.3	9.3%
	Transverse	*18.4 (2.7)*	0.98	0.4	2.0%	1.00	0.4	1.9%	1.00	0.2	1.1%	0.90	0.7	4.0%
Knee right	Sagittal	*73.3 (3.2)*	1.00	0.2	0.2%	1.00	0.2	0.3%	0.92	0.9	1.3%	1.00	0.2	0.3%
	Frontal	*11.8 (1.9)*	0.68	1.1	9.4%	0.72	1.0	8.7%	0.83	0.8	6.8%	0.15	1.8	**15.3%**
	Transverse	*18.9 (2.8)*	0.93	0.7	4.0%	0.91	0.8	4.4%	0.77	1.4	7.2%	0.78	1.3	6.9%
Ankle right	Sagittal	*46.3 (16.0*	0.95	3.4	7.4%	0.97	2.9	6.2%	0.93	4.3	9.3%	1.00	1.0	2.2%
	Frontal	*18.5 (3.2)*	0.69	1.8	9.8%	0.56	2.1	**11.5%**	0.80	1.4	7.7%	0.83	1.3	7.2%
	Transverse	*21.3 (4.8)*	0.83	2.0	9.4%	0.81	2.1	9.7%	0.63	2.9	**13.6%**	0.81	2.1	9.9%
Hip left	Sagittal	*46.9(4.7)*	1.00	0.2	0.5%	1.00	0.2	0.5%	1.00	0.1	0.3%	0.90	1.2	2.6%
	Frontal	*15.0 (1.6)*	0.91	0.5	3.3%	0.90	0.5	3.2%	0.90	0.5	3.3%	0.40	1.3	8.5%
	Transverse	*17.2 (3.5)*	0.98	0.5	2.8%	1.00	0.8	4.5%	1.00	0.3	1.6%	0.90	0.9	5.2%
Knee left	Sagittal	*75.2 (9.3)*	1.00	0.5	0.7%	1.00	0.6	0.8%	0.98	1.2	1.6%	0.99	0.9	1.2%
	Frontal	*10.4 (2.8)*	0.60	1.8	**17.1%**	0.70	1.5	**14.9%**	0.86	1.0	**10.0%**	0.84	1.1	**10.9%**
	Transverse	*19.7 (4.5)*	0.99	0.4	2.1%	0.99	0.4	2.2%	0.97	0.7	3.8%	0.95	1.0	5.3%
Ankle left	Sagittal	*41.0 (3.7)*	0.92	1.0	2.5%	0.98	0.6	1.4%	0.96	0.7	1.8%	0.98	0.5	1.2%
	Frontal	*17.4 (4.5)*	0.87	1.6	9.1%	0.86	1.7	9.6%	0.70	2.4	**14.0%**	0.95	1.0	5.9%
	Transverse	*19.8 (3.6)*	0.68	2.0	**10.3%**	0.83	1.5	7.5%	0.75	1.8	9.2%	0.88	1.3	6.3%
Mean Sagittal		49.4 (5.9)	0.98	0.8	1.8%	0.99	0.7	1.4%	0.97	1.1	2.2%	0.96	0.7	1.7%
Mean Frontal		13.4 (2.4)	0.79	1.1	8.0%	0.80	1.1	7.8%	0.86	1.0	6.7%	0.63	1.2	9.0%
Mean Transverse		19.0 (3.7)	0.91	0.9	4.4%	0.93	0.8	4.3%	0.87	1.0	5.2%	0.90	1.1	5.4%
Mean Pelvis		12.5 (2.4)	0.99	0.1	0.8%	0.99	0.1	0.7%	1.00	0.0	0.4%	0.91	0.2	2.8%
Mean Hip		26.9 (2.8)	0.94	0.5	2.5%	0.95	0.5	2.6%	0.96	0.3	1.8%	0.76	1.1	5.2%
Mean Knee		34.8 (4.1)	0.86	0.8	5.6%	0.89	0.8	5.2%	0.89	1.0	5.1%	0.78	1.1	6.6%
Mean Ankle		27.4 (5.9)	0.82	2.0	8.1%	0.84	1.8	7.7%	0.80	2.3	9.3%	0.91	1.2	5.5%

Bold values indicate SEM% > 10%, ROM: range of motion, SD: standard deviation, ICC: intraclass correlation coefficient, SEM: standard error of measurement, SEM%: standard error of measurement in proportion to the mean.

3. Assessment of Accuracy in Different Test Movements

The functional calibration movements that provided the highest accuracy were the tilted and extension movements, as well as walking at a self-selected speed (mean RMSE for all joints, respectively: 2.5°, 2.3°, 2.2°). Since these calibration movements provided a comparable performance for the functional calibration of the IMU, the mean accuracy with the walking functional calibration was computed for the measurement of the four gait test movements, namely walking, ascending or descending one step, and stepping over an obstacle, when considering the lateral axis of the reference frame perpendicular to the gravity vector (as similar accuracies were obtained when considering it parallel to the axis of the segment rotation during the functional calibration). The mean accuracy obtained with the functional calibration movements retained is presented in Table 4. The RMSE and ΔROM were both smaller than 6° for ascending a step and smaller than 13° for descending it. For stepping over an obstacle, the RMSE reached 13° but the ΔROM had maximum values of only 4°. Notably, the accuracy of the IMU measurements were higher in walking than in other gait movements, where larger inaccuracies were observed, especially for a step ascent (both ankles in the sagittal plane and back leg hip in the frontal plane), for a step descent (all joints of both legs), and for stepping over an obstacle (all joints of both legs excluding the pelvis), although the error was generally within 5° and only rarely exceeded 10°.

Table 4. Accuracy obtained for each joint, in the 3D, in the four test movements: walking, stair ascent, stair descent, and obstacle crossing.

		Mean ROM (°)				Mean RMSE (°)				Mean ΔROM (°)			
		Walking	Stair Ascent	Stair Descent	Obstacle Crossing	Walking	Stair Ascent	Stair Descent	Obstacle Crossing	Walking	Stair Ascent	Stair Descent	Obstacle Crossing
	Pelvis Sagittal	12.8	18.4	24.4	18.9	0.9	1.2	1.7 *	1.4	0.5	0.8	1.3	1.3
	Pelvis Frontal	6.8	13.7	20.0	13.3	0.9	2.2*	2.6 *	1.3	0.5	0.7	1.9 *	1.0
	Pelvis Transverse	17.9	20.0	47.6	27.1	1.4	1.5	3.5 *	2.0	0.2	0.6	0.8 *	0.6
	Hip Sagittal	50.4	83.5	69.7	104.5	2.1	2.8	7.8 *	7.3 *	1.2	1.0	2.6 *	1.5
	Hip Frontal	14.0	18.1	30.8	17.4	2.4	4.3 *	4.9 *	3.5 *	2.1	3.8	3.1	2.9
	Hip Transverse	18.4	20.6	30.0	21.4	2.2	3.1	4.2 *	4.5 *	2.0	2.2	3.0	2.8
	Knee Sagittal	73.3	98.7	101.6	119.2	3.6	5.3	12.8 *	10.2 *	1.3	4.3	2.3	5.3 *
Front	Knee Frontal	11.8	13.6	15.3	14.4	2.7	4.2	4.2	4.4	2.5	4.7	3.0	6.5 *
	Knee Transverse	18.9	19.1	25.3	27.8	2.4	3.8	5.4 *	5.1 *	1.9	4.2	4.0	5.6 *
	Ankle Sagittal	46.3	45.4	76.5	48.4	2.1	5.7 *	7.9 *	4.3 *	1.6	5.4	4.6	6.8
	Ankle Frontal	18.5	21.6	25.1	21.9	2.5	2.4	4.0 *	3.3	2.7	2.3	3.8	3.0
	Ankle Transverse	21.3	18.4	24.5	21.2	2.4	2.3	3.1	3.6 *	4.1	4.4	1.9	3.0
	Hip Sagittal	46.9	83.2	65.6	65.9	2.3	3.5	5.9 *	5.1 *	0.9	1.1	2.1	1.1
	Hip Frontal	15.0	20.2	29.9	19.6	1.8	5.2 *	4.0 *	3.3 *	1.5	4.8 *	1.9	3.6
	Hip Transverse	17.2	23.7	29.8	22.0	2.0	3.2 *	4.7 *	3.1 *	1.7	2.5	2.7	1.4
	Knee Sagittal	75.2	94.1	103.5	129.6	3.2	5.3	11.3 *	12.3 *	1.0	3.6	2.6	5.1 *
Back	Knee Frontal	10.4	16.2	16.3	15.3	2.7	3.7	4.4 *	4.0	2.7	2.8	4.0	4.9
	Knee Transverse	19.7	19.8	24.0	27.5	2.3	4.1 *	5.3 *	4.9 *	1.3	2.9	4.1 *	3.3
	Ankle Sagittal	41.0	50.8	70.8	43.9	2.9	4.5	9.0 *	7.2 *	1.0	6.4 *	3.5	4.6
	Ankle Frontal	17.4	18.4	22.1	22.7	2.2	2.3	3.5 *	3.9 *	2.4	2.2	3.2	2.8
	Ankle Transverse	19.8	17.8	22.1	31.8	1.9	2.3	3.9 *	5.0 *	1.8	3.6	2.2	10.5 *

ROM: range of motion, RMSE: root mean square error, ΔROM: absolute difference in range of motion, *: significant difference in the post hoc test against the walking task.

Discussion

The main objective of this study was to compare the accuracy and reproducibility of lower limb joint angles computed from IMUs following different functional calibration methods. The study showed that applying a functional calibration movement before IMU-based lower limb kinematic assessment allowed for a fairly accurate measurement of gait movements. Except for the squat calibration movement, only small discrepancies were observed between functional calibration movements during a walking task, with a peak mean error of 3.6° for any joint in any plane of movement. Overall, the absolute reproducibility was similar for the three planes, but relative reproducibility was higher in the sagittal plane, with a mean standard error of measurement of less than 1.1° observed between multiple repetitions of the same functional calibration movement. A comparable overall performance was observed for different calibration movements, although each movement reported variable merits for different joints and planes of movement. Although the highest accuracy was observed in straight walking with a mean error of 2.2°, more complex gait movements tended to provide larger but limited errors, with a mean error of 3.5° for a step ascent, 5.4° for a step descent, and 4.7° for crossing an obstacle.

1. Accuracy of Different Calibration Methods during Straight Walking

The accuracy reported in this study during straight walking ranged from 1.1° to 3.6° for RMSE and from 0.2° to 3.4° for ΔROM, which is comparable to the mean error below 3° reported when using marker clusters on segments[127] rather than markers on anatomical landmarks [128-130]. Indeed, both methods reported different joint kinematics and accounted

differently for errors of markers placement, soft tissue artefacts, and biomechanical model calculations [127]. The functional calibration of the optical system did not influence the accuracy, indicating that the reference frame obtained for each segment with the functional calibration movements were close to the optical reference frame. As the magnetometer was not used in the AHRS algorithm, the DRIFT was controlled to be acceptable (mean of 2.3°) for such short experiments. The drift was slightly higher in the more distal joints, probably due to the higher speed of the movements [131, 132]. The drift was slightly lower in the sagittal plane, probably because the drift in this plane was better compensated by the sensor fusion algorithm. Although a mean error under 2° has been obtained on a single-joint movement [117], the accuracy obtained with our multi-joint model is acceptable for most clinical gait applications [112].

While the tilted and extension calibration movements provided a higher accuracy in the hip and ankle kinematics compared to the knee, walking calibration movements reported a higher accuracy for distal joints, whatever the walking speed. This observation can be supported by (1) a greater variability in the knee kinematics during the tilted and extension movements compared to straight walking and (2) higher accelerations of the distal relative to the proximal segments during walking. This observation also showed that the reference frame for each segment can be equally determined via a rotational movement recorded by the gyroscopes or via a translational movement recorded by the accelerometers contained in each IMU. The accuracy obtained in slow walking also validates the use of this functional calibration movement in similar conditions, which is often encountered in pathological gaits or in older adults [133].

The lower accuracy reported for the knee and ankle via the squat calibration movement could be explained by the lower movement amplitude of the shank and foot segments during this calibration movement. Indeed, the smaller amplitude-to-noise ratio probably resulted in an erroneous definition of the reference frame, leading to kinematic crosstalk [134]. This result also showed that functional movements exploring a wide range of segment orientation tended to provide more accurate segment reference frames.

2. Reproducibility of Calibration Movements

Reproducibility was excellent in 65% of the tested joints and motion planes, good in 24%, acceptable in 10%, and poor in 1 observation out of 84. Concerning differences between calibration movements, the walking calibration movement produced the highest reproducibility and SEM% for the ankle, while the other functional calibration movements produced higher reproducibility indices for the pelvis and hip joints. The lower reproducibility at the ankle for the segment-rotation-based movements could be explained by the difficulty in reproducing movements purely in the sagittal plane. This observation also supports previous results showing that the variable position of the foot affects the functional calibration when using different static postures [116]. The use of more guidance or more repetitions of the calibration movements could improve the reproducibility by (1) avoiding parasitic movements of the feet out of the sagittal plane and (2) decreasing the impact of any parasitic movement on the definition of the rotation axis. However, in order to limit the complexity and burden of the functional calibration movements, the walking calibration movement remains a remarkably convenient alternative since it offers a good to excellent reproducibility (though

lower than other movements for the proximal joints), with a very simple and ecological movement. Caution may be needed for subjects having an impaired walking pattern, e.g., a subject walking with the feet pointing outwards.

A higher reproducibility was observed in the sagittal plane compared to the frontal plane. This could be explained by the higher range of motion in the sagittal plane during walking, leading to more kinematic crosstalk in the other planes measured and/or by the fact that the functional calibrations mainly generated segment movements in the sagittal plane. The combination of the higher variability and smaller ROM in the frontal plane during walking led to a higher SEM% in this plane, as also shown in upper limb anteroposterior reaching tasks [124]. Higher SEM% values inevitably require higher changes to detect meaningful functional changes, e.g., after therapy. The reproducibility of calibration movements in the frontal plane should be explored for the assessment of functional outcomes involving larger movements in the frontal plane.

3. Accuracy across Different Gait Movements

More complex gait movements tended to provide larger errors than a peak mean RMSE of 3.6° and a peak mean ΔROM of 3.4° for straight walking. Indeed, the peak mean errors obtained in the sagittal plane for a step ascent of 3°, 5°, and 5° for hip, knee, and ankle, respectively, correspond to errors in elevation angles of 5°, 4°, and 4° previously reported for the same joints [135]. Similarly, the peak mean ΔROM of 6.4° obtained for the stair ascent and of 4.6° for the stair descent are comparable to the errors previously reported for healthy subjects (peak error of 4.1° for a stair ascent and 4.8° for a stair descent) [136]. Therefore, before implementing inertial sensors in a complex, real-life

context, the accuracy should be established in such a context rather than extrapolated from simpler gait movements recorded in controlled lab conditions.

4. Limitations and Perspectives

This study focused on healthy adults and this could be a limitation in case the functional calibration movements proposed here would be used with patients with a limited range of motion or who have parasitic movements that may hinder an accurate and reproducible calibration movement. The transferability to the elderly or to patients with motion disabilities should be assessed in further studies.

The IMU magnetometer was voluntarily omitted in this study in order to avoid ferromagnetic disturbances. The recordings in this study were limited in time due to the short time required to execute the investigated movements. The drift resulting from longer records [49] could be limited by using the IMU magnetometer or algorithms that constantly fuse the segment's angular velocity and linear acceleration via known kinematic relations between segments [137].

Although a high accuracy for the lower limb joint angles has been obtained by using only the gyroscope signals, our methods could be improved by also accounting for the segment accelerations [113, 138], which can be used to locate the joint centers and improve the robustness of the segment orientations [139]. Another approach consists in using a hinge joint model and kinematic constraints to develop automatic or so-called "plug and play" calibrations [113, 140]. This less restrictive method may facilitate clinical

applications where patients with motion disabilities cannot be expected to perform precise prescribed calibration movements.

Conclusions

This study documents the high accuracy of IMU-derived lower limb joint angles during walking using several functional movements for the sensor-to-segment calibration. Functional movements requiring larger segmental angular amplitudes provided more accurate segmental reference frames and led to a higher accuracy regarding the kinematics of the adjacent joints. Alternatively, the higher linear accelerations generated at distal segments during a walking functional calibration also led to a higher reproducibility for distal joints, in comparison to the functional calibration based on the principal rotational axes. The walking, tilted, and extension functional calibration movements were shown to be three equivalent options for the gait movement examined in this study. In addition, for examining walking, the walking functional calibration may be superior because it involved very limited material and instruction complexity, which strengthens its interest in uncontrolled environments.

Chapter 2. Potential of using locomotor outcomes for clinical use

Lower limb kinematics improvement after genicular nerve blockade with patients suffering from knee osteoarthritis: a milestone study using inertial sensors

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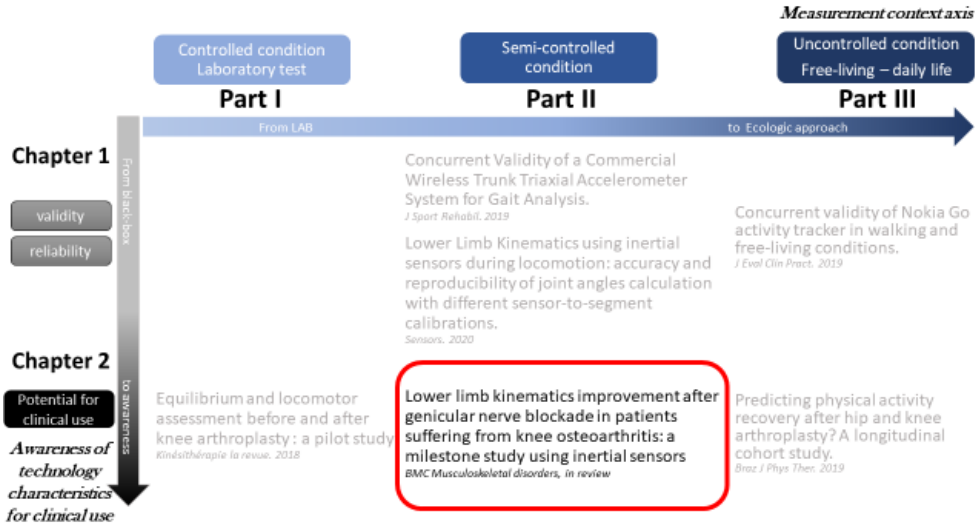


Lower limb kinematics improvement after genicular nerve blockade with patients suffering from knee osteoarthritis: a milestone study using inertial sensors

Abstract

Purpose

Genicular nerve blockade is a possible treatment for patients with knee osteoarthritis. Pain relief and improvement in functioning is expected. This procedure could be of major interest for patients in low-income countries where total knee arthroplasty is not available for the population. This study aimed at assessing the immediate benefits on pain, gait, and stairs kinematics after a genicular nerve blockade in patients suffering from knee osteoarthritis in Cameroun.

Methods

A prospective study was carried out in 26 subjects in Cameroun. A genicular nerve blockade was performed on fourteen women with painful knee osteoarthritis grade 2-4. Lower limb joint angles were recorded with inertial sensors before and one hour after injection. Patient-reported outcomes of pain and perceived difficulty were collected, as well as 10 meters and 6 minutes walking tests. A reliability analysis of inertial sensors was performed on a sample of 12 healthy subjects by calculating the intraclass correlation coefficient and the standard error of measurement.

Results

Pain and perceived difficulty decreased significantly ($p<0.001$). Cadence increased significantly in stairs climbing (upstairs: +7.7 steps/min; downstairs: +7.6 steps/min). There was an improvement for hip sagittal range of motion during gait (+9.3°) and pelvis transverse range of motion in walking upstairs (-3.3°). Angular speed range of the knee in the sagittal plane and of the

hip in the frontal plane increased significantly in stairs descent (+53.7°/s, +94.5°/s).

Conclusion

This study quantified improvement of gait and stair climbing in the short term after a genicular nerve blockade in patients suffering from knee OA in Cameroon. This is the first study objectifying this effect, through wearable sensors.

Keywords: inertial sensor, gait, knee osteoarthritis, Genicular nerve blockade, biomechanics.

Levels of evidence: Therapeutic study – level II

Lower limb kinematics improvement after genicular nerve blockade with patients suffering from knee osteoarthritis: a milestone study using inertial sensors

Introduction:

Knee osteoarthritis (OA) affects one third of the population above 65 years [141, 142]. In sub-Saharan Africa, the prevalence may reach up to 33% of the population above 35 years [143]. Pain, local joint swelling, stiffness, and difficulties in the activities of daily living (ADL) are the main symptoms [58].

The goal of knee OA treatment is pain relief and improvement of physical function. One would assume that by using pain-relief therapies, there would almost automatically be improvement in function, but this is not necessarily the case [144]. The lack of a significant correlation between the decrease in perceived pain and the objective improvement of their functional capabilities [145] makes the assessment of the latter essential. Previous studies have assessed the functional quantitative changes of non-surgical pain relief treatment such as oral medications [145, 146], intra-articular injections of hyaluronic acid [147-149], or nonsteroidal anti-inflammatory drugs [150, 151]. They showed positive effects on lower extremity joint kinematics, gait parameters and knee-related functional status on the short term [150, 151], or the long term [147, 148]. However, Shrader et al. underlined that although the relief of knee pain is sufficient to enhance gait function in osteoarthritis of the knee, it is insufficient to enhance stair-stepping function [150].

In the past decade, genicular nerve blockade (GNB) and radiofrequency ablation (RFA) appeared as relevant alternatives in the treatment of chronic knee OA pain [152-155]. These procedures are based on the selective inhibition of the consistent sensitive nerves supplying the knee joint capsule, which suppresses the related nerve impulses. This leads to expect a knee pain relief and functional improvement [152, 154, 155]. In comparison

to methods that relieve pain by a peripheral action (intra-articular corticosteroids infiltrations, viscosupplementation, etc.) or central (oral analgesics), there is a probable inhibition of proprioceptive impulses. However, only 5 out of the 11 to 13 sensory nerves innervating the joint capsule of the knee are blocked precisely to be effective enough on the pain without completely suppressing proprioceptive influences [156]. These techniques are ambulatory, minimally invasive, with a high potential of pain relief in one single session [152, 157-159]. The average pain relief at three months follow-up after a GNB-RFA is 67% improvement from baseline knee pain score, and 95% of these patients still describe pain relief at 6 months [159]. Such procedure could be beneficial for patients in sub-Saharan African settings where total knee arthroplasty (TKA) is difficult to access for the population.

Although previous studies have assessed the effects of GNB and RFA on self-reported measure of pain and function [152, 154, 160-162], little is known about the functional quantitative changes in gait or stair climbing after this treatment. The hypothesis whereby pain relief results in gait improvement should be verified. The studies assessing intra-articular injection were achieved through a motion capture (MoCap) laboratory [147, 150, 151]. This first instrumentation method, a MoCap laboratory, allows for an objective assessment, but is challenging to make use of in low income countries, in particular because of its extensive price, the electrical network and climatic conditions. Another instrumentation method consists of technology based on inertial measurement units (IMU), which are low cost portable electronic devices. They consist of an accelerometer, a gyroscope and often a magnetometer, which enables them to record kinematic data (velocity, acceleration, orientation). Such technology is therefore an opportunity to

answer specific research questions in resource-limited settings. Although those sensors have an acceptable validity in comparison to MoCap laboratory [163], the discriminative capacity to detect differences after treatment in population with knee OA has rarely been studied, in particular in out-lab settings [50].

To our knowledge, no study has assessed the quantitative improvement of locomotion after a GNB, especially using inertial sensors. We hypothesized that wearable sensors could detect the quantitative functional effects of peri-articular injection of genicular nerve on gait and stair climbing in patients suffering from knee osteoarthritis in Cameroun. The aim of this therapeutic pilot study was to assess the ability of inertial sensors to detect differences in kinematics of gait and stairs climbing after a GNB in patients suffering from knee osteoarthritis.

Methods :

1. Study design

This interventional study was conducted from September to November 2019 at Centre Hospitalier Dominicain Saint Martin de Porres in Yaoundé, Cameroun. The Central Region Ethics Committee for Human Health Research (Yaoundé, Cameroon) approved the study protocol (agreement number: CE 0-771/CRERSHC/2019) and each patient provided written informed consent prior inclusion in this study.

2. Participants

A convenience sample of 26 adults participated in this pilot study. Consecutive patients who presented themselves at the investigators' consultation within the study period with painful knee OA, who did not respond to conservative therapy, were considered for the study. Radiographic confirmation of knee OA (Kellgren-Lawrence) by a radiologist was required. Patients were included if they suffered from knee pain (Numeric rating scale (NRS) >5/10) for more than three months, not relieved by conservative treatment (oral medication, intra-articular injections with corticoids and viscosupplementation), with a radiological confirmation of tibio-femoral OA grade 2 to 4. Exclusion criteria included other connective tissue diseases that affected the knee, skin lesion on the knee, steroid or hyaluronic acid injection therapy during the previous 3 months, knee surgery scheduled in the next three months, anticoagulant medication use, unbalanced diabetes mellitus or hypertension and patients unable to walk. Fourteen adults were recruited to

participate in the interventional procedure and twelve healthy adults were recruited by an advertising poster for a reliability assessment (Table 1).

Table 1 : Characteristics of participants

	Knee OA	Healthy	t-test
	Mean (SD)		p-value
Gender (F/M)	14/0	7/5	-
Age (years)	64.5 (11.3)	50.6 (11.9)	0.787
Height (m)	1.61 (0.05)	1.69 (0.07)	0.010
Weight (kg)	87.4 (17.2)	71.5 (10.1)	0.184
Disease start (months)	50.1 (41.2)	-	-
Bilateral pain (n)	7		
	Median [25-75]		
Kellgren-Lawrence	3 [2-4]	-	

OA : Osteoarthritis, SD : standard deviation, F : female, M : male, n : number of subjects, [25-75] : interquartile range

3. Experimental protocol

a. Interventional procedure

A single treatment session was performed for each of the 14 patients. In case where the patient displayed bilateral knee pains, both knee were treated. The patient was placed in a supine position with a pillow under the popliteal fossa. No premedication or sedatives were administered. Under sterile conditions, the GNB with updated targets [156, 158] was performed with fluoroscopic guidance. At each injection site, skin and soft tissues were anesthetized with 1 mL 1% lidocaine. The five nerves were targeted as referenced above (Figure 1) [156, 158]. For the superior-lateral genicular nerve (SLGN), a 10cm 22-gauge radiofrequency (RF) cannula was advanced percutaneously towards the superior edge of the lateral femoral condyle until the tip touched the bone on the anterior posterior (A-P) view. Then the C-arm

was rotated to have a true lateral view, with both condyles superimposed. The needle tip was adjusted to fit the target area located at the junction between the superior edge of the lateral condyle and the posterior femoral cortex. For the superior-medial genicular nerve (SMGN), the RF cannula was advanced towards the superior edge of the medial condyle until the tip touched the bone on AP view. Subsequently, on the lateral view, the tip of the cannula was adjusted to fit in front or just above the adductor tubercle. For the inferior-medial genicular nerve (IMGN), the RF cannula fitted at the confluence of the medial tibial shaft and the tibial flare in the A-P view, and the midpoint of the tibia in the lateral view. The recurrent fibular nerve (RFN) was targeted on a longitudinal line drawn below the Gerdy's tubercle (GT), at a point located 1cm below the inferior edge of the GT. The RF cannula was inserted at that point and advanced until the tip touched the bone. For the infrapatellar branch of the saphenous nerve (IPBSN), the treatment target was the longitudinal line connecting both following transversal lines, 4 cm medially to the patellae apex: the transversal line passing by the patellae apex and the one passing by the top of tibial tuberosity. The RF cannula was inserted longitudinally at the proximal edge of the treatment line and advanced deeply in the subcutaneous tissue until the distal edge of the treatment line. For each of the 5 targeted nerves, after

verification of the correct needle placement, a total of 1mL of lidocaine plus 20 mg of triamcinolone was injected.

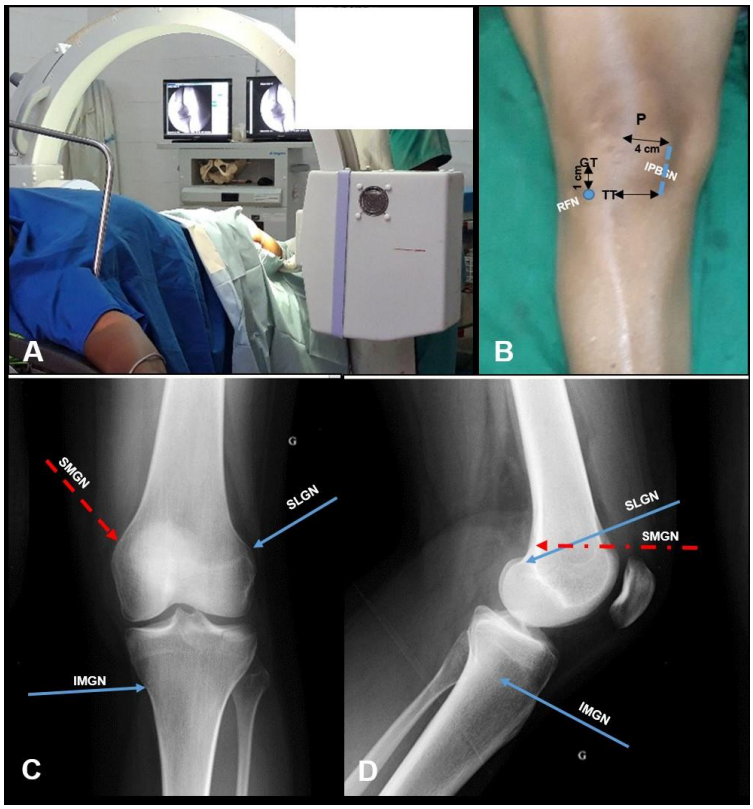


Figure 1. Anatomical targets for fluoroscopic guided genicular nerve blockade. (A) Installation (B) Landmarks for infrapatellar branch of the saphenous nerve (Dashed blue line represents the treatment line) and recurrent fibular nerve (blue point) targeting. (C) Anterior-Posterior X-ray view of the knee. Landmarks of Cannula placement for targeting the Superior medial genicular nerve (Dashed red arrow), superior lateral genicular nerve (Upper blue arrow) and inferior medial genicular nerve (lower blue arrow). (D) Landmarks for targeting the same three nerves on the lateral view of the knee. SLGN, superior lateral genicular nerve; SMGN, superior medial genicular nerve; IMGN, inferior medial genicular nerve; IPBSN, infrapatellar branch of saphenous nerve; RFN, recurrent fibular nerve; P, patella; TT, tibial tuberosity; GT Gerdy's tubercle

b. Assessment

Assessment was performed one hour before, and one hour after the interventional procedure by an independent evaluator (JL). All the participants attended the hospital center for the primary data collection session. They were asked to complete the Patient-reported outcome measures (PROM) and their pain intensity after the testing protocol using Numeric Rating Scale (NRS).

The testing protocol included 5 locomotor tasks. The tasks were demonstrated by the operator and were performed in the same order:

A) walking ten meters at self-selected speed;
B) walking ten meters at higher speed;
C) ascend stairs;
D) descend stairs;
E) walking freely for six minutes.

After the assessment, their perceived difficulty during the 5 test tasks was assessed on a NRS scale (0-10). Afterwards, participants went to the surgery room for the GNB of the painful knee(s). The same testing protocol was performed one hour after the intervention. Pain intensity was assessed individually in the cases where the infiltration was performed in both knees. The same assessment was performed two times in 12 healthy adults who did not undergo the intervention, by removing the sensors between the consecutive sessions.

4. Equipment

The time to perform 10 meters was measured with a standard chronometer. The distance covered during the six minute walking (E) was assessed with a pedometer (GEONAUTE ONWALK).

To assess the lower limb joint kinematics, seven wearable IMUs; (x-IMU, x-io Techologies, UK) were attached by means of a semi-elastic belt to seven body parts: the waistline at the level of the fifth lumbar vertebra (L5), the middle of the thighs, the middle of the shanks, and at the dorsal side of the feet [163]. Each IMU included a tri-axial accelerometer (full scale ±6g), a gyroscope (±2000°/s) and a magnetometer (±8.1G) that recorded at sampling frequency of 128 HZ. The IMUs were connected to a computer by means of a Bluetooth connection. Custom application based on open source software was used to record the IMU data (C# program, github.com/xioTechnologies).

5. Outcomes

   a. Patient-reported outcome measure (PROM)

Pain intensity was measured by a Numeric rating scale (NRS). Function was assessed by the Knee Osteoarthritis Outcome Score (Koos) [8]. As pain and perceived difficulty are different constructs [164, 165], perceived difficulty during movement was assessed by another NRS scale (0-10).

b. Objective functional assessment of locomotion

Clinical outcome

Walking speed (m/s) was assessed twice on a 10 meter-track, once at self-selected speed, and once at higher speed. Walking endurance was assessed during a 6 minutes free walking test. Results are expressed in distance (m).

Inertial sensors – Kinematics outcome

Each task was segmented with semi-automatic threshold methods based on accelerometer signals (flat zone detection, and peak detection) [47]. Three gait cycles were normalised on 0-100 points and averaged. A cycle begins with the contact from the foot until the next contact of the same foot. A combination of vertical shank acceleration and hip and knee angular movement were used to detect those events [47, 166].

The parameters were cadence (step/s), stride time (s), joint range of motion (ROM in degree) and angular speed range (SPEED in °/s). Joint angles of both legs were calculated by a validated method, using the walking functional sensor-to-segment calibration [163]. Three-dimensional joint kinematics of the pelvis, hip, knee, and ankle were calculated based on the recommendation of the international society of biomechanics [6]. Instantaneous three dimensional angular velocity was calculated by the finite derivative. ROM and SPEED were computed as the difference between the maximum and the minimum in the average gait cycle.

6. Statistical analysis

Differences betweenpre-injection and post-injection conditions were performed with the two tailed paired t-test. The Wilcoxon signed rank test was used for PROM and for variables that failed the test for normality. P-level was fixed $p < 0.05$.

Reliability of the clinical and kinematics outcome was performed on the healthy subjects according to a method described by Wagner [124] using the intraclass correlation coefficient (ICC) and the standard error of the measurement (SEM). ICC consistency parameters were calculated in a 2-way mixed model. SEMs estimate the non-systematic variance. As a measure of within-subject variability among repeated trials, the SEM expresses the measurement error in the same units as those of the original measurement.

Statistical analysis was performed using SPSS (version 25, IMB Corporation, Amonk, NY, USA).

Results

1. Patient reported outcome measure

The score (median [25-75 interquartile range) for KOOS subscale was 54 [39-72] for symptoms, 56 [34-68] for pain, 47 [44-63] for activities of daily living, and 44 [33-74] for quality of life.

The pain and the perceived difficulty during the functional tasks decreased significantly after the intervention ($p<0.001$) (Figure 2).

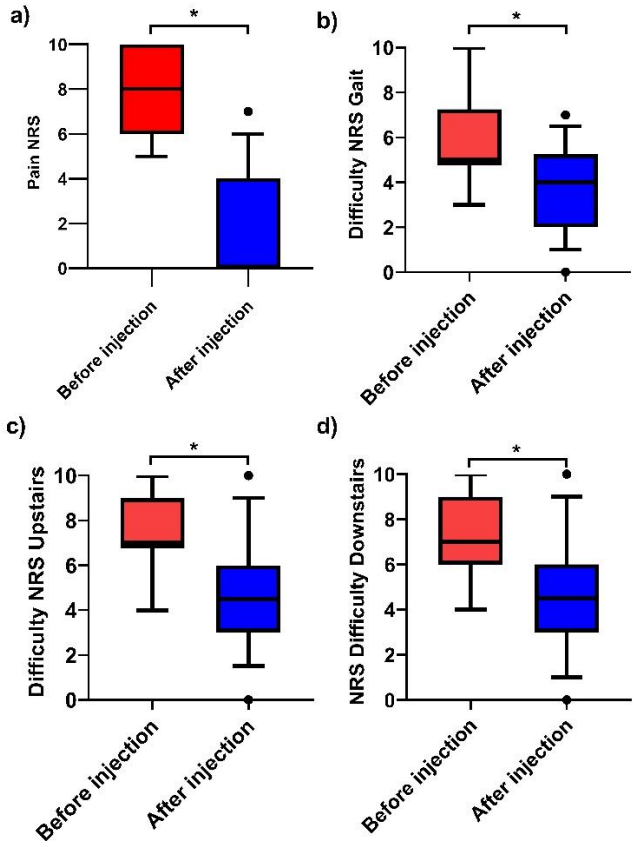


*Figure 2: Patient reported outcome during functional tasks: (a) Pain reported on the NRS (most affected knee), (b) Perceived difficulty during gait, (c) Perceived difficulty during ascending upstairs, (d) Perceived difficulty during descending downstairs, * indicate a significant difference ($p<0.001$)*

2. Objective functional assessment of locomotion

Clinical outcome

The impact of the intervention on quantitative parameters of locomotion is shown in Table 2. Self-selected walking speed and high walking speed increased significantly after the intervention (mean difference of 0.15m/s and 0.17 m/s respectively). Walking endurance performance measured by the 6 minute free-walking test improved significantly (mean difference of 58 m).

Inertial sensors – Kinematics outcome

Cadence and stride time during gait were not significantly affected by the injection, whereas cadence and stride time during ascending and descending stairs evolved significantly towards healthy subject group values (respective mean increase 7.7 steps/min, -0.40 s, 7.6 steps/min, -0.35s). During gait, sagittal and transverse hip ROM of the most painful side increased significantly of 9.3° and 3.5° respectively. For the hip, the increase in sagittal ROM is manifested by an increase in hip extension at 50% of the gait phase (Figure 3), while there is a shift in ankle sagittal trace in the swing phase of gait (50-100% of the gait phase). There was no significant difference for knee ROM in the sagittal plane.

For ascending stairs, only pelvis transverse ROM decreased significantly by 3.3°, while no significant ROM differences were observed for descending stairs. Graphs on joints angles for ascending and descending stairs are visible in Additional file 1.

All significant differences were higher than the SEM calculated on the reproducibility test on the healthy subjects.

Angular speed range (SPEED) for the hip and pelvis in walking and ascending stairs increased (Table 3). For descending stairs, hip SPEED in the frontal plane increased significantly by 150 % (mean difference of 53.7°/s). Knee SPEED in the sagittal plane also increased significantly by 123% (mean difference 94.5°/s).

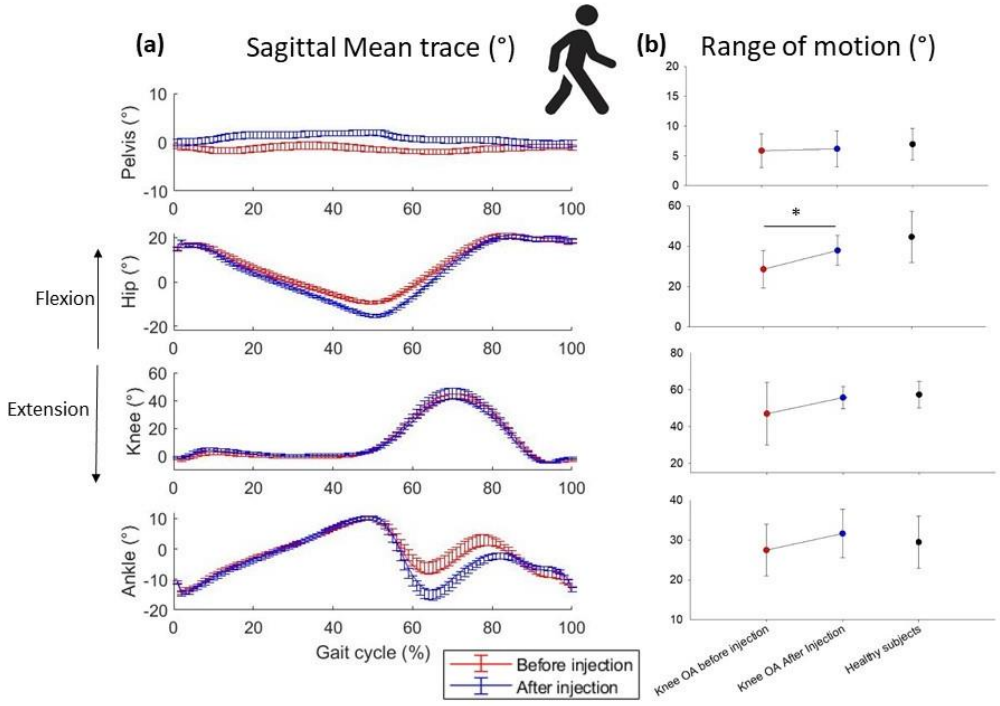


*Figure 3: Joint angle in sagittal plane of all subjects for pelvis, hip, knee and ankle during gait: (a): Sagittal mean trace. Error bar display standard error. (b) Range of motion, * indicate significant differences ($p<0.05$) between post-injection and pre-injection. The black dots on the right represent the healthy subjects group.*

Table 2 : Clinical and inertial sensor ROM results in functional activities before and after genicular blockade

			Pre	Post	Paired t-test		Reliability
			Mean (SD)		p value	diff	SEM
6 min walk	Distance (m)		286 (137)	319 (127)	**0.037**	58	36
Walking	High speed (m/s)		1.07 (0.35)	1.19 (0.35)	**0.004**	0.17	0.12
Walking	Self-selected speed (m/s)		0.81 (0.27)	0.96 (0.27)	**0.005**	0.15	0.11
	Stride time (s)		1.20 (0.14)	1.15 (0.07)	0.248	-0.05	0.03
	Cadence (step/min)		50.7 (5.8)	52.4 (3.4)	0.319	1.7	1.5
	Pelvis ROM (°)	Sagittal	5.9 (2.8)	6.2 (3.0)	0.484	0.3	1.5
		Frontal	5.3 (2.2)	6.2 (3.2)	0.321	0.9	1.6
		Transverse	7.5 (3.4)	8.4 (3.5)	0.152	0.9	1.9
	Hip ROM (°)	Sagittal	28.6 (9.3)	37.9 (7.3)	**0.004**	9.3	4.6
		Frontal	13.6 (5.2)	16.6 (9.1)	0.287	2.9	3.3
		Transverse	15.0 (6.5)	18.4 (7.3)	**0.041**	3.5	3.5
	Knee ROM (°)	Sagittal	47.0 (17.1)	55.8 (6.0)	0.094	8.8	7.1
	Ankle ROM (°)	Sagittal	27.5 (6.5)	31.6 (6.1)	0.244	4.1	6.2
Upstairs	Stride time (s)		1.84 (0.79)	1.45 (0.42)	**<0.001**	-0.4	0.1
	Cadence (step/min)		36.4 (10.7)	44.1 (9.8)	**<0.001**	7.7	3.6
	Pelvis ROM (°)	Sagittal	12.0 (6.9)	9.4 (4.3)	0.063	-2.6	0.7
		Frontal	12.0 (4.8)	11.5 (5.7)	0.646	-0.6	1.6
		Transverse	17.9 (6.1)	14.5 (5.7)	**0.035**	-3.3	2.3
	Hip ROM (°)	Sagittal	51.7 (9.8)	48.7 (8.7)	0.349	-3.1	2.1
		Frontal	23.7 (7.1)	20.8 (6.4)	0.242	-2.9	4.0
		Transverse	26.0 (7.8)	21.5 (7.7)	0.074	-4.4	3.8
	Knee ROM (°)	Sagittal	67.9 (10.4)	69.9 (12.7)	0.638	1.9	2.9
	Ankle ROM (°)	Sagittal	37.3 (14.4)	27.3 (7.6)	0.080	-10	4.0
Downstairs	Stride time (s)		1.66 (0.91)	1.31 (0.41)	**0.005**	-0.35	5.1
	Cadence (step/min)		41.5 (12.1)	49.1 (12.0)	**0.005**	7.6	0.2
	Pelvis ROM (°)	Sagittal	9.6 (2.7)	9.1 (4.3)	0.485	-0.5	1.2
		Frontal	10.6 (4.7)	9.4 (4.5)	0.383	-1.3	0.9
		Transverse	17.8 (9.0)	15.6 (7.3)	0.215	-2.2	0.8
	Hip ROM (°)	Sagittal	26.5 (7.4)	24.5 (7.5)	0.244	-2.0	3.6
		Frontal	14.2 (5.3)	16.3 (5.7)	0.394	2.1	2.1
		Transverse	26.4 (9.3)	25.3 (10.3)	0.484	-1.2	2.4
	Knee ROM (°)	Sagittal	61.0 (14.3)	63.6 (16.1)	0.722	2.6	2.7
	Ankle ROM (°)	Sagittal	52.3 (16.5)	46.3 (12.4)	0.206	-6.0	2.1

The data presented are those from the most painful leg. SD : standard deviation, SEM : standard error of measurement, paired t test : difference between pre-injection parameters and post-injection parameters , ROM : Range of motion

Table 3 : Inertial sensor SPEED results in functional activities before and after genicular blockade

			Pre	Post	Paired t-test		Reliability
			Mean (SD)		p value	Diff	SEM
Walking	Pelvis SPEED (°/s)	Sagittal	59.4 (25.8)	65.7 (29.1)	0.590	6.3	17.4
		Frontal	59.2 (26.7)	71.1 (32.8)	0.133	11.9	8.8
		Transverse	58.7 (17.3)	61.8 (23.8)	0.152	3.1	7.8
	Hip SPEED (°/s)	Sagittal	191.5 (63.8)	245.7 (46.6)	**0.002**	54.2	29.7
		Frontal	137.0 (55.1)	166.4 (80.6)	0.233	29.4	27.6
		Transverse	180.7 (69.2)	226.4 (105.6)	**0.002**	45.7	34.9
	Knee SPEED (°/s)	Sagittal	518.5 (179.1)	593.3 (97.4)	0.244	75.1	64.1
	Ankle SPEED (°/s)	Sagittal	316.8 (129.1)	364.5 (104.8)	0.138	47.7	66.0
Upstairs	Pelvis SPEED (°/s)	Sagittal	65.2 (27.2)	65.2 (25.2)	0.997	0.0	7.2
		Frontal	59.3 (22.3)	73.4 (25.5)	0.646	14.0	19.1
		Transverse	79.8 (22.4)	83.4 (22.0)	**0.035**	3.6	12.9
	Hip SPEED (°/s)	Sagittal	251.1 (45.2)	278.7 (53.1)	0.349	27.6	12.6
		Frontal	138.5 (48.9)	156.5 (37.8)	0.215	18.0	21.2
		Transverse	205.5 (73.5)	226.2 (78.8)	0.074	20.6	26.4
	Knee SPEED (°/s)	Sagittal	436.1 (109.9)	492.6 (81.1)	0.638	56.6	44.2
	Ankle SPEED (°/s)	Sagittal	354.5 (93.5)	340.3 (101.3)	0.080	-14.2	24.1
Downstairs	Pelvis SPEED (°/s)	Sagittal	70.7 (30.3)	92.9 (40.6)	0.137	22.2	11.1
		Frontal	76.6 (39.0)	83.1 (30.1)	0.542	6.5	15.4
		Transverse	101.0 (33.9)	114.0 (24.9)	0.190	13.0	16.7
	Hip SPEED (°/s)	Sagittal	172.7 (55.3)	221.9 (64.4)	0.063	49.2	40.6
		Frontal	109.3 (36.3)	163.0 (63.0)	**0.015**	53.7	15.9
		Transverse	192.2 (85.1)	229.3 (79.4)	0.158	37.1	57.9
	Knee SPEED (°/s)	Sagittal	407.0 (139.0)	501.4 (93.9)	**0.018**	94.5	37.3
	Ankle SPEED (°/s)	Sagittal	414.0 (137.7)	462.7 (123.7)	0.232	48.7	39.8

SD : standard deviation, SEM : standard error of measurement, paired t-test : difference between pre-injection parameters and post-injection parameters, SPEED : Angular speed range

Discussion

The most important finding of the present study was that the GNB improved gait kinematics of patients with knee OA immediately. Beyond the subjective improvement in pain and difficulty during the tasks assessed with NRS scale, the assessment using low cost wearable sensors quantified the improvement in gait, ascending and descending stairs.

Although the treatment was focused on the knee, the hip range of motion increase during gait was the most noteworthy difference observed. As joint kinematics is speed-dependent during gait [167] and that the speed was not greatly changed, we can conclude that the increase in ROM is effectively related to the treatment, and not an effect due to acceleration of the gait. Previous studies already showed that changes in gait mechanics in the knee joint have an effect on the ROM of the ankle and hip joints [168, 169]. This result can be explained by the fact that these three joints operate as a kinetic/kinematic chain while walking [168, 170]. It means that problems with one joint are biomechanically related to problems in the others [171, 172]. Skwara et al obtained similar results combined with an increase in hip and knee range of motion after an intra-articular injection [149].The lack of significate improvement for knee range of motion in our sample could be partially explained by the broader variability at baseline. However, there is lack of consensus in the ability of pain therapy to improve knee range of motion. Detrembleur et al also found no improvement with oral medication [146], while Mehta et al found significant improvement of knee ROM after intra-articular injection of corticoids and xylocaine [151]. Pain reduction obtained in those studies itself may therefore not always be sufficient in improving ROM. It is important to note that muscle strength plays an important role in physical

functioning [173] and that pain relief might not be enough to improve joint mechanics in some patients.

As stair climbing is suggested to be the first affected task in individuals with knee OA [174] with increased hip ROM, and decreased knee and ankle ROM [175], its assessment besides gait is important. GNB has no impact on this latter kinematic outcome. This means that this treatment is not successful in modifying motor strategy that tends to reduce the ground reaction force moment arm by ambulating with more trunk/hip flexion, less knee flexion, and less ankle dorsiflexion [175]. Asay et al. found that individuals with the same pain levels but with different degree of severity of arthrosis displayed similar stair climbing patterns [176], except for trunk flexion which was not assessed in this study. Those results could explain that the pain decrease did not result in modification of stair ambulation mechanics.

Interestingly the cadence increased and stride time decreased more in stair activities than in gait after the GNB, which means that the overall velocity increased. Moreover, the stride time difference is above the minimal clinically meaningful change of 0.2 s according to Oh Park et al [177]. This result is consistent with the greater improvement of perceived difficulty during stairs (Figure 2). We cannot distinguish if the speed increase was mainly in the stance or swing phase of stairs negotiation as we did not record the foot-off events [166]. But as the subjects increased their joint angular speed at the knee in the sagittal plane and at the hip in the frontal plane when descending stairs, one can assume that patients lowered their body faster during the stance phase [126]. This partially explains the strategies to improve the cadence [178].

Few studies assessed stair kinematics in OA [50], probably due to the complexity of an experimental setup. However, some studies support that there is no particular benefit in measuring more than gait for an indicator of ambulatory functional status [179], our data does not support this statement as the results deliver different insights in motor behaviour. Yet stairs climbing is a high expectation for people receiving treatment for OA [14, 173]. Larger ranges of knee flexion angle and knee flexion moment are required during this task [180, 181] and are therefore more challenging for this population. Furthermore, stair climbing is a single leg activity, where the entire weight of the subject is supported on a single leg and has to withstand the forces of propelling the body upward and forward to the next step [181].

To our knowledge, this is the first study assessing the gait and stairs kinematics after a genicular nerve blockade. It is the second study assessing a clinical population in stairs with inertial sensors [136], and the first one in the African continent. The accessibility of inertial sensors has the potential to increase the clinical understanding of the biomechanics and pathomechanics of the lower limb during daily life activities. It might help surgeons and therapists to integrate scientific findings into clinical examination and management of patients with lower extremity dysfunction [182]. The combined assessment of pain and quantitative, objective outcome of movements appears to be an opportunity to assess the relevant treatments for this population. The inertial sensors allowed to ease the assessment of joint behaviour. In fact, the patients of this study used to wear long dresses that make it difficult to assess movements, even qualitatively. Out of a dedicated laboratory, it is difficult to ask patients to walk half-naked in a hospital. Inertial sensors are helpful in this regard.

Though the final goal of these interventions is to obtain long-lasting improvements, this study examined only the immediate effects of the GNB and found that it was effective in relieving knee OA pain and improving gait kinematics one hour after the intervention. These results are interesting for pain physicians because the GNB is usually performed as a prelude to the RFA which allows long lasting results. Therefore, assuming that the subsequent radiofrequency ablation would inhibits (by thermocoagulation) the same nerves as the local anaesthetic injected during the prognostic GNB, but for a longer duration, one could envision that the observed benefits last over time. Moreover, the addition of corticoids to the local anaesthetic prolongs the effects of GNB [152, 160], which may be a relevant alternative for the treatment of knee OA pain in poor areas. The results are relevant for physicians in developed countries as well, where GNB and RFA are increasingly performed on patients with intractable knee OA pain who do not qualify for a TKA [153, 183]. Though all the studies assessed the benefits of GNB-RFA on pain and function up to one year after the intervention, no previous study assessed the effects on gait kinematics. Further research is expected to assess the duration of improvements found in this study.

The ecological settings of this study are to be highlighted. Although inertial sensors are intended to be used in out-of-lab settings, most of the studies performed in OA population were still conducted in a laboratory environment [50]. The 6 minutes walking test was adapted with a low-cost sensor to the clinical context, and allowed to easily quantify the improvement in walking long distance. The patients didn't have to move to a specialised laboratory for the kinematic analysis as the research took place directly in the clinic, using the own buildings' stairs and corridors. It required few materials and

standardization. Although this aspect could be criticized, the low standard error of measurement assessed in the healthy subjects, the same operator for assessment [184], as well as the practical aspect of the study did not diminish the validity of the results. The use of inertial sensors in the GNB seems an added value to represent the whole picture of functional improvement in gait. Clinicians should keep these points in mind when planning, or assessing treatment in knee OA patients. Future studies could assess patients in their everyday environment.

The improvement of motor performance implies more factors than pain alone. The patient's overall health, the level of strength, abnormalities of the other joints of the lower extremities or spine, or pain avoidance behaviour could be responsible for movement limitations. Eighty percent of our sample presented chronic low back pain at the time of measurement, whereas the other confounding factors were not assessed.

Limitations

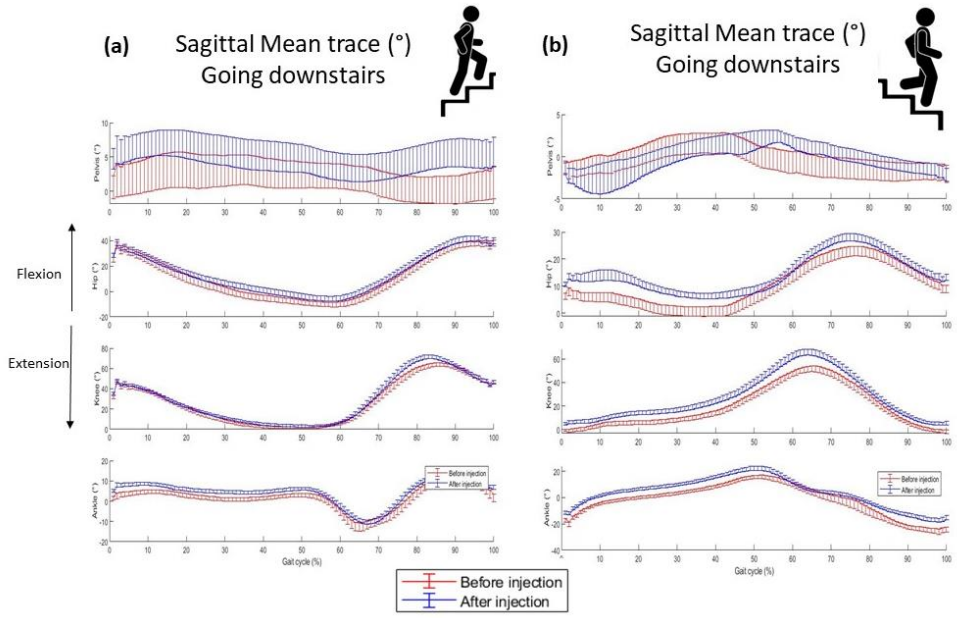
There are some limitations in the current work. First, the exclusive female participant group could be seen as a limitation. However, it is common in Cameroon hospitals that there is a majority female consultation for OA. This could be due to cultural or economic factors. Moreover, the healthy group composed of males and females does not match the OA group.

Second, given the pilot characteristic of this study, data were collected only in the short term and the sample size was low. This limits the clinical applicability and the generalizability of results. The results could only be seen

as potential trends. Ongoing work will assess the long-term maintenance of the improvement in a larger population.

Conclusion

In conclusion, this study showed improvement in hip range of motion during gait, and cadence during stair climbing immediately after a genicular nerve blockade in patients suffering from knee OA in Cameroon. Future studies should look at the maintenance of the benefit of GNB and RFA over the longer term. This is the first study objectifying this effect, through wearable sensors. Inertial sensors could be used to detect functional differences after pain relief therapies. This study has the potential to guide clinicians for the choice of injection techniques for OA management.



Additional file 1. : Joint angle of pelvis, hip, knee and ankle in the sagittal plane during stairs climbing: (a): Ascending stairs. (b) Descending downstairs: Error bar display standard error

Lower limb kinematics improvement after genicular nerve blockade with patients suffering from knee osteoarthritis: a milestone study using inertial sensors

Part III. Free-living conditions

Chapter 1. Instrument validation and psychometrics properties

Concurrent validity of Nokia Go activity tracker in walking and free-living conditions.

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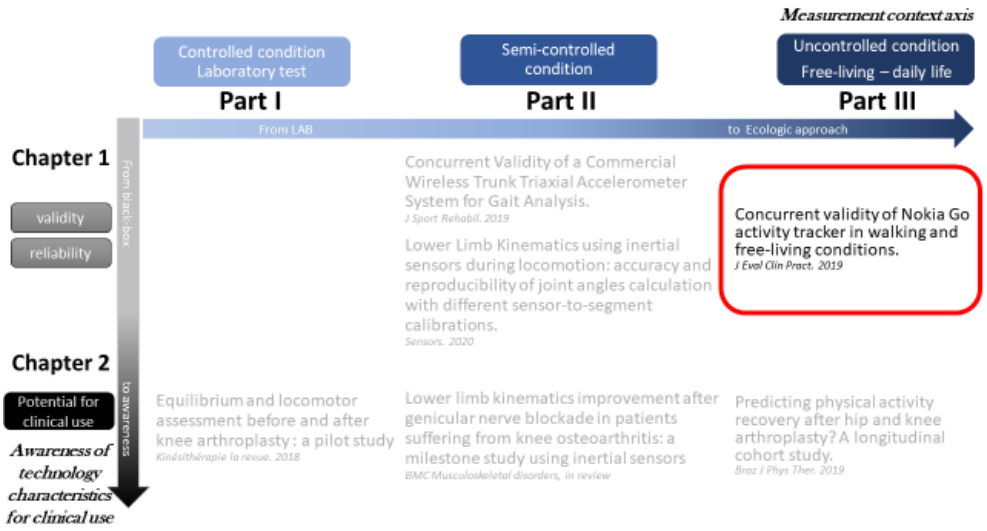

Abstract

Rationale, aims, and objectives

Consumer-based activity trackers aim at quantifying physical activity in a wide range of contexts. Nevertheless, they need to be validated before they are confidently used. This study assessed the concurrent validity of the Nokia®Go against reference devices, according to different sensor locations, in two measurement conditions: during a walking task and during a 24-hour free-living condition.

Methods

We examined the agreement between devices and between locations in the number of steps and total sleep time by using intra-class correlation coefficient and Bland-Altman method.

Results

In the walking task, the agreement is good to excellent for steps between the Nokia®Go and the reference device. In the free-living condition, there is a systematic underestimation of steps in comparison to the ActiGraph. Excellent agreement was found between locations. The device worn at the hip indicated the lowest number of steps and the device located at the dominant wrist, indicated the greatest number of steps.

Conclusions

There are high discrepancies in step count between devices due to the different type of activities in daily life. The Nokia®Go may be confidently used for step counting during pure walking tasks, at different locations. However, the lack of

concurrent validity with ActiGraph call for caution regarding their use in daily living conditions.

Keywords: activity trackers, validation study, free-living, step count

Introduction

Physical activity (PA) and sleep are two critical determinants of physical, social and mental well-being [185]. PA is defined as any bodily movement produced by skeletal muscles that results in energy expenditure[186]. It covers recreational, professional and household activities. Its quantification is possible through questionnaires, but the majority of the population tends to overestimate its PA level [187]. Step count is a simple, objective and intuitive mean to quantify PA. The strengths and limitations of step counting, characteristics et localization of step counters and perspectives in this domain have been thoroughly reviewed [188].

Even if polysomnography is the gold standard to assess sleep patterns, these can also be quantified through actigraphy [189]. The patient and caregiver burden are much lower with this approach, which also enables to assess sleep duration and quality for longer periods and in ecological conditions. Therefore, actigraphy-based sleep assessment is a highly interesting method for assessing sleep both at patient and population level.

Activity trackers, such as the Nokia®Go, can approximate PA and sleep duration, respectively through the number of steps and total sleep time. While initially developed for a wide consumer-based use, they could also be hypothetically used for research and medical purposes. However, these devices need to be validated in order to be confidently trusted for research, or in a clinical setting.

Validity may differ depending on the sensor location and measurement. Our objective was to assess the validity of the Nokia®Go against reference devices (concurrent validity), with various sensors locations and in two

measurement conditions: semi-controlled during walking and uncontrolled during free-living conditions.

Methods

Study sample

Healthy adults were recruited through advertisements within the Université catholique de Louvain for the walking condition and via personal contacts for the free-living condition. Prior to testing, all subjects claimed no pathology affecting their ambulatory capacity. The ethics committee of our university approved the study protocol and each subject provided informed consent before participating.

Activity trackers

Ten Nokia®Go were used for this study. The tracker contains a triaxial accelerometer and can be worn with a clip on the waist or in a band on the wrist. There are waterproof (34.5 x 9.4 mm, and a weight of 9 g) and have a long autonomy (up to 8 months battery life).

Three Nokia®Go (Nokia Corporation, Espoo, Finland) were worn on the dominant wrist, non-dominant wrist and on the hip on the non-dominant side. The Geonaute OnWalk 50 (Decathlon, Villeneuve-d'Ascq, France) was used as the gold standard to measure the number of steps during the walking task. Its reliability has been previously checked in comparison to manual step counting (unpublished data). The Geonaute was worn on the non-dominant side hip. The ActiGraph wGT3X-BT (Actigraph Corporation, Pensacola, FL, USA) was used as the comparison device in the free-living condition to measure the number of steps and total sleep time, and was worn on the non-dominant

wrist, as recommended for sleep assessment and to match the comparison with the watch Nokia®Go. This ActiGraph wGT3X-BT was chosen because it is widely used and validated for the assessment of PA [190].

Testing procedure

In the walking task, subjects were asked to walk an even 1570-meter outdoor circuit (dry dirt and Finnish track) at self-selected speed, one time.

In the free-living conditions, participants wore the devices during 24h. They could accomplish their daily tasks freely and independently. They were requested to abstain from cycling and swimming during the test period to avoid damaging the ActiGraph and to decrease the influence on step measurements during cycling.

The activity of the participants was measured during 24h. The total number of steps and total sleep time measured by the monitors worn on the wrist were recorded.

Statistical analysis

Validity was assessed through different methods: bias (mean difference between trackers x 100 / mean gold standard), intra-class correlation coefficient (ICC; absolute agreement, two-way random, single measures, 95% confidence interval), limits of agreements (Bland Altman method). ICC was interpreted according to Shrout and Fleiss [125]. Those methods were applied to assess (1) the concurrent validity of the Nokia®Go against Gold Standard and (2) the reliability of the Nokia®Go between different locations.

Results

Twenty-four subjects were included in the walking task and 60 subjects were assessed in free-living conditions. Characteristics of participants are shown in Table I.

Table I : Characteristics of subjects

	Walking condition Total (n=24) Mean (SD)	Daily living condition Total n (n=60) Mean (SD)
Gender, M/F	13/11	31/29
Age, years	23.4 (1.3)	39.4 (12)
Height, cm	175.8 (8.4)	171.3 (13.5)
Weight, kg	68.4 (11.8)	74 (20.11)
Dominance, R/L	22/2	59/1
Non-wear time, min	-	10.7 (7.8)
Sport practice , Yes/No	-	16/44
Number of steps	1969 (107)	11351 (3038)
Time (min)	17.1 (1.1)	1440 (0)

M : male, F : Female, SD : standard deviation, R : Right dominance, L : left dominance, Number of steps is the number calculated by the gold standard

1. Concurrent validity

In the walking task, the agreement was good-to-excellent, with ICC ranging from 0.62 to 0.84. The bias for the number of steps did not exceed 0.9%. Limits of agreement are ranging from -11.8% to 10.2% (Table II).

In the free-living condition, agreement was poor-to-fair with ICC ranging from 0.31 to 0.43. The bias for the number of steps range from 35 to 44%, depending on the tracker location. There was a systematic underestimation by the Nokia®Go in comparison to the ActiGraph. Upper limits of agreement are consequently up to 155.8%.

Concerning total sleep time, agreement was fair at the dominant side (ICC=0.53) and good at the non-dominant side (ICC=0.65). Systematic bias was 4.1%, with wide confidence intervals (respectively 46% and 41% for dominant and non-dominant sides).

2. Comparisons between locations

We found excellent agreement for the number of steps for both for the walking task and the free-living condition (Table II).

The bias was under 0.8% for the walking task. Limits of agreement are ranging from -9.6% to 10.9%. In the free-living condition, the comparison between non-dominant wrist and hip, the bias was low (5%), while the highest difference in bias was between the dominant wrist and the hip (13.6%). The device worn at the hip indicated the lowest number of steps. The device located at the dominant wrist, indicated the greatest number of steps.

For total sleep time, the systematic bias was 0.1%.

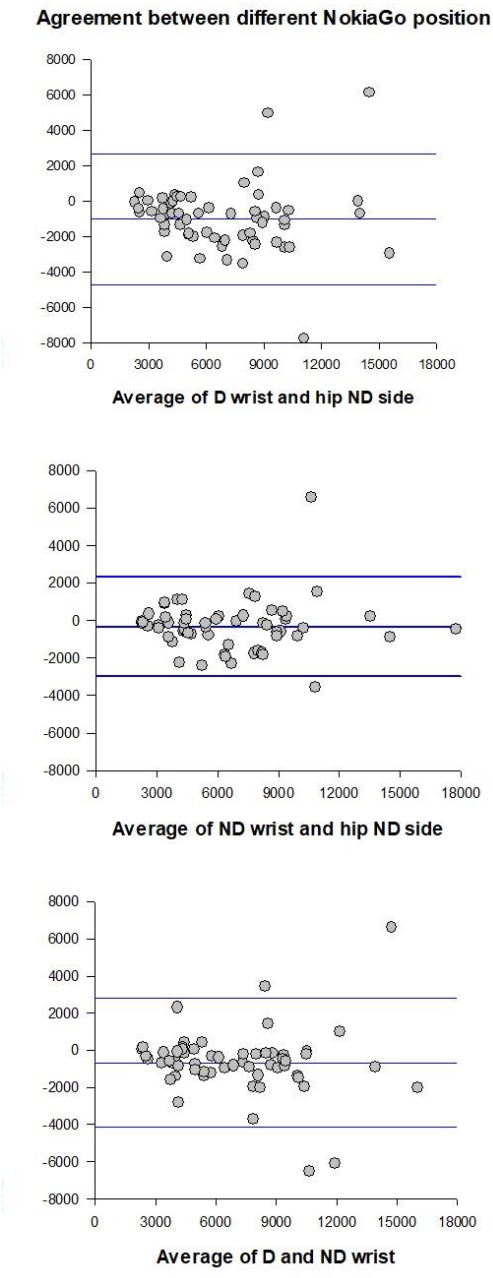

Figure 1 Bland-Altman plot with 95% limits of agreement for the comparison between different Nokia Go locations

Table II : Validity of Nokia go

Part 1 : Against reference device

	Number of subjects	24						60							
		Walking task						**Free-living condition**							
	Location	**Nokia go**	**Geonaute**				**Limits of agreement**	**Nokia Go**	**ActiGraph**				**Limits of agreement**		
		Mean (SD)		**ICC (95% CI)**	**Bias 1**	**Mean difference (Geonaute -Nokia go)**	**Lower**	**Upper**	**Mean (SD)**		**ICC (95% CI)**	**Bias 1**	**Mean difference (Actigraph-Nokia Go)**	**Lower**	**Upper**
Number of steps	**D**	1971 (141)		0.84 (0.65-0.93)	0.0%	2	-7.4%	7.4%	7395 (3347)		0.43 (-0.09-0.76)	-35.0%	3957	-5.7%	112.7%
	ND	1984 (143)	1969 (107)	0.62 (0.46-0.89)	0.8%	-15	-11.8%	10.2%	6725 (3269)	11351 (3038)	0.36 (-0.08-0.72)	-41.0%	4627	4.3%	133.3%
	Hip	1987 (137)		0.81 (0.61-0.91)	0.9%	-18	-8.5%	6.6%	6387 (3312)		0.31 (-0.09-0.65)	-44.0%	4964	-0.4%	155.8%
Total sleep time (min)	**D**								465 (89)	447 (126)	0.53 (0.32-0.69)	4.1%	-18	-48.4%	40.4%
	ND								464 (95)		0.65 (0.47-0.77)	4.1%	-18	-43.5%	35.6%

Part 2 : Between locations

	Comparison between locations	1st Mean (SD)	2nd Mean (SD)	ICC (95% CI)	Bias 2	Mean difference (2nd-1st)	Limits of agreement Lower	Limits of agreement Upper	1st Mean (SD)	2nd Mean (SD)	ICC (95% CI)	Bias 2	Mean difference (2nd-1st)	Limits of agreement Lower	Limits of agreement Upper
Number of steps	D vs ND	1971 (141)	1984 (143)	0.85 (0.48-0.88)	-0.7%	-13	-9.6%	10.9%	7395 (3347)	6725 (3269)	0.84 (0.73-0.91)	9.0%	-670	-55.9 %	37.8 %
	D vs H	1971 (141)	1987 (137)	0.91 (0.79-0.96)	-0.8%	-16.4	-5.1%	6.8%	7395 (3347)	6387 (3312)	0.81 (0.61-0.89)	13.6 %	-1007	-63.6 %	36.3 %
	ND vs H	1984 (143)	1987 (137)	0.80 (0.59-0.91)	-0.2%	-3	-8.7%	8.9%	6725 (3269)	6387 (3312)	0.91 (0.86-0.95)	5.0%	-337	-44.4 %	34.3 %
Total sleep time (min)	D vs ND								465 (89)	464 (95)	0.79 (0.68-0.87)	0.1%	0	-24.9 %	24.9 %

SD : standard deviation, D : dominant side ND : non dominant side, H : hip non dominant side, ICC : Intraclass correlation coefficient (two way random, absolute agreement with a 95% confidence interval), Mean difference is the difference between the second device and the first device (2-1), Bias 1: Mean percentage error of bias : mean difference activity tracker-gold standard x 100 / mean gold standard, Bias 2 : Mean percentage error of bias : mean difference first device-second device x 100 / mean second device.

Discussion

In this cross-sectional, validation study, we explored whether the consumer-based watch Nokia®Go is valid for step counting, in semi-controlled and daily living conditions, and for sleep duration measurement.

In the walking task, the validity against a validated consumer-based pedometer was good to excellent for the Nokia®Go, regardless from location. Previous research had shown that, contrarily, when subjects walked faster than 0.8m/s, which is the case here, positioning the pedometer at the waist tended to produce lower mean errors than at the wrist position.[191] The accelerometer settings and algorithms behind the step counting function of the Nokia®Go could be responsible for this difference. Unfortunately, the manufacturer does not share these data. Nevertheless, in standardized conditions, the Nokia®Go is therefore valid, if, and only if, walking is the only task assessed. Using other activity trackers and pedometers, other authors have found that, for various activities performed in free-living conditions, there are discrepancies in the assessment of the number of steps [192]. These PA monitors employ different step-detection strategies which is responsible for the variability in step counting [193]. The high systematic bias found here is not acceptable (>10%) [194], the limits of agreement are too broad, and prevents interpreting the absolute step count value.

In research setting, a high validity is essential to compare PA at a group level [42]. Given our findings, it cannot be assumed that the absolute step value obtained by the Nokia®Go is similar to other activity trackers. However, in clinical practice, the activity trackers are mainly used to compare the subject

with himself, using the same device. Our results suggest that the Nokia®Go may be used for clinical purpose, at different wearing locations.

The most popular location to wear a tracker is the wrist [195]. However, caution must be taken as our results showed that wrist-worn sensor on the dominant side gives a highest number of step than on hip. A large amount of research has shown that trackers worn close to the centre of mass of the body exhibit a better validity than trackers worn on the wrist, especially during free-living conditions [196, 197]. Nevertheless, recently, Toth et al. have shown, using real world video recording, that some devices were more accurate when placed on the wrist rather than the hip, and that, the overall most accurate device was an ankle-worn pedometer, the StepWatch [198]. At the wrist, more measurement error can occur, due to more variation in the way the arms are used in free-living conditions, as for example, when ironing. This error of measurement can be reduced if the device is worn at the non-dominant side [199].

Regarding sleep time, the Nokia®Go has a low systematic bias. However, the confidence interval of the accuracy of measurement is very wide, and it is therefore wise not to recommend it for an individual-centred assessment. Even if this level of accuracy is comparable to other consumer-based activity trackers, it is insufficient to recommend its use in a clinical setting [200]. However, other previous works have shown that the location of the tracker is not critical for accurate sleep measurement [189, 201], suggesting that a wrist-worn device could validly and reliably assess some important sleep parameters This aspect should be clarified by further research.

Our study presents some strength, such as having a large and heterogenous sample in terms of age range and PA profile, illustrated by the large standard deviation in the number of steps. The ecological validity of our study is high as it was performed during 24 hours under free-living conditions. Therefore, our results can be extrapolated to real world conditions.

Some limitations of our work should be acknowledged. As we only studied healthy subjects, our study sample performed the walking task in the same amount of time. As step count accuracy is affected by walking speed [202, 203], a more heterogeneous sample may lead to different results. Although the ActiGraph is a widely accepted reference for PA measurement, it is not the gold standard for step count in free-living conditions. However, given our settings and the high number of subjects assessed, it was not feasible to use hand counted steps across one day. Algorithm which detect non-locomotion periods could be used in future studies [204].

This is the first Nokia®Go validation study. The Nokia®Go is valid for step counting in walking tasks, among healthy subjects, however it cannot be compared to other trackers in free-living conditions. This tracker is suitable for consumer use, but caution is necessary when devices are located at different locations and when subjects perform physical activities that do not involve body part where the tracker is located. Future work is needed to determine the number of days needed to get a reliable measure on an individual basis, the minimal detectable change for step counting using this device and, finally, involving the device in interventional studies. Furthermore, it should also be validated among disabled patients before it can be reliably used in this context.

Chapter 2. Potential of using locomotor outcomes for clinical use

Predicting physical activity recovery after hip and knee arthroplasty? A longitudinal cohort study.

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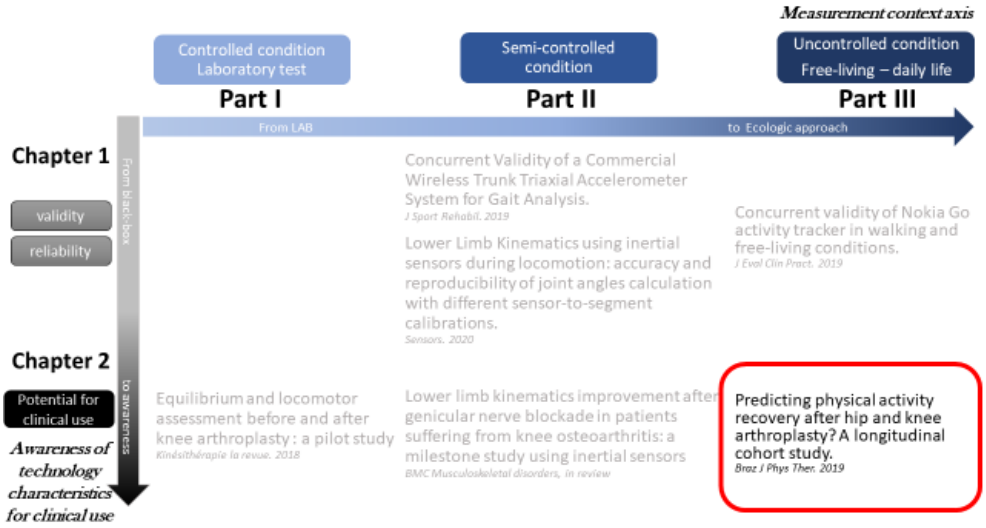

Predicting physical activity recovery after hip and knee arthroplasty? A longitudinal cohort study.

Abstract

Background: Recovery of physical activity (PA) after telerehabilitation following knee and hip arthroplasty (TKA-THA) has rarely been studied. An improved understanding of PA recovery is needed, as it could be influenced by many factors such as age, gender or pre-operative physical function.

Objectives: To assess PA recovery weekly for 3 months after TKA-THA and to determine perioperative factors that could help predict PA recovery at 3 months.

Methods : From one week before until 3 months after surgery, 132 patients wore a fitness tracker continuously. Each patient received personalized and daily exercises and feedback through a tablet. Before and after surgery, patient-reported outcome measures of symptoms, pain, activities of daily living and quality of life were recorded. A one-way repeated-measure ANOVA was used to assess the time effect on step count for each post-operative week. To predict the absolute step count at 3 months post-surgery, a backward multiple linear regression was used.

Results: Patients reached their pre-operative PA level at week 7, with no significant additional improvement by 3 months post-surgery. Pre-operative step count, the number of days using crutches and pre-operative symptoms explained 35 % of the variability of step count at 3 months.

Conclusion: This patient population receiving telerehabilitation reached their pre-operative PA level at 7-week post-surgery with no further improvement over the subsequent 5 weeks. The PA level at 3 months could be predicted by

pre-operative step count, duration of crutches use, and pre-operative symptoms.

Keywords: Exercise; Arthroplasty; Fitness tracker; Wearable; Telemedicine.

Predicting physical activity recovery after hip and knee arthroplasty? A longitudinal cohort study.

Introduction

Total hip arthroplasty (THA) or total knee arthroplasty (TKA) are surgical interventions for end-stage symptomatic osteoarthritis (OA). In Belgium, a total of 49771 hip and knee arthroplasties were performed in 2016 (26736 THA, 23035 TKA [56]) and in the US the numbers are expected to increase significantly until 2030 [57, 205].

Physical activity (PA) is defined as any movement produced by skeletal muscles that requires energy expenditure. PA is beneficial for physical, social and mental health [185], and is a major concern for public health policies [206]. Recovery of PA in the early postoperative phase is still unclear as studies have showed conflicting results. Schotanus et al.[207] observed that PA post TKA improves steeply during the first 4 postoperative days and continues to improve towards the 6-week assessment, however, it remained below preoperative levels. Conversely, Twiggs et al [208] recorded step counts similar to preoperative values at the 6 week post-operative time point. For later stages in the recovery, a systematic review by Arnold et al (373 patients) found that compared to prior to surgery there are negligible changes in PA at 6 months after THA and TKA, but larger changes 1 year postoperatively[66] . Although THA and TKA improve pain and function[185, 209] , it is not clear if PA is also enhanced in comparison to pre-operative level[66] . Moreover, PA post-surgery remains lower than in people without OA[210] .

PA assessment can be conducted through validated self-report questionnaires [211]. Although such questionnaires are convenient, they are subject to patient recollection and therefore recall bias [212]. In addition, change in PA after joint arthroplasty has mostly been assessed as a "snapshot",

at two or three time-points [66], failing to convey the complexity of the recovery. Paxton et al recommended the use of fitness trackers, as accelerometry-based measurements at multiple post-surgical time points during rehabilitation [210]. These fitness trackers enable objective PA quantification over long periods of time, calculating the number of steps per day [213]. These devices can also gather detailed PA data that physical therapists cannot otherwise observe [214].

Although the optimal approach for rehabilitation following TKA and THA is still unclear, telerehabilitation has demonstrated its potential to maximize patient recovery in a cost-effective way [215-217]. Telerehabilitation appears to be effective and comparable to standard rehabilitation for the improvement of physical function and pain [218]. Technological developments combining fitness tracker and tablet use are promising for providing telerehabilitation and for monitoring daily activity [219].

To develop effective strategies to improve PA after hip and knee arthroplasties, it is necessary to thoroughly explore preoperative predictors of changes in PA after surgery [66]. Some studies have already assessed the influence of predictors such as age and gender on PA in patients with knee or hip OA [220, 221]. Female gender [221], old age and a low level of functional performance [220] were associated with lower levels of PA. However to our knowledge, studies assessing the influence of predictors on recovery after TKA and THA are scarce [222]. Twiggs et al. [208] recently observed a gender effect in PA in the pre-op period but not at 1 and 6 week after TKA. In their systematic review, Veenhof et al.[220] observed conflicting evidence for the association between age, gender and the recovery level of PA in patients with knee OA. Twiggs et al.[208] also showed that quality of life, body mass index

(BMI) and preoperative step count predicted PA recovery in patients at 6 weeks post TKA [208]. A better understanding of the association between age, gender, surgery type, PA level and PA recovery after surgery may help to tailor telerehabilitation programs to the patients' needs.

The aim of our study was (1) to assess PA recovery each week after TKA or THA surgery, distinguishing between male and female populations, treated by telerehabilitation and, (2) to explore predictors of PA recovery every week up to 3 months after the surgery.text

Methods

1. Study design

The data for this longitudinal cohort study are from the MORE Clinical Study (Clinical Trial Number : B117201732334), a pilot study part of a national program to create a framework for mobile health application in the Belgium healthcare system [223]. During this study, patients were monitored after knee or hip arthroplasty by a commercially available mobile-health application (Move Up cvba, Belgium) for 3 months after surgery. This protocol was conducted, recorded and reported in compliance with ISO 14155:2011 Clinical investigation of medical devices for human subjects – Good Clinical Practice and ICH-GCP guidelines. In this report, we used data for one arm of study, all patients receiving the same treatment.

2. Setting

Patients had TKA or THA performed between January 2017 and February 2018 in one of three Belgian hospitals. They were given a wearable fitness tracker and a computer tablet one week before surgery. They were

instructed to continuously wear the tracker upon receipt, except during surgery and to synchronize it to the tablet by means of a Bluetooth connection every 7 days.

The ethics committee of Université catholique de Louvain (Brussels, Belgium) approved the study protocol (agreement number: B403201523492) and each patient provided written informed consent to the use of their anonymized data for scientific use.

3. Participants

The study population consisted of 146 adults aged 18 years and over, undergoing a hip or knee arthroplasty, discharged to home environment and able to independently perform daily life activities. Participants had to reside in Belgium during the rehabilitation period, to be able to give voluntary, written informed consent, and to receive no other physical therapy care other than the telerehabilitation program. Participants were identified and recruited by the clinical investigators and their research staff of each hospital.

Exclusion criteria were: pre- or post-operative procedures that might interfere with the rehabilitation, scheduled surgery in the 3 months following the reference surgery, significant neurological conditions that might interfere with the rehabilitation program, psychiatric disorders or active alcohol/drug abuse, incomplete medical record data either before the surgery or at 3 months post-operative period, a rehabilitation period of less than 6 weeks or an interruption in the data collection of the tracker.

4. Variables

The primary, dependent outcome variable is PA, represented by the amount of steps recorded each day with the use of a fitness tracker. PA was measured during one week before surgery to the end of the rehabilitation. To have representative and reliable data, PA was averaged on a sample of 5 to 7 consecutive days for every analyzed period [224]. The mean pre-operative step count was calculated over a 5 to 7 day period depending on the available recorded data. The mean post-operative step count was calculated for each post-operative week. The ratio between the step count at 3 months and the step count in the week before surgery was calculated and expressed as a percentage.

The following patient information was recorded pre-operatively as independent variables: age, gender, type of surgery (hip or knee arthroplasty) and patient-reported outcome measures (PROM) of symptoms, pain, activities of daily living and quality of life within the 2 weeks before surgery. In the post-operative phase we recorded: the number of days of anti-inflammatory drugs intake, the number of days using crutches and PROM 3 month after surgery. PROM were measured by adapted validated scales for the study population. Patients with knee surgery completed the Knee Injury and Osteoarthritis Score (KOOS) while those with hip surgery completed the Hip Injury and Osteoarthritis Score (HOOS) questionnaire using the tablet. These PROMs consist of 42 questions for the KOOS and 40 for the HOOS and investigate five different subscales: symptoms, pain, activities of daily living, sports and recreational activities and quality of life [225]. Scores for each subscale range from 0-100 with higher scores representing better function. The sports and

recreational activities subscale were not used for this study as improvement in these areas was not expected within 3 months of surgery.

5. Data sources/measurement

PA was measured by a commercial fitness tracker (Nokia®Go, Nokia Europe, Issyles-Moulineaux, France). This device was selected for its low cost, long battery life, and waterproof properties [226]. The validity of step count measurement has been previously assessed in comparison to reference devices for walking and free-living conditions [227]. For walking, the bias was limited, while for the free-living condition, PA was underestimated in comparison to data from the ActiGraph device [227].

The monitor was worn on the patient's wrist (side based on patient's personal preference). Compliance with monitor wear was assessed by self-reported take off time and post hoc examination of data. Patients were aware of their daily step count as displayed by the tracker.

The self-administered questionnaires (KOOS or HOOS) were completed electronically through the tablet as well as the recording of the number of days of anti-inflammatory drugs intake and the number of days using crutches.

6. Rehabilitation program

The rehabilitation was implemented through a mobile health solution via a tablet. The patient's progress was monitored daily by physical therapists who offered daily personalized advices and adapted exercises. The daily step count, pain and stiffness reported by the patient, as well as the patient's baseline

characteristics were used to adapt advices and exercises. No specific behavior change principle was used at this stage, but physical therapists monitored for overloading.

The rehabilitation program was also adapted according to patients' expectations and reported pain level during their daily activities and exercises.

7. Statistical methods

Descriptive statistics were used to describe the study population. PA recovery was first assessed by the slope of the linear regression between the weeks (X) and weekly mean step count (Y). Subsequently a one way Repeated ANOVA Measure (RM) was used to assess the time effect (comparing each postoperative week (1 to 11) data with pre-surgery data) in all patients (THA and TKA together). In addition, based on the potential influence of gender [221] and type of surgery [222], the time effect (RM ANOVA) was also assessed for 4 subgroups: Women with knee surgery, Women with hip surgery, Men with knee surgery, Men with hip surgery, using a multiple comparison post hoc test with Holm-Sidak correction. To examine the association between dependent variables (PA recovery assessed by absolute step count at 3 weeks, 6 weeks; 3 months post-surgery) and independent variables, a multiple linear regression was used. The independent variables were the subgroups (Women with knee surgery, Women with hip surgery, Men with knee surgery, Men with hip surgery), age, PROM before-surgery (Symptoms, Pain, Activities of daily living and quality of life), mean PA before surgery (number of steps), the number of days of anti-inflammatory drugs intake, and the number of days using crutches. As described by Freke et al.[228], to limit over-fitting data, a backwards stepwise regression model was performed, using the Akaike

Information Criterion; where variables were removed one by one to achieve the optimum fit for the model, while penalizing for complexity (more variables). This method was automated using SPSS software (IBM statistics SPSS 25). The adjusted regression coefficient of all independent variables entered into the multivariable model were displayed (with 95% confidence intervals) together with the adjusted R2 for each model, to indicate the strength of the model in explaining the variability in the dependent variable. A statistical significance of ≤ 0.05 was used for the overall model, and the association between independent variables within the model and the absolute step count.

Results

1. Participants

From the 146 participants, 11 were excluded because of gaps in data collection (n=3) or a rehabilitation period of less than 6 weeks (n=8). Therefore 132 patients (42% females) were included in the data analysis. The mean age was 62 years (standard deviation: 10). The total proportion of knee and hip arthroplasty was equal (50% vs 50%). Patients took anti-inflammatory drugs on average for 25 days and required the use of crutches for 31 days. The 1 week preoperative and 3 months postoperative PROM data for each surgery and gender categories are presented in Table 1 and Figure 1. Data for the PROM were significantly improved in each subgroup at 3 months post-surgery compared to pre-surgery ($p<0.05$).

Table 1 : Participants

	Variables	Women/knee arthroplasty	Women/hip arthroplasty	Men/knee arthroplasty	Men/hip arthroplasty
	Number of subjects	32	24	34	42
	Age (SD)	60 (11)	64 (9)	63 (9)	62 (11)
Number of days of	**NSAID use (SD)**	29 (25)	29 (42)	28 (24)	18 (24)
	Crutches use (SD)	32 (18)	37 (19)	24 (22)	31 (19)
Before surgery	**Step count (SD)**	4609 (2229)	3742 (2475)	4777 (3960)	5237(2948)
	PROM Symptoms (SD)	53 (16)	44 (16)	58 (17)	56 (21)
	PROM Pain (SD)	45 (12)	45 (17)	55 (20)	51 (18)
	PROM ADL (SD)	51 (14)	48 (17)	61 (20)	53 (17)
	PROM QoL (SD)	31 (13)	34 (13)	40 (18)	41 (17)
3 months after surgery	**Step count (SD)**	4117 (1709)	4865 (4130)	5658 (3346)	5780 (2305)
	PROM Symptoms (SD)	68 (14)*	76 (16)*	67 (16)*	81 (16)*
	PROM Pain (SD)	70 (18)*	78 (17)*	79 (13)*	84 (16)*
	PROM ADL (SD)	74 (14)*	80 (15)*	82 (13)*	81 (17)*
	PROM QoL (SD)	51 (19)*	57 (19)*	61 (16)*	70 (21)*

Data are presented as mean (standard deviation),PROM : patient-reported outcome measures, for hip arthroplasty (KOOS), for knee arthroplasty (HOOS) (0-100 scale, higher scores indicating better function), ADL : activities of daily living, QoL : quality of life, SD : standard deviation, * : statistical difference between **before and after surgery** (* : $p<0.05$)

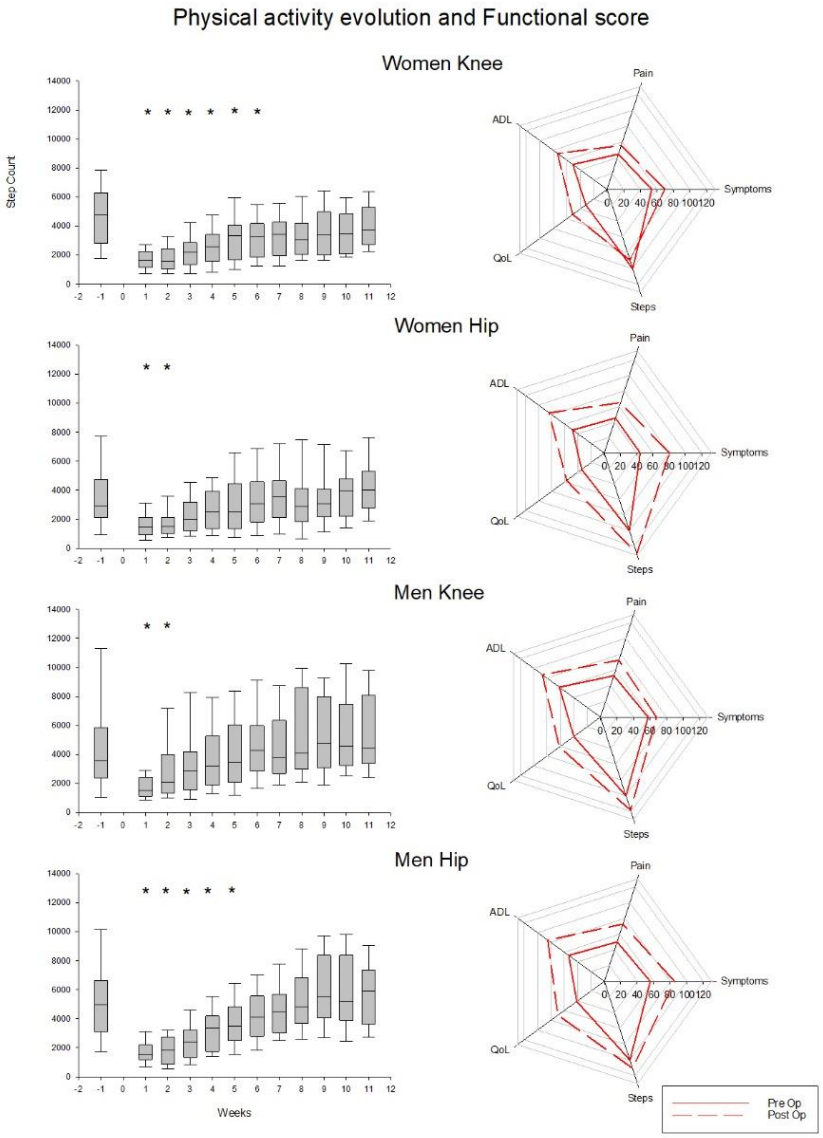

Figure 1: Physical activity progression for the 12 weeks post-surgery and PROMs scores and step count for each subgroup before and 3 months after surgery. On the box plot, * represents statistically significant difference with pre-op step count. On the radar plot, actual PROM scores are presented for each time while step are presented in percentage from the preoperative step count (100%)

2. PA recovery

The time effect for PA, i.e. the mean step count per week, is presented in Table 2 and illustrated in Figure 1. All patient groups presented a significant reduction in PA during the first weeks compared to the pre-operative value. Speed of PA recovery varied between groups, as illustrated by the step count increase per week (Figure 1).

Starting from 7 weeks after surgery, PA values were similar to pre-operative PA values. However, considering the wide 95%CI in the subgroups (Table 2), not every patient recovered to their pre-operative activity level. Finally, none of the subgroups showed significant additional improvement between 7 weeks and 3 months post-surgery.

Women with hip arthroplasty were the most sedentary group pre-operatively (mean [95% confidence interval] step count of 3742 [2751:4732]). They presented no significant difference with pre-operative step count at week 3 post-surgery, with no additional improvement beyond that point. Women with knee arthroplasty had the slowest recovery (241 step increase/week), while men with knee and hip arthroplasty respectively reached their pre-operative PA level at weeks 4 and 5 before reaching a plateau.

3. Predictors of PA recovery

Results of the stepwise backward regression are presented in Table 3. For the step count at 3 and 6 weeks, pre-operative step count, the number of days using crutches and the score on the pre-operative quality of life subscale on the PROM explained respectively 32 and 43% of the variability. For the step count at week 11, pre-operative step count, the number of days using crutches

and the score on the pre-operative symptoms subscale of the PROM explained 35% of the variability.

The pre-operative step count variable was the strongest predictor at week 11 (26.6%) while the pre-operative symptoms (change in R2: 5.2%) and crutches use (change in R2: 3.4%) had less weight in the model.

Age, sub group (1 to 4), symptoms, pain, activities of daily living (PROM) and number of days using anti-inflammatory drugs were not significant and therefore not included in the regression model ($p>0.05$).

Table 2 : Multiple comparison post hoc test with Holm-Sidak - time effect across category based on gender and surgery type

		All		**Women/knee arthroplasty**		**Women/hip artroplasty**		**Men/knee arthroplasty**		**Men/hip arthroplasty**	
Number of subjects		132		32		24		34		42	
		Mean	*95% CI*	*Mean*	*95% CI*	*Mean*	*95% CI*	*Mean*	*95% CI*	*Mean*	*95% CI)*
Pre-op step count		4695	(4189,5200)	4609	(3836,5381)	3742	(2751,4732)	4777	(3536,6017)	5237	(4345,6128)
Slope regression : Step count increase/week		365	(321,408)	241	(187,294)	288	(171,404)	401	(315,486)	474	(397,550)
		diff	*95% CI*	*Diff*	*95% CI*	*diff*	*95% CI*	*diff*	*95% CI*	*diff*	*(95% CI)*
Difference between preoperative level and week	1	**-3082**	(-2445,-3515)	**-2839**	(-2033,-3736)	**-2136**	(-965,-3174)	**-3040**	(-1656,-4423)	**-3420**	(-2579,-4462)
	2	**-2675**	(-1979,-3162)	**-2818**	(-1975,-3690)	**-1829**	(-472,-3051)	**-1932**	(-384,-3479)	**-3248**	(-2392,-4305)
	3	**-2050**	(-1276,-2615)	**-2305**	(-1404,-3233)	-1254	(-294,-2668)	-1208	(550,-2966)	**-2592**	(-1673,-3712)
	4	**-1531**	(-753,-2110)	**-1864**	(-883,-2872)	-858	(680,-2262)	-712	(1025,-2449)	**-1874**	(-985,-3062)
	5	**-1126**	(-323,-1720)	**-1380**	(-339,-2448)	-536	(1022,-1962)	-401	(1405,-2207)	**-1455**	(-510,-2601)
	6	**-657**	(-188,1292)	**-1315**	(-364,-2293)	-92	(1625,-1677)	167	(2045,-1711)	-743	(305,-1993)
	7	-443	(358,-1091)	-1203	(-202,-2232)	261	(2090,-1434)	152	(1928,-1624)	-421	(576,-1620)
	8	-59	(824,-772)	-1152	(-124,-2207)	-101	(1828,-1896)	986	(2933,-961)	341	(1429,-947)
	9	250	(1114,-469)	-907	(-189,-1831)	102	(2011,-1673)	865	(2736,-1006)	942	(2077,-393)
	10	270	(1178,-421)	-733	(-41,-1949)	521	(2521,-1275)	1202	(3106,-702)	624	(1781,-653)
	11	416	(1263,-289)	-422	(500,-1484)	1021	(3169,-923)	881	(2683,-921)	643	(1691,-605)

Differences highlighted in bold are the statistical significant difference with pre-op step count level. Negative difference means lower step count value than pre-op, positive difference means higher step count value than pre-op. diff : mean difference with pre-op step count, 95% CI : confidence interval for difference

Table 3 : Stepwise backward regression: adjusted estimates (95% CI) of the association between dependent variables and physical activity level post arthroplasty

Outcome (PA)	Age	Category	PROM pre-op				Behavior		PA	Adjusted R2 (p-value)
			Symptoms	Pain	ADL	QoL	NSAID use	Crutches use	Steps pre-op	
Step count week 3	na	na	na	na	Na	32.1 (6.4,47.7) (p=0.015)	Na	-30.4 (-49.5,-11.3) (p=0.002)	0.31 (0.17,0.44) (p<0.001)	0.324 (p=0.015)
Step count week 6	na	na	na	na	Na	49.9 (14.8,73.2) (p=0.004)	Na	-43.9 (-65.7,-22.1) (p<0.001)	0.45 (0.29,0.60) (p<0.001)	0.431 (p=0.004)
Step count week 11	na	na	32.3 (2.8,61.7) (p=0.032)	na	Na	na	Na	-28.9 (-55.6,-2.3) (p=0.034)	0.51 (0.32,0.69) (p<0.001)	0.352 (p=0.034)

pre-op : pre operative, ADL : activities of daily living (PROMS), QoL : Quality of life (PROMS), NSAID : non-steroidal anti-inflammatory drug, PA : physical activity (stepcount), na : no association

Discussion

The main results of this study were that individuals post TKA and THA recover rapidly to their pre-operative PA levels over the first six weeks after surgery, which was followed by a plateau until the end of the follow-up period at 3 months post-surgery. The level of PA at 3 months post-surgery was mostly predicted by the preoperative PA level, the number of days of using crutches and the pre-operative symptoms as measured with a PROM.

PA recovery

We observed a fast recovery of PA in the first weeks, followed by a steady state at the pre-operative level until the end of the follow-up period. Function and pain improved over time from the preoperative levels but were not associated to PA increase. Post-operative PA recovery varied depending on gender and type of surgery. Women with hip surgery recovered faster but were also the most sedentary group at baseline. Although women with knee surgery didn't fully recover to their pre-operative PA level, there was no significant difference between step count at 3 months and the preoperative step count. They recovered at a slower pace and exhibited the lowest functional score (PROMS) after 3 months. Those results suggest that this patient subpopulation could benefit from better guidance in the recovery process.

In our sample, the 7 week time point appears to be a milestone in the patient's recovery process, followed by a steady state. Crizer et al.[222] also observed a gradual PA recovery for patients with THA and TKA, with a slower recovery for TKA. Those patients regained their baseline PA after 9 weeks [222] while patients post-THA recovered to their baseline at week 5. The faster recovery in our sample could be partly attributed to the telerehabilitation

program and the PA awareness developed by the patients [229]. But in the current study, a control group with standard of care is lacking. Our sample also had higher baseline PA than in the study by Crizer et al.[222] with a PA baseline for all of our patients at 4695 steps while their PA baseline was 2682 steps for THA and 1988 steps for TKA.

Self-reported measurements of pain and function improved significantly post-surgery. This result is in agreement with studies on the effectiveness of arthroplasty for reducing symptoms [215]. But PA did not increase relative to pre-operative level. Similar findings demonstrating this disassociation have been demonstrated in several studies [230, 231]. PROM are central to orthopedic research, clinical practice and quality control [232], But they provide only one aspect of functional recovery. Other objective markers of successful rehabilitation are therefore needed. While pre-operative PA level could be a good indicator of returning to normal physical habitus after surgery, the association of PA with the return to professional and recreational activities, patient specific goals [233] and participation according the International classification of Functioning, Disability and Health need to be addressed in future research [234, 235].

Predictors of PA recovery

Our second aim was to explore predictors of PA recovery every week up to 3 months after TKA and THA. The PA recovery at 3 and 6 weeks could be predicted by preoperative PA level, the number of days using crutches and the pre-operative score on the quality of life subscale of the PROM. While the PA at 3-months could be predicted mostly by the preoperative PA level, the number of days using crutches and the pre-operative score on the symptoms

subscale of the PROM. At week 11, preoperative PA level is the most important predictor. It means that a higher preoperative PA level, less days using crutches and less preoperative symptoms predict a better PA outcome.

Twiggs et al.7 also observed a relationship between the physical component of the SF12 (SF12 PCS) and 6 week post-operative step count and explained 75% of the PA level at 6 weeks with preoperative step count, gender and score on the SF12 PCS [208]. It means that such PROM results could be useful in the step count target in rehabilitation. However, Crizer et al.[222] observed in 589 patients that lower extremity functional scale (LEFS) scores were only weakly correlated with step count and that the correlation became increasingly weak with time after surgery, with no clear correlation at 12 weeks postoperatively [222]. This lack of correlation between preoperative symptoms and PA level observed in our study could be surprising. Past experiences with illness, personality, coping styles, family and cultural norms, and current interpersonal interactions are examples of factors that interact to influence PA level [236]. Surprisingly, age was not retained in our model while Toogood et al observed an age effect by a categorical comparison (<70 years versus > 70 years) [237]. This discrepancy could be explained by the statistical method as well as by differences in our patient sample.

The number of days using crutches seems to be associated with PA and could represent an objective quantitative measure of disability. While this assistive technology may be beneficial for individuals with OA, their impact on mobility and community participation depend on the user's ability [238]. Caution is needed in the understanding of the relation between PA and crutches use, because of its potential impact on the accuracy of the activity

tracker, which may be reduced when using crutches [239]. However, a recent study on step recording validity in subjects walking with crutches reported a small bias, unlikely to affect our study results [240].

Perspectives

Previous research observed few changes in PA evolution at two or three time points after arthroplasty [66]. In contrast, a few recent research investigated recovery speed and found different results regarding pre-surgery PA level [207, 208, 222]. Our study adds more information in the emerging field of continuous activity monitoring. Moreover, it is the first study to provide activity data from patients, which performed their rehabilitation through a tele rehabilitation application. However, as we did not have a control group, have no data on co-morbidities and functional recovery, it is not possible to conclude on its effectiveness on PA change.

Our results provide information allowing the development of personalized step count goals for patients, based on their pre-operative activity status and characteristics. Step count is not a new marker, but its accessibility to healthcare providers is increasing. The relevance of this measure in the 3 months acute phase could contribute to adapt treatment frequency and intensity in the recovery of pre-operative PA level. The use of relevant and objective, continuous or high-frequency digital measures of clinical efficacy and disability (pain severity, PA, sleep quality, activity pattern, etc.) could have a positive influence on the rehabilitation protocols and lead to more personalized care. Their potential for clinical decision making in personalized rehab should be evaluated in future research.

Measurement of PA level at further time points may prove informative. Indeed, in our sample, the PA level was low in comparison to healthy individuals and PA guidelines [241]. The main goal of surgery is pain relief, but expectations in a public health perspective could be to increase PA level in the long term. Attention to social and community participation, in the context of a multidisciplinary approach to post-operative care, could be paramount in improving PA [234].

Furthermore, the device used in our study provides the patient feedback on their activity level and allow a comparison with their pre-operative PA level. This technology will allow to define patient-specific step goals to reach in the post-operative period and assess their clinical utility in PA improvement in future studies [229].

Limitations

Our study has some limitations. First, we did not control for comorbidities, BMI or social support which could influence PA [221] and we lacked information such as the pre-operative use of crutches or working status. Secondly, we reach the conclusion that PA is unchanged from the pre-operative situation at 3 months, but changes to PA are difficult to detect [242] and depend on the outcome measure. An increase in PA would require 6 to 12 months to be clearly highlighted [242]. Moreover, the variability of the step count estimates in our sample limit the interpretation. Considering that a whole subgroup recovered their pre-operative PA level at week 11 could be a mistake as the confidence intervals are wide.
We also compared our results with other studies but comparison with other studies is difficult as the step tracking method differ. Indeed, our fitness tracker

(Nokia®Go) has an excellent agreement with the reference device in a walking task but systematically underestimates step count in comparison to the Actigraph wGT3X-BT in a free living task (41% difference) [227].

Conclusion

In conclusion, our patient population receiving telerehabilitation reached their pre-operative PA level at 7-week post-surgery and subsequently maintained a steady state until 3 months post-surgery. Women with knee arthroplasty recovered slower and experienced more pain and symptoms. The PA level at 3 months could be moderately predicted by pre-operative step count, duration of using crutches post-surgery, and pre-operative symptoms level.

Predicting physical activity recovery after hip and knee arthroplasty? A longitudinal cohort study.

General discussion

Thesis summary

This thesis aimed to investigate mobility outcomes among patients with knee osteoarthritis in a clinical context, as well as semi-controlled and free-living conditions. In the first part, the assessment of locomotor outcomes in a clinic context after arthroplasty was not conclusive due to a lack of adherence to the evaluation. The main cause for refusal and drop-out was the distance between the hospital and the patients' home. The second part of the thesis has been carried out in semi-controlled conditions. First, the validity of inertial sensors (the tools needed to assess the semi-controlled condition) was evaluated in comparison to gold standard technologies. Afterwards, the potential of wearable sensors was tested for the assessment of an innovative therapy for treatment of knee pain in Cameroun. The wearable sensors allowed to show improvement of mobility after innovative therapy during gait and stairs climbing. In the last part of the thesis, an application of sensors (activity trackers) was explored in free-living conditions. The device was used continuously during 3 months on patients which had received an arthroplasty associated to a telerehabilitation treatment. The recovery of their physical activity level to the pre-operative level was achieved around the 7th week post-operatively. This thesis opens perspectives for the development of the use of wearable sensors in outdoor conditions.

Validity and clinical relevance of sensor-based mobility outcomes

The lack of quantitative tools to measure mobility outcomes in ecological settings motivated the validation of technologies and their implementations in ecological settings. Considering the methodological and technological challenges raised by this choice, a step-by-step approach towards uncontrolled conditions was chosen.

Psychometrics properties: reliability and validity

First, the psychometrics properties of used technologies were assessed. Their validity was assessed in comparison to gold standard measurement systems. As discussed in chapter 1, the validity of sensor-based outcomes depends on many factors such as the task assessed, the duration of the experiment, the plane of movement, the subject or the calibration procedure. The inertial sensor (BTS G-WALK®) displayed **excellent concurrent validity** for measuring **spatio-temporal parameters of gait**, but the results differed depending on the outcome used. Single and double support accuracy determination depend on the identification of the final toe off contact event of gait. The **identification of this event** is especially difficult in **pathological populations** [243] due to the irregular trunk acceleration patterns, the use of walking aids and the low gait speed. During **daily life**, this error could be higher because of the subjects' natural behaviour (variating the direction of progression and repeated stopping and starting). Consequently, the performance of the algorithm is expected to decrease in uncontrolled conditions weakening the clinical applicability [243].

Afterwards, a custom built system based on open source inertial sensors was used to record joint angles. **Clinically acceptable errors** (<5°) [112] were obtained using several functional calibration movement procedures. It confirmed that inertial sensors can be used confidently for walking in ambulatory settings. However, we had to consider **constraints**, such as short measurements in time. We chose not to use the magnetometer which might have enabled longer recordings because the use of this component in sensor fusion made recordings sensitive to ferromagnetic perturbations in the environment [244]. **Calibration procedures** have to be as easy as possible, requiring little expertise. The focus of automatic calibration requires **further development**, for example by implementing a method to automatically determine which sensor has been attached to which body segment [245], so that the patient can attach them autonomously. Easy sensor-to-segment calibration, such as the walking calibration used in this thesis, may allow for automatic calibration without burdensome procedures.

Finally, we explored the validity of the consumer-based watch "Nokia go" in semi-controlled and daily living conditions to measure **the number of steps**. This device was found **valid** in standardized conditions (walking), but discrepancies were found in free-living conditions, between the "Nokia go" and the reference system. Moreover, its validity is questioned in tasks implying **upper limb movement** (when the device is worn at the wrist) or in **impaired walking** (as for example during walking with crutches) [240]. However, in clinical practice, the activity trackers are mainly used to track the activity of the subject itself, using the same device (relative comparison). Therefore, they are still **suited for consumer-based self-monitoring** (e.g. in behavior change

interventions) rather than to evaluate research outcomes (objective comparison) [246].

Clinical potential of the sensor-based mobility outcome

Considering all those constraints for valid use, we then explored the clinical potential of the mobility outcome measures with patients suffering from OA. The first study was performed in a **clinical context, using a baropodometric treadmill** to assess temporal parameters of gait and postural outcomes. Due to the low rate of participation of subjects with arthroplasty (6%), insufficient **feasibility** was provided for potential larger future studies [247]. The main reason for refusal or dropout raised by patients was the distance between the hospital and their home. Therefore, the clinical interest of the baropodometric treadmill for TKA management was not supported by our results. Few changes were observed, possibly due to the **standardized environment** of the study: walking at a steady state on a treadmill was not a challenge for the patients pre- nor post-operatively. This result constitutes a second motivation to use ambulatory technologies to assess patients (1) at home (2) in more challenging daily life situations.

Therefore, we assessed the **efficacy of an innovative treatment** (pain relief in knee OA patients) in a clinical and more challenging context. The peri-articular injection of genicular nerve is a low invasive technique of major interest for people who are still suffering after TKA or for the populations that have difficult access to arthroplasties (such as is the case in Cameroon). The inertial sensors allowed to quantify kinematic improvement after injection for **hip ROM** during **gait** and **pelvis ROM** in **ascending stairs**. This is the first study in Cameroon to objectify short term kinematic improvement in OA

population, which opens new perspectives for lowly funded hospitals that had lesser access to quantified measurement techniques before.

In all studies, **gait speed and cadence** varied: i.e. after a knee arthroplasty, after an injection, or depending on the walking surface. Although these variables are already being taken into account by physiotherapists in clinical practice, the assessment of them in ecological settings raise multiple questions that will be addressed in the next section.

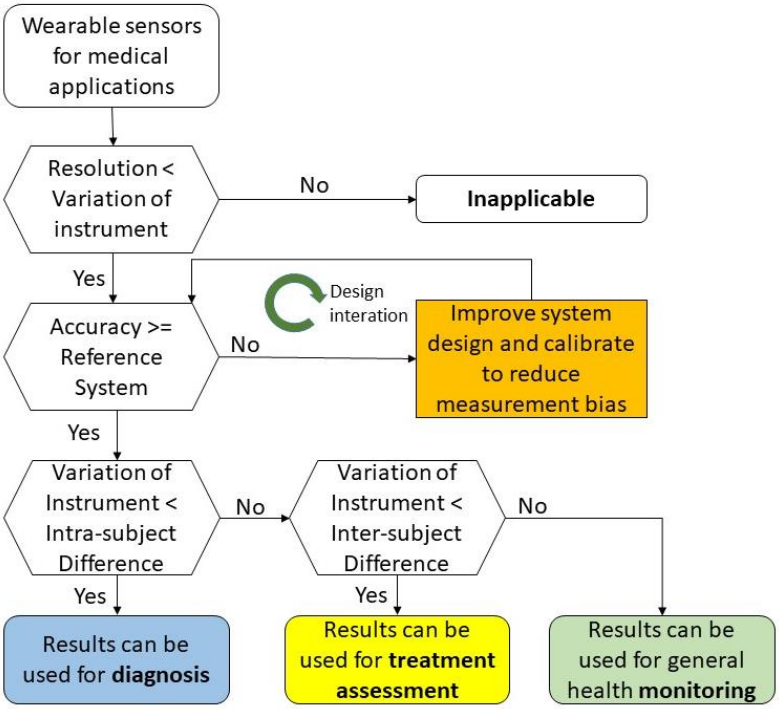

Figure 1: Impact of wearable sensor data quality on clinical applications. The condition for the measurement result to serve for diagnosis is the strictest, as diagnosis requires both high precision and high accuracy from the measuring instrument. (Accuracy refers to the closeness of a measured value to a standard or known value. Precision refers to the closeness of two or more measurements to each other. Precision is independent of accuracy). If the measurement uncertainty is higher (i.e., the precision of the system is lower) than the intersubject difference, the differences shown in the measurement results cannot be trusted to differentiate a patient group from a control group [248]. <: inferior, >= superior or equal. (Resolution is the "smallest change in a quantity being measured that causes a perceptible change in the corresponding indication) Adapted from Chen et al 2016

The clinical relevance of measurement arises from its discriminative power. If the instrument does not achieve high accuracy and precision (Figure 1), such as an **activity tracker**, it can still be used for **monitoring.** Moreover, this device is currently easily **accessible** for everyone. Within the framework of this thesis, the device is particularly interesting as it can reflect behavior in free-living conditions, and can measure continuously (as opposed to the conditions in Part I and II, where subjects were measured for only one minute). The use of this activity tracker for behavioral change or medical treatment is increasing [249, 250]. In our thesis, the tracker was used in a longitudinal cohort study to track **patient recovery** during 3 months after arthroplasty. The sensors allowed to **objectify** the recovery of pre-operative physical activity (PA) level over the first six weeks after surgery. The fact that this **information was not retained by physiotherapists without technology[53],** combined with the **sedentarity** of this population [221], are factors increasing its clinical relevance to manage people with knee OA. If you cannot measure PA, you cannot improve it. This opens perspectives to develop behavioural strategies in order to improve PA level for people with OA in the long term [251, 252]. In this respect, wearable sensors create opportunities to optimise at-distance-care [253]. Feedback on mobility data may motivate better self-management as well as increase exercise adherence [219, 254].

Challenges regarding ecological assessment

Unsupervised assessments of mobility variables in ecological settings may in the future become an important aspect of clinical decision making. However, there are many challenges to be met.

1st challenge: interpretation and comparison of data in standardized settings with high variability data from unsupervised settings

Cadence was modified in almost each clinical context studied, in certain circumstances highlighting a potential benefit of a treatment. This outcome is related to **walking speed**. Walking speed is the simplest outcome to monitor functional status and health in a wide range of populations [255]. A slower speed is associated with greater mortality, morbidity and fall risk [255]. However, this outcome is affected by many environmental factors and displays a wide variability in free-living context. It thus raises questions about its interpretation. Indeed, a supervised assessment involving only one trial delivers average gait speed for a subject, whereas a continuous unsupervised evaluation might include hours of walking and deliver a wide variability of gait speeds [256] (Figure 2).

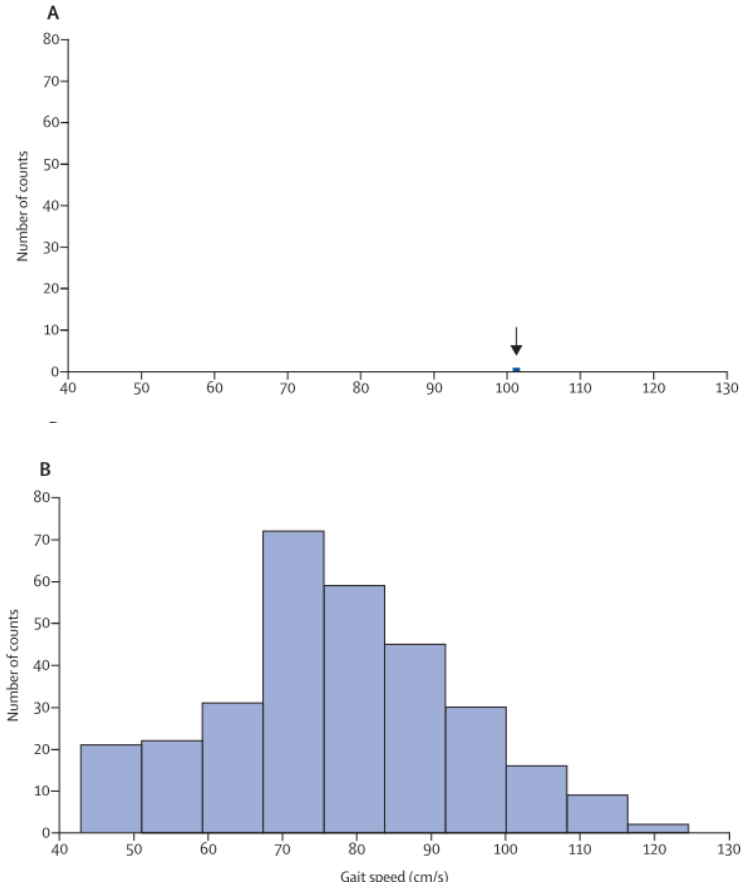

Figure 2: Gait speed measurements based on evaluation in the laboratory and in the daily living environment. Upwards, a single mean speed value is displayed for the subject while the scope of his speed is displayed on the histogram downwards. From Warmerdam et al.[256]

The variability in unsupervised settings reflects the adaptive performance of a subject to multiple environmental factors, as observed in our study on cobble stones in Part II. The drivers of this variability often remain unknown as it is almost impossible to standardize the environment. Dual tasking, stress, walking surface, noise, traffic, distractors, obstacles are some of the various factors that influence gait speed [257-259].

Nevertheless both approaches, i.e. standardized and ecological, can provide valuable information for mobility assessment. The **interpretation and comparison** of the information coming from both approaches is necessary for therapists. How to interpret and compare single speed data with many data obtained from a distribution of gait speeds (Figure 2)? Are the extremities of the distribution representative of disability? These are relevant questions that future studies could address by deeper analysis of the optimum state (the best 10% of an individual's distribution) and the lower range of an individual's distribution throughout the day [256].

Despite these questions, it is highly interesting to underline that research is already ongoing towards long term study of free-living gait speed. A consortium of 34 universities, hospitals and industry is developing a low cost, wearable digital technology to monitor and measure real life walking speed [260]. They are focusing on 5 different patient groups in 12 different sites across Europe and hope, with this promising research project, to drive the future of gait speed assessment and potentially guide practices.

2nd challenge: technical limitations in real life situations

Initially, the research project's objective was to measure behaviour related to knee instability as well as to measure the quality and quantity of movement. However, the experience of knee instability is an unpredictable event [261, 262], that would require **long duration of recording**, in order to be able to detect such unpleasant event in kinematic records. However, it was not technically feasible to reach this objective, due to technical limitations (drift over time with the IMU algorithm and therefore duration of recording limited to one minute). Continuous recording using AHRS will require further

improvements in algorithms so that the magnetic compensation is optimized for longer recording periods [110]. Moreover, a long duration of recording implies **huge amounts of data**, coming from multiple sensors. It can be challenging to identify numerous sequences of **comparable data in real life situations** [263] to grasp quality of movement. The development of automatic data processing is a critical problem related to large real life activities, as well as noise and artefacts [264].

3rd challenge: contextual factors affecting mobility variables

In Part II, we explored the semi-controlled condition which appears as a **trade-off** between the standardized and ecological approach. It allowed to examine tasks under conditions comparable to those in which the individual with OA typically functions. However, once more, many factors contributed to the signals acquired and part of them cannot be controlled as the data was acquired in a non-controlled condition [264]. The **distractors and unforeseen events** rendered necessary for us to perform measurements several times to increase confidence for data interpretation. This again highlights the **tension** between the need for control and the need to preserve the essence of the phenomenon under investigation [256].

In order to interpret the mobility variables, we would need **to be aware of all potential environmental factors that could affect** them. Nevertheless, in Part II -Chapter 2, we were not able to standardize and measure the weather conditions or the environmental noise when walking on outdoor surfaces such as grass or cobblestones. Other research teams have chosen another approach by **creating a real controlled environment** inside a laboratory. The KITE Lab in Toronto built an instrumented lab in order to create an icy/snowy

environment that allowed them to test the safety of walking with snow boots [265]. This option remains expensive and is not applicable for management of OA in clinical practice. A similar approach is performed in a **living lab** [30, 266] where intensive research is performed in smart homes to promote healthy aging and autonomy in elderly. These homes are equipped with multiple sensors in order to characterize the interaction between the subject and its environment.

Virtual reality (VR) is another interesting approach, with a much lower cost than building a specific instrumented lab. VR as a technology allows to **standardize the visual flow** of the subject and to **provoke perturbations and facilitators** to help subjects to perform a task [267-269]. Unpredicted perturbations in VR can increase the face validity of experimental models of fear, anxiety and avoidance [270]. This technology has already been tested in the treatment of several pathologies, such as phobia [271] or low back pain [272]. VR could also be used for mobility assessment with a **treadmill**, or in a **dedicated space**. Stride time or step length are two spatio-temporal parameters used to assess mobility. A preliminary study revealed an increase in step length when walking on a treadmill in a VR environment, compared to a treadmill conditon at the same speed without VR (*article in preparation with A.Lheureux*- Figure 3-a). For this experiment, the patients were equipped with **inertial sensors** at both limbs. This raises perspectives for assessment and treatment in ecologically valid situations of daily living, with speed standardization. Other research aimed to measure and improve specific gait parameters in a virtual game with optical cueing in a market store (*work in progress Thesis Nguyen-Moens*- Figure 3-b). The use of visual cueing in an ecological environment combined with the measurement of gait patterns

through positional trackers, might help tailor relevant treatments (called *"serious games"*) to increase i.e. step length [273]. Therefore, these last VR approaches may relevantly help interpreting the mobility variables, in relationship with several contextual factors.

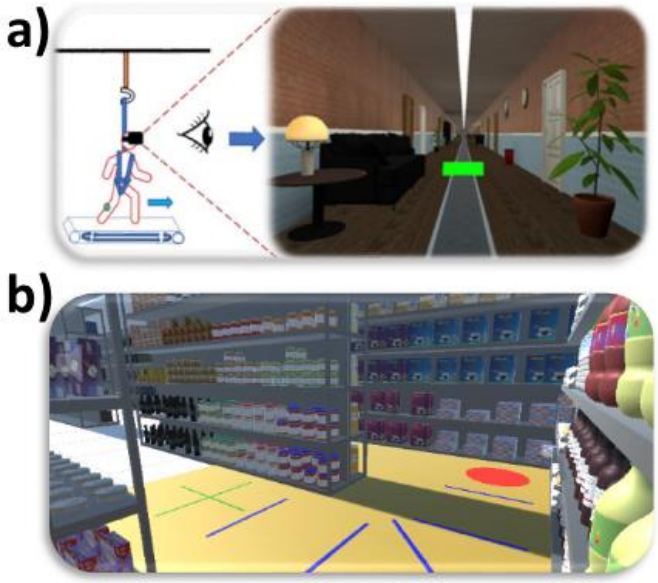

Figure 3. Combination of treadmill walking and immersive virtual reality (a) Patient walking on the treadmill wearing a non-weight bearing security harness attached to the ceiling and an immersive virtual reality – what the patient sees in VR: a long hallway moving forward at the same speed as the speed of the treadmill. Image from A.Lheureux. (b) Market store environment to improve gait behavior of Parkinson disease patients. The blue lines are indicators to increase step length and help navigation. Image from H. Nguyen and L. Moens.

Perspectives regarding technology and methods

As highlighted in the introduction, mobility is a multidimensional construct which is challenging to measure, especially in unsupervised settings. Therefore, we suggest several technological perspectives and methods for a better capture of mobility.

1st perspective: artificial intelligence to improve the validity of sensors and to answer clinical questions

As highlighted in Chapter 1 of this thesis, **validity of sensors** can vary depending on the sensor site or the activities performed by the subject. The detection of the toe off event in the gait cycle e.g. may be improved by using **machine learning (ML) methods** [108, 248], as it was tested in a master's thesis on stroke patients walking with two sensors attached to the ankles (*unpublished data – Student Thesis M.Schinkus*) while walking with the Wheeleo cane [274].

Algorithms that can accurately estimate walking cadence and speed for long recording of daily life walking were recently developed and validated in various environments [275] and pathologies [276]. However, the **distinction between walking periods and other activities of daily living** is needed. The combination with algorithms that classify the different activities of daily living could provide a valid functional performance for each activity performed by the subject in daily life [277, 278].

Overworked therapists need easy and fast reliable information to manage treatment and make informed clinical decision [279, 280]. They are used to take advantage of their work experience in the recognition of

pathological movement pattern and for fast-thinking reasoning [281]. How can we acquire robust data, in a limited quantity, that represents the pathomechanics or the improvement after a treatment? To decrease the variability of sensors, new **options for data analysis must be operated** [256]. Gait is a complex movement which cannot be resumed to a single variable such as gait velocity or ROM [282]. The mapping of a multivariate temporal pattern and the identification of disorders are obstacles to gait analysis in clinics. Artificial intelligence and ML techniques appear as opportunities to model the complex structure of human gait in a holistic way [283]. The **association of IMU and ML** techniques can use gait data to provide a cost-effective and non-intrusive means of monitoring locomotor function [284, 285]. This could **"automatize" functional assessment [286]** by **reducing the dimensionality** of huge amounts of data [108, 287, 288], that a therapist is apprehending with his eyes and his work experience. To illustrate this, we attempted to represent the kinematic improvement observed in the study performed in Cameroon by using ML feature selection algorithms for walking and climbing stairs (Figure 4). We used a mathematical process (a correlation-based feature selector - CFS –Appendix 1) to determine the best predictors of the improvement in the 14 knee OA patients. Such an approach could help to find limited numbers of variables that best represent the subjective improvement observed by therapists, and the pain relief experienced by patients.

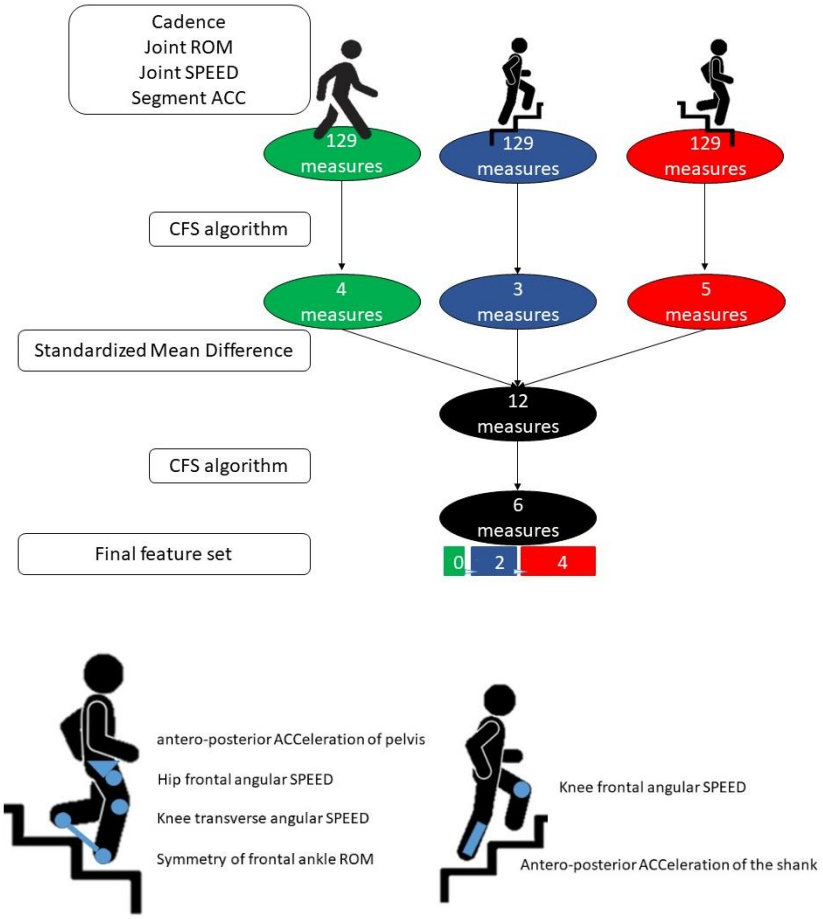

Figure 4: Procedure and sequential reduction of measurements using CFS. 129 measurements representing cadence, joint range of motion and joint speed of the pelvis, hip, knee and ankle, as well as linear segment accelerations (ACC) of pelvis, thigh, shank and foot were used as input for the CFS algorithm. The algorithm selected the best measure to predict the classification of the patient in the class "before injection" or "after injection". A top of 12 measures discriminated improvement of performance after injection. The top four most sensitive measures in SMD were measured during walking downstairs (based on the Standardized mean difference). The second reduction in measures from the three tasks delivered 6 measures coming from walking downstairs and walking upstairs. A combination of ROM, SPEED and ACC represents the functional improvement best.

One of the main challenges in ML studies is that the number of subjects is often too low, which in turn leads to an **overfitting** of data and limits the applicability to new data. This is true in the analysis performed in Cameroon as well as in all the other classical statistical analysis performed in the framework of this thesis. The sample sizes were small and conclusions from the studies are therefore reflective of the participants and the healthcare settings. Transferability of results should thus be considered with caution. Another limitation is the question of the **clinical validation of algorithms.** Some authors chose to correlate algorithms to clinical scales [289, 290], but at the same time, they developped the need to improve the assessment beyond the limitations of the clinical scales. The support of clinicians in the development and qualitative validation of algorithms appear as opportunities when no base reference for comparison is available.

2nd perspective: finer reflection of multiple dimensions of mobility

Quantity of movement In Part III of this thesis, we measured physical activity by the number of steps and found no physical activity (PA) increase during the 3 months after surgery. The heterogeneity in used metrics for arthroplasty populations contributes to contradictory results in similar studies [66, 242, 291]. However, according to the World Health Organisation, physical activity is a **multidimensional behaviour** that consists of four dimensions: Frequency, Intensity, Time and Type (F.I.T.T.) Two studies demonstrated that persons with knee OA are not necessarily only more sedentary [291, 292]. Different **types** of PA (stair climbing, short walking periods and short periods of sedentary time) were different according to BMI, whereas overall physical

activity and sedentary time were not discriminant. These studies highlighted the importance to use F.I.T.T. instead of general overall physical activity.

Quality of movement. We focused on kinematic representations of mobility, measured by spatio-temporal parameters, ROM, angular SPEED, etc. As high knee joint-loading increases the risk and progression of knee OA [293], **joint forces and moments** are also relevant variables to be measured. The combination of wearable sensors [294] and **in-shoe pressure sensors** make such measurements feasible [295, 296]. Segment orientations and fixed segment lengths can be used as an input in a linked-segment model to obtain joint positions [297].

Inertial sensors belong to the **family of wearables** which are defined as "*portable devices that transform a physical quantity in a quantitative or qualitative quantity allowing to illustrate or interpret a patient state*". Other kinds of sensors might help in monitoring several variables that are useful for care: blood pressure, oxygen saturation, body temperature, stress arousal, respiratory frequency [298], etc. Those sensors are embedded in wireless sensor networks that consist of multiple sensor nodes, which are organized in a network performing the functions of communications, computations and sensing. Investigations in the area of **wireless sensor network** are exploding [299] as their potential to monitor different pathologies and assist caretakers is huge. However, the combination of all those sources of information in an integrated approach is another issue.

3rd perspective: integrated approach by triangulation between different sources of information: quantitative measures, subjective experience and ethnographic observation

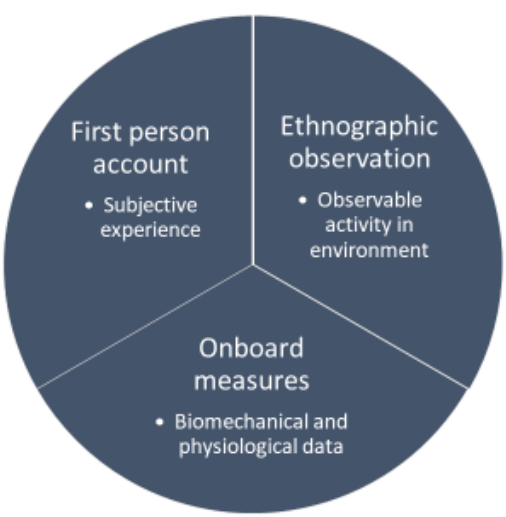

Figure 5: An integrated approach by triangulation of three categories of data sources to characterise mobility performance in real life: on-board measures, subjective experience and ethnographic observation. Inspired from ecological research on Parkinson Disease by Parry 2017 [300].

In this thesis, we focused only on **biomechanical data** to represent mobility. As we know that fear of falling when walking on a slippery surface, or on pavement, could affect gait behaviour [301] or that cultural factors or weather conditions may have a major impact on physical activity behaviour [221], measurement of other dimensions is needed as well. An integrated approach by **triangulation** of articulated data sources (patient input, observable in environment, biomechanical and physiological measurements – Figure 5) may be used to disentangle key elements contributing to the emergence of a patient's locomotor behaviour through the course of **complex, daily life situations** [24, 300]. For example, the interpretation of a freezing of gait pattern

associated to stress in public could only be performed through motion measurement in combination with objective markers of stress (as heart rate variability or galvanic skin response) with the patient's experience [263].

The meanings and relationships of objective movement measurement can be explored with **subjective outcomes** related to the patient, as pain or fatigue. Subjective outcomes can also be measured in free living conditions thanks to smartphone and the **Ecological Momentary Assessment** (EMA) [302]. EMA involves repeated sampling of subjects' current behaviours and experiences in real time, in subjects' natural environments. EMA aims to minimize recall bias, maximize ecological validity, and allows the study of micro processes that influence behaviour in real-world contexts. The combination of EMA [302] with mobility variables was already performed in **complex problems** such as in drug research [303] or patients with apathy or schizophrenia [304]. In multiple sclerosis, one recent study revealed the dynamic aspect of fatigue symptoms across and within days [305] and highlighted the need for real life measures for the management of complex symptoms.

The stakeholders of the technologies: A third conceptual axis:

This thesis was illustrated as a two-axis exploration towards (1) ecological settings and, (2) the knowledge of technology characteristics for clinical use in the road map (Figure 8 in introduction). However, 3D analysis can deliver more information than 2D. That is why it is interesting to develop a **third conceptual axis**, as a perspective of this work. Indeed, the developed technologies to measure patients will have to be used by different stakeholders with different needs: patients, therapists and the health system (Figure 6).

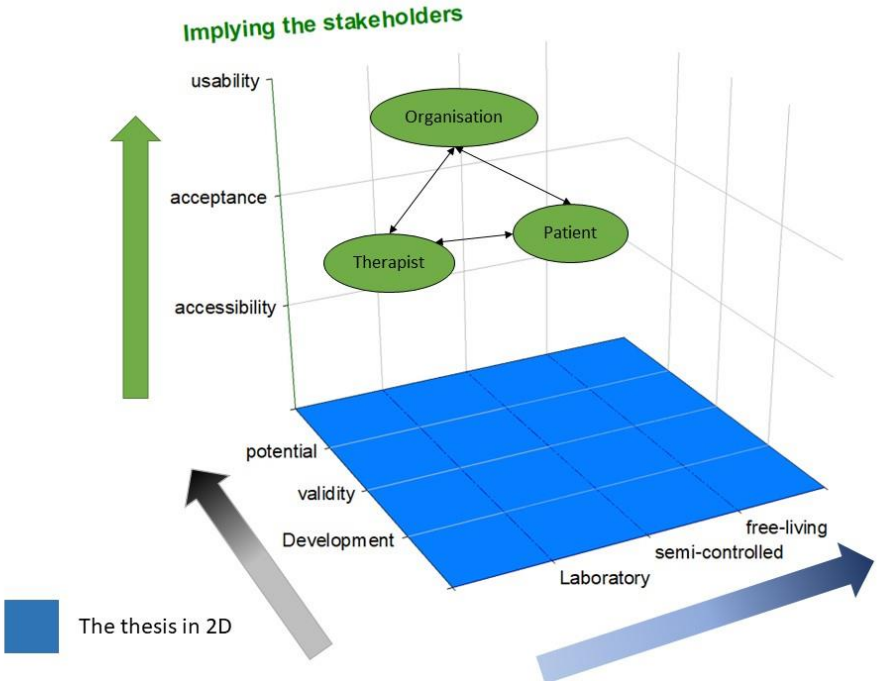

Figure 6: A third investigation axis: accessibility, acceptability and usability by stakeholders. Patients, therapists and the organisations as well as their relationships can be impacted by the wearable technology in the future.

Needs and concerns of patients

In the pilot study performed in a clinic in Part I, we observed low participation due to the distance between the hospital and the home of the patients. The development of portable devices may allow the increase of **accessibility** of mobility assessment for patients in general and in particular for those living far away from health providers and for those living in low income countries.

Several measurements were also taken from patients before and after knee arthroplasty **at their home or at their work place** with inertial sensors (*unpublished data – Figure 7*). At this stage, measurements were taken in attendance of the researcher. However, as sensors are miniaturized and embedded in the smartphone or the watch of the patient, those patients could assess themselves remotely and send their data to their therapists [306-308]. Cadence assessment through smartphone is an actual topic of the development of the telerehabilitation platform with which we collaborated. Those patients appreciated the approach of their daily life problems and were interested in such connection of their health state to their therapists but were concerned with other issues such as **privacy** [250].

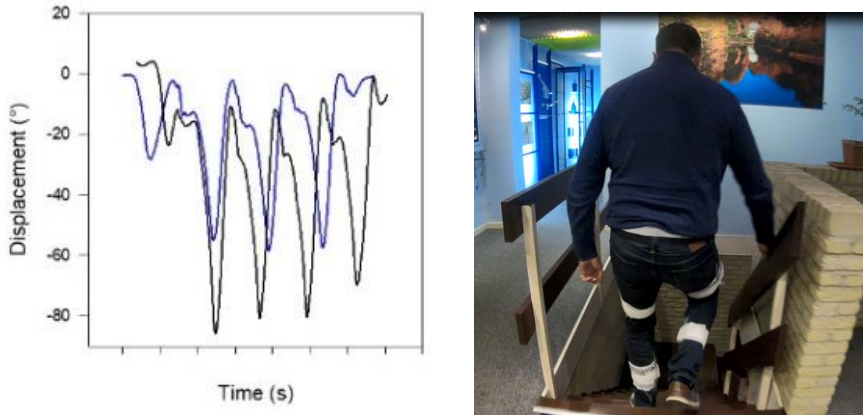

Figure 7: Joint angular displacement of the knee in the sagittal plane of one subject during stair descent in his environment before and 2 months after total knee arthroplasty. The same stairs was used for both assessments.The blue line represent the data before the surgery, the black line represent the data after the surgery. This measurement required only two sensors.

If a continuous monitoring is to be attained, sensors must be comfortable to wear. The **wearability** of sensors is one of the challenges for use by patients [109]. The use of textile sensors, embedded in clothes [309], and the development of miniaturised batteries could increase their use and mitigate this challenge.

The availability of health data, such as mobility variables, will change the **relationship between patients and therapists**. The health agenda expects that patients receive more insights about their health status and that decision making moves from the traditional biomedical care system to **shared-decision making** [310]. This paradigm shift involves the removal of physicians from their position of authority with patients playing a passive role in care to putting the patient at the centre.

Needs of Therapists

In order to have therapists use sensor-based mobility outcomes for their patients, they have to be inclined to use technology (acceptance). **Acceptance** research has been on understanding the user's rationale behind "*how*" and "*why*" individuals adopt, retain or abandon new technologies. Their goal is to understand usage as the dependent variable. Acceptance is mediated by many social, cognitive and psychological factors [54] and multiple models, as the **Unified Theory of Acceptance and Use of Technology**, were proposed. That study is interesting because it suggests the direction for future research and design. For example, we interviewed 8 physiotherapists to understand why they are not keen on using a new rehabilitation robotic device (*unpublished data – Thesis Blanckaert-Bonmariage).* These interviews revealed that the amount of training courses was perceived as insufficient, that the informatics bugs had a negative impact on the self-efficacy perception and that the perceived effort to install the patient on the robot was still too high [311](*poster presentation at Rehabweek – Toronto*). Moreover, despite a randomised controlled trial that proved the effectiveness of the robot [312], therapists had a perception of ineffectiveness with their patients. That kind of information would allow to carry on the innovation process with this technology, as well as with the one engaged in this thesis with wearable sensors.

Some sensor-based mobility outcomes are already being used by therapists in practices, but it is unknown to what different purposes they are being used. That is why we are studying the actual measurement of three outcomes of this thesis (spatio-temporal parameters, physical activity, joints angles) by Belgian physiotherapists. An **online survey** was designed to depict their frequency of use, their target population as well as their perspective for

technology development. The survey is available in Appendix 2. Preliminary results on 148 French speaking and 123 Dutch speaking physiotherapists suggest that **80 % of them assess joint range of motion**, mainly for patients after surgery (89%), **but only 20% by means of technology**. It means that most of them do it with the bare eye. **66% assess spatio-temporals parameters of gait**, 31% by means of technology. **Concerning physical activity, only 20% of physiotherapists** assess this outcome in their practice, while 90% amongst them use this variable to educate or motivate their patients with chronic conditions to move *(unpublished results - Thesis Mauclet L. Monhonval P. Hubert G.).* To clarify the needs and expectations of therapists towards objective movement assessment with sensors, in-depth interviews will be performed with some of the responders of the survey. In what way can technology help, how can it be improved, what are the needs that caregivers have when it comes to such technology? How should technology be designed to best meet caregiver needs, what barriers and opportunities exist for technology use among caregivers?

This next step is called the assessment of **usability**. Usability is the capacity in human functional terms to allow the effective, efficient, and satisfying use with which specified users, with adapted education and support, can achieve specific goals in particular environments [313-315](ISO 9241-11:2018). This implies studying the sensors and the software in the hands of physiotherapists in order to better understand their needs regarding software friendliness, continuous education, etc. The development of a **user-friendly software** for the use of our customized sensor network, as well as **automatic reports** with the relevant gait measures are perspectives for development of this thesis.

Needs of the health system

Given the financial pressure on the health system, a clear clinical utility of technologies needs to be demonstrated. For example, for movement analysis in semi-controlled setting, multiple issues need to be resolved, such as (1) whether movement analysis is more beneficial than visual gait analysis and (2) whether the increased time, effort and expense of mobility analysis affects treatment decision making and functional outcome. The path to increase the role of movement analysis lies in proving its value through case studies demonstrating its **cost-effectiveness**. The cost of a conventional gait analysis lies at roughly 250€, whereas the cost of an analysis with our customized sensor system is likely to be much lower. Apart from the cost, improving the education of clinicians in the understanding of abnormal results will maybe force clinicians into recognizing the limitations of visual observation, especially when costly or invasive treatments are involved [13].

Conclusions

This thesis went beyond the mobility measurements performed in laboratories and explored new horizons by analyzing challenging situations of daily life, such as stair climbing, walking outdoor or even continuously measuring mobility during the day. An innovative approach of wireless inertial sensors was used to record joint angles unobtrusively. After validating the sensors and the easy-to-use gait calibration against gold standard technologies, it was confirmed that the sensors can be used to assess gait. These technological instruments and the ecological settings brought to light improvements after treatment for patients suffering from OA. For example, we found that hip and pelvis kinematics improved after a knee injection in Cameroon. Furthermore, we also found that the physical activity level of patients after knee arthroplasty went back to the preoperative level at the 7th postoperative week thanks to a telerehabilitation program.

These findings are promising to improve monitoring of patients in their daily life and for moving towards data-based healthcare. A multidisciplinary approach is and will remain necessary to face the methodological challenges raised by an assessment in ecological settings as well as to continuously improve the validity of sensors. It is unknown how many pieces the puzzle contains, but with this thesis, the first pieces have been placed on the board.

List of abbreviations

ΔROM	Absolute Difference in the Range Of Motion
2MWD	Two-minute Walk Distance
2MWT	Two-minute Walk Test
3D	3 Dimensions
6MWT	Six-minute Walk Test
ABS	Acrylonitrile Butadiene Styrene
ADROM	Ankle Dorsiflexion Range Of Motion
AHRS	Attitude Heading Reference System
B-IPQ	Brief Illness perception questionnaire
BF	Biceps femoris
CdP	Centre de pression
COM	Center Of Mass
CRF	Centre Reine Fabiola
DRIFT	Drift
EMA	Ecological Momentary Assessment
EMG	Electromyography
FSD	Forward Step-down Test
GL	Gastrocnemius lateralis
HOOS	Hip Injury and Osteoarthritis Score
ICC	Intraclass Correlation Coefficient
ICF	International Classification of Functioning, Disability and Health
IMU	Inertial Measurement Unit
KOOS	Knee Injury and Osteoarthritis Score
LEFS	Lower Extremity Functional Scale
LLMP	Lower Limb Movement Patterns
MDC	Minimal Detectable Change
MFP	Modified Foot Position
OA	Osteoarthritis
PA	Physical activity
PCA	Principal Component Analysis
PROM	Patient-Reported Outcome Measure
RMSE	Root mean square error
ROM	Range of motion

SEM	Standard Error of Measurement
TA	Tibialis anterior
THA	Total hip arthroplasty
TKA	Total knee arthroplasty
TRL	Technology Readiness Level
TSK	Tampa Scale of Kinesiophobia
VL	Vastus lateralis
VR	Virtual reality

Appendix

Appendix 1: CFS algorithm

A correlation-based feature selector (CFS) was used to determine the best predictors for the groups (before injection/after injection). CFS is a simple filter algorithm that ranks feature subsets (set of variables) according to a correlation based heuristic evaluation function

$$Ms = \frac{\text{k rcf}}{\sqrt{k + k(k-1)\, rff}}$$

where Ms is the heuristic "merit" of a feature subset S containing k features, rcf is the mean feature-class correlation (f ∈ S), and rff is the average feature-feature intercorrelation. The numerator of Equation can be thought of as providing an indication of how predictive of the class a set of features are; the denominator of how much redundancy there is among the features.

The algorithm is based on the hypothesis that a good feature subset (set of variables) is one that contains features highly correlated with (predictive of) the class, yet uncorrelated with (not predictive of) each other [316]. For example, hip frontal angular speed was highly correlated with the class "before-after injection" that was coded as 0-1. Whereas variables correlated with hip frontal angular speed were removed from the feature subset to avoid redundancy. Variables were normalized to 0-1 before the application of the algorithm.

The space of attribute subset was searched by greedy hill climbing [317], started at any point and searched in both direction (forward and backward). The search was terminated when 5 consecutive non-improving nodes were reached,

converging to a local optimum. All feature selection were performed with Weka Software [318].

Appendix 2: Questionnaire: Use of measurement technologies of mobility by Belgian physiotherapists

1ère partie : l'utilisation des technologies d'évaluation du mouvement

Activité physique journalière

1) Dans votre pratique, évaluez-vous l'activité physique journalière (nombre de pas par jour) de vos patients ?
   a. Oui
   b. Non (Allez directement à la question 11)

2) Si oui, pour quel type de patient ?
   a. Post-chirurgie (PTH, PTG, LCA, etc.)
   b. Maladie chronique (SEP, fibromyalgie, lombalgie chronique, BPCO, etc.)
   c. Traumatologie (entorse, fracture, tendinopathie, etc.)
   d. Autre (précisez)

3) Si oui, à quelle fréquence ?
   a. Moins d'une fois par mois
   b. Plusieurs fois par mois
   c. Plusieurs fois par semaine
   d. Plusieurs fois par jour

4) Dans le cadre de ma prise en charge, la mesure de l'activité physique journalière m'est utile pour :
   a. Anamnèse
   b. Diagnostiquer
   c. Évaluer / Réévaluer
   d. Traiter (lors des séances)
   e. Éduquer
   f. Motiver votre patient
   g. Autre (précisez)

5) Utilisez-vous des technologies d'évaluation de la mobilité afin d'évaluer ce paramètre (activité physique journalière) chez vos patients ?
 Par technologie, nous entendons tout appareil permettant de fournir des données au kinésithérapeute afin d'évaluer son patient, aussi bien dans en

cabinet que dans l'environnement de ce dernier. Cela peut être sous la forme de montres connectées, d'applications de santé sur smartphone ou encore de capteurs mobiles.

a. Oui □
b. Non □ (Allez directement à la prochaine partie)

6) Si oui, laquelle utilisez-vous ?

a. Applications smartphone
b. Montres connectées
c. Appareil podomètre
d. Autre (précisez)

7) Si oui, comment avez-vous été amené à utiliser des technologies de mesure de mobilité ?

a. Formation
b. Avis de collègues
c. Intérêt personnel pour les technologies
d. Demande du patient
e. Autre (précisez)

8) Si oui, quels sont les bénéfices que vous retirez de l'utilisation de technologie ?

a. Gain de temps
b. Gain d'argent
c. Meilleure évaluation
d. Meilleur traitement
e. Meilleure communication des données entre le patient et vous
f. Meilleure communication des données entre spécialistes de la réadaptation
g. Aucun
h. Autre (précisez)

9) Comment abordez-vous la question de l'activité physique avec votre patient ?

a. En lui demandant s'il fait du sport
b. En lui demandant le nombre approximatif d'heures par semaine où il est essoufflé / rouge, où son cœur bat plus fort
c. En lui demandant ses loisirs / activités
d. En lui demandant de remplir un questionnaire sur celle-ci
e. Autre (précisez)

10) Qu'est-ce qui vous encouragerait à évaluer l'activité physique de vos patients de manière plus régulière avec ces technologies ?

a. Meilleur rapport coût/bénéfice
b. Plus de connaissance (séminaire, formation, cours, etc.)
c. Tester ces technologies au préalable
d. Preuve de l'efficacité
e. Demande du patient
f. Autre (précisez)

11) Si vous étiez un jour amené à devoir évaluer ce paramètre (activité physique journalière), de quelle manière procéderiez-vous ?

a. En demandant à vos patients (exemple : description d'une journée type ? Quelles sont vos activités ? etc.)
b. À l'aide de technologie (applications, capteurs, etc.)

Angle articulaire

1) Dans votre pratique, évaluez-vous les angles articulaires (au niveau du membre inférieur) de vos patients ?

a. Oui
b. Non (Allez directement à la question 11)

2) Si oui, pour quel type de patient ?

a. Post-chirurgie (PTH, PTG, LCA, etc.)
b. Maladie chronique (SEP, fibromyalgie, lombalgie chronique, BPCO, etc.)
c. Traumatologie (entorse, fracture, tendinopathie, etc.)

3) Si oui, à quelle fréquence ?

a. Moins d'une fois par mois
b. Plusieurs fois par mois
c. Plusieurs fois par semaine
d. Plusieurs fois par jour

4) Dans le cadre de ma prise en charge, la mesure de l'angle articulaire m'est utile pour :

a. Anamnèse
b. Diagnostiquer

c. Évaluer/Réévaluer
d. Traiter (lors des séances)
e. Éduquer

5) Utilisez-vous des technologies d'évaluation de la mobilité afin de mesurer ce paramètre (angle articulaire) chez vos patients ?
a. Oui
b. Non (Allez directement à la prochaine partie)

6) Si oui, laquelle utilisez-vous ?
a. Applications smartphone (exemples : iHandy Level)
b. Caméra / photo (pour visualiser les ralentis de vidéos par exemple)
c. Capteurs mobiles
d. Inclinomètre
e. Autre (précisez)

7) Si oui, quels sont les bénéfices que vous retirez de l'utilisation de technologie ?
a. Gain de temps
b. Meilleure évaluation
c. Meilleur traitement
d. Meilleure communication des données
e. Autre (précisez)

8) Si oui, comment avez-vous été amené à utiliser des technologies de mesure de mobilité ?
a. Formation
b. Avis de collègues
c. Intérêt personnel pour les technologies
d. Demande des patients
e. Autre (précisez)

9) Si non, avec quoi évaluez-vous ce paramètre (angle articulaire) ?
a. Goniomètre
b. Instinct / à l'œil
c. Autre (précisez)

10) Qu'est-ce qui vous encouragerait à évaluer l'angle articulaire de vos patients de manière plus régulière avec ces technologies ?

a. Meilleur rapport coût/bénéfice
b. Plus de connaissance (séminaire, formation, cours, etc.)
c. Tester ces technologies au préalable
d. Preuve de l'efficacité
e. Demande du patient
f. Autre (précisez)

11) Si vous étiez un jour amené à devoir évaluer ce paramètre (angle articulaire), de quelle manière procéderiez-vous ?
a. Goniomètre
b. À l'œil / instinct
c. À l'aide de technologies (applications, capteurs, inclinomètre, etc.)

Paramètres spatio-temporels de la marche/course (cadence, vitesse, longueur du pas, symétrie, etc.)

1) Dans votre pratique, évaluez-vous les paramètres spatio-temporels de la marche/course de vos patients ?
a. Oui
b. Non (Allez directement à la question 11)

2) Si oui, pour quel type de patient ?
a. Post-chirurgie (PTH, PTG, LCA, etc.)
b. Maladie chronique (SEP, fibromyalgie, lombalgie chronique, BPCO, etc.
c. Traumatologie (entorse, fracture, tendinopathie, etc.)
d. Autre (précisez)

3) Si oui, à quelle fréquence ?
a. Moins d'une fois par mois
b. Plusieurs fois par mois
c. Plusieurs fois par semaine
d. Plusieurs fois par jour

4) Dans le cadre de ma prise en charge, la mesure des paramètres spatio-temporels de la marche/course m'est utile pour :
a. Anamnèse
b. Évaluer/Réévaluer

c. Diagnostiquer
d. Traiter (lors des séances)
e. Éduquer
f. Autres (précisez)

5) Utilisez-vous des technologies de mesure de la mobilité afin d'évaluer ce paramètre (paramètres spatio-temporels de la marche/course) chez vos patients ?

a. Oui
b. Non (Allez directement à la prochaine partie)

6) Si oui, laquelle utilisez-vous ?

a. Applications smartphone
b. Capteurs mobiles
c. Montres connectées
d. Vidéo
e. Chronomètre
f. Tapis roulant intelligent
g. Autre (précisez)

7) Si oui, quels sont les bénéfices que vous retirez de l'utilisation de technologie ?

a. Gain de temps
b. Meilleure évaluation
c. Meilleur traitement
d. Meilleure communication des données
e. Autre (précisez)

8) Si oui, comment avez-vous été amené à utiliser des technologies de mesure de mobilité ?

a. Formation
b. Avis de collègues
c. Intérêt personnel pour les technologies
d. Demande du patient
e. Autre (précisez)

9) Si non, avec quoi évaluez-vous ce paramètre (paramètres spatio-temporels de la marche/course) ?

a. Instinct / à l'œil
b. Autre (précisez)

10) Qu'est-ce qui vous encouragerait à évaluer les paramètres spatio-temporels de vos patients de manière plus régulière avec ces technologies ?

a. Meilleur rapport coût/bénéfice
b. Plus de connaissance (séminaire, formation, cours, etc.)
c. Tester ces technologies au préalable
d. Preuve de l'efficacité
e. Demande du patient
f. Autre (précisez)

11) Si vous étiez un jour amené à devoir évaluer ce paramètre (paramètres spatio-temporels de la marche/course), de quelle manière procéderiez-vous ?

a. À l'œil / l'instinct
b. À l'aide de technologies (vidéo, capteur, etc.)

<u>2ème partie : perception / avenir</u>

1) Selon vous, les innovations technologiques suivantes ont-elles un impact sur les soins médicaux et la santé à ce jour ? (Par impact nous entendons que les technologies de mesure seraient de plus en plus utilisées dans le monde de la santé de manière positive et utile)

a. Applications médicales (par application médicale, nous entendons toutes applications sur smartphone qui récoltent toutes des données sur la mobilité du patient comme le nombre de pas par jour, le nombre de km parcourus, etc.)

i. Tout à fait d'accord
ii. D'accord
iii. Pas d'accord
iv. Pas du tout d'accord
v. Non concerné

b. Montres connectées

i. Tout à fait d'accord
ii. D'accord
iii. Pas d'accord
iv. Pas du tout d'accord
v. Non concerné

c. Capteurs mobiles (nous entendons par capteur mobile, toutes technologies permettant d'évaluer le patient et de récolter des données sur la mobilité de celui-ci.)
   i. Tout à fait d'accord
   ii. D'accord
   iii. Pas d'accord
   iv. Pas du tout d'accord
   v. Non concerné

2) Si vous deviez classer les différents freins à l'utilisation des technologies par ordre d'importance, quel serait votre classement ? (De 1 (le plus grand frein) à 4 (le plus petit frein))
   a. Le manque de confidentialité des données
   b. Le prix
   c. Le manque de fiabilité (par fiable, nous entendons la confiance que vous pouvez avoir dans les mesures prises par les technologies)
   d. Le manque de formation
   e. La durabilité (entretien, maintien dans le temps)
   f. La portabilité
   g. La disponibilité/accessibilité
   h. Le fonctionnement sous toutes conditions
   i. Le manque de temps pour évaluer

3) Voyez-vous d'autres améliorations dans les technologies de mesures de la mobilité ciblées dans cette enquête pour augmenter leur performance et leur utilisation ?
Question ouverte : . . . . . .
. . . . . .
. . . . . .

4) Si vous deviez classer les différentes améliorations à apporter pour vous amener à utiliser ces technologies de mesure par ordre d'importance, quel serait votre classement ?
   a. Augmentation de la précision
   b. Augmenter la formation
   c. Meilleur rapport coût/bénéfice
   d. Augmentation de la disponibilité/accès
   e. Plus grande demande de la part de vos patients
   f. La durabilité (entretien, maintien dans le temps)
   g. La portabilité

h. La disponibilité/accessibilité
i. Le fonctionnement sous toutes conditions
j. Influence des collègues qui utilisent déjà / aussi des technologies

5) Au vu de la progression des technologies ces dernières années, celles-ci pourraient devenir des outils incontournables à l'évaluation des patients dans les années à venir. Pensez-vous que ces nouveaux outils changeront la prise en charge des patients ?

a. Tout à fait d'accord
b. D'accord
c. Pas d'accord
d. Pas du tout d'accord
e. Non concerné

<u>3ème partie : caractéristiques du sujet</u>

1) Dans quel type de structure travaillez-vous ? (Possibilité de cocher plusieurs choix)

a. Cabinet privé
b. Cabinet de groupe
c. Centre de rééducation
d. Hôpital universitaire
e. Hôpital régional
f. Maison de repos
g. Autre (précisez)

2) Type de prise en charge :

a. ...% de mes patients ont une prise en charge post-chirurgicale (curseur)
b. ...% de mes patients ont une prise en charge post-traumatique (curseur)
c. ...% de mes patients ont une prise en charge chronique (curseur)
d. ...% de mes patients sont hospitalisés/ internés
e. ...% de mes patients sont vus en ambulatoire
f. ...% de mes patients sont vus à leur domicile

3) Quel est votre nombre d'années d'expérience ?

a. Curseur avec le nombre d'années exactes

4) Dans quelle partie de la Belgique travaillez-vous ?
   a. Anvers
   b. Brabant flamand
   c. Brabant wallon
   d. Bruxelles
   e. Flandre occidentale
   f. Flandre orientale
   g. Hainaut
   h. Liège
   i. Limbourg
   j. Luxembourg
   k. Namur
   l. À l'étranger

5) Où avez-vous fait vos études ?
   a. Flandre – université
   b. Flandre – haute école
   c. Wallonie – université
   d. Wallonie – haute école
   e. Étranger (précisez)

6) Dans quel domaine exercez-vous la plupart du temps ?
   a. Autant de % en ortho
   b. Autant de % en kiné sportive
   c. Autant de % en neuro
   d. Autant de % en cardio-respi
   e. Autant de % en gériatrie
   f. Autant de % en pédiatrie

7) Si vous désirez vous exprimer davantage sur le sujet, vous pouvez nous laisser votre mail ou numéro de téléphone (ainsi que votre nom) dans la zone de texte ci-dessous.
Vos données resteront confidentielles et ne seront utilisées que pour les besoins de l'enquête.
 E-mail :
 Numéro de téléphone :

Appendix 3: Accompanying thesis: article 1

Effects of ankle dorsiflexion limitation on lower limb kinematic patterns during a forward step-down test: A reliability and comparative study

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Appendix 3

Abstract

Background: Altered lower limb movement patterns (LLMP) during the forward step down (FSD) test have been studied in people suffering from knee instability. However, ankle dorsiflexion range of motion (ADROM) seems to be related to LLMP but no causal inference has been defined between those variables.

Objective: Our goals were to evaluate (1) psychometric quality of the FSD test in healthy people and (2) the influence of experimental restricted ADROM on LLMP.

Methods: Kinematics were measured by a motion capture system. Angular displacement and speed were calculated as well as center of mass (COM) and knee linear displacement. Forty-two healthy participants; firstly, performed the test three times to assess reliability, and secondly the same test in an experimental condition limiting the ADROM.

Results: Reliability was excellent for all parameters (ICC: 0.75–0.99, SEM%: 2.0–34.0%, MDC%: 5.7–94.3%). ADROM limitation significantly decreased Knee flexion range of motion (ROM) (3.8), increased Hip flexion ROM (6), Hip adduction ROM (6.1), Pelvis drop ROM (3.5), Pelvic rotation ROM (3.1). No significant effect was found for Hip rotation ROM.

Conclusions: LLMP was affected by this experimental ADROM limitation in healthy people. As this limitation is often encountered in post-traumatic ankle sprain patients, clinicians should consider this point during FSD assessment test.

Keywords: Lower limb, movement pattern, functional tests, kinematics, biomechanics

Introduction

Altered Lower Limbs Movement Patterns (LLMP) have been mainly studied in people suffering from lower limb disorders [319, 320] such as knee instability [321, 322] in order to improve our understanding of etiopathology and consequences of musculoskeletal disorders of the lower limb. The alteration of these LLMP has been used as risk factor for musculoskeletal injury [323, 324]. Also the analysis of these patterns provides outcomes to assess the benefit of treatment [325].

Kinematics analysis as gait analysis has been widely accepted as an objective measure of physical function, allowing researchers and clinicians to better understand biomechanical alterations and to identify specific LLMP [326-328]. However, analysis of LLMP in complex dynamic motions such as gait, running, or stairs negotiation is difficult in clinical practice. Therefore, some easy tests have been developed such as the Forward Step Down (FSD) test. This functional and clinical test is easy to administer in a clinical setting and had been used in previous studies to assess LLMP [329-332].

LLMP can be influenced by many factors like vision [333], speed of movement, height [334], lower extremity alignment or mechanical limitation of range of motion (ROM) [335, 336]. A good knowledge of the influence of all factors is necessary to interpret correctly the FSD test results. A main factor that can altered LLMP is the limitation of ankle dorsiflexion ROM (ADROM), resulting from post-traumatic arthrosis of the tibiotalar and/or talocrural joints, frequently encountered after an ankle sprain [337]. A recent study [338] showed the relationship between limitation of ADROM and lower limb kinematics during a step down test. Therefore, those authors recommend

assessing ADROM before screening LLMP [336, 338]. A systematic review demonstrated that a reduced range of motion was present in individuals presenting with dynamic knee valgus when compared to controls. However they stated that cause and effect relationship among ADROM limitation and quality of movement during the FSD test remain challenging.[339]

For a correct interpretation of this clinical FSD test, we need to know (1) the psychometric properties, such as reliability, responsiveness (2) and the cause and effect relationship between ADROM and LLMP.

Our goal was to evaluate; (i) the reliability and responsiveness of this usual clinical FSD test in healthy subjects, and (ii) the influence of ADROM limitation by a modified foot position (MFP) on LLMP. Moreover, an exploratory analysis using kinematics and a dynamical system approach [340] will address the clinically observed LLMP during the FSD test for a better understanding of the influence of ADROM limitation on kinetic chain.

Methods

1. Subjects

Forty-two healthy participants were recruited from the student population of our university. The inclusion criteria were (1) to be free of recent lower limb injuries (in the last six months) and (2) to be able to descend a 15 cm high step. Lower limb dominance was established by noting the limb of choice when the participant was asked to kick a ball. The ethics committee of our university approved the study protocol (agreement number: B403201523492) and informed consent was obtained from subjects prior to testing.

2. Procedure: Task and Instructions

The participants were asked to stand relaxed barefoot on both legs on the 15 cm high step with arms folded across the chest. They performed the stair descent task at a self-selected pace until they touched the ground with the heel and then returned to the starting position on both legs to restore balance (Figure 1b). During the test, they were instructed to maintain their visual focus directly in front of them at eye level. The foot position was standardized at the top of the step and participants were asked to keep the heel of the supporting foot flat. All participants received the same instructions provided by a short demonstration video. They trained before recording to avoid a learning effect. After, they completed 6 trials of 10 consecutives cycles starting with the

dominant and alternating with non-dominant leg as supporting leg (control condition). Each trial was separated by a two-minute break.

Then, they achieved the same protocol with a modified foot position (MFP) by means of a 2 cm high platform under the middle and fore supporting foot in order to decrease ADROM and to decrease ADROM (Figure 1a). The platform's height was chosen to be feasible by large population and to reduce ADROM (MFP condition)[341].

3. Motion capture

All tests took place in a motion-analysis laboratory during one single session. Kinematic analysis was performed with the Elite motion capture system (BTS Bioengineering, Italy) with 8 high speed cameras at 200 Hz. Twenty-two reflective markers were placed on specific anatomical landmarks according to The Helen Hayes model (Figure 1c) [342] and allowed computation of the angular displacement of pelvic, hip, knee and ankle in the 3D planes. The knee frontal and transversal plane were not considered due to cross-talk [343].

We also recorded sacrum marker 3D displacement estimated as an approximation of body center of mass (COM) [344] during the whole cycle. Besides, we recorded supporting knee center medial displacement during the descent phase (between leg off and floor contact (Figure 1b)). Knee center was defined as the midpoint between femoral condyles.

All markers were placed by the same examiner who also measured the anthropometric variables and ADROM with the weight bearing lunge test [345].

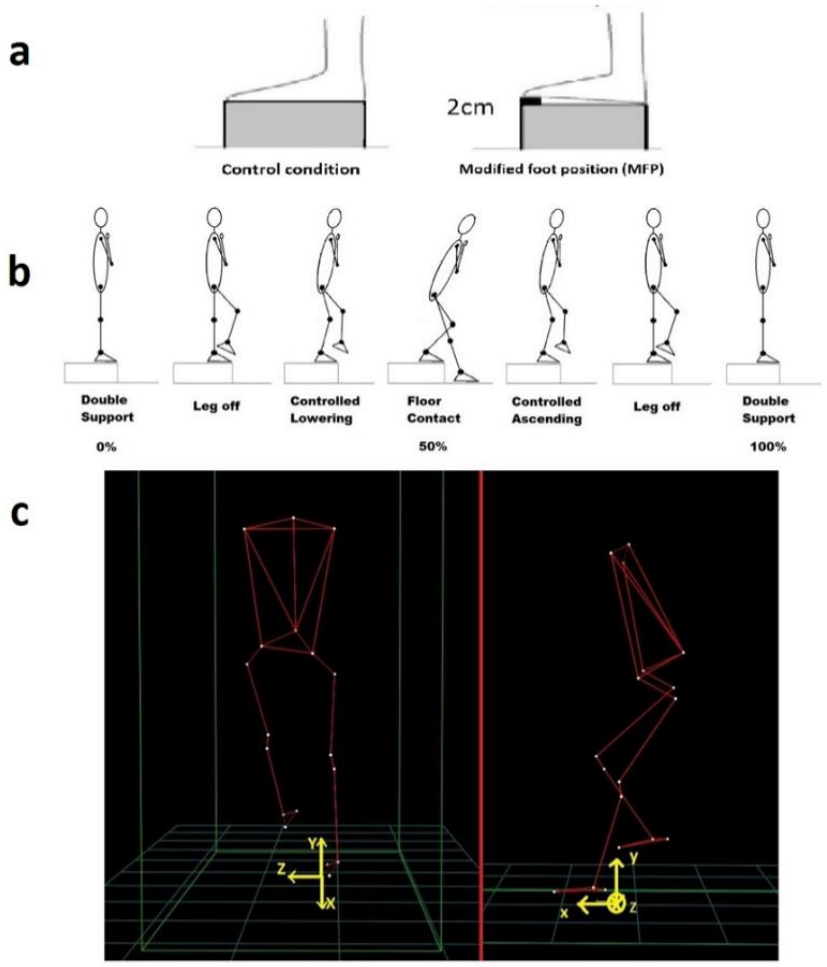

Fig 1. a: Illustration of the control and the modified foot position. b: Illustration of the step-down task and cycle composition. c: Illustration of the Helen Hayes model and the coordinate axis. Frontal view (left) and lateral view (right). X: anterior compared to the step descent axis, Y: vertical, Z: lateral.

Appendix 3

4. Parameters

Five consecutives cycles were selected and normalized in 100% of the cycle. 0% matches the beginning of descent task and 100% matches the return to the starting position standing on both legs. We calculated the average time of the 5 consecutive cycles to know movement speed.

Mean ROM and Mean SPEED were calculated for each angular displacement per task (5 consecutive cycles) per subject.

For mediolateral knee displacement, we assessed medial knee displacement only during the descent phase to depict the same data as Whatman et al.[346] instead of the lateral body weight shift between double support and single support.

5. Statistical analysis

(1) 1srt objective: psychometric properties

Reliability

Reliability of our kinematics parameters was performed on the three trials according to a method described by Wagner [124] using the intraclass correlation coefficient (ICC) and the standard error of the measurement (SEM). ICC consistency parameters were calculated in a 2-way mixed model using the SPSS software (version 23.0 for Windows, SPSS INC., Chicago, IL, USA).

SEMs estimate the non-systematic variance. As a measure of within-subject variability among repeated trials, the SEM expresses the measurement error in the same units as those of the original measurement. Measurement

error was also expressed as the SEM%, the within subject standard deviation as a percentage of the mean [347].

Responsiveness

This parameter indicated sensitivity to a change in outcome measures and was assessed from the SEM using the minimal detectable change (MDC). MDC95 represented the change in variables falling outside of the measurement error and the magnitude of change necessary to exceed the measurement error of two repeated sessions as a specified confidence interval (CI) of 95%.

The MDC was independent of the unit of measurement by expressing it as a percentage (MDC %) [347].

(2) Influence of ADROM limitation

Comparison between original test and MFP

Paired t-tests were used to compare the ROM, SPEED and displacement of knee and sacrum between the initial test and the test with MFP. In case of failed normality test, we used a Wilcoxon Signed Rank Test. The effect size with the standardized mean of difference was calculated to compare the magnitude of difference between tests.

(3) Exploratory analysis

We used the kinematics and dynamical systems approach proposed by Spinelli to address coordination and neuromuscular control of lower limb kinetic chain during control and MFP condition. Phase-plane graphs displayed angular displacement (x-axis) and instantaneous angular velocity (y-axis).

Appendix 3

Results

1. Subjects

Characteristics of subjects are shown in Table 1.

Table 1 : Characteristics of subjects

	Total (n=42) Mean (SD)
M/F	20/22
Age, years	22.8 (2.1)
Weight, kg	63.6 (9.3)
Height, m	172.1 (7.6)
Dominance, R/L	37/5
Lower limb length D/ND, cm	87.4 (4.3) / 87.5 (4.1)
Ankle ROM, D/ND, cm	12.9 (2.1) / 13.1 (2.1)

M : male, F : Female, SD : standard deviation, R : Right, L : Left, D : Dominant limb, ND = Non dominant limb, ROM : Range of motion (Weight Bearing Lunge Test)

2. Psychometric properties

Reliability (Table 2).

The ICC, SEM% and MDC% are presented in Table 2. Reliability is interpreted according to Shrout & Fleiss [125]. The reliability for the ROM and SPEED was excellent with ICC ranging from 0.75 to 0.98 for ROM and from 0.75 to 0.96 for SPEED. Regarding displacement, the reliability was excellent with ICC ranging from 0.81 to 0.96.

The SEM value for ROM ranged from 0.6 to 1.3° while it ranged from 4.99 °/s to 11.17°/s for SPEED. Besides that, the SEM value for displacement ranged from 0.3 cm to 1.2 cm. MDC% value for displacement are above 40% except for sacrum vertical displacement.

Statistical results on non-dominant limb were similar and are consequently not presented.

3. Ankle dorsiflexion range of motion limitation

The results between control and MFP are presented in Table 3 and illustrated in Figure 2 and Figure 3. Completion time increased significantly in MFP by 0.6 seconds. The average ADROM limitation was of 5.6°. All ROM variables increased significantly except for hip transverse ROM. Knee flexion ROM decreased of 3.8°.

Displacement significantly increased for the medial displacement of the knee during the descent phase by 0.3cm. Antero-posterior displacement of the sacrum increased in the MFP condition.

SPEED decreased for the ankle and knee in the MFP while it increased for the hip in the frontal plane and the pelvis in the sagittal and transversal plane.

Appendix 3

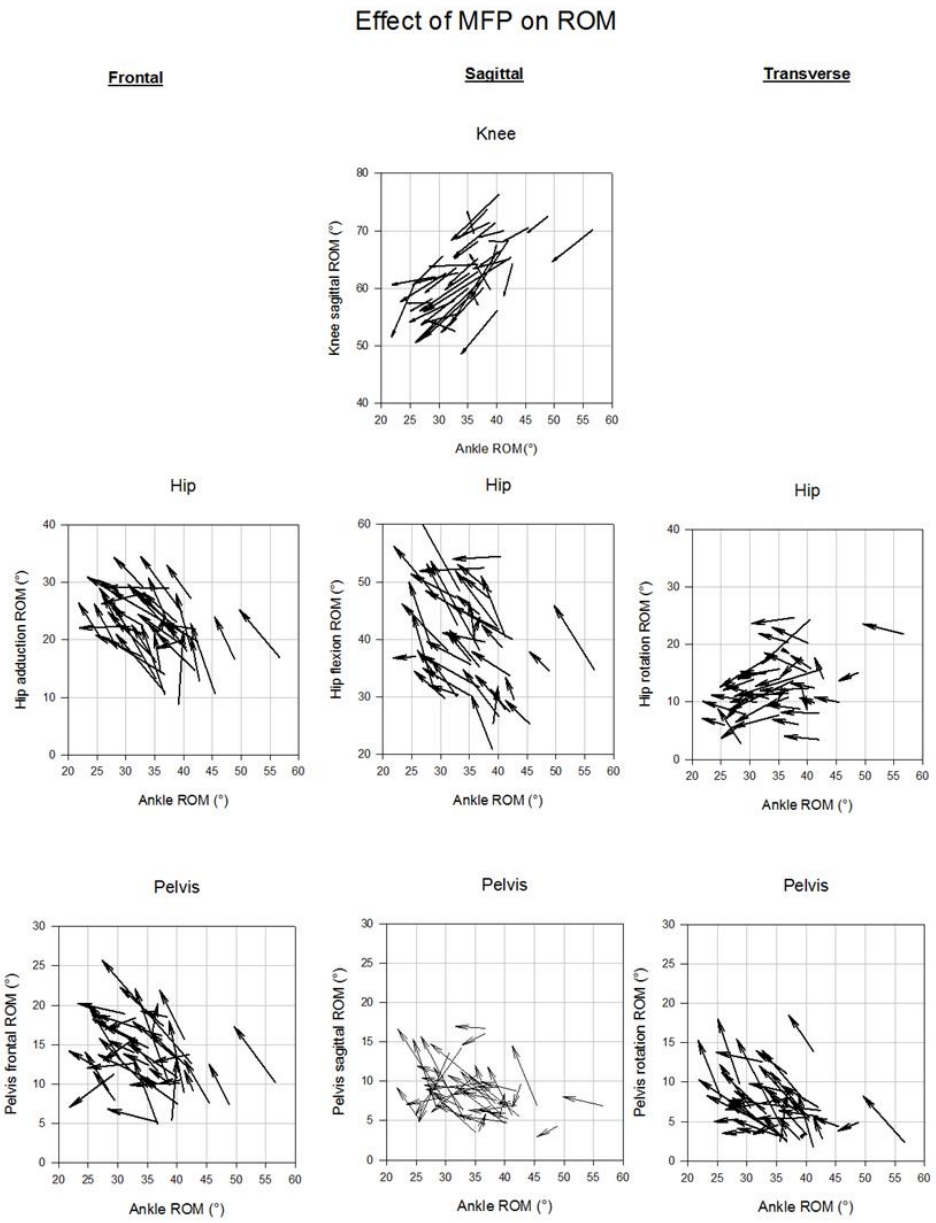

Fig 2. Effect of the modified foot position (MFP) on ROM on the dominant limb of all subjects. While knee and hip motion evolved in a typical way for all subjects, pelvis motion was not equally affected by the MFP.

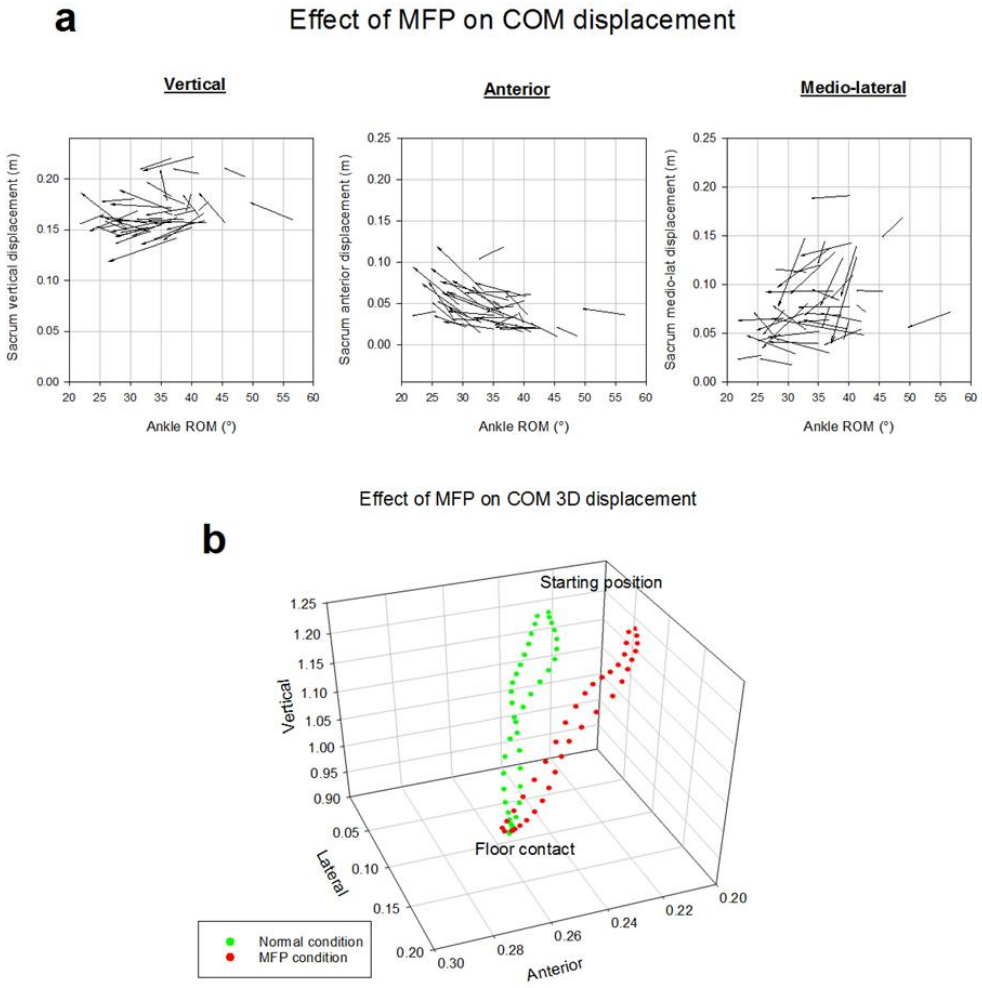

Fig 3. a: Effect of the modified foot position (MFP) on COM displacement during the entire cycle on the dominant limb on all subjects. Medio-lateral displacement was reduced for a certain number of subjects while some others were not affected by the MFP

b: Effect of the MFP on COM 3D displacement (cm) for one subject on dominant limb. X: anterior compared to the step descent axis, Y: vertical, Z: lateral. COM starting position is more posterior and towards supporting limb. We could have observed that the COM start more posteriorly in the MFP condition than in the normal condition.

4. Exploratory analysis

Exploratory results of coordination to address LLMP in control and MFP condition are illustrated in Figure 4 by means of phase plane graph [340]. We could observe the decrease in ankle and knee ROM and SPEED, and the increase in hip ROM and SPEED.

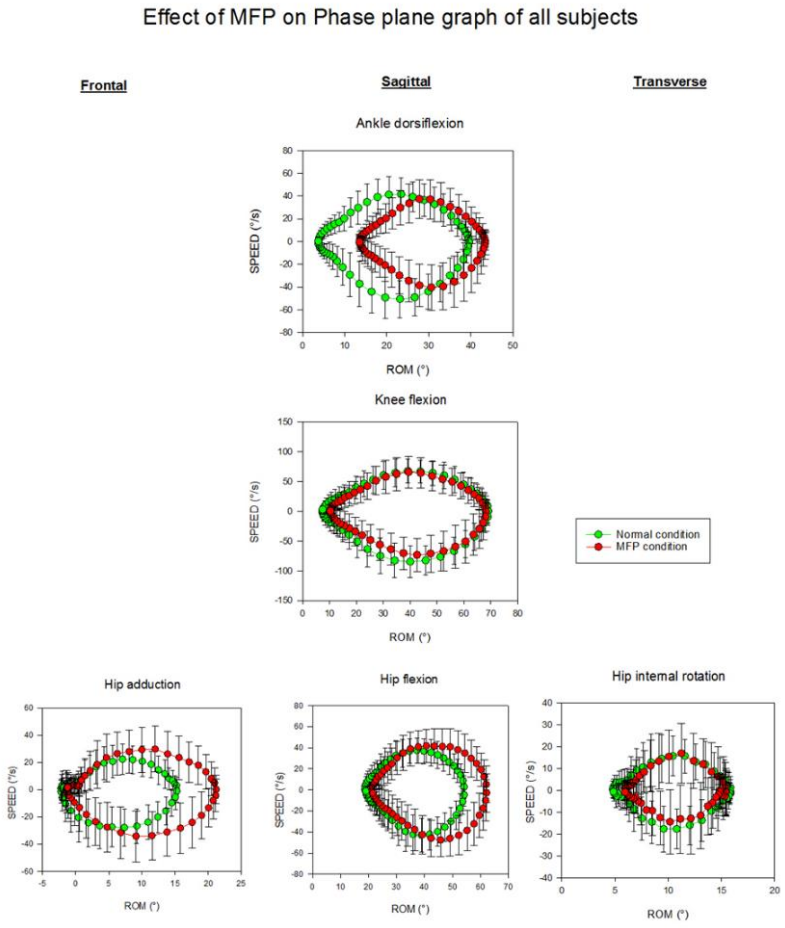

Fig 4: Effect of the MFP for all subjects on ROM and SPEED during the mean of 5 cycles. The upper part of the graph depicts the descent phase while the lower part depict the ascent phase. Pelvis graph are not presented because of high variability.

Table 2 : Reliability of step down task on dominant limb

		Dominant limb					
			Reproducibility	Measurement error		Responsiveness	
		Mean (SD)	ICC (95%CI)	SEM	SEM%	MDC	MDC%
	Time (s)	3.4 (0.9)	0.97 (0.95-0.98)	0.15	4.6	0.42	13
	ROM (°)						
Ankle	Sagittal	36.9 (5.7)	0.98 (0.96-0.99)	0.9	2.4	2.5	6.7
Knee	Sagittal	63.6 (6.0)	0.95 (0.92-0.97)	1.3	2.0	3.6	5.7
Hip	Sagittal	36.8 (7.9)	0.97 (0.96-0.99)	1.2	3.3	3.4	9.3
	Frontal	19.2 (5.3)	0.97 (0.95-0.98)	1.0	5.0	2.7	13.9
	Transverse	12.5 (4.9)	0.98 (0.97-0.99)	0.6	4.9	1.7	13.7
Pelvis	Sagittal	8.1 (3.3)	0.85 (0.76-0.92)	1.3	15.6	3.5	43.2
	Frontal	11.9 (3.8)	0.95 (0.91-0.97)	0.9	7.2	2.4	20.1
	Transverse	5.6 (2.9)	0.85 (0.76-0.92)	1.1	19.8	3.1	54.9
	SPEED (°/s)						
Ankle	Sagittal	116.3 (27.6)	0.91 (0.85-0.95)	8.3	7.1	23.0	19.7
Knee	Sagittal	186.8 (40.2)	0.92 (0.87-0.96)	11.2	6.0	31.0	16.6
Hip	Sagittal	100.4 (26.2)	0.96 (0.93-0.98)	5.4	5.3	14.9	14.8
	Frontal	69.9 (26.5)	0.96 (0.93-0.97)	5.6	8.0	15.6	22.3
	Transverse	51.2 (18.3)	0.92 (0.87-0.96)	5.1	9.9	14.1	27.5
Pelvis	Sagittal	34.2 (16.6)	0.88 (0.80-0.93)	5.7	16.7	15.8	46.3
	Frontal	50.2 (20.7)	0.94 (0.90-0.97)	5.0	9.9	13.8	27.6
	Transverse	30.6 (18.7)	0.75 (0.58-0.86)	9.4	30.7	26.0	85.1
	Displacement (cm)						
Supporting Knee	Medial descent phase	0.9 (0.7)	0.81(0.69-0.89)	0.3	34.0	0.8	94.3
Sacrum	Medio-latéral	8.0 (3.6)	0.89 (081-0.94)	1.2	14.8	3.3	41.0
	Antero-posterior	3.5 (1.4)	0.86 (0.76-0.92)	0.5	14.7	1.4	40.7
	Vertical	16.7 (1.9)	0.96 (0.93-0.98)	0.4	2.4	1.1	6.5

Time : trial duration (in seconds); ROM : range of motion (in degrees); SPEED : velocity (in degrees/s); Displacement(in centimeter); SD : Standard deviation; ICC : Intraclass correlation coefficient; SEM : Standard error of measurement, expressed in theu unit; SEM% : Standard error of measurement, expressed as a percentage; MDC : Minimal detectable change expressed in the unit; MDC% : Minimal detectable change, expressed as a percentage

Table 3 : Effet of the modified foot position on the dominant limb

		Dominant limb			
		Normal Mean (SD)	**MFP Mean (SD)**	**SMD (SD)**	**pvalue**
	Time (s)	3.1 (0.8)	3.7 (1.2)	-0.6 (0.8)	***
	ROM (°)				
Ankle	Sagittal	37.1 (5.8)	31.6 (6.4)	5.6 (3.1)	***
Knee	Sagittal	63.9 (5.6)	60.2 (6.5)	3.8 (4.0)	***
Hip	Sagittal	37.0 (7.8)	43.0 (8.3)	-6.0 (3.8)	***
	Frontal	19.3 (5.4)	25.6 (4.1)	-6.1 (3.7)	***
	Transverse	12.9 (5.1)	12.4 (5.1)		NS
Pelvis	Sagittal	7.8 (3.2)	9.6 (3.5)	-1.8 (4.2)	*
	Frontal	11.9 (3.7)	15.4 (4.3)	-3.5 (3.6)	***
	Transverse	5.5 (2.8)	8.7 (4.0)	-3.2 (2.9)	***
	SPEED (°/s)				
Ankle	Sagittal	122.1 (28.2)	96.1 (32.4)	26.4 (27.5)	***
Knee	Sagittal	195.2 (42.3)	162.7 (46.8)	33.2 (35.6)	***
Hip	Sagittal	105.4 (27.1)	106.6 (26.6)		NS
	Frontal	72.9 (30.8)	87.1 (28.2)	-13.4 (27.2)	**
	Transverse	53.3 (18.9)	53.9 (19.1)		NS
Pelvis	Sagittal	34.9 (15.4)	42.9 (24.8)	-8.0 (23.2)	*
	Frontal	52.5 (23.8)	56.1 (19.3)		NS
	Transverse	29.9 (14.9)	38.2 (16.4)	-7.9 (11.5)	***
	Displacement (cm)				
Supporting Knee	Medial descent phase	0.8 (0.8)	1.1 (0.9)	0.3 (0.8)	*
Sacrum	Medio-latéral	8.4 (4.2)	7.5 (3.3)		NS
	Antero-posterior	3.5 (1.5)	4.9 (2.4)	-1.3 (1.9)	***
	Vertical	16.6 (1.8)	16.6 (2.3)		NS

Time : trial duration (in seconds); ROM : range of motion (in degrees); SPEED : velocity (in degrees/s); displacement (in centimeter) ;SD : standard deviation; [25-75%] : interquartile range ; SMD :standardized mean of difference * p<0.05; **:p<0.005 ; ***;p<0.001; NS : p>0.05

Discussion

The primary objective of this study was to assess the reliability and responsiveness of the FSD. First of all, the reliability was excellent. The results for ROM were like previous studies, which assessed similar task [329, 346, 348, 349]. Concerning the knee medial displacement during the single leg stance phase, reliability was excellent, whereas other studies found also excellent within-day reliability and moderate between-day reliability [346]. Due to high SEM% (34%) and MDC% (94%), we could add that its measurement is not easily readable due to a relatively small-scale and large variability. This information could prevent us to draw conclusions about real knee medial displacement change following injury or rehabilitation. More research about this parameter is needed to give information to the clinicians which tend to use this criterion as an index of step descent [350-352]. As highlighted by Jones et al [353], observational rating of knee displacement may not be an appropriate substitute for a variable of interest: the knee abduction angle. They suggested that the apparent knee medial collapse was rather a pelvis lateral shift than knee displacement because the tibia remained relatively vertical.

Secondly, we observed that MFP decreasing ADROM significantly affects many ROM and SPEED parameters. Indeed, an effect on almost all parameters was observed. The lack of effect by the MFP on hip rotation was unexpected. We expected to have an effect such as Bell-Benje et al [336] who observed a decreased hip internal rotation in an opposite experimental set up in which they elevated heel during FSD. They suggested that hip internal rotation ROM will increase during a loading test that requires an increase in the hip flexion angle. In contrast to their hypothesis, we found a significant increase in hip sagittal ROM while hip rotation ROM remained the same. This

is probably due to the difference in the MFP used in these both studies, i.e. in plantar flexion and dorsal flexion respectively.

Rabin et al. also looked into the relationship between ADROM and lower limb kinematics without determining a cause-and-effect relationship [338]. Our MFP condition confirmed this effect on knee, hip and pelvis motion. This increase in pelvic ROM was consistent in all subjects in the frontal plane while the use of pelvis compensation in sagittal and transverse plane was more variable across subjects (Figure 2).

In this study, sacrum displacement was also assessed [350] that can be approximated as body COM [344]. The elevation of fore and mid-foot increases the COM anterior displacement while the ADROM did not affect it (Figure 3a). This could be explained by the initial posterior position of COM in the MFP condition. An illustration of COM displacement is presented in Figure 3b.

Since the FSD test is supposed to help detect aberrant movement pattern, it is important to understand the "normal" movement. The FSD test consist of lowering the body center of gravity from a certain height by functionally shortens the limb [354, 355]. Pelvic frontal and hip adduction movement occur to lower the swing phase limb while increases anterior pelvic tilt and forward trunk flexion move the center of gravity forward and maintain balance. As visual rating scales consider pelvic deviation and hip adduction as "poor" movement quality [350], it is important to notice that those movements are "normal" body adaptation to task constraints.

The ankle constraint in our MFP condition transformed motor pattern from an ankle strategy to a hip strategy as described by Horak et al. [356]. It reduced one degree of freedom of the body [357] and caused motor adaptation by increasing multiplanar motion on proximal joints. The multiplanar increases enhance the kinetic chain function in closed chain activity as step descent [170, 358]. Kinetic chain concept suggests that if the end of an overlapping rigid segment system if fixed, the application of an external force causes each segment to receive and transfer force to the adjacent segment. As a result, movement at any joint will produce a predictable movement pattern at all other joints [359].

Lower limb misalignment during dynamic task is the result of misalignment of the COM and the geometrical center of foot support [360]. As alignment is affected by ADROM limitation, previous authors recommend clinicians to assess it before screening LLMP [336, 338]. However, no clear ADROM cutoff value, neither guidelines concerning normal movement pattern exist to define aberrant movement pattern. Moreover, some authors standardize the FSD test by choosing step height relative to the subject's lower limb length [329]. In clinical practice, these standardizations are not always workable because of equipment availability. A perspective might me to study the relationship between anthropometry and task constraints in order to describe the behavior change in healthy subjects [25].

Previous studies hypothesized that angular speed motion might be a sensible outcome in pathological condition [361, 362]. Reduced knee angular velocity had already been observed in patients suffering from patellofemoral pain [362]. In our study, SPEED parameters evolved in the same way as ROM.

We proposed to use phase plane graph to get information about neuromuscular control of a segment [340] and highlight functional adaptation to the entire kinetic chain. We observed that ankle and knee speed decrease occurs mainly during the ascent phase. This approach has the potential to increase our understanding of LLMP.

LLMP result from complex interactions between task components, degree of freedom availability, subject anthropometry, automatic and voluntary control, motivation as well as instructions understanding. Even if altered LLMP can be useful to identify and to rehabilitate, other parameters were also determinant for prevention of musculoskeletal disorders of the lower limb such as the mechanical quantification of stress [363]; during daily life, work, sport combined with psychological anxiety, stress, as well as life style factors [364]. Furthermore, our study was achieved in a motion analysis lab. Whereas available method for clinicians is real-time visual observation [365], a perspective may be to quantify the same parameters with affordable new technologies such as the Inertial Measurement Unit [366]. Such approach will probably increase clinicians' abilities to measure LLMP on the patient spot and in their daily practice.

Limitations: The effect of our ADROM limitation is not confined to a mechanical constraint. Indeed, the ankle position slightly modified body orientation and therefore sensory input to the somatosensory system (vestibular and foot receptors)[360]. It is unknown if our experimental model is representative of dorsiflexion limitation following ankle injuries. Our constraint does not reproduce the somatosensory loss [367] or overactive EMG activity which can be present in some patients with posttraumatic ankle sprain

[368] Sensory information is interpreted by the CNS to generate appropriate equilibrium reaction [360] and can change LLMP without mechanical limitation of ROM. Further researches on other musculoskeletal disorders or even surgery might help to increase our understanding of LLMP in ankle pathologies.

Moreover, it is difficult to exclude other effects such as fatigue even if we provide a two minutes' break between each test. A randomization of trial's order might have been appropriate.

Unfortunately, our model could not assess trunk motion nor three-dimensional knee motion as well as foot pronation. Finally, only a relatively young population was assessed while other authors show an age effect including greater frontal and transverse plane motion [369, 370].

Conclusions

In conclusion, reliability of the kinematic parameters of the FSD test in our healthy sample were excellent. Kinematics results were affected by an ADROM experimental limitation except for hip rotation ROM. This is a demonstration of cause-and-effect relationship between ADROM limitation and LLMP in FSD test as knee, hip and pelvis motion were affected. Even if comparative studies are needed with post-traumatic ankle sprain patients, the clinicians should already take ADROM limitation into account when interpreting LLMP during FSD test.

Appendix 4

Appendix 4: Accompanying thesis: article 2

Variations in muscle activity patterns when walking in everyday environments: the effect of different surfaces

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Appendix 4

Abstract

Purpose: To examine variations in lower limb EMG activity when walking on different outdoor surfaces and characterize the different potential motor strategies.

Methods: Forty healthy adult participants walked at self-selected speed over asphalt, pavement and grass. Participants then walked on an indoor treadmill at the gait speed observed for each outdoor condition. EMG activity of the vastus lateralis (VL), tibialis anterior (TA), biceps femoris (BF) and gastrocnemius lateralis (GL) muscles was recorded. The duration and intensity (root mean square) of EMG burst activity was calculated.

Results: Walking on grass resulted in longer TA burst duration than other outdoor surfaces. Walking on pavement was associated with increased intensity of TA and VL activation compared to the indoor treadmill condition. The variability of EMG intensity for all muscle groups tested (TA, GL, BF, VL) was greatest on grass and lowest on asphalt.

Conclusion: Healthy adult subjects vary muscle activity patterns in response to the different qualities of outdoor walking surfaces. Ongoing development of ambulatory EMG methods will be required to support ecologically tailored gait retraining programs.

Keywords: electromyography, wearable devices, gait, ambulatory assessment, walking surfaces

Introduction

The use of surface electromyography (EMG) in gait laboratories is an established technique in clinical evaluation of gait disorders [371]. By measuring variations in electrical activity of muscle fibres, EMG provides insight into the timing and intensity of muscle bursts during human gait [372]. EMG can assist clinicians to examine pathological (e.g. paresis or spasticity) as well as functional (e.g. adaptive or compensatory) changes in patterns of muscle activity during gait [373]. Evaluations carried out in gait laboratories however, provide only a snapshot of a person's locomotor profile[374], and the extent to which these measures are predictive of further injury or falls risk may be questionable[375, 376]. As such, there is a broad clinical demand for methods which support movement analysis in everyday environments [374, 377].

Recent advances in wearable sensor technology provide the possibility of recording EMG signals under ecologically valid conditions [248, 378-380]. A small number of studies have attempted to characterize motor behaviour using these techniques. For example, Roy et al. have used wearable EMG to classify daily life movements (feeding, locomotion...) and distinguish clinical features (tremor, dyskinesia...) in patients with neurological disorders [381, 382]. Studies examining patterns of muscle activation during walking in ecological contexts however remain scarce [383]. In order to develop pertinent clinical approaches for examining muscle burst timing during everyday walking, ongoing work is needed to understand how different patterns of muscle activation correspond with particular environmental conditions[300].

Importantly, previous experimental studies have illustrated changes in lower limb EMG activity on different walking surfaces. For example, a general

prolongation in muscle burst duration is observed on uneven [384] and slippery [385] conditions. In the present study, we examine patterns of lower limb muscle activity vary across different terrains typically encountered in day to day walking. The first surface, asphalt, is highly regular. The second, pavement, is more irregular with greater surface roughness, in addition to the presence of macroscopic particles (e.g. sand) causing variations in the coefficient of friction [386]. The third surface, grass, has comparatively lower rigidity at the surface layer and potential variations in surface roughness from the ground. Our hypothesis was that variations in the duration and intensity of EMG activity would be more apparent on the comparatively irregular pavement and grass surfaces, than on the more regular asphalt surface. Given that lower limb muscle activity is related to walking speed [387], we also carried out further testing at corresponding gait speeds on an indoor treadmill. This measure was used to verify that differences in EMG patterns seen between an outdoor walking surface and the associated treadmill trial were specific to the surface rather than gait speed.

Methods

Participants

A convenience sample of forty healthy participants (17 female, 23 male; age 22.8±2.2 years, height 175 ± 6 cm) were recruited (see supplementary material S7). The study was approved by the university ethics committee (agreement number: B403201523492). Written consent was obtained from subjects prior to testing.

Experimental setup

EMG data was recorded using a 1 kHz telemetry system (BTS FREEEMG 1000, Italy). The surface of the skin was shaved and cleaned prior to electrode placement. Pre-gelled electrodes were placed on the vastus lateralis (VL), tibialis anterior (TA), biceps femoris (BF) and gastrocnemius lateralis (GL) of the non-dominant leg (determined using item 1 of the Waterloo Footedness Questionnaire)[388] following the recommendations established by Perotto [389]. These muscles were chosen for their role in sagittal plane mechanics, important in forward locomotion.1 Outdoor gait speed was measured using a GPS watch (Garmin Forunner 110). The accuracy of this device was previously assessed over a distance of 100m against a manual stopwatch (bias: 0.005 ms-1, limits of agreement: -0.13 ms-1, 0.14 ms-1).

Experimental procedure

Participants walked at spontaneous gait speed, with their habitual footwear on 3 flat outdoor surfaces: an asphalt road (A), grass (G) and pavement (P) (see supplementary material S1). Each was instructed to walk forwards, in a straight line, from a designated location to a fixed target in front of them. The order of trials was randomly allocated yielding one of six possibilities (AGP, APG, GAP, GPA, PAG, PGA). All outdoor trials were carried out under favorable weather and light conditions. The chosen pathways were in less frequented areas so as to limit interactions with other pedestrians or optical flow effects. As this experiment was conducted in an outdoor campus environment there was a continuous, ambient background noise, common to urban environments (birds, distant traffic...). Following this, participants walked on an indoor treadmill at the different gait velocities measured in the outdoor

conditions. EMG signals were recorded for a duration of 60 seconds in each condition.

EMG signal processing

Raw EMG signals were rectified, and the linear envelope processed using a 5th order low-pass Butterworth filter with a cut-off frequency of 20Hz. For each condition, 20 successive bursts of EMG activity were identified using the method described by Van Boxtel [390] (see supplementary material S2). Durations for individual muscle bursts were then collated for each condition. Following this, root mean square (RMS) values for each burst were calculated using the following equation (1):

$$\text{RMS} = \sqrt{\frac{\sum_{i=0}^{n=1} X_i^2}{n}} \quad (1)$$

Where i= frame at sample i, X: EMG signal, n=number of selected frames.

The mean and the coefficient of variation (CV) were calculated for each parameter on the 20 successive bursts (burst duration, RMS). CV is a measure of relative variability and was calculated using the following equation (2).

$$\text{CV} = \frac{standard\ deviation}{mean} \quad (2)$$

Statistical analysis

Differences in gait speed for walking on the outdoor surfaces were examined using repeated measures analysis of variance (RM-ANOVA). Mean and CV values for EMG burst duration and RMS were similarly examined using one way RM-ANOVA (outdoor surface condition with 3-levels: asphalt,

pavement, grass). Where data was not normally distributed, non-parametric equivalents (by ranks) were used. Multiple comparisons during post-hoc analysis were corrected using the Bonferonni or Dunn method (with Asphalt as control condition) where required. Paired t-tests were used for comparisons between walking on outdoor surfaces and walking on a treadmill at matched gait speed. The threshold for statistical significance was set at p=0.05. Statistical analyses were performed using SPSS and Sigmaplot.

Results

Gait speed across outdoor surfaces

Spontaneous gait speed varied across outdoor surfaces (DF=2, F=1.35, p=0.010) with subjects walking at a mean speed (standard deviation) of 1.33ms-1 (0.19ms-1) on asphalt, 1.31ms-1 (0.22ms-1) on grass and 1.2ms-1 (0.31ms-1) on pavement. Post-hoc testing confirmed that gait speed was greater on asphalt than pavement (p=0.008).

Muscle activity patterns across outdoor walking surfaces

Mean duration of TA activity varied according to surface (DF=2, chi-square= 10.364 p=0.006) with a median value for mean burst durations of 0.68s on asphalt, 0.72s on grass and 0.68s on pavement. Type of surface also had a significant effect on CV of burst duration for VL (DF=2, chi-square= 6.867, p=0.032) and TA (DF=2, chi-square= 6.606, p=0.037,) with CV values increasing between asphalt, grass and pavement respectively. Post-hoc testing indicated that CV of duration of VL activity was greater on pavement than on asphalt (p=0.026).

Variation of EMG intensity similarly changed from one surface to another. Repeated measures ANOVA yielded significant differences for CV of RMS values in each muscle group ($p<0.001$). Post-hoc testing indicated that CV of RMS values was greater on grass than on asphalt ($p<0.001$ for each muscle), and that CV was greater on pavement than on asphalt ($0.002<p<0.011$ for different muscles tested).

See figure 1A summarizes CV of RMS values for each muscle group across the outdoor surfaces. The table in supplementary material S3 provides further details on these results.

Muscle activity patterns comparison between outdoor surfaces and indoor treadmill condition

EMG intensity when walking on pavement was different to that observed at matched gait speed on an indoor treadmill. Median value of mean RMS for VL was 0.099mV on pavement compared to 0.079mV on the treadmill ($p=0.015$). Median value of for mean RMS for TA was 0.099mV on pavement compared to 0.089mV on a treadmill ($p=0.006$).

For walking on grass, BF median intensity was 0.18mV, greater than the 0.15mV observed on the treadmill ($p=0.004$). The CV of intensity was greater on grass than on the treadmill for the ensemble of muscles tested (VL $p=0.002$, BF $p=0.013$; TA $p<0.001$; GL $p=0.011$). The median CV of GL burst duration was also greater on grass at 0.15s than in the treadmill condition at 0.11s ($p=0.02$).

No differences were observed between walking on asphalt and walking on the treadmill were observed for any of the parameters tested. Figure 1 panels

B–D illustrate significant findings while tables S4-S6 in the supplementary materials provide further detail of statistical comparisons between outdoor walking surfaces and indoor treadmill walking at corresponding gait speed.

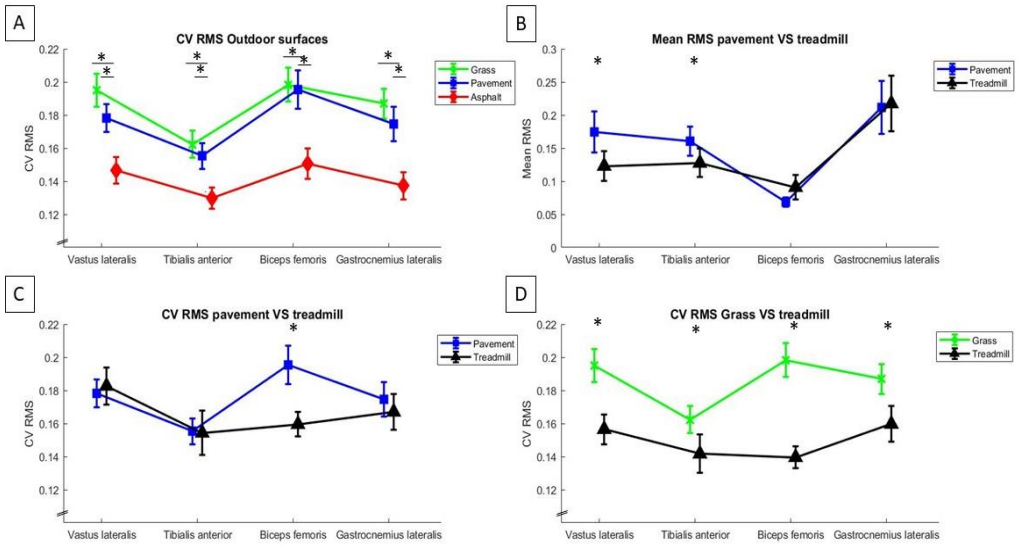


*Figure 1. Intensity of EMG burst activity on outdoor surfaces and treadmill. (A) Repeated Measures ANOVA between outdoor surfaces :significant differences for CV of RMS values in each muscle group for outdoor surfaces, (B-C-D) Paired t-test between outdoor surface and treadmill at matched speed: significant differences for mean RMS values for vastus lateralis and tibialis anterior between pavement and treadmill conditions, significant difference for CV of RMS for biceps femoris between pavement and treadmill conditions, significant difference for each muscle group for CV of RMS values between grass and treadmill conditions. (RMS: root mean square, CV: coefficient of variation, * indicate a significant difference (p<0.05), error bar represents standard error)*

Discussion

Consistent with the initial hypothesis, EMG activity changed on different outdoor walking surfaces. Our principal finding was that the CV of EMG intensity (RMS) varied for each muscle group tested (VL, BF, TA, GL; figure 1A).

Changes in muscle activity in response to outdoor walking surfaces

Walking on pavement resulted in increased RMS values of EMG activity in TA and VL. Upon contact with the ground, TA and VL work together, ensuring stability from heel strike into the weight transfer following initial contact[371]. Increased intensity of their activity on pavement is likely to reinforce dorsiflexion and knee extension through these points in the gait cycle. These results may be compatible with another study which noted specific kinematic changes for the knee and ankle in healthy subjects and patients with Parkinson's disease (PD) walking on a cobblestone surface [391]. An increase in knee sagittal amplitude and a stable ankle joint may be considered as adaptation strategies to this irregular surface and may be evocative of divergent concentric or eccentric muscle contractions across the knee and ankles. Whether these changes in EMG intensity serve to compensate for hardness (and corresponding shock upon foot placement) or irregularity in the form of the surface remains to be determined.

Walking on grass resulted in longer burst duration for TA. Increased EMG duration through stance phases has been associated with strategies for enhanced gait stability. For example, TA burst duration increases on slippery walkways, with GL-TA cocontraction inhibiting unintended variations in ankle range[385]. In addition, TA also supports dorsiflexion, thus ensuring foot

clearance. However, as TA is typically active through the entire duration of the swing phase[371], it is less likely that increased burst duration observed on grass would be associated with this phase. The CV of EMG intensity was consistently greater for grass than the other surfaces in all muscles. Instead of a fixed change in muscle firing patterns, walking on grass appears to involve punctual changes in muscle activation. This finding reflects the small, continuous variations in camber and roughness of grassy outdoor surfaces. Punctual modifications in EMG intensity at the level of the knee and ankle are potentially associated with changes in surface regularity through the antero-posterior and medio-lateral axes respectively [392].

The absence of significant differences in EMG activity between asphalt and the indoor treadmill condition underscores the relatively consistent properties of this outdoor surfaces.

EMG data from everyday walking environments for improved patient care

Improving gait stability in everyday settings implies identifying and training adaptive neuromuscular responses. The changes in EMG activity we observed appear consistent with mechanisms to ensure gait stability. We presume that this ability to regulate EMG activity would be diminished in people with pathology of the locomotor system. People with recurrent ankle injuries have difficulties managing frontal plane dynamics [393], they would likely exhibit deficient EMG firing patterns on surfaces which induce lateral instability. Older adults have a tendency for increased agonist-antagonist coactivation during stance phases [394]. Age related decline in gait function would potentially involve stereotypical patterns of cocontraction to maintain joint stability in anticipation of surface irregularity.

A potential benefit of ecological EMG data is the ability to verify how neuromuscular exercises used in rehabilitation are generalized into everyday life situations. Gait retraining after a musculoskeletal lesion typically involves progressions combining bilateral and unilateral exercises with varying degrees of surface instability and dynamic movement. However, these interventions are not necessarily associated with changes in kinematic patterns during laboratory gait analyses [395]. EMG data from everyday life situations may prove more sensitive to the different mechanisms (e.g. feedforward movement processing, proprioceptive neuromuscular correction) contributing to adaptive patterns of muscle activation.

In addition to diagnostic and evaluative purposes, wearable EMG sensors have the potential to become viable tool in gait rehabilitation. Previous clinical based studies suggest that task oriented biofeedback approaches effectively support motor learning. For example, EMG biofeedback training in patient with neurological conditions has been associated with significant improvement in joint power, stride length and gait speed when compared with gait conventional rehabilitation methods [396, 397]. One might envisage that real-time feedback regarding patterns of muscle activity over varying terrains may provide a particular advantage in consolidating novel motor strategies for enhanced gait stability in daily life environments.

Limitations and ongoing development of ambulatory gait analysis systems

Limits of this study may be attributed to both methodological choices as well as current technological barriers. From a study design perspective, a further experimental condition whereby subjects were required to walk at a specific gait speed over each outdoor surface would improve the validity of

direct comparisons in EMG parameters across the surfaces. Using a counterbalanced trial order as opposed to randomised trial order may have further reduced potential biases related to sequencing of the conditions. Further to this, the addition of further EMG sensors (e.g. peroneus) may have provided greater insight into muscle activation serving to provide support against lateral instability. In order to be used as a truly valid ecological gait assessment tool, EMG sensors must be coupled with additional sensors (accelerometers, foot switches) capable of distinguishing the different phases of the gait cycle[398]. This will considerably improve functional interpretation of variations in muscle activity. EMG signal processing also remains a time-consuming process. Software facilitating data analysis is needed for clinicians.

Conclusions

Healthy adult subjects modify patterns of lower limb muscle activity according to the characteristics of outdoor walking surfaces. We propose that improved understanding of EMG activity across different terrains can be used in planning and evaluating the effectiveness of gait rehabilitation. Ongoing collaborative work between physiotherapists and medical engineers is necessary to further develop EMG methods for gait analysis in everyday environments.

Key messages

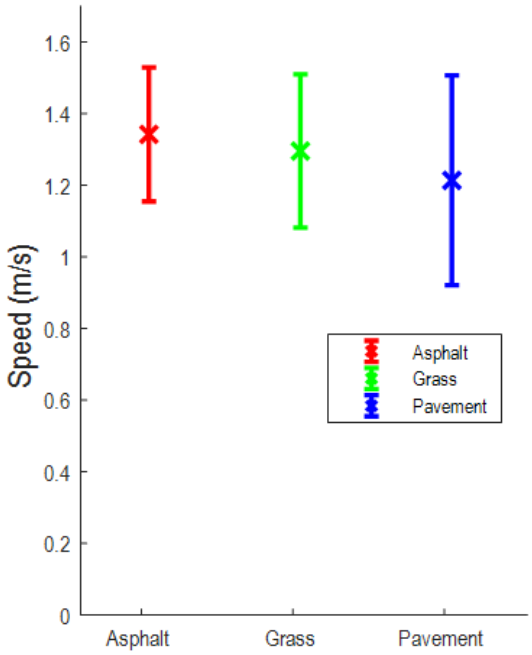
What is already known on this topic

With the development of onboard sensor technology, it is now possible to record EMG data during everyday situations. Still, these methods are not sufficiently mature for use in clinical gait assessment [374]. A central problem is interpretation of this this data in non-standardised conditions[264].

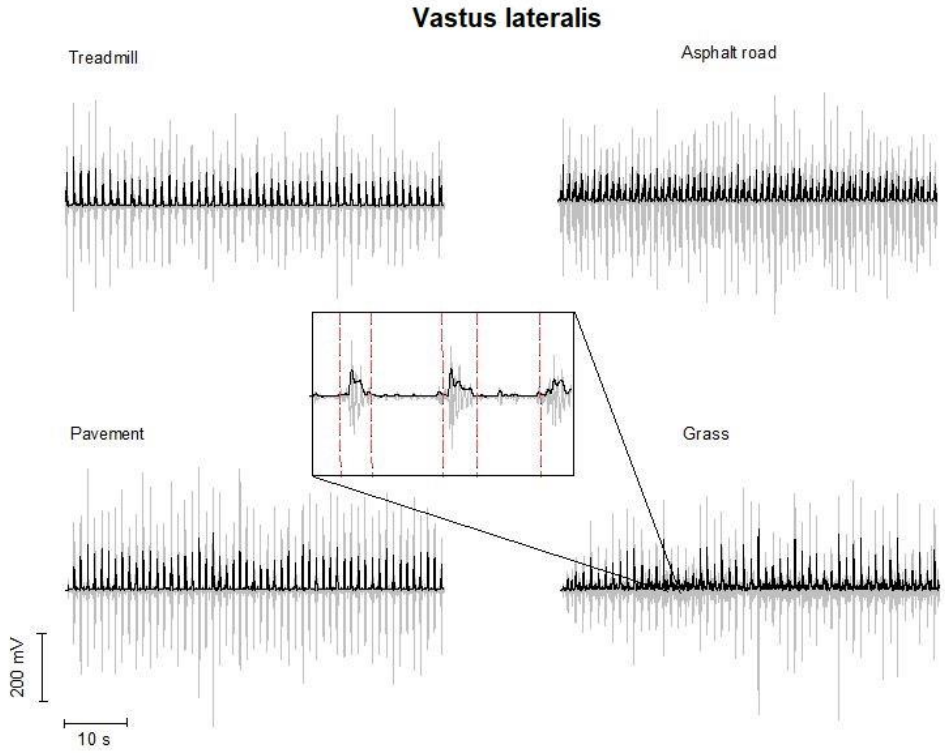
What this study adds

This study demonstrates that patterns of muscle activation change over different walking surfaces. Modifications in burst duration or intensity may reflect the specific properties of that surface while increased variation of these parameters punctual changes in surface quality. Onboard EMG sensors could be used to evaluate muscular responses during gait.

Supplementary materials



Supplementary material 1. Gait speed across outdoor surfaces



Supplementary material 2: Typical trace from vastus lateralis muscle from one subject in the different conditions. The central window illustrates the selection of burst of EMG activities

Appendix 4

Table S3 : Muscular strategies on outdoor surfaces

			Outdoor Surfaces					
			Asphald road (A)	Grass (G)	Pavement (P)	ANOVA - p value	Post-hoc p value	
							Asphalt VS Grass	Asphalt VS Pavement
		Walking speed (ms^{-1})	1.33 [1.23-1.46]	1.28 [1.17-1.36]	1.19 [0.94-1.44]	**0.010**	0.546	0.006
Intensity : RMS (mV)	Mean	Vastus lateralis	0.089 [0.056-0.233]	0.089 [0.055-0.201]	0.099 [0.059-0.194]	0.53	-	-
		Tibialis anterior	0.105 [0.071-0.190]	0.112 [0.069-0.171]	0.099 [0.073-0.194]	0.53	-	-
		Biceps femoris	0.066 [0.038-0.112]	0.061 [0.040-0.099]	0.053 [0.037-0.100]	0.51	-	-
		Gastrocnémius lateralis	0.108 [0.044-0.407]	0.102 [0.514-0.321]	0.091 [0.045-0.199]	0.53	-	-
	Coefficient of variation	Vastus lateralis	0.13 [0.11-0.18]	0.19 [0.15-0.23]	0.17 [0.15-0.21]	**<0.001**	<0.001	0.004
		Tibialis anterior	0.13 [0.10-0.15]	0.16 [0.12-0.19]	0.14 [0.12-0.19]	**<0.001**	<0.001	0.005
		Biceps femoris	0.13 [0.11-0.18]	0.19 [0.15-0.24]	0.18 [0.16-0.24]	**<0.001**	<0.001	0.002
		Gastrocnémius lateralis	0.13[0.09-0.18]	0.18 [0.15-0.23]	0.16 [0.13-0.20]	**<0.001**	<0.001	0.008
Activation duration (s)	Mean	Vastus lateralis	0.47 [0.33-0.83]	0.47 [0.36-0.79]	0.50 [0.33-0.81]	0.43	-	-
		Tibialis anterior	0.68 [0.51-0.82]	0.72 [0.59-0.82]	0.68 [0.57-0.83]	**0.006**	0.006	0.019
		Biceps femoris	0.60 [0.47-0.66]	0.56 [0.48-0.67]	0.55 [0.49-0.65]	0.28	-	-
		Gastrocnémius lateralis	0.64 [0.49-0.74]	0.68 [0.55-0.77]	0.68 [0.54-0.81]	0.79	-	-
	Coefficient of variation	Vastus lateralis	0.13 [0.08-0.18]	0.15 [0.10-0.19]	0.15 [0.09-0.21]	**0.032**	0.187	0.020
		Tibialis anterior	0.13 [0.09-0.18]	0.14 [0.09-0.18]	0.15 [0.11-0.18]	**0.037**	1.000	0.073
		Biceps femoris	0.15 [0.12-0.19]	0.17 [0.13-0.22]	0.17 [0.15-0.23]	0.15	-	-
		Gastrocnémius lateralis	0.15 [0.11-0.18]	0.15 [0.12-0.19]	0.15 [0.12-0.20]	0.85	-	-

Data are presented in Median (25-75% quartile range), A= Asphalt road, G= Grass, P = Pavement

Table S4 : Muscular strategies on asphalt VS treadmill

			Asphalt road	Treadmill - asphalt	paired two tailed p value
Intensity : RMS (mV)	Mean	Vastus lateralis	0.089 [0.056-0.233]	0.087 [0.057-0.181]	0.13
		Tibialis anterior	0.105 [0.071-0.190]	0.110 [0.068-0.190]	0.44
		Biceps femoris	0.066 [0.038-0.112]	0.047 [0.033-0.092]	0.42
		Gastrocnémius lateralis	0.108 [0.044-0.407]	0.097 [0.046-0.313]	0.63
	Coefficient of variation	Vastus lateralis	0.13 [0.11-0.18]	0.13 [0.12-0.17]	0.42
		Tibialis anterior	0.13 [0.10-0.15]	0.13 [0.09-0.16]	0.76
		Biceps femoris	0.13 [0.11-0.18]	0.13 [0.12-0.16]	0.76
		Gastrocnémius lateralis	0.13[0.09-0.18]	0.14 [0.11-0.15]	0.3
Activation duration (s)	Mean	Vastus lateralis	0.47 [0.33-0.83]	0.47 [0.35-0.80]	1
		Tibialis anterior	0.68 [0.51-0.82]	0.63 [0.49-0.76]	0.53
		Biceps femoris	0.60 [0.47-0.66]	0.57 [0.47-0.68]	0.83
		Gastrocnémius lateralis	0.64 [0.49-0.74]	0.65 [0.57-0.78]	0.14
	Coefficient of variation	Vastus lateralis	0.13 [0.08-0.18]	0.12 [0.09-0.16]	0.78
		Tibialis anterior	0.13 [0.09-0.18]	0.15 [0.11-0.18]	0.24
		Biceps femoris	0.15 [0.12-0.19]	0.17 [0.13-0.22]	0.73
		Gastrocnémius lateralis	0.15 [0.11-0.18]	0.13 [0.09-0.17]	0.15

Data are presented in Median (25-75% quartile range)

Table S5 : Muscular strategies on grass VS treadmill

			Grass	Treadmill - grass	paired two tailed p value
Intensity : RMS (mV)	Mean	Vastus lateralis	0.089 [0.055-0.201]	0.088 [0.047-0.171]	0.21
		Tibialis anterior	0.112 [0.069-0.171]	0.104 [0.061-0.164]	0.15
		Biceps femoris	0.061 [0.040-0.099]	0.049 [0.031-0.089]	0.07
		Gastrocnémius lateralis	0.102 [0.514-0.321]	0.084 [0.040-0.285]	0.17
	Coefficient of variation	Vastus lateralis	0.19 [0.15-0.23]	0.15 [0.12-0.17]	**0.002**
		Tibialis anterior	0.16 [0.12-0.19]	0.13 [0.09-0.15]	**0.013**
		Biceps femoris	0.19 [0.15-0.24]	0.14 [0.11-0.16]	**<0.001**
		Gastrocnémius lateralis	0.18 [0.15-0.23]	0.14 [0.12-0.18]	**0.011**
Activation duration (s)	Mean	Vastus lateralis	0.47 [0.36-0.79]	0.48 [0.36-0.79]	0.73
		Tibialis anterior	0.72 [0.59-0.82]	0.69 [0.52-0.79]	0.23
		Biceps femoris	0.56 [0.48-0.67]	0.59 [0.51-0.72]	0.31
		Gastrocnémius lateralis	0.68 [0.55-0.77]	0.65 [0.49-0.81]	0.73
	Coefficient of variation	Vastus lateralis	0.15 [0.10-0.19]	0.14 [0.10-0.17]	0.53
		Tibialis anterior	0.14 [0.09-0.18]	0.16 [0.09-0.18]	0.79
		Biceps femoris	0.17 [0.13-0.22]	0.16 [0.11-0.20]	0.25
		Gastrocnémius lateralis	0.15 [0.12-0.19]	0.11 [0.07-0.17]	**0.02**

Data are presented in Median (25-75% quartile range)

Table S6 : Muscular strategies on pavement VS treadmill

			Pavement	Treadmill - Pavement	paired two tailed p value
Intensity : RMS (mV)	Mean	Vastus lateralis	0.099 [0.059-0.194]	0.079 [0.047-0.152]	**0.015**
		Tibialis anterior	0.099 [0.073-0.194]	0.089 [0.061-0.135]	**0.006**
		Biceps femoris	0.053 [0.037-0.100]	0.040 [0.032-0.102]	0.48
		Gastrocnémius lateralis	0.091 [0.045-0.199]	0.087 [0.038-0.285]	0.07
	Coefficient of variation	Vastus lateralis	0.17 [0.15-0.21]	0.17 [0.13-0.22]	0.12
		Tibialis anterior	0.14 [0.12-0.19]	0.13 [0.11-0.18]	0.14
		Biceps femoris	0.18 [0.16-0.24]	0.15 [0.13-0.18]	**0.004**
		Gastrocnémius lateralis	0.16 [0.13-0.20]	0.15 [0.11-0.21]	0.13
Activation duration (s)	Mean	Vastus lateralis	0.50 [0.33-0.81]	0.49 [0.36-0.75]	0.64
		Tibialis anterior	0.68 [0.57-0.83]	0.68 [0.52-0.84]	0.97
		Biceps femoris	0.55 [0.49-0.65]	0.61 [0.52-0.74]	0.07
		Gastrocnémius lateralis	0.68 [0.54-0.81]	0.71 [0.52-0.80]	0.71
	Coefficient of variation	Vastus lateralis	0.15 [0.09-0.21]	0.13 [0.09-0.19]	0.06
		Tibialis anterior	0.15 [0.11-0.18]	0.15 [0.09-0.19]	0.78
		Biceps femoris	0.17 [0.15-0.23]	0.17 [0.12-0.20]	0.17
		Gastrocnémius lateralis	0.15 [0.12-0.20]	0.15 [0.10-0.19]	0.13

Data are presented in Median (25-75% quartile range)

Table S7 : Subject's characteristics

Gender	Age (years)	Weight (Kg)	Height (cm)
Women (n=17)	22.53 (2.21)	59.29 (6.9)	167.53 (4.91)
Men (n=23)	23.04 (2.16)	75.75 (9.66)	180.96 (6.79)

Data are presented in Mean (standard deviation)

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