Dottorato di Ricerca in Bioingegneria
XIX Ciclo, Anno Accademico 2005-2006

Sede Amministrativa: Università degli Studi, Bologna
Sede Consorziata: Istituto Universitario di Scienze Motorie, Roma

MECHANICS OF THE UPPER PART OF THE BODY DURING LOCOMOTION

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I wish to thank and dedicate this thesis

to my Mum,

to the best supervisor I have ever met in my life, Claudia,

and to my best friend, Caffeine:

without their support this thesis would have never been possible!

(da *Vorrei* di Francesco Guccini)

"…vorrei restare per sempre in un posto solo
per ascoltare il suono del tuo parlare
e guardare stupito il lancio, la grazia, il volo
impliciti dentro al semplice tuo camminare…"

Ringraziamenti

Era il 25 giugno 2002 quando, quasi per sbaglio, ho messo piede in quello che sarebbe stato il mio futuro di cinque lunghi anni. Passato, presente e futuro di ricordi ed emozioni, di visi ed occhi che fra queste mura ho incontrato e che non dimenticherò.

Voglio dedicare la mia tesi a tutte le persone che hanno preso parte, volenti o nolenti, attivamente o passivamente, a questo progetto.

Ovviamente le prime parole di ringraziamento vanno ai miei genitori, *Luisa* e *Giancarlo*, il loro supporto ed il loro modo di crescermi nella lealtà, nella lotta, insegnandomi a credere sempre in ciò che volevo, sono stati fondamentali per farmi arrivare dove sono ora. Con loro non posso ovviamente dimenticare mio padre *Agostino* ed i miei nonni, *Claudio*, *Rosina*, *Peppe* e *Francesca*, purtroppo scomparsi da troppi anni e troppo presto. A questo piccolo gruppo di persone dal cuore smisurato devo tutto quello che sono e tutto quello che ho. Insieme a loro devo ringraziare il mio primo punto di riferimento maschile della mia vita, *Leonardo*. A suo fianco *Dorothy* per aver addolcito mio fratello e per avermi fatto conoscere *Mikol* e tutta la *combriccola del Buru*. A tutti loro, un sentito e profondo grazie per avermi regalato pomeriggi indimenticabili e serate di feste dimenticate!

Durante questo cammino di studio e lavoro sono stato affiancato da un nutrito numero di persone e personaggi: *Giorgio*, *Pietro*, *Ugo*, *Mounir*, *Andrea Linetti*, *Paolo*, *Marco Io(sa)*, *Valentina e Domenico*. Con loro *CoCoCo Andrea*, *CoCoCo Valentina*, *CoCoCo salva.motta*, *ProfMassimo* ed il MITICO *ProfNicola*. Persone superlative per cui svegliarsi la mattina del sabato è sempre stato e sarà sempre un piacere! Per ultimo, ma non in ordine di importanza, ringrazio *Giuseppe* colui che mi ha accolto col suo proverbiale sorriso quel giorno d'estate di 5 anni fa e che, qualche mese dopo mi ha passato come fossi un testimone a *Claudia*, ora e per sempre la mia supervisor. Lei e con lei il *Professor Cappozzo,* che per una volta voglio chiamare *Aurelio*, sono colpevoli di avermi fatto crescere dentro il senso della ricerca, l'amore per questo lavoro ed a loro devo le splendide esperienze avute durante questi tre anni di dottorato e le

laute cene che abbiamo divorato insieme. Tra le esperienze più eccitanti non posso dimenticare Glasgow ed il dipartimento di fisiologia applicata dove ho avuto il piacere di collaborare con *Andrea Macaluso*, a cui bacio le mani. Con lui i miei coinquilini: *Waheed* l'egiziano dalla pinta facile, *Sandy* il Mauriziano, la mia kompagna di partito *Senija* e il grande e insuperabile *Luca*, perché a Glasgow si può imparare anche il Siciliano e non solo l'Inglese!

Torniamo a Roma, ma rimaniamo sulla vita quotidiana. Grazie a *Davide* per aver sopportato le mie lamentele ed i miei avanti e indietro nel corridoio alla ricerca del grafico giusto per soddisfare Claudia! A proposito di coinquilini, non posso non ringraziare il padrone della pensione sociale Via Carletti, il mitico e che più roscio non si può *Gatto Al*. Casa che ha visto visi e voci di chi ho amato prima o poi andar via, mi scuserà Claudio per la citazione ma ci stava tutta. Altri volti sono rimasti e fra loro ringrazio con tutto il cuore *Valentina*, *Ester*, *Paolo*, *Silvia, Francesa la Turinés e tutta la brigata Carletti*.

Ho lasciato l'ultima parte dei ringraziamenti agli Amici, allo zoccolo duro, a quelli che mi conoscono da secoli. Agli apocalittici (più riserva): *Marco*, *Damiano*, *Mattia* e *Franco*, amicizie su cui ho sempre potuto contare e su cui so di poter contare, per sempre. Con loro voglio ringraziare *Paola* e la sua presenza sempre preziosa nella mia vita, *Cristiano* e *Fabrizia* perché solo a guardarli viene voglia di amare, le dolcissime *Antonella* e *Laura* due bionde fuori dalle righe, le *due Francesca* per il loro *sorriso*, e la ormai Bolognese di adozione *Benedetta*, e con lei il grande *Petkov*. E poi *Giulia*, voce fuori dal coro, amica di tanti anni e di grande importanza. A *Marco* un ringraziamento doppio per avermi sopportato nel 1997 quando siamo diventati Periti, nel 2003 quando siamo diventati Ingegneri ed ora nel 2007 finalmente Dottori! Più di 12 anni di amicizia e studio insieme!

Quel 2002 è stato importante anche per la conoscenza di un nutrito gruppo di spagnoli che voglio ringraziare anche solo per il ricordo che ho di loro e per il piacere che ho nel vederli anche se solo per un fine settimana l'anno: *Marta Rubia*, *Clara*, *Marta Morena*, *Aida*, *Alfonso*.

Anche se sembrerà anacronistico e strano volevo ringraziare *Mariangela* che molti di voi hanno avuto il piacere di conoscere. Con lei *Daniela* e *Nadia*, tre

donne che mi hanno fatto crescere ed amato ognuna a modo loro. A loro dovete i sorrisi e le lacrime di tutto l'amore che ho speso e che ho da spendere. Grazie!
Anche questo come la laurea è un passaggio importante nella mia vita. Tanto più che dopo questi 5 anni lascerò il mio posto di lavoro, la mia scrivania, le mie chincaglierie. Qui allo IUSM lascio un pezzo di cuore, l'amore per le persone con cui ho lavorato è enorme e mi fa bene pensare di essere stato così fortunato da essermi imbattuto, quasi per sbaglio, in tanta gente quel 25 giugno 2002. A tutto lo IUSM un grazie di cuore, un enorme grazie dal mio enorme cuore. Esco come entro: col sorriso!
A tutti voi compagni di questa avventura il passo di una delle canzoni più belle del Maestro:

"…a te dedico queste parole da poco
che sottendono solo un vizio antico
sperando però che tu non le prenda come un gioco,
tu, ipocrita uditore, mio simile...
mio amico..."

Ciao!

Sommario

Introduzione. Uno degli obiettivi principali della analisi del movimento è fornire informazioni quantitative per la valutazione della abilità locomotoria di un individuo. Nella analisi di un determinato atto locomotorio, il problema generale della valutazione della abilità può essere ricondotto al problema particolare della valutazione della abilità di muovere il proprio corpo da un punto A ad un punto B nello spazio. Tradizionalmente, l'abilità di un soggetto viene valutata confrontando la strategia messa in atto dal soggetto in analisi con un riferimento definito "normale" stabilito osservando una popolazione di controllo. Questo tipo di approccio, tuttavia, non tiene in considerazione l'impossibilità intrinseca per alcuni individui di effettuare movimenti "normali", a dispetto della loro abilità nel compiere un dato atto motorio, vista la presenza di limitazioni che colpiscono il sistema neuro-muscolo-scheletrico. L'ipotesi su cui si basa l'approccio alternativo proposto nella presente tesi è che l'abilità locomotoria di un soggetto sia strettamente legata alla affidabilità della strategia motoria e che possa essere descritta in termini di mantenimento della stabilità e simmetria e semplicità di movimento del corpo durante il cammino.

Obiettivo: Sulla base dell'approccio appena descritto, obiettivo generale della tesi è stato dunque di proporre un metodo per la valutazione della abilità motoria con particolare riferimento a processi deteriorativi. E' largamente riconosciuto che il mantenimento della stabilità durante il cammino viene facilitato da una maggiore coordinazione dei segmenti che costituiscono la parte superiore del corpo e che il naturale processo di invecchiamento si ripercuote in importanti deficit nella suddetta coordinazione. Inoltre, i movimenti della parte superiore del corpo racchiudono importanti elementi direttamente connessi alla simmetria del pattern motorio del cammino, simmetria che si riflette in traiettorie quasi-sinusoidali dei segmenti che la costituiscono. In accordo con queste considerazioni, la abilità locomotoria è stata valutata in termini di affidabilità della strategia motoria adottata da un

soggetto focalizzando l'attenzione nella analisi della meccanica della parte superiore del corpo durante il cammino in piano.

Materiali e metodi. La natura simmetrica e quasi-sinusoidale dei movimenti della parte superiore del corpo suggerisce la possibilità di utilizzare l'analisi armonica per descrivere i movimenti lineari dei segmenti che la costituiscono. Prima di effettuare questa analisi, è stato affrontato un problema metodologico con lo scopo di sviluppare un metodo capace di validare l'ipotesi di periodicità del cammino in piano, ipotesi che può essere considerata valida solo in prima approssimazione poiché, strettamente parlando, il cammino può essere considerato un fenomeno pseudo-periodico. Per questo motivo è stato effettuato uno studio sperimentale al quale hanno preso parte dieci soggetti giovani sani ai quali è stato richiesto di camminare prima su un percorso rettilineo di 20 metri e successivamente su un percorso di 8 metri a tre differenti velocità di progressione auto-selezionate: lenta, normale ed elevata. Sul corpo del soggetto sono stati distribuiti un insieme di marcatori la cui posizione istantanea è stata ricostruita utilizzando un sistema stereofotogrammetrico, mentre gli istanti di contatto piede-suolo sono stati identificati utilizzato un tappeto strumentato. Il movimento della parte superiore (senza arti superiori), della parte inferiore e di tutto il corpo è stato rappresentato utilizzando la cinematica di tre cluster di marcatori. L'aperiodicità del cammino è stata stimata selezionando un insieme di variabili che fossero affini a diversi tipi di energia: cinetica lineare e rotazionale e potenziale gravitazionale ed elastica. Da queste grandezze è stato stimato un indice J(t) che fosse capace di descrivere lo "stato meccanico" della parte del corpo di riferimento (p.e. parte superiore ed inferiore del corpo).

L'analisi armonica delle traiettorie di punti localizzati a livello di testa, spalle e pelvi può essere utilizzata per identificare due pattern del movimento umano: un pattern intrinseco caratterizzato dalla perfetta simmetria rispetto ai piani anatomici e considerato specifico di una popolazione, ed un pattern estrinseco che rappresenta la deviazione rispetto a tale simmetria ed è stato considerato specifico di un singolo individuo o di un sotto gruppo di individui facenti parte di una specifica popolazione. Per verificare tali ipotesi è stata effettuata una sperimentazione alla quale ha preso parte un gruppo di undici donne di età compresa fra 75 ed 85 anni alle quali è stato richiesto di camminare su un percorso ovale della lunghezza di 20m (circuito composto da due rettilinei di 8m) a velocità auto-selezionata confortevole. La sperimentazione è stata effettuata in tre sessioni distanti sei settimane l'una dall'altra al fine di esaminare l'affidabilità del metodo proposto sia a breve che a lungo termine. In particolare, si è verificato che il pattern intrinseco fosse ripetibile sia intra- che inter-soggetto, mentre il pattern estrinseco fosse ripetibile solo intra-soggetto. Infine l'analisi armonica, una volta verificata la sua affidabilità nella determinazione dei due pattern del movimento, è stata utilizzata per valutare l'abilità motoria in 20 soggetti giovani e 20 soggetti anziani sani. L'analisi è stata eseguita non solo studiando le traiettorie dei punti descritti ma anche le loro accelerazioni, al fine di esplorare la possibilità di sostituire il sistema stereofotogrammetrico con una strumentazione più economica e portabile come, ad esempio, i sensori inerziali.

Risultati e discussione. Dall'analisi della pseudo-periodicità emerge che i soggetti giovani hanno un cammino pseudo-periodico e che questa caratteristica è ancor più marcata quando si analizza il movimento della parte superiore del corpo. Inoltre è stato provato che, se i test vengono effettuati in laboratori di analisi del movimento di piccole dimensioni, dovrebbe essere sempre effettuato un controllo di aperiodicità.

L'analisi armonica si è mostrata affidabile per l'identificazione di due pattern del movimento umano in donne anziane sane: un pattern intrinseco, caratteristico dell'intera popolazione investigata. Il pattern estrinseco, caratteristico di un individuo o di un sottoinsieme di individui appartenenti alla stessa popolazione investigata. Il pattern intrinseco ha mostrato una eccellente ripetibilità sia intra- che inter-soggetto (coefficiente di correlazione multipla compreso fra 0.82 e 0.99), mentre il pattern estrinseco ha mostrato una eccellente ripetibilità intra-soggetto (coefficiente di correlazione multipla compreso fra 0.70 e 0.90) ed una minore ripetibilità inter-soggetto ad indicare la specificità individuale del suddetto pattern.

Il confronto fra soggetti giovani e soggetti anziani ha evidenziato le differenti strategie motorie adottate dai due gruppi per controllare la stabilità della parte superiore del corpo. Come atteso, la velocità di cammino era ridotta nei soggetti anziani e come riportato anche in letteratura ad una diminuzione della velocità sono associate minori oscillazioni a livello della pelvi soprattutto in direzione antero-posteriore e medio-laterale. Viceversa, non sono state trovate differenze statisticamente significative fra i due gruppi a livello della testa. Questo risultato indica l'abilità di entrambi i gruppi nel controllare la stabilità della testa durante il cammino. Questa ipotesi è rafforzata dai risultati ottenuti mediante l'analisi delle accelerazioni: in entrambi i gruppi sono state riscontrate minori accelerazioni a livello della testa rispetto a quelle registrate a livello della pelvi. Nel confronto fra i due gruppi, mentre non sono state trovate differenze a livello della testa, i soggetti anziani avevano le accelerazioni a livello della pelvi minori rispetto alle corrispettive registrate nel gruppo dei soggetti giovani.

Conclusioni. I risultati di questa tesi dimostrano l'affidabilità della analisi armonica nel descrivere la cinematica della parte superiore del corpo in soggetti anziani a partire dai movimenti di punti definiti a livello di testa, spalle e pelvi. Tale analisi fornisce elementi per discriminare due pattern del movimento umano: intrinseco ed estrinseco. La determinazione di questi pattern può essere utilizzata per fornire informazioni utili circa la strategia motoria adottata da un soggetto e permette di identificare limitazioni fisiche e di descrivere il loro ruolo nella determinazione di limitazioni funzionali in soggetti anziani.

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CHAPTER 1 Introduction

1.1 Background of the problem

The assessment of gait function can be essential in monitoring the recovery or decline of motor function. In fact, gait function can be considered to be a major determinant for independent motor function, and the amount of walking during the day is indicative for the level of physical activity. Different pathologies can affect gait and balance and a wide range of clinical disciplines involved in treatment of disturbances in balance and gait (Odding et al., 2001). Alterations of the motor capacity, particularly those that reflect in balance and locomotion disturbances, can crucially affect the quality of life of an individual and his/her social cost. In addition to that, they correlate with disability and mortality (Exton-Smith, 1977; Tinetti, 1986; Rubenstein, 1993; Williams, 1998; Williams et al., 2004). Therefore, any intervention aimed at further improving the assessment of an individual's motor ability is considered strategically important in our society (Schultz, 1992).

In the clinical practice, an individual's motor ability is assessed using the so-called performance tests. These tests, that use interviews or direct observation of the patient (referred as semi-quantitative or quantitative) or both (Berg et al., 1992; Reuben et al, 1992; Guralnik et al., 1994; Tinetti et al., 1995; Means, 1996), have exhibited a good operative validity. The advantages of these methods are the low cost, the fact that they are well accepted by the subjects, and the fact that they can be administered in different environments. They can discriminate subjects with from subjects without disturbances; however, they are unable to discriminate subjects located at the edges of the functional spectrum. In addition, they are unable to discriminate conditions in which the use of compensatory strategies allows for the maintenance of a good functional level. They certainly supply insufficient information for the identification of the cause of the functional limitation and their reliability can certainly be improved.

Despite a great deal of research in this field has been carried out, the state of the art of knowledge is generally recognised as disappointing (Bronstein et al., 1996; Tideiksaar, 1997; Masdeu et al., 1997; Rubenstein et al., 2001; Cappozzo, 2002). This state of affairs has been denounced by a "position statement" of the World Health Organisation, which identifies a possible solution to the problems involved in a closer collaboration between doctors and engineers.

This would seem to allude to the use of the movement analysis laboratory. In fact, the biomechanical analysis of suitably selected motor tasks, carried out using stereophotogrammetry, dynamometry, electromyography, indirect calorimetry and multi-segment human body modelling may, in principle, provide relevant, thorough, and objective information. Human movement analysis aims to gather information, for an individual performing a motor task, concerning the movement of the whole-body centre of mass, the relative movement between adjacent bones (joint kinematics), the forces and couples exchanged with the environment (external loads) or between body segments (intersegmental loads), body segment energy variations and muscular work. These analyses require the reconstruction of the instantaneous position and orientation (pose) of systems of axes that are, in principle, embedded in the bones under analysis (bone embedded technical frames), relative to a laboratory frame. To this purpose stereophotogrammetric systems are most commonly used. These allow for the reconstruction of the 3-D position of markers attached to the surface of body segments in each sampled instant of time.

The motor ability of an individual is assessed analysing the above listed variables when the subject is performing a "standardised" motor task, such as raising from a chair, step ascent, running, walking or jumping. In this thesis the attention has been focused on the investigation of the most common and basic locomotor act: level walking (intended as walking along a linear pathway and at constant speed of progression). Level walking can be considered one of the fundamental motor skills performed by humans, where the term "fundamental" suggests that this skill provide the foundation for the learning of other, more specialised, movement skills.

Habitual upright walking is a characteristically human trait that provides a unique set of physiological challenges, and the potential for a loss of balance when performing an apparently simple task is considerable. In particular, the challenge of remaining upright when walking and the ramifications of impaired postural control are particularly evident in presence of pathology of the vestibular system and with advancing age.

The investigation of this motor act plays a crucial role in the assessment of the motor ability of an individual. An impairment in the body function and structure of the locomotor system can heavily affect the daily life of an individual and can lead to limited activities and restricted participations, which might turn into disability.

Falls and instability are among the most serious problems facing the elderly population and the analysis of human movement in general and of the locomotor acts such as walking in particular have an important impact in the analysis of the elderly people daily life. They are major causes of morbidity, mortality, immobility, and premature nursing home placement.

According to a recent report by the Center for Disease Control and Prevention, the number of deaths due to falls for people over the age of 65 rose 55 percent between 1993 and 2003. In 2003 alone, more than 13,000 elderly men and women died from fall-related injuries. A fall also constitutes a marker condition of an underlying reversible medical or psychosocial problem that can be corrected. Several studies reported that approximately one third of older people fall at least once a year, with many suffering multiple falls (Campbell et al., 1981; Prudham and Evans, 1981; Tinetti et al., 1988; Blake et al., 1988; Nevitt et al., 1989; Lord et al., 1994; Luukinen et al., 1995; Menz et al., 2000; Hill et al., 1999; Bergland et al., 2003). In addition, falls are the leading cause of injury-related death and hospitalisation in persons aged 65 years and over (Baker and Harvey, 1985; Menz et al., 2000), resulting in disability, restriction of activity and fear of falling - all of which reduce quality of life and independence (Tinetti et al., 1994; Hill et al., 1996; Maki et al., 1991; Menz et al., 2000). The most locomotor act related to falls is walking.

Between 50% and 70% of falls occur during walking (Cali and Kiel 1995; Berg et al., 1997; Norton et al., 1997; Mbourou et al., 2003). One of the objectives of human movement analysis is to underlying the mechanisms of this failure of the postural control system when walking. View in this context, irrespective of the strength of the associations between tests of standing balance and walking ability, it is clear that experimental methods must be specifically developed to provide accurate and valid information on the specific tasks being investigated.

1.2 Assessment of an individual's motor ability

The importance to assess the motor ability of an individual is one of the primarily objectives of human locomotion. In order to evaluate the motor ability of an individual a set of physical variables has to be either measured or assessed during the execution of a selected motor task. In the present thesis this evaluation can be done through the analysis of the motor strategy adopted by the subject to progress from a point A to a point B in the space.

1.2.1 Motor strategy

A motor act is accomplished through the summation of musculo-skeletal functions, under the control of the central nervous system. The way these functions are combined in terms of timing and relative weighting, together with the objectives that are pursued through their combination, pertains to the motor strategy. A strategy is chosen by a given individual among those that are consistent with the structural and functional constraints of his/her neuro-musculo-skeletal system, and that tend to maximise the effectiveness of the motor act in terms of maintenance of balance, mechanical load generated by or acting on tissues, and energy expenditure (Cappozzo, 1983). Thus, the motor strategy adopted is determined by the motor functional status of the subject.

1.2.2 Classical approach for motor ability assessment

Traditionally, in gait analysis the motor strategy adopted by a subject is observed by comparing it to a normal reference established using a control population. However, the overall strategy of the movement does not take into consideration the objective impossibility for the patient to achieve a "normal" motion trajectories since the patient's neuro-muscular-skeletal system is, in some cases, irreparably impaired. Nevertheless, the patient can accomplish a high level of ability in walking relative to his/her locomotor system. However, the quality of the gait performed by an individual depends not only on the functional and structural constraints imposed by the subject's locomotor system and, given this locomotor system, but also on the subject ability to put in to action an affective motor strategy. Thus, in the present thesis an alternative approach is proposed.

1.2.3 An alternative approach for motor ability assessment

An individual, potentially, can execute the same motor task with a different strategy according to circumstances. However, it has also been mentioned that the motor strategy is chosen compatibly with the constraints imposed by the neuro-musculoskeletal system. These constraints depend, in general terms, by available muscle strength, range of motion of the joint, and ability to maintain balance. If an impairment and/or any structural and/or functional decay in the neuro-musculo-skeletal system occur, then such constraints become stricter, i.e. the variety of motor strategies available to the subject diminishes, and, possibly, no choice at all may be allowed. The availability or not of different motor strategies to the subject pertains to the so-called "physical functional reserve" (Pendergast et al., 1993).

Level walking evaluation should be approached in terms of both assessment of the resources that the subject employs to accomplish the locomotor act and the quality to perform the locomotor act. View in this context, the assessment of the motor ability of an individual can be directly related to the reliability of the locomotor act and to its consistency with relevant aesthetic canons. In this context the word "reliability" is used in the engineering sense, that is, as the probability that the function be performed satisfactorily under given circumstances. The proposed approach is based on the assumption that the more the subject is able to successfully complete the motor task, the more the motor task is reliable. Thus, the problem of the evaluation of the motor ability of an individual can be assessed through the quantitative analysis of the reliability of the locomotor act. To tackle the problem of assessing the reliability of a locomotor act it has been related, in the case of level walking, to the following aspects (Cappozzo, 1984): (1) symmetry and simplicity of the movement, and (2) the maintenance of stability. The first aspect pertains the aesthetics of walking, the second to the reliability of the locomotor act.

The concept of stability (the state or quality of being stable) has been defined in a different ways in physics, bioengineering and physiological literature. The Oxford dictionary defines "stable" as "firmly fixed or established, not easily to be moved or changed". This definition can be easily applied to standing, as when subjects are requested to stand still, the aim is to keep their velocity equal to zero. However, this definition is not useful when considering dynamic tasks such as walking. As walking inherently requires displacement of the body, velocity of the body's centre of mass cannot be equal to zero. However, if we request a subject to walk at a constant velocity, his/her body's centre of mass acceleration will be (hypothetically) equal to zero. Thus, walking instability can be defined as the amount by which acceleration is not zero, and therefore with the smoothness of the displacement trajectories of the body centre of mass.

In summary, the approach proposed in this thesis aims to evaluate the individual's motor ability analysing the reliability of the locomotor act performing by the subject. Since the word reliability is difficult to evaluate the analysis of a locomotor act can be studied observing the motor strategy adopted by the subject to progress toward destination in terms of both the assessment of the resources employed to accomplish the locomotor act and the quality of the result.

1.3 Target population(s)

1.3.1 Rationale

The experimental and analytical methods developed in the framework of the present thesis are intended for being used especially in a preventive medicine context, and in particular for the assessment of motor functional status of elderly individuals. The reasons for such a choice will be apparent from the following sections, which evidence the clinical and social impact of functional assessment in the elderly.

1.3.2 The reduction of physical functional reserve in the elderly

Ageing is characterised by the development of concurrent pathologies that may have a synergetic effect (Guralnik et al., 1994). In particular, the ageing process is accompanied by a natural and progressive decay of the systems responsible for motor, sensory and/or cognitive abilities, which determines a progressive reduction of the physical functional reserve. Therefore, on the basis of the considerations reported previously, functional assessment plays a relevant role for the elderly. On this ground, Kane (1993) wrote, «The operative word in geriatrics is function».

For a better understanding of the mechanisms responsible for the aforementioned functional reserve reduction, the most relevant causes associated with it will be briefly analysed in the following.

According to the study of Wollacott (1993), ageing is accompanied by:

- alterations in the organisation of muscular response synergies;
- a reduction in muscular strength and tone, also related to skeletal muscle mass loss;

- increasing frailty of the skeletal system;
- alterations in the sensory systems which contribute to balance (somato-sensory, vestibular and vision);
- a reduction of the adaptation capacity (e.g. adaptation of the postural response to a required change in the motor task); and
- a reduction of the capacity of integrating postural and voluntary activity.

The issue of muscular strength decline in the elderly has been addressed in a considerable number of studies. In particular, it was shown that a decrease in voluntary strength becomes apparent since the age of 60 (see the review by Porter et al., 1995). In fact, strength decline from 25 to 65 years old was found to be about 30% (Schultz, 1995). Elderly in the age range 70-80 were found to score, on average, 20 to 40% less than young adults, and even greater reductions (50% or more) were found for the very old ones. This strength reduction was found to depend from alterations in the nervous system and in the motor unit properties, such as changes in size, number and proportions of muscle fibre types. The above differences between young and elderly groups of adults were consistently less for the eccentric type of muscle action (lengthening) than during either isometric or concentric contractions, suggesting that, for a given absolute load, the intensity of muscular effort in the elderly would be less for an eccentric condition than for a concentric one. According to this, the elderly may find some relative advantage in using muscles under lengthening situations.

Significant variations in muscle histology, morphology and performance and a reduction in the conduction speed of peripheral nerves were also shown (e.g. Larsson et al., 1978). However, whether and how such morphological changes affect functional activity has not been fully studied yet. Furthermore, an age decline in the capacity of rapidly developing ankle joint couples was evidenced (Thelen et al., 1996).

A still open issue about the above muscular strength decline associated with age is the extent to which it can cause motor disability. In fact, evidence was presented that joint couples needed to carry out Activities of Daily Living (ADL) do not limit, in general, the ability to execute such activities, as these couples are often not large (Schultz, 1992). For example, it was shown that, in rising from a chair, smaller couples than the maximal available, also when rising from low seats, are required (Alexander et al., 1997). Accordingly, it was suggested that older adults may not necessarily use motor strategies that minimise couple requirements (Schultz, 1992).

From all the above considerations, it can be inferred that while the decline in strength may well explain some kind of functional limitations in the elderly, disabilities in executing ADL should be attributed to other problems. On this ground, reduced range of joint motion (e.g. Walker et al., 1985) and disturbances in balance control mechanisms may play an important role. As already mentioned above, such balance problems are due to age-related alterations in the relevant sensory systems. They entail a reduced ability to control the variations in height of the centre of mass of the human body and of its single segments. To evaluate the importance of these alterations, it should be kept in mind that two main processes (Wollacott, 1993) represent the basis for motor independence, i.e. balance control and the ability of integrating balance adjustments in the context of on-going voluntary movements (e.g. taking an object, lifting a weight or walking).

1.3.3 Social impact of studies dealing with functional assessment in the elderly

Distinguishing the “physiological” ageing process illustrated in the previous section from disease is possible through the function–oriented approach described previously and reported by Schultz (1992). Elderly individuals for whom this “physiological” ageing process does not imply invalidating disturbances or abnormalities of the neuro-musculo-skeletal system, and who do not suffer from other forms of disability and/or pathology, can be defined as “healthy”. In Italy and in many other countries, healthy elderly represent over 20% of the population. These elderly, although healthy in the aforementioned sense, are at risk of disablement. As an example of that, the results of a study examining the functional status of non-institutionalised persons over 65 in the United States (Leon and Lair, 1990, through Schultz, 1995) can be considered. Of the 27.9 million elderly persons in the Unites States, 12.9% were found to have difficulties with at least one of a number of ADL-related motor tasks analysed. In particular, 8.9% had difficulties in bathing, 7.7% in walking and 5.9% in bed or chair transfers. The rate of difficulty increased progressively over 65, climbing sharply in the 80-ies.

As a further example, consider that a relevant percentage of the healthy elderly individuals (at least 30% in Italy, and comparable data are reported in other countries) incur every year in falls with physically and/or psychologically disabling effects. Falls occur, in most of the cases, in the domestic environment, where it may be assumed that “tricky” structures have been avoided. This latter circumstance suggests the hypothesis that such falls are to be ascribed to factors intrinsic in the physiological and/or pathological status of the subject involved (Tinetti et al., 1995). The rehabilitation programme which should follow such falls entails an important economic burden for the National Institutes of Health.

1.4 Objectives of this thesis

The main objective of this thesis, as stressed in the previous paragraphs of this chapter, is the assessment of the motor ability of elderly individual through the analysis of walking on a level surface. For the purpose of this analysis, human locomotion can be seen as the lower limbs that transport the upper part of the body. View in this context the human body can be divided in to two subsystems, the lower part of the body, which can be related to the lower limb movements, and the upper part of the body, which can be related to the head, trunk, and upper limbs movements. The pelvis, moreover, can be seen as the platform which sustains the upper body, and which, during the locomotion, ensures the upper body the needed stability.

The attention of researchers has mainly focused on the study of the lower limb motion. However, there are a number of complementary investigations involving detailed analyses of head and trunk movements (Cappozzo, 1981; Hirasaki et al., 1999; Nadeau et al., 2003; Mulavara et al., 2002; Cromwell 2003), which provide information for the assessment of ontogenetic (Assaiante and Amblard 1993; Bril and Ledebt 1998; Baumberger et al., 2004; Gill et al., 2001), pathological (Pozzo et al., 1991; Hirasaki et al., 1993; Mamoto et al., 2002), and deteriorative phenomena (Winter, 1991; Pozzo et al., 1991; Gill et al., 2001; Hirasaki et al., 1993; McGibbon and Krebs, 2001; Menz et al., 2003; Paquette et al., 2006). This part of the literature review is deeply reported in Chapter 2.

Keeping in mind the assumptions that the motor ability of an individual can be assessed by observing the level walking motor strategy, and that this can be obtained by considering symmetry and simplicity of the movement and the maintenance of dynamic stability, a further hypothesis in this thesis is that this information cam be obtained by looking only at the upper part of the body movements.

As previously seen the biomechanical analysis, carried out using a complex instrumentations and experimental protocols might be awkward to be used in clinical practice. To overcome this obstacle, the mechanics of the upper part of the body can be investigated using few and simple measures in association with a mechanical model with the least possible number of degree of freedom. In this thesis the upper body will be modelled using a three-segment model, the movement of which is obtained from the movement of three points located approximately at the head, upper trunk, and pelvis levels. Moreover, the linear displacement of the above mentioned points, will be represented in the frequency domain, as suggested by the cyclic nature of human walking. This analysis is expected to allow to identify two superimposed movement patterns: an intrinsic pattern characteristic of a specific population, and an extrinsic pattern characteristic of an individual or of a sub-group of individuals in a given population (Cappozzo, 1981; Pecoraro et al., 2004) The detection of these patterns, finally, is expected to provide information about the motor strategy adopted by a subject and to allow for the identification of physical impairments and for the description of their role in determining mobility limitations.

To verify the above assumptions, short- and long- term reliability of the harmonic analysis to identify both the intrinsic and the extrinsic patterns in healthy women aged 75-85 will be investigated. In particular, the hypothesis that will be tested are that, for this selected population sample, the intrinsic pattern was both intra- and inter-subject repeatable, while the extrinsic pattern was only intra-subject repeatable.

A methodological issue, preliminary to the harmonic analysis, has also been dealt with: the test of the validity of the hypothesis of periodicity of gait. Since human walking is a biological phenomenon, in fact, the assumption that gait data pattern is periodic is simplistic, and, strictly speaking, it can only approximate periodicity and it may be referred to as pseudo-periodic (Stokes et al., 1999). This pattern emerges from gait initiation and is ultimately clearly identifiable only during steady-state pace. Since such pace is reached after negotiating some steps (Miller and Verstraete, 1999), short walking pathways, as found in many gait laboratories, might be one of the causes that contribute to the pseudo-periodicity in the recorded data. For this reason, in this thesis particular attention will be focused on the analysis of gait periodicity, and a method will be proposed for the assessment of the gait pseudo-periodicity. The information provided by this method is expected to be useful for two reasons: from a heuristic point of view, it allows an insight into a possible methodological, external, cause of the variability of gait strides (Chau et al., 2005), and, from a practical standpoint, it allows for a control of the consistency of the observed gait stride with the hypothesis of steady state.

CHAPTER 2 State of the art

2.1 The control of stability during level walking

Activities of daily life require an individual to move in his/her environment. Locomotion is an unique feature of the animal kingdom. According to the environment in which they live, animals use different means of propulsion: terrestrial mammals move using legs. Locomotion allows individuals to meet others, to find better food and a better climate, to pursue a prey or to escape an impeding danger. For most humans, the typical mode of locomotion is to walk from one place to another place. The fact that the gait of humans are bipedal and that an individual progress for the major period with one foot (during walking), no feet (during running) or two feet in contact (during standing position) with the ground creates a major challenge to their balance control system. Human walking can be divided in to three main phases:

- The initiation of the whole body movements. This phase has been defined by Nissan and Whittle (1990) as following: "When a subject changing his/her mechanical condition from standing at rest to the cyclic movement of walking".
- The progression forward at steady state pace. This phase is reached after a certain number of steps and is identified when the subject walks at constant mean speed of progression.
- The termination of the whole body movement. This phase can be considered the mirror image of the gait initiation in terms of centre of mass and centre of gravity trajectories of the subject (Winter, 1995). The subject starting from a steady state pace commence to decelerate till reaching the standing position.

These three phases are characterised by a continuous movement of the lower limbs generated by the individual to transport the upper part of the body and, thus, to progress toward a target destination. During these movements, the first goal of the subject is to maintain the stability in order firstly to avoid falls. In addition, the individual performing the locomotor act has to be able to meet any changes in the demands of the environment such as obstacles or steps.

Walking is a natural way of moving. For this reason the act of walking seems to be a simple locomotor act, however, maintaining stability during progression is a particularly challenging task for the human postural control system. To maintain stability during a dynamic task the first objective of the postural control system is to keep the centre of gravity of the body safely within the base of support. However, although this principle can be easily applied to standing, it is conceptually difficult to apply it during dynamical locomotor act, such as walking. This is due to the fact that the projection of the centre of gravity is outside the area of the base of support for a large proportion of the walking cycle. This area is provided by the foot (or the feet) which are in contact with the ground at the particular instant of time. There are three main reasons that make maintenance of stability a particularly challenging task for the human postural control system:

1. Two-thirds of the body mass are located at two-thirds of body height above the ground.
2. For a major period of the walking cycle the body is supported by a single limb.
3. During the gait cycle the vertical projection of the centre of mass passes outside the base of support provided by the foot in contact with the ground (Winter, 1995).

Starting from the first phase previously defined, the subject initiates to walk. At the beginning the subject is in standing position and therefore has both feet in contact with the ground. During this phase the projection on the ground of the body centre of gravity is contained within the base of support. Thus, the maintaining of stability is simple achieved. When the subject starts to walk and when one of the two feet leaves the ground the centre of gravity of the body is made to fall forward outside of the base of support. In this period of time the base of support is provide by the only foot which is in contact with the ground (Winter, 1991). For this reason gait initiation is an unstabilising event since in order to progress forward the individual initiate a loss of balance and commence a forward fall, and it is only by the correct placement of the controlateral foot that a complete loss of balance is avoided. Gait initiation has received only limited attention in literature, however, it is widely accepted that many falls in older people happen when the subjects walk in particular along short distances (Ashley et al., 1977).

The first phase ends after a certain number of steps, when the subject achieves the self selected comfortable cadence and reaches the steady state pace (steady state is reached when the mean speed of progression is constant). There is a little agreement as the number of steps necessary to reach steady state gait, and this might be due to the fact that different researchers based their conclusion on different biomechanical parameters (Miller and Verstraete, 1999; Forner-Cordero et al., 2006). However, there is an agreement about the fact that the steady state is reached no sooner than two consecutive steps.

When the steady state phase is achieved, the only period of time in which the centre of gravity passes closest to the base of support is when it passes forward along the medial border of the foot. Even during the 10% double support stance periods, body feet are not flat on the ground. Thus, looking at these considerations, during the stride period the body is in an inherent state of instability (Winter, 1990).

The challenge of remaining upright when walking and the ramifications of impaired postural control are particularly evident in subjects with loss in peripheral sensory and in with advancing age. People with diabetic peripheral neuropathy commonly report balance difficulties when walking (Cavanagh et al., 1992) and have a markedly increased risk of experiencing falls (Richardson et al., 1992). Similarly, community studies reveal that approximately one third of older people fall at least once a year, with many suffering multiple falls (Campbell et al., 1981; Prudham and Evans 1981; Tinetti et al., 1988; Blake et al., 1988; Nevitt et al., 1989; Lord et al., 1994; Luukinen et al., 1995). Falls are the leading cause of injury-related death and hospitalisation in persons aged 65 years and over (Baker and Harvey 1985), resulting in disability, restriction of activity and fear of falling - all of which reduce quality of life and independence (Maki et al., 1991; Tinetti et al., 1994; Hill et al., 1996).

Numerous studies have addressed the changes in the gait pattern of the elderly compared with those of the younger adult. The majority of these studies have concentrated on basic outcome measures such as the temporo-spatial parameters (i.e. stride length, stride frequency, and walking speed) and the variability of those measures. Usually in literature changes in these parameters are related to fall, mobility, and post-fall anxiety, suggesting that a degeneration of balance control is combined with a general loss of muscle strength. The maintenance of equilibrium is extensively studied also by observing the lower limb strategies adopted to progress forward and to sustain the upper part of the body. Conversely, the influence of the upper part of the body for the analysis of the stability during dynamical and postural tasks has not been studied much or at least not recently. However, several studies reported that the analysis of the oscillations of the upper body along the medio-lateral and the antero-posterior directions may provide information about how the nervous system deals with an increase in the level of equilibrium difficulty. In particular the stabilisation of the head is referred as a natural frame of reference since it contains the two most important perceptual system for detection of self-motion relative to space: the visual and the vestibular system.

In fact, it has been demonstrated that both visual (Hollands and Marple Horvat 2001) and somatosensory (Zehr and Stein 1999) information influences the control of the postural stability during dynamic tasks, such as walking. However, several studies have reported that in absence of vision the vestibular system plays an important role for maintaining stability when the subject perform various locomotor tasks in a previously seen target trials (Fitzpatrick et al. 1999; Jahn et al. 2000). This finding suggests that vestibular system plays an important role since its information contributes to maintain the head stable during dynamic tasks and to enable successful gaze control as reported by Pozzo and colleagues (1990) in a young healthy subject population. In addition the vestibular system can be used to generate a general postural responses that provide a stable reference frame (Pozzo et al., 1995).

The importance of the vestibular system has also been reported in studies of chronic vestibular-deficient patients. In literature several studies reported that the vestibular system plays an important role in the successful completion of locomotor activities. However, although the subjects with bilateral vestibular deficits tend to walk slower and to change direction earlier than healthy, they are able to walk successfully over a short distance without vision (Cohen 2000; Glasauer et al. 1994; Tucker et al. 1998).

The results reported in literature (Lee et al., 1975; Berthoz et al., 1975; Lestienne et al., 1977; Patla, 1991) shown that the importance of the visual input is required when an individual is walking in a challenging environment and it is particularly important when negotiating discrete obstacles placed along the walking path. However, the role of vision becomes less important when a subject walk in optimal performance in which the absence of vision is replaced by the other sensory inputs, such as vestibular system, which provide sufficient information to successfully walk from one location to another.

Finally, the input provided by the peripheral sensory system contributes has been reported to be particularly important when subjects perform more dynamic tasks. In particular, when walking, the nervous system relies on peripheral sensory input to develop and continually update an egocentric reference frame by detecting the relative positions and displacements of the body, particularly the legs and feet. When peripheral sensation is impaired, this reference frame is incomplete and may cause the subject to be less able to control body stability when walking. Alternatively, the individual may compensate for this impairment by actively seeking out additional sources of sensory information (Spaulding et al., 1994) or altering the mechanics of their walking pattern (Beggs, 1991; Spaulding et al., 1995). Therefore, the biomechanical analysis of the upper part of the body movements during walking can be used as an index for the evaluation of dynamic stability of an individual and for the assessment of the maintenance of the equilibrium.

2.2 Temporo-spatial parameters as descriptors of walking stability

A single gait cycle is called stride and starts from the contact of the foot under investigation and ends at the contact of the same foot. It can be divided over two steps: the instant of time in which the contro-lateral foot touch the ground (with respect with the limb under investigation) identifies both the end of the first step and the beginning of the second step. An example of a gait cycle is depicted in Figure 2.1.

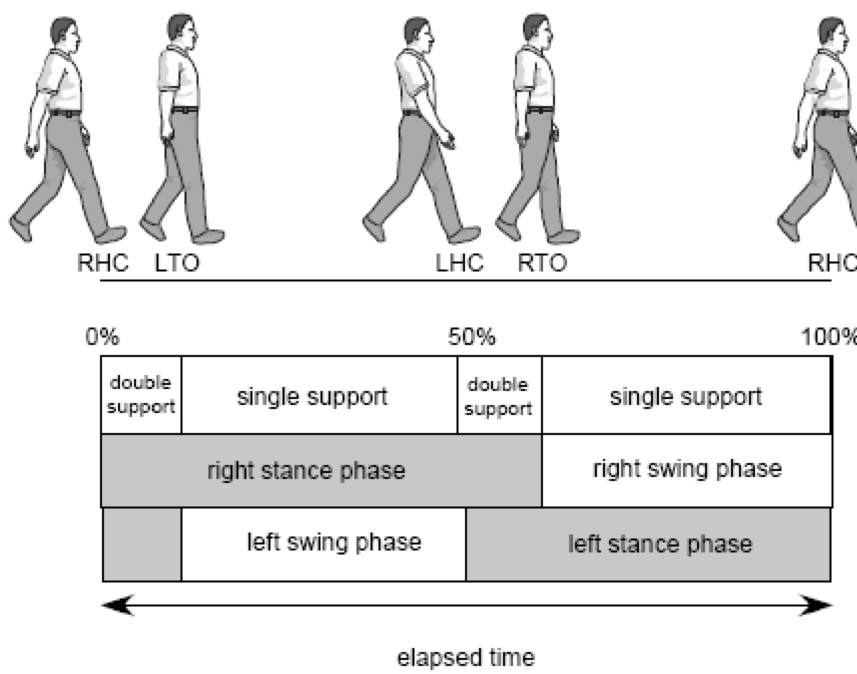

Figure 2.1 The gait cycle. RHC: right heel contact, LHC: left heel contact, RTO: right toe-off, LTO: left toe-off (from Menz 2003).

Each gait cycle can be divided in two phases depending if the limb under investigation is in contact or not with the ground. These phases are called respectively stance phase and swing phase. During the swing phase the limb under investigation is progressing forward in anticipation of the next foot contact, whereas during the stance phase the limb under investigation is in contact with the ground. Moreover, stance phase can be divided in to double and single limb support phases, depending to the number of feet in contact with the ground at that particular instant of time (Inman, 1966; Winter, 1991).

From these definitions a certain number of basic temporo-spatial parameters of gait can be obtained according to the time taken and the distance covered during a single gait cycle. The attention in literature has generally been focused on the following parameters:

- stride/step period which is the time taken to perform a complete stride/step;
- stride/step length which is the distance covered by the subject during a stride/step period;
- stride/step frequency which is the number of stride/step taken per second.

From these parameters it is possible to calculate the mean walking speed of progression which can be computed as the product of stride length and stride frequency. These parameters can be assessed using a relatively unsophisticated instruments. For example in their study Sekiya and colleagues (1997) proposed to use a long piece of paper with ink-satured material attached to the toe and heel of the subjects' shoes.

Many authors have suggested that the temporo-spatial parameters of gait can be used as a global measures of the success of a walking task, and therefore they can be used as an indicator of walking stability (Andriacchi, 1977; Imms et al., 1981; Friedman et al., 1988; Wolfson et al., 1990). Several studies have proposed the temporo-spatial parameters of gait and in particular walking velocity as an index to distinguish normal and pathological gait and to compare different populations such as young and elderly healthy and frail subjects (Tang and Woollacott, 1996; Cappozzo, 1983; Zijlstra et al. 2004). This approach is based on the assumption that the more stable an individual is, the faster he/she can walk. Some authors reported that a slower gait occurs in patients with neurological disorders affecting the central (Alexander, 1996; Thaut et al., 1999) or peripheral (Dingwell and Cavanagh, 2001) nervous system, muscular (Lohmann Siegel et al., 2004) or orthopaedic (Powers et al., 2006) disorders, and with normal aging (Alexander, 1996; Anstey et al., 1996).

Focusing the attention on the age-related changes the most reported and consistent finding of these studies is that older people walk more slowly than young people (Lord et al., 1996; Imms and Edholm, 1981; Murray et al., 1964; Murray et al., 1969; Finley et al., 1969; Cunningham et al., 1982; O'Brien et al., 1983; Hagemon et al., 1986; Elble et al., 1991; Dobbs et al., 1992; Dobbs et al., 1993; Oberg et al., 1993; Fransen et al., 1994; Buchner et al., 1996; Lajoie et al., 1996; Bohannon et al., 1997). This has been found to be a function of both a shorter step length (Lord et al., 1996; Murray et al., 1964; Murray et al., 1969; Finley et al., 1969; Hagemon et al., 1986; Elble et al., 1991; Oberg et al., 1993; Fransen et al., 1994; Lajoie et al., 1996; Crowinshield et al., 1978; Winter et al., 1990; Ferrandez et al., 1990) and increased time spent in double limb support (Lord et al., 1996; Murray et al., 1969; Finley et al., 1969; Winter et al., 1990; Ferrandez et al., 1990). These reported findings revealed that the elderly subjects walking speed is associated with a decreasing in the lower limb joint mobility (Olney et al., 1994; Judge et al., 1996; Kerrigan et al., 2001) and, in general, the slower speed adopted by elderly subjects with respect young subjects can be associated to a limiting impairments that avoid them to walk at fast speeds. In addition, the same findings can be observed comparing elderly with a history of falls who walk slower than elderly without risk of fall (Kerrigan et al., 2000). Thus, if it is assumed that older people and people with risk of falls are inherently less stable than younger people, it would seem that the measurement of temporo-spatial parameters of gait would provide a useful indicator of walking stability.

However, when comparing elderly people with and without risk of fall some authors found that slow gait predicts future falls (Luukinen et al., 1995; Bergland et al., 2003), whereas others studies found that it does not (Lord et al., 1996; Maki, 1997; Hausdorff et al., 2001a; Hausdorff et al., 2001b). This kind of contradiction is also detected when comparing young subjects with elderly subjects. In fact, when walking speed is constrained for example using a treadmill, some authors found that age-related biomechanical differences persisted (DeVita and Hortobagyi, 2000; Kerrigan et al., 2000), whereas others authors found that there are no differences in terms of walking speed in deteriorative phenomena (Alexander, 1996; Kerrigan et al., 1998).

In literature, the reduction of walking speed and step length, and the increase of the duration of the double limb support phase have generally been thought to indicate the adoption of a more conservative, or less destabilising gait (Woollacott and Tang, 1997; Prakash and Stern 1973; Sudarsky and Ronthal 1992). If it is assumed that older people are inherently less stable than younger people, it would seem that the measurement of temporo-spatial parameters of gait would provide a useful indicator of walking stability. However, there is a fundamental paradox in the way in which these age-related differences have been interpreted. Reduction in walking speed and step length, and increased duration of the double limb support phase of walking have generally been thought to indicate the adoption of a more conservative, or less destabilising gait (Woollacott and Tang, 1997; Prakash and Stern 1973; Sudarsky and Ronthal 1992). Thus, on the one hand, temporo-spatial changes observed in older subjects are thought to indicate diminished stability, while on the other hand these changes are thought to be strategies to improve stability.

The alternative interpretation looking at the basic temporo-spatial parameters is that elderly subjects and patients with gait disabilities walk at slow speed because this characteristic reflects a “safer” or “more cautious” gait strategy (Winter et al., 1990; Courtemanche et al., 1996; DeVita and Hortobagyi, 2000). The same pattern can be observed when young healthy subjects walk naturally slow down during walking on slippery surfaces like ice.

Elderly subjects slow down also when their stability is challenged, such as when negotiating obstacles (Hahn and Chou, 2004) or walking over irregular terrain (Menz et al., 2003; Richardson et al., 2004). In fact, elderly people who walk faster may be more likely to trip and fall than those who walk more slowly (Pavol et al., 1999 and Pavol et al., 2001).

Human walking is not only a progressive movement. Whereas, the vast majority of research on human gait has focused on movements in the antero-posterior direction (Winter, 1991), movements in the medio-lateral direction have a particular importance for maintaining balance control (Krebs et al., 2002; MacKinnon and Winter 1993; Patla et al., 1999; Thorstensson et al., 1984). In terms of temporo-spatial parameters of gait another parameter can be introduced, the step width. This parameter is obtained by measuring the medio-lateral distance between the two feet when both of them are in contact with the ground (i.e. at the foot strike instant of time), and is important in determining the base of support covered by the feet in contact with the ground at the beginning and at the end of a gait cycle. Several investigators have suggested that the step width can be a measure of balance control during gait (Blanke and Hageman, 1989; Maki, 1997). A major finding reported in the literature is that if a subject walks with a narrow step width the centre of gravity has a greater tendency to be displaced lateral to the stance foot, leading to an increased risk of loss of balance in the lateral direction (Speers et al., 1998). Therefore, it appears that the placement of the feet in the medio-lateral plane direction is one of the primary determinants of the frontal plane stability of the body during walking (Townsend 1985; MacKinnon and Winter, 1993; Bauby and Kuo, 2000). It has been suggested that increasing step width may be an attempt to improve walking stability, particularly in older people with weak hip abductor muscles (Rigas, 1984). In addition, increasing step width will provide a larger area for the centre of gravity to travel medial to the foot when weight is transferred from one limb to the other, thereby decreasing the likelihood of loss of balance in the lateral direction.

However, the paradox evident in the measurement of walking stability arises again when evaluating stride width. While older people walk with a larger step width than younger people, and increased step width is a predictor of falls (Gehlsen and Whaley, 1990), this gait alteration is also thought to improve stability (Murray et al., 1969; Finley et al., 1969; Sudarsky, 1990). Recently, it has also been suggested that increasing stride width is an indicator of fear of imbalance in older people, and may actually be associated with an increased risk of falling. Until when this paradox is not resolved, it would appear that assessing stride width in isolation does not provide a useful measure of walking stability.

In conclusion, the basic temporo-spatial parameters have generally used to indicate the more conservative gait adopted by the elderly subjects to walk (Winter, 1991; Menz et al., 2003). This assumption can be seen from two different point of view: it could be assumed that a reduction in walking speed in elderly adult has related with poor walking stability, because of their balance control mechanism does not enable them to progress forwards at a normal pace. Conversely, it could be argue that a slowly walking older adult is stable, as they do not place excessive demands on their diminished postural control system. Therefore, it is evident that while temporo-spatial parameters of gait provide a very limited picture of the stability of the body during the performance of a locomotor act.

2.3 The basic mechanics of the human walking

Cappozzo in his thesis defined locomotion as "the action with which the entire bulk of the animal's body moves through aerial, aquatic and terrestrial space". For an individual the action to move the body from a point A to a point B through the space is achieved by coordinated movements of the body segments employing an interplay of internal and external forces. Being the internal forces the muscular forces whereas being the external forces the inertial, gravitational, and frictional forces.

Since the XVII century the human locomotion has been analysed using the Galileo's methods by Giovanni Alfonso Borelli. His most famous opera is called "De Motu Animalium" (On the Movement of Animals, Rome 1680-1681) in which he sought to explain the movements of the animal body on mechanical principles. For this opera he got the crown of the father of biomechanics. Borelli explained the movement of the human body as following: “when man stands he is certain that his legs, resting on the floor, form the isosceles triangle ABC (Figure 2.2), and, at the same time, nature causes and produces many circular movements from which walking derives.”. Thus, as Borelli reported, the progression during human locomotion is possible due to the alternatively rotations of the lower limbs (in Figure 2.2 depicted as BA and CA) which cause a movement of the whole machine R which push towards K.

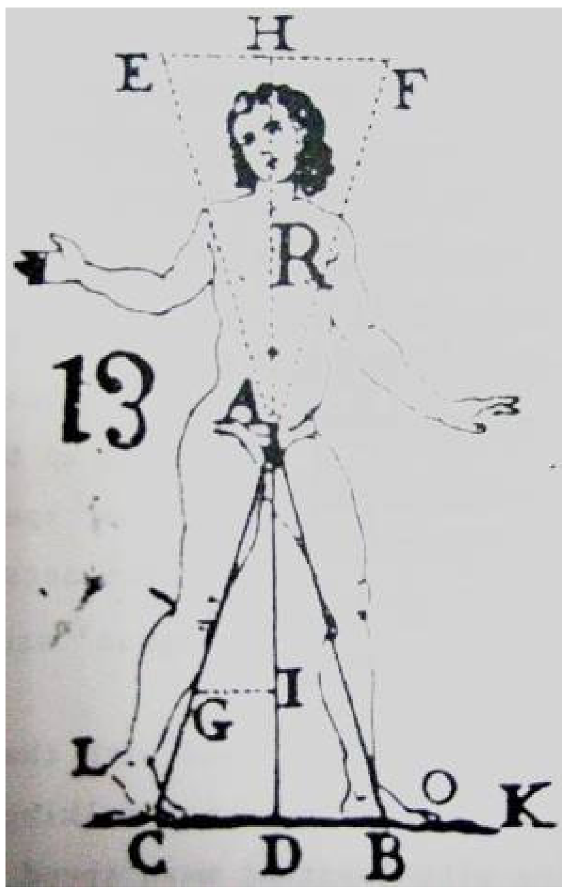

Figure 2.2 The human body depicted by Borelli and reported in the *“Du motu animalum”* (1680-1681).

For the purposes of the human movement analysis, the body can be effectively divided in to two sub-system: the lower limbs and the upper part of the body (head, trunk, pelvis and arms). The former transport the latter. The lower limbs, which can be considered the *transporters*, move rhythmically inducing oscillations of the upper part of the body (Berthoz and Pozzo, 1994; Grossman et al., 1988; Winter, 1991), which can be considered the *transported*.

As reported by Saunders and colleagues (1953) the rhythmic pattern of motion of the lower limbs is such to minimise the mechanical energy exchange between the lower part and the upper part of the body. However, as emphasised by Cappozzo and colleagues (1978), the movements of the trunk and the head also contribute to the reduction of this energy exchange. As reported in that study the relative movements of these segments in a coordinated fashion with the pelvis minimise the energy variation. In fact whether the head and the trunk moved rigidly with respect to the pelvis the magnitude of the energy variation increased. The relevance of this observation is more evident at high speed of progressions. This behaviour is related to the necessity of protecting the system contained in the head from excessive mechanical stimulus. For example a reduction of the accelerations at the head level obviously correspond to a reduction of the accelerations the vestibular apparatus was exposed to. The importance derived by this attenuation will be explained in the next section of this paragraph. As a matter of fact, although successful locomotion is deceivingly simple-looking, it is generally accepted that the maintenance of postural stability requires a complex coordination of the whole body, including information from of visual, vestibular, and somatosensory inputs, such as from skin and muscle, that are integrated by the central nervous system to produce an appropriate motor response (Patla et al., 1992). Even if it is proved that those systems play an important role in maintaining stability and in avoiding loss of balance, most of the studies which investigate locomotion are focused on the lower part of the body whereas, the influence of the upper part of the body to the locomotor strategies has not been studied much or at least not recently.

For an insight about the role of the lower body and the upper body in maintaining stability during walking this section provides an overview of the literature and it is divided in to three main sections. The first section provides an overview of the role of the lower body to sustain and to progress forward the whole body. In particular the six determinants of gait proposed by Saunders and colleagues (1953) will be explained. The second section provides a comprehensive literature review of the investigations which have involved a detailed analyses of upper part of the body during locomotion. Finally, the third section describes the age-related changes focusing the attention on the upper part of the body during walking.

2.3.1 Mechanics of the lower part of the body

In this section a brief analysis of the lower part of the body will be reported. In fact, even if this thesis focused the attention on the mechanics of the upper part of the body, it is interested to analyse how the circular movement of the lower limbs influences the oscillations of the head and trunk. Thus, this paragraph focuses the attention primarily on the study published by Saunders, Inman and Eberhart (1953) more than 50 years ago and well reprise and elucidate by Della Croce and colleagues (2001).

Saunders, Inman and Eberhart (1953) defined six major determinants of gait as functions occurring during normal gait which served to minimise and smooth the vertical and lateral displacement of the centre of mass. They introduce firstly the most simple case of a bipedal system which is called “compass gait”. This “virtual” model is characterise only by the flexion and extension of the hip joint. The lower extremities are represented by rigid levers without foot, ankle, and knee mechanisms. In addition there is no intermediary pelvis. For these reasons, along the sagittal plane the centre of gravity of the body draws a series of intersecting arcs whose amplitude is considerably greater than that in normal gait. The “compass gait” model is depicted in Figure 2.3.

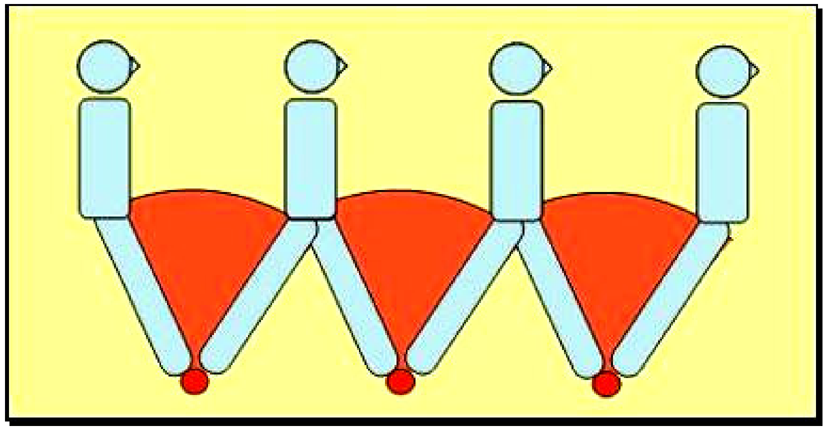

Figure 2.3. Compass gait: a crude approximation of the model of human gait kinematics. Modified from Saunders et al. (1953)

Saunders and colleagues (1953) then analysed the three dimensional movement of the lower body and of the centre of mass and identified six major determinants of gait that are added to the "compass" gait model. The first two determinants are related to the motion of the pelvis: (1) pelvic rotation and (2) pelvic list (referred also as pelvic obliquity). These two determinants are depicted in Figures 2.4 and 2.5.

During each gait cycle the pelvis rotates around the vertical axis forward on the swing phase, and then rotates relatively backward during stance phase. This first component contributes to a smoother trajectory of the body's centre of gravity, more harmonious transitions between successive "arcs" in "compass gait", and thus to a decrease in energy consumption. In addition, the pelvis rotates relative to the horizontal plane (second determinants), and in particular it tilts downward in order to increase the effective leg length at the toe-off and heel-strike phases. Due to this rotation the magnitude of the trajectory of the centre of mass is minimised.

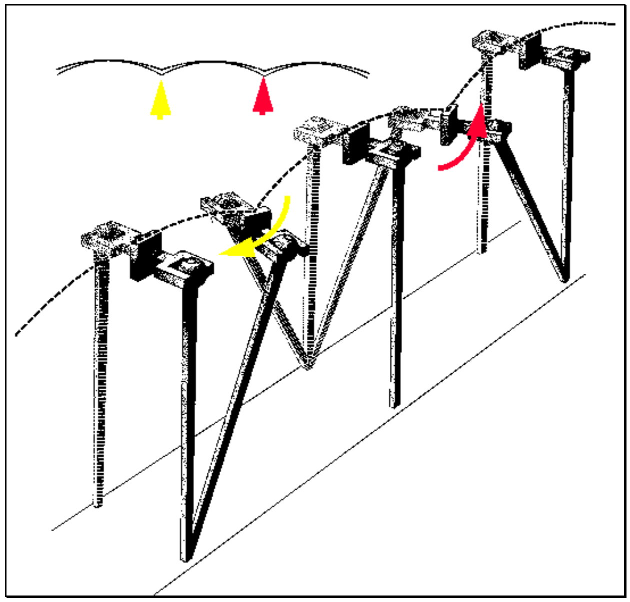

Figure 2.4. The influence of pelvic rotation flattens the arcs of the pathway of the center of gravity. Modified from Saunders et al. 1953.

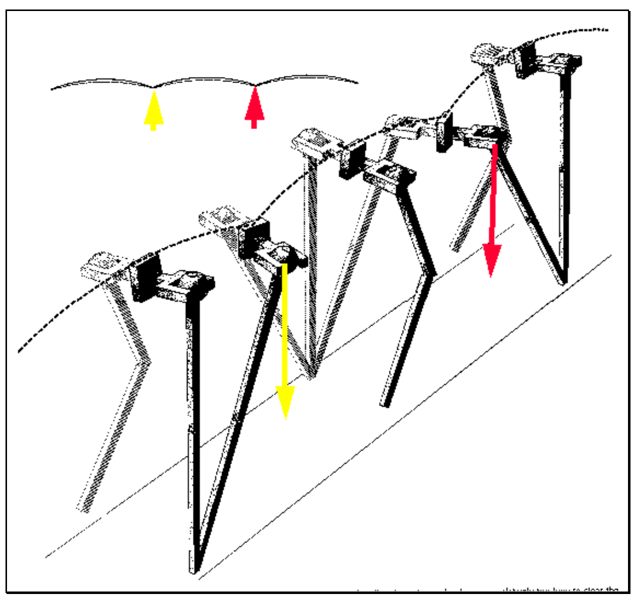

Figure 2.5. The effects of pelvic tilt on the non-weight-bearing side further flatten the arc of translation of the centre of gravity. Modified from Saunders et al. 1953.

The third component is related to the flexion of the knee (Figure 2.6). The trajectory of the pelvis becomes smooth, and the flexion of the knee during the swing phase helps to conserve the energy expenditure. This is due to the effective shortening of the pendulum. This determinants is strictly related with the fourth one: the knee flexion of the stance leg.

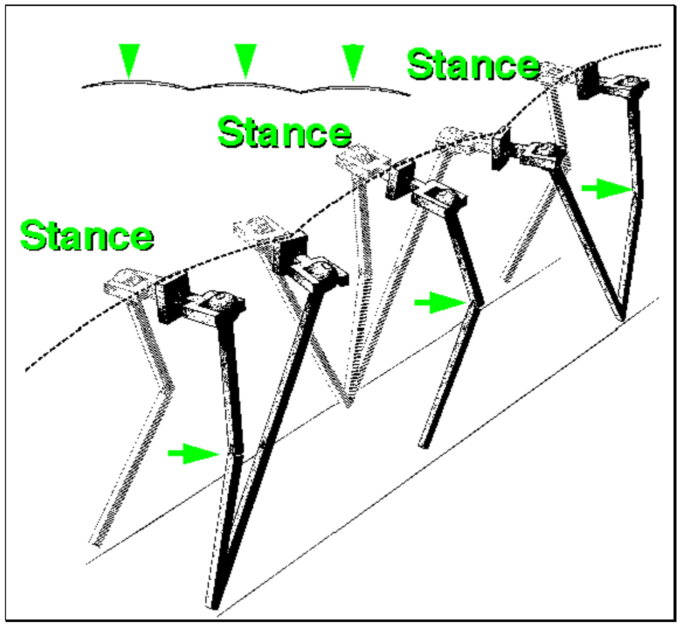

Figure 2.6. The effects of knee flexion combined with pelvic rotation and pelvic tilt achieve minimal vertical displacement of the center of gravity. Modified from Saunders et al. 1953

The fourth and fifth gait determinants encompass foot and knee mechanisms. The combined influence of ankle and knee joints results in the smoothing out of abrupt inflections at points where the curves, described by arcs, intersect and which show a forward progression of the centre of mass (in Figure 2.7 the fourth and five determinants are shown).

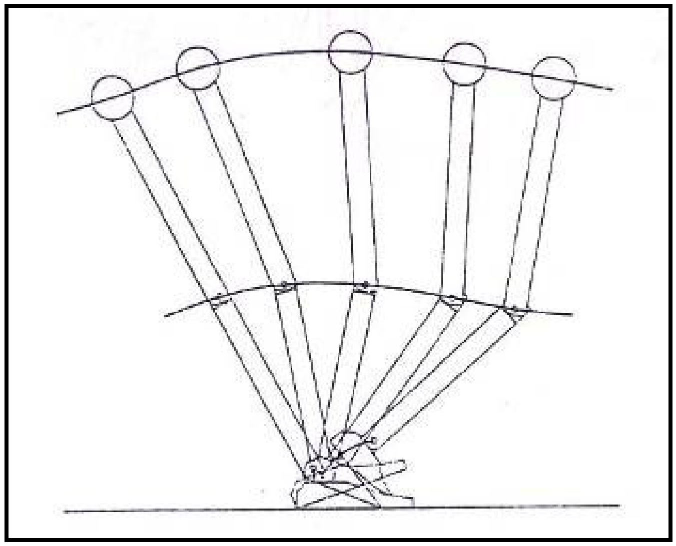

Figure 2.7. The fourth and fifth gait determinants: foot and knee mechanisms. From Saunders et al. 1953.

Finally, the sixth determinant is over concerning the lateral displacement of the pelvis. During gait, the centre of mass of the body is displaced laterally over the weight-bearing leg twice during the cycle of motion. This is accomplished by the horizontal shift of the pelvis or by relative abduction in the hip. Due to the tibiofemoral angle, excessive lateral displacement is corrected (Figure 2.8).

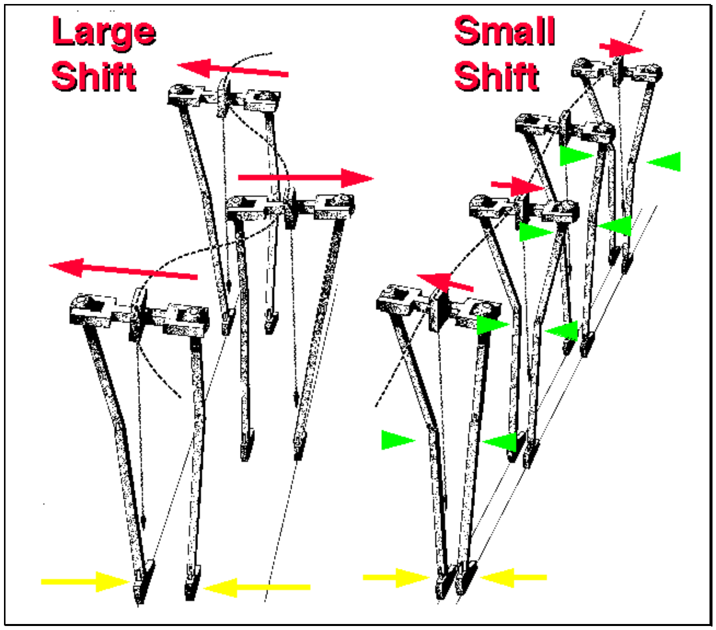

Figure 2.8. The influence of a tibiofemoral angle and of the adduction at the hip joint, corrects excessive lateral displacement of the pelvis. Modified from Saunders et al. 1953.

The above reported determinants of gait are one of the major findings in human movement analysis. In the later years in particular the first three determinants have long been considered as the principal mechanisms which specifically reduce the vertical displacement of the centre of mass from the predicted compass gait model (Saunders et al., 1953; Corcoran, 1971; Inman VT et al., 1981; McMahon et al., 1984; Lehmann et al., 1990; Perry, 1992; Inman et al., 1994; Rose et al., 1994; Gonzalez et al., 1994; Whittle, 1996; Pease and Quesada, 1996; Kerrigan et al., 1998; Malanga and DeLisa, 1998; Esquenazi and Talaty, 2000).

However, the concept that these determinants reduce the vertical displacement of the centre of mass had not been questioned formally until recently by Gard and Childress (1997). Using a model predicting the isolated effect of pelvic obliquity on centre of mass displacement (Gard and Childress, 1997), they concluded that pelvic obliquity reduces the vertical displacement of the centre of mass by, at most, only 2–4 mm. They also investigated knee flexion in stance phase (Gard and Childress, 1999), and again, contrary to interpretations of Saunders and colleagues (1953), pointed out that the knee is flexed only slightly when the centre of mass is at its highest in mid-stance, so that it reduces the maximum height of the centre of mass by only a few millimetres.

Later in the years a group led by Della Croce and Kerrigan examined a refined view of the determinants of gait. They concluded that the heel rise (foot/leg combination) plays a major role in the reduction of the vertical displacement of the centre of mass during a gait cycle (Kerrigan et al., 2000). In addition, they reported that the pelvic rotation is a minor determinant of gait since the reduction of the vertical displacement of the centre of mass is accounted for about 12% of the total centre of mass vertical displacement reduction with respect to the compass gait model (Kerrigan et al., 2000). Summarizing, the resulting effect of these major determinants is fundamental to progression. They clear a pathway for the advancing limb and reduce the displacement of the centre of mass during walking in both horizontal and vertical planes, and, as a consequence, reduce the energy consumption.

According to Saunders and colleagues (1953), the “bipedal gait” can be considered as the driving force behind the human locomotion. The significant role played by the lower limbs and by the pelvis in walking determined that locomotion and the gait cycle has been generally described starting from the lower limbs’ movement’s analyses. However, this thesis aims to evaluate the capacity of a subject observing the mechanics of the head and the trunk substain on a moving platform, the pelvis.

2.3.2 Mechanics of the upper part of the body

Despite the fact that the attention of researchers has mainly focused on studying lower limb motion, there are a number of complementary investigations involving detailed analyses of head and trunk movements (Cappozzo, 1981; Hirasaki et al., 1999; Nadeau et al., 2003; Mulavara et al., 2002; Cromwell 2003). Several studies have reported that, despite the fact that the head has a range of motion of ±20 degrees in all planes with respect to the trunk in a static state (Winters et al., 1990), the head and trunk segments are stabilised with respect to the environment with a precision of a few degrees while performing various motor acts including jumping, walking in place, walking, running, hopping, and tasks which require to maintain the equilibrium on a beam or a moving platform (Pozzo et al., 1990; Pozzo et al., 1991; Hirasaky et al., 1999).

During walking on a level surface or on a treadmill, the head is stabilised both in translation and in rotation providing a stable reference frame for gaze (Pozzo et al., 1990). One of the benefits of the head stabilisation is in its use as an inertial guidance platform to help provide a stable reference frame in order to coordinate body motion during complex movements (Berthoz and Pozzo, 1990). In addition the head stabilisation during locomotion is important in preserving visual acuity. As reported in the previous section, the coordination of the upper body segments helps to the necessity of reducing the magnitude of perturbations to the sensory systems located in the head, especially at high speeds of progression (Cappozzo, 1981). In addition, the gaze is well stabilised also by the restriction of head angular motion which helps to reduce the ocular compensation necessary to maintain gaze during walking on a fixed target (Grossman et al., 1988). Looking at these observations the stabilisation of the head plays an important role in economy of walking.

Also the trunk plays an important role for two reasons: first it represents over the 50% of the total mass of the body and thus must be sufficiently stabilised to enable the body to smoothly progress forward during gait, and, second, the location of the body centre of mass lies within the lower trunk, just posterior to the ombelicus. This observation makes the analysis of the lower trunk movements as an useful indicator of the centre of mass movements (Whittle, 1997). The most important role played by the trunk is to minimise the motion of the head by attenuating gait-related oscillations, presumably to regulate the inputs to the visual and vestibular apparatus (Kavanagh et al., 2005; Prince et al., 1994; Winter, 1991). Three-dimensional kinematic studies reveal that in the medio-lateral and vertical planes, increasing walking speed causes an increase in displacements at the level of the waist, but head movements remain relatively stable (Cappozzo 1981; Thorstensson 1984), and this suggests that a considerable amount of motion is attenuated by the trunk. In addition the trunk plays an important role also in steering (Patla et al. 1999). During the gait cycle the trunk minimises the angular displacement of the upper body relative to vertical (Cromwell et al. 2001; Thorstensson 1984), and, in addition it acts to attenuate gait-related oscillations from impacting on head motion during walking (Cappozzo 1981; Ratcliffe and Holt 1997).

The inherent neuro-mechanical dynamics of the trunk associated with the reflexes from vestibular, proprioceptive, and ocular sources, play complementary roles in achieving head control. Previous researches have shown that upper body accelerations are gradually attenuated from the pelvis to the head in the transverse plane. This process could be associated with the top–down recruitment of paraspinal muscles during walking (Prince et al. 1994). This finding supports the suggestion that the coordinated motion of the upper body may result in the trunk operating as a low-pass filter for the attenuation of gait related harmonics (Kavanagh et al. 2004; Menz et al. 2003b; Winter et al. 1990). In fact, a coordinated movement of the head and trunk is fundamental for maintaining stability during walking (Cappozzo, 1981; Hirasaky et al., 1999; Kavanagh et al., 2005; Cromwell et al., 2002), and it is also an energy saving mechanism, since the mechanical cost of walking would be higher if head and trunk moved rigidly with the pelvis (Cappozzo, 1981). To this purpose, in normal circumstances, the upper body motor control aims at minimizing the head accelerations. Given the oscillatory nature of segmental movement a uniform speed is difficult to be attained and, thus, a quasi sinusoidal trajectory, with the same period as the step cycle, is used to grant for minimal acceleration (Kavanagh et al., 2004).

Hence, the trunk is considerable not just a “passenger unit” on top of the locomotor apparatus represented by the lower limb (Winter et al., 1990), but it is suggests that it plays an important dynamic role so as to ensure the head, and hence the visual platform, remains stable attenuating the accelerations during walking (Winter, 1995; Menz et al., 2003; Prince et al., 1994). This is achieved either through passive mechanical damping or through the active control of the paraspinal muscles. Due to this consideration appears that the motion of the upper part of the body is not less important than the motion of the lower part of the body.

2.4 Age-related changes in walking mechanics

It is well recognised that normal ageing is associated with changes in function of sub-components of musculo-skeletal and neurological systems such as, visual and vestibular systems, peripheral sensation and muscle strength, which contribute to postural stability (Lord and Ward, 1996; Kaplan et al., 1985; Kokman et al., 1977; Thornbury and Mistretta, 1981). Consequently, ageing may manifest as a measurable deficit in the maintenance of walking stability. Many studies suggested that age-related changes in the overall walking patterns can be interpreted as associated with the search for a safer gait (Braune and Fischer 1895-1904; Inman, 1966; Andriacchi et al., 1977; Imms and Edholm 1981).

With advancing age there is an important gradual and progressive lost of the function of the vestibular system in terms of: reduction of the vestibular cells (Rosenhall, 1973), reduction in the diameter of vestibular nerve fibres (Bergstrom, 1973) and formation of scar tissue in the vestibular epithelia (Rosenhall and Rubin, 1975). These changes have been associated with increased sensitivity to caloric stimulation (Bruner and Norris 1971), reduced gain of the vestibulo-ocular reflex (Wall et al., 1984) and a decreased ability to perceive postural vertical (Brocklehurst et al., 1982). Therefore, a detailed analysis of the upper part of the body can be used to address the age-related effects in maintaining the control of balance during walking. In addition, despite the fact that the influence of the upper part of the body to the locomotor strategies has not been studied much or at least not recently a several number of biomechanical investigations have attempted to address the age-related issues of balance control when walking by conducting detailed analysis of trunk and head movements (Cappozzo, 1981; Hirasaky et al., 1999;

Kavanagh et al., 2005; Cromwell et al., 2002). In response to the observation that head movements are minimised during gait in normal healthy subjects, a number of authors have head motion as an indicator of walking balance in normal children (Bril and Ledebt, 1998), children with cerebral palsy (Holt et al., 1999), older people (Winter, 1991; Hirasaki et al., 1993) and patients with vestibular disturbances (Pozzo et al., 1991; Hirasaki et al., 1993; Takahashi et al., 1988; Grossman et al., 1990).

Focusing the attention on the aging process, the first finding reported by many studies is that age-related changes in the overall walking patterns can be interpreted as associated with the search for a safer gait (Woollacott et al., 1997; Prakash and Stern 1973; Sudarsky 1990; Sudarsky and Ronthal 1992). As discussed previously, the most consistent finding of these studies is that older people walk more slowly than young people, a function of both a shorter step length and increased time spent in double limb support. These temporo-spatial differences appear to be a direct result of variation in self-selected walking speed because when old and young people are instructed to walk at a specified velocity, no significant differences are apparent (Jansen et al., 1982).

Analysing the upper body movements, the major finding is that older individuals walk with conservatively lower pelvis accelerations than observed in younger people. As opposite, older individuals have the accelerations at the head level higher than young individuals. This result suggests that in older subjects there is a decline in the ability to control the movement of the upper body (Winter, 1991; Kavanagh et al., 2004). The same pattern has been detected by analysing the head movement in the frequency domain: Hirasaky and colleagues (1999) observed higher acceleration frequencies in older individuals than in young individuals. This pattern of motion can be associated to a signal regularity which has been studied also in older individuals characterised by an increased risk of fall (Menz et al., 2003). In addition, the accelerations at the head and at the pelvis level were more irregular in older people than those found in young people along both the vertical and the antero-posterior directions (Menz et al., 2003). Similarly, during walking older subjects, in comparison to young subjects, tend to exhibit lower pelvis root mean square accelerations (Menz et al., 2003b), and differences in amplitude and timing of peak accelerations of the head and trunk during critical parts of the gait cycle (Kavanagh et al., 2004).

Winter (1991) computed the antero-posterior accelerations of head and trunk by double differentiation of displacement data obtained via video-based motion analysis. Results revealed that elderly subjects had significantly lower hip accelerations and higher head accelerations during gait than young subjects. Furthermore, the elderly subjects were less able to attenuate antero-posterior head accelerations compared to the young group. On the basis of these results, it was suggested that either the elderly have diminished trunk balance control, or that the gain of the vestibular system in the older individual may be reduced. The impact of this diminished gain is that the vestibular system would only sense head motion when larger accelerations occur. Using accelerometers, Yack and Berger (1993) were able to discriminate between elderly subjects with and without a history of walking instability on the basis of the ratio of odd to even harmonics. This measure, known as an index of smoothness, was applied also by Menz and colleagues (2003) to examine differences in head and pelvis accelerations between young and elderly groups walking over different surfaces. Although no significant differences were found in index of smoothness calculations, elderly subjects exhibited significantly reduced walking velocity and step length, a finding which is consistent with the view that the elderly use a more cautious, or conservative, gait pattern to reduce trunk accelerations and maintain balance (Woollacott and Tang, 1997).

However, Kavanagh and colleagues (2005) reported that there is a fundamental requirement of the postural system to prioritise the control of head motion over the motion of other more inferior segments, then the subsequent motor output for the head may not differ in terms of signal complexity with age. One potential consequence of this, is that for the ageing system, endeavouring to maintain head control may be at the expense of coordinating movement of other body segments, such as the trunk, which may exhibit significant changes in complexity with age. Examining the relative differences in accelerative dynamics and coupling between the head and trunk may provide further insight into the coordination strategy employed by the elderly in order to maintain stability during gait.

CHAPTER 3 Mechanical models of the upper part of the body

3.1 Introduction

This chapter includes two main sections. The first section reports an overview of the models proposed in the literature to analyse the mechanics of the upper part of the body. The second section illustrates the mechanical and the mathematical models devised for this thesis.

In this thesis the attention is focused on the linear displacement and acceleration of each body segments. Conversely, several studies have addressed the analysis on the roto-traslational kinematics and kinetic of the head, trunk, and pelvis, involving the estimates of the joint angles and of the displacement of the centre of mass of each segment. For this purpose, the upper part of the body has been subdivided into three body segments: the head, the trunk, and the pelvis. Despite most of the authors, moreover, have defined the trunk as a single segment (Thorstensson et al., 1984; Bates et al., 1977; Brandell, 1973; Carlson et al., 1988; Elliott and Blanksby, 1979; Elliott and Blanksby, 1979b; Elliott and Roberts, 1980; Elliott and Ackland, 1981; Morgan et al., 1991; Thorstensson et al., 1982; Wank et al., 1998; Williams and Cavanagh, 1986; Williams and Cavanagh, 1987; Williams et al., 1991), it can be also divided in to a subsection such as lower and upper trunk.

Usually, joint angles are calculated both relative to the motion between two adjacent body segments and with respect to an external reference frame, whereas, the translational movements of each segment are calculated only with respect to the external reference frame. The three-dimensional analysis of these data involves modelling of the body segments as rigid bodies, and kinematic data are calculated by referring the rotation about three orthogonal axes of one particular body segment with respect to a reference set of axes. The reference set of axes, usually, corresponds to the laboratory (global) set of axes. However, it can be replaced by a reference frame taking into account the direction of progression. Further information on the modelling procedures that have commonly been used in three-dimensional analysis of the upper body and the lower body can be found in the papers of Davis and colleagues (1991, 1996), Cappozzo (1984; 1991), and Kadaba (1990).

3.2 Overview of the models proposed in the literature

The most used among the models developed in the last years is the one proposed by Winter and colleagues in the early 90s (Winter et al., 1990; Winter, 1995; MacKinnon and Winter, 1993). For the purpose of kinetic and kinematic gait analysis, Winter and colleagues modelled the upper body as an inverted pendulum that explains the muscular activation patterns involved in the maintenance of walking stability (Winter et al., 1990; Winter, 1995; MacKinnon and Winter, 1993). This model is based on the assumption that the primary task of walking stability is to control the movement of the head, arms, and trunk. However, these segments are simplified and represented using a single section, referred to as the HAT segment (i.e. head, arms and trunk) which rotates around the axis of the hip joint in the sagittal and frontal planes (see Figure 3.1 and 3.2 respectively).

Using this model, Winter and colleagues have shown that a significant “unbalancing” moment of the HAT segment occurs in the sagittal plane, as the horizontal acceleration of HAT causes it to flex forwards during weight acceptance and extend backwards during propulsion. To neutralise this movement, the body responds by providing a counteracting hip moment, which in normal individuals stabilises the trunk such that angular accelerations of the HAT segment are very small. The major conclusion to be drawn from this model is that walking stability in the sagittal plane is primarily provided by hip flexor and extensor muscles.

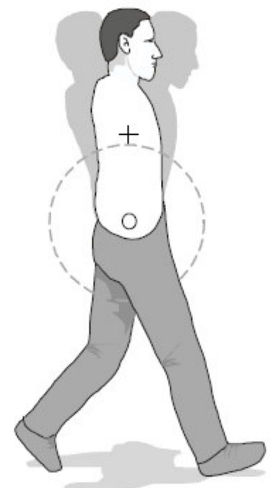

Figure 3.1. Inverted pendulum biomechanical models of walking (sagittal plane) (from Menz 2003).

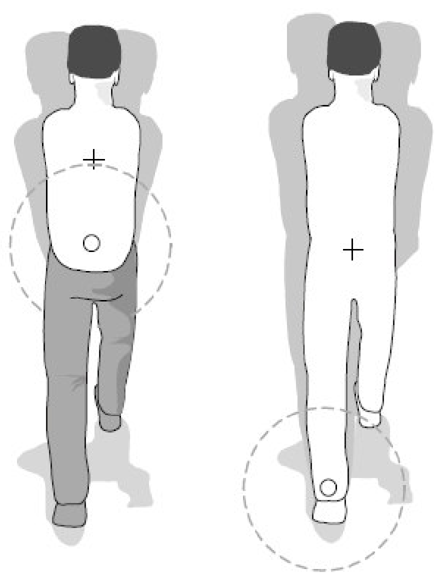

Figure 3.2. Inverted pendulum biomechanical models of walking (frontal plane) (from Menz 2003).

In the frontal plane, two inverted pendulum systems exist; one in which the HAT segment rotates around the hip joint and one in which the total body rotates around the subtalar joint of the stance foot. The tendency for the HAT segment to fall laterally during single limb stance is counteracted by the abductor and adductor muscles of the hip and the invertor and evertor muscles of the subtalar joint. Results from normal subjects using this model suggest that the stability of the body is primarily determined by the placement of the foot in the transverse plane after the HAT segment has fallen forwards and medially towards the swing limb, and highlights the importance of hip abductor muscles in walking stability. The invertor-evertor muscles of the subtalar joint were found to have only a negligible role in stabilising the trunk in normal walking (Winter et al., 1990; MacKinnon and Winter, 1993).

The model proposed by Winter and colleagues delineate the contributions of joint moments to the motion of the upper body during walking. However, the validity of the model has not been fully established and presents inherent limitations due to the simplifications of the multiple joints into single segments. The most significant limitation of the inverted pendulum model is the partial view of the role played by the vestibular and visual systems. In fact, the upper body is represented using only one segment since the authors have suggested that the vestibular system plays a minor role in the maintenance of balance during gait (Winter, 1995). However, it has been demonstrated that both visual (Hollands and Marple Horvat, 1996) and vestibular (Pozzo et al., 1990) information influences the control of the postural stability during dynamic tasks, such as walking. The head stabilisation has been shown to be significantly impaired in patients with vestibular hypofunction (Pozzo et al., 1991; Hirasaki et al., 1993; Takahashi et al., 1988; Grossman and Leigh, 1990) who regularly report difficulties in maintaining balance when walking. In addition, the assessment of walking stability using this model requires complex kinetic and kinematic instrumentation, and is therefore limited to assessments of walking on even ground in laboratory settings. Using the inverted pendulum model Miller and colleagues (1996) studied the mechanical energy expenditure during walking representing the upper body using a single segment with one set of principal moments of inertia and one centre of mass.

An important scientific contribution on the analysis of the head stability and the role played by the vestibular and the visual system was given by the team led by Pozzo. In their studies Pozzo and colleagues (1990, 1991) analysed the upper body movements during various locomotor tasks in different populations modelling the upper body in to three segments: head, neck, and trunk. The motion of these segments has been analysed using the reconstructed trajectories of ten retro reflective markers. The head is defined by the line that connect the lower border of the eye socket and the meatus of the ear, called the Frankfort plane (De Beer, 1947). The neck is defined using markers placed at the level of the seventh cervical vertebrae and on the trapezium. The trunk is defined using markers placed on the tubercle of the iliac crest.

Differently, Cromwell (2002) studied the upper body movement focusing the attention on the angular motion of the head segment with respect to the trunk along the sagittal plane. In this study the authors used the stereophotogrammetric system to reconstruct the trajectories of spherical reflective markers to define the following segments: the head defined using the markers placed at the apex of the skull and the junction between the sixth and seventh cervical vertebrae. The trunk defined using the marker placed at the junction between the sixth and seventh cervical vertebrae, and the marker placed at the lumbosacral interspace. The marker at the apex of the skull was fixed to a headband worn securely on the subject’s head, whereas, the markers placed on to the cervical and lumbar were taped directly to the skin.

In further studies, Cromwell (2003) represented the upper body using a three-segment model, in which the head segment is divided in to two subsegments: the head and the neck. The head defined using the markers placed on the apex of the skull and posteriorly over the atlanto-occipital articulation segment. The neck defined using the marker placed at the atlanto-occipital articulation, and the marker placed on to the junction between the sixth and seventh cervical vertebrae. This mechanical model, with the equivalent marker set, was recently used by Laudani and colleagues (2006) for the analysis of the upper body in the sagittal plane during gait initiation.

Some researcher did not investigate the upper body analysing the movement of the different segments composing it. For example, Dingwell and colleagues (2001, 2006) studied the movement of the upper body using only an infrared retro-reflective marker attached in to the skin over the first thoracic vertebrae. The objective of this study was to analyse the dynamic stability. The authors suggested that, since the motions of each segment directly affect the motions of all segments (Zajac, 1993), examining the dynamics of any single segment over a sufficiently long period of time, it is theoretically sufficient to capture the full complexity of the dynamics of the whole system (Sauer et al., 1991; Kantz and Schreiber, 1997). Holland and colleagues (2004) analysed only the head movement using the segment which passes between the markers placed on top of an helmet and the marker placed back at the seventh cervical vertebrae. Whereas Hirasaky and colleagues (1999) studying the relation between the speed of progression and the vertical translation of the head during walking on a treadmill. For this analysis they modelled the upper body defining head and trunk as rigid bodies using eight markers placed on the headband (head) and four markers on the small plate attached to the chest (trunk).

The motion of the upper body segment was measured using a video based motion measurement system also by Mulavara and colleagues (2002). Head and trunk motion was measured with six passive retro-reflective markers: three fixed on to a helmet worn on the head and three on a T-shaped vest worn on the upper torso. The helmet markers were positioned such that one coincided with the vertex, one was positioned just above the right tragion and the third was in the occipital area. On the trunk one marker approximated the position of the tip of the seventh cervical spinous process, the other two markers were placed so that they were on opposite sides equidistant from the midline of the body at the level of the tenth thoracic vertebrae. Whereas Bent and colleagues (2004) analysed only the part of the trunk between the eight and the twelfth thoracic vertebrae using three markers placed on the medial forehead on the back of the trunk.

Although the relevance of spine movements in walking is widely recognised (Gracovetszky, 1997), only a few studies on human locomotion have proposed a detailed model of the trunk. More recently the lumbar spine, lower thoracic, and pelvis movements, were analysed quite thoroughly during walking (Whittle and Levine, 1997; Vogt and Banzer, 1999; Callaghan et al., 1999; Crosbie et al., 1997a; Crosbie et al., 1997b) and were put in relation with the co-ordination mechanisms of gait.

Nadeau and collegues (2003) detailed the trunk as reported in Figure 3.3. In this study, the authors represented the motion of the upper body during walking in backward and forward directions using a stereophotogrammetric system. The trajectories of the following markers were analysed: five markers were placed onto the skin faced the spinal processes of the seventh cervical vertebra, the sixth and twelfth thoracic vertebrae, and the third and fifth lumbar vertebrae. In addition the head motion was analysed using two markers placed on both sides over the head (mastoid) and the pelvis using two markers placed on to the posterior iliac spine of the pelvis.

Another study that collected data of spine movements was published by Frigo and colleagues (2003). The analysis of spine movements during walking could be used as a reference for studies on pathological populations. In this paper they proposed a model able to schematise the relationship between spine mobility and gait. The location of the markers on the spine was specifically designed for the present work, and was aimed at capturing the relevant spine morphology (Figure 3.4).

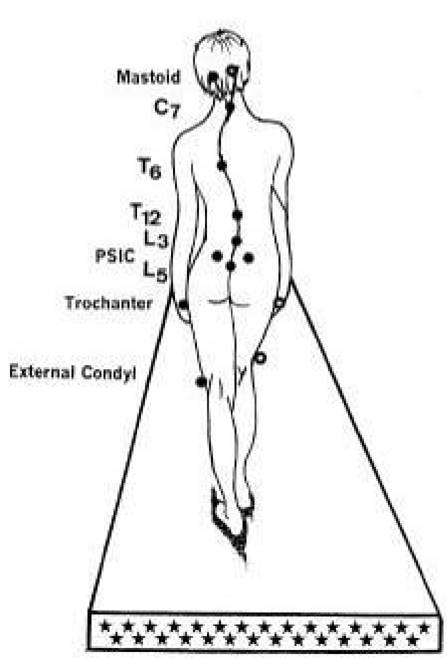

Figure 3.3. Arrangement of the 13 markers fixed to the subject and used by Nadeau and colleagues (2003) to model the upper part of the body.

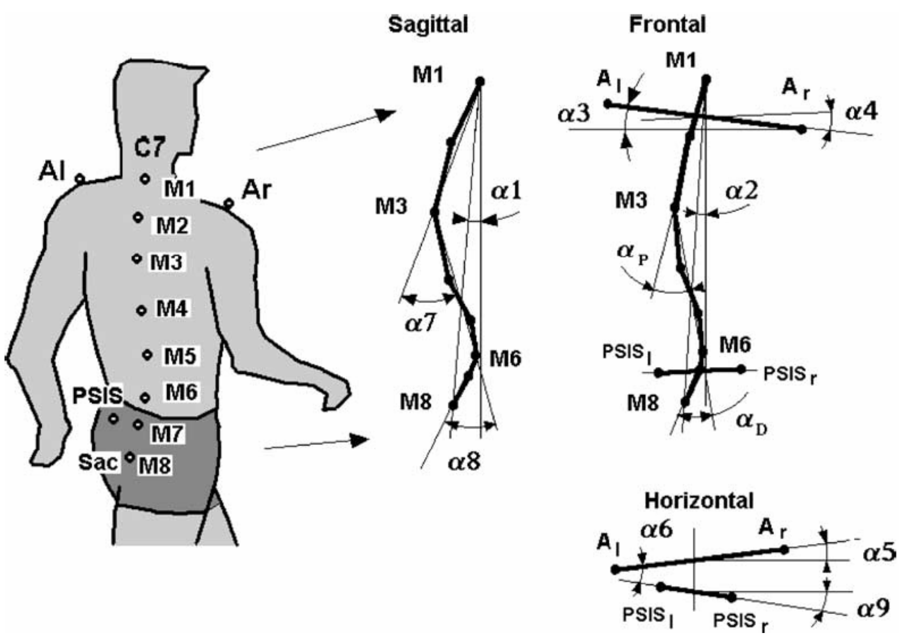

Figure 3.4. Marker placement in the study of Frigo and colleagues (2003).

Another study that collected data of spine movements was published by Frigo and colleagues (2003). The analysis of spine movements during walking could be used as a reference for studies on pathological populations. In this paper they proposed a model able to schematise the relationship between spine mobility and gait. The location of the markers on the spine was specifically designed for the work presented by Frigo and collegues (2003), and was aimed at capturing the relevant spine morphology (Figure 3.4).

A TV-based motion analyser (ELITE system, BTS-Milan, Italy) and a piezoelectric force platform (Kistler, Winterthur, Switzerland) were used to collect three dimensional coordinates of retroreflective markers and ground reaction forces respectively.

Another approach has been published in which the whole spine segmental movements (from seventh cervical vertebrae and the sacrum) in the frontal and sagittal planes were analysed (Syczewska et al., 1999). In this study fourteen retro-reflective markers were placed on the back of the subjects: eight over the spinal processes of vertebrae: C7, T4, T7, T10, T12, L2, L4 and S2 (Figure 3.5); two over the posterior superior iliac spines; two over the acromions.

Using this model Syczewska and colleagues (1999) concluded that, even if there are small inter-segmental movements there is a movement behaviour of the spine as a rigid body during gait. This finding is in agreement with the results reported in the literature, at least within a few degrees.

Finally, Van Emmerik and colleagues (2005) modelled the upper body using three segments: the head defined using three markers attached to a headpiece fit on the subject's head; the trunk defined by three markers, one aligned with seventh cervical vertebra and the other two near the bottom of the rib cage, and the pelvis defined using the markers attached to the left and right posterior aspect of the ilium (between iliac crest and superior iliac spine) and the sacrum.

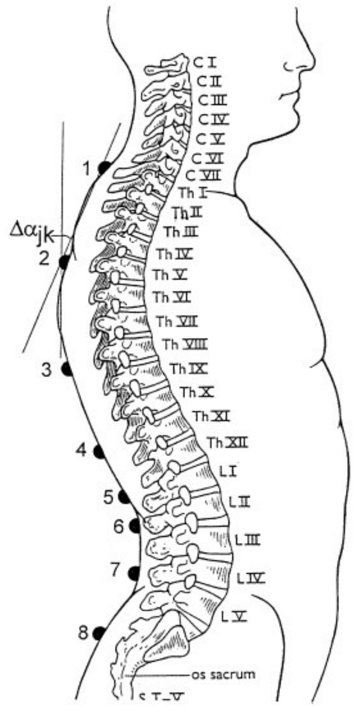

Figure 3.5. Diagram showing the angle calculations and numbering of the markers placed along the subject's spine.

In the latest years, a new methodology has been proposed for substitute the stereophotogrammetric system: the accelerometers. Measures of the upper body accelerations has been identified as a possible measure of balance during walking (Yack and Berger, 1993; Auvinet et al., 2002) and in literature several authors have applied accelerometers in order to evaluate physical strain (Bussmann et al., 2000), activity levels (Bussmann et al., 1998; Bussmann et al., 1998b; Hendelman et al., 2000; Jakicic et al., 1999; Verbunt et al., 2001) and muscle power (Thompson and Bemben, 1999). In addition, the equipment is easy to use, is wearable by the subjects/patients, it is possible to carry the entire instrumentation everywhere, also in places such as hospitals, and does not require fixed frames of reference.

However, so far methodological problems may be one reason for the limited use of accelerometers in gait analysis and must be solved in order to test validity and reliability of the method. Moe-Nilssen (1998; 1998b) only recently reported a procedure for elimination of the unwanted variability caused by the gravity component using a mathematical algorithm. This study presented the test-retest reliability analysis of the trunk acceleration (the accelerometer was placed at the third lumbar spinal vertebrae using an elastic belt). Kavanagh and colleagues (2004; 2005) used two triaxial accelerometers to measure head and trunk accelerations during each walking trial (Figure 3.6). The head accelerometer was attached over the occipital pole of the participants head via a firm fitting elastic headband, whereas the trunk accelerometer was placed over the third lumbar spinal vertebrae using rigid sports tape. This is because the trunk accelerometer was positioned close to the centre of gravity of the body, on the third lumbar vertebrae. The same protocol and the same instrumentation were used in several studies by Menz and collegues (2003).

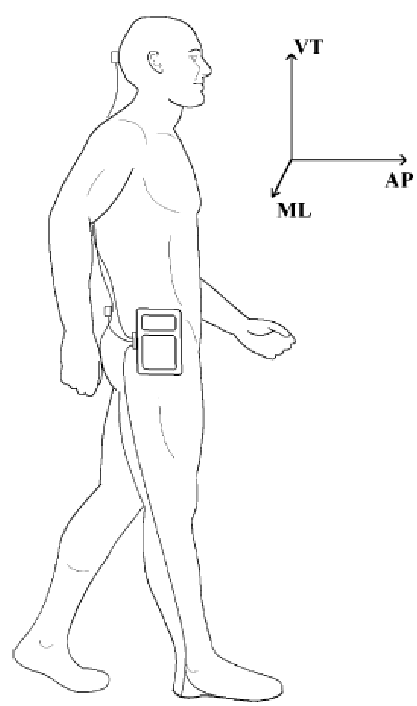

Figure 3.6. Instrument placement and acceleration sign convention proposed by Kavanagh and colleagues (2004).

3.3 The model proposed in this thesis

3.3.1 Minimum Measured-Input Models

As reported in the first chapter one of the problem to tackle in the present thesis was to define simple mechanical and mathematical models to observe the movements of the upper part of the body. To accomplish this objective, the model of the upper body has been based on the experimental approach proposed by Cappozzo (2002), called Minimum Measured Input Model (MMIM), which aims at obtaining the maximum number of information in terms of outputs of the model with the minimum number of variable in terms of inputs of the model.

The MIMM models are based on the measure of a minimum number of biomechanical variables. These quantities should be acquired using an experimental apparatus least perceivable to the test subject and of a moderate cost. Since data thus obtained do not necessarily lend themselves to straightforward interpretation in terms of functional status assessment, be it at local or global level, models of the neuro-musculoskeletal system able to output richer, physiology-related, and thus easier to interpret, information must be developed. To this purpose, all a-priori available information and, in particular, the invariant aspects of both the modelled system and the motor task, must be exploited and incorporated in the models. This implies that some of the complexity of the experimental apparatus may be transferred to the model. Due to these characteristics, these models were named "Minimum Measured-Input Models" (MMIMs).

One of the strengths of this approach is the reduction of complexity in the experimental set-ups and protocols that usually make the biomechanical approaches not practically suitable for application to subject-specific evaluation in clinical practice. In the next sections, a brief analysis of the Minimum Measured Input Model is reported.

As illustrated in Chapter 1, biomechanics can provide, through the use of complex experimental apparatuses quantitative mobility-related information,. Besides the intrinsic complexity of such apparatuses, the information extracted through them may result somehow redundant or indeed incorporate elements not useful for the mobility evaluation of the individual involved. Therefore, although biomechanical analysis may surely represent a tool to build up clinical knowledge, such knowledge needs to be further defined, understood and organised, and an interdisciplinary effort is required to translate the biomechanical techniques into clinical protocols. On the basis of the above considerations, the number of measured variables within the biomechanical test should be minimised. Furthermore, such variables should be acquired using an experimental method entailing a “basic”, low-cost, experimental apparatus. Since experimental data obtained with such a method do not necessarily lend themselves to straightforward interpretation in terms of mobility, these data should be fed as input to a model of the neuro-muscolo-skeletal system, or of a portion of it, able to output information more directly interpretable in mobility terms. To this aim, such a model should already incorporate all the information known a-priori about the subject and the motor task. This implies that the complexity of the biomechanical experimental apparatus has been transferred, to the maximal feasible extent, to the model used to process the experimental data. This type of models allows minimising the number of measured variables needed for the assessment of the functional status of an individual, i.e. the number of variables the model itself receives as input. For this reason, these models have been named Minimum Measured-Input Models. Firstly, the MMIM has to be associated to the experimental method and thus these following steps have been carried out:

i. The selection of a suitable motor task, according to the criteria illustrated in a previously.

ii. The measurement of biomechanical variables, i.e. the quantitative analysis of the above motor task, performed using tools typical of a biomechanical laboratory.

iii. The use, for the above quantitative analysis, of a properly designed experimental apparatus, which results least perceivable to the test subject and “economic” for the experimenter.

According to the structure reported in Figure 3.7, in general, a MIMM:

1. receives as input few variables, measured using the experimental apparatus;
2. embodies the invariant aspects of the modelled system, in particular the functional and structural constraints pertaining to the neuro-musculo-skeletal portion involved in movement;
3. embodies the invariant aspects of the motor task, in particular the kinematic and dynamic constraints which pertain to its execution;
4. organises the measured information, using also the above a priori knowledge about the neuro-musculo-skeletal portion and the motor task;
5. eventually estimates model parameters, on the basis of the above organisation;
6. elaborates all the above information through a certain representation of the neuro-musculo-skeletal portion modelled, thus obtaining a description of movement kinematics and dynamics; and
7. outputs physiology-related, and thus easier to interpret in functional terms, information, based on the above kinematic and dynamic description.

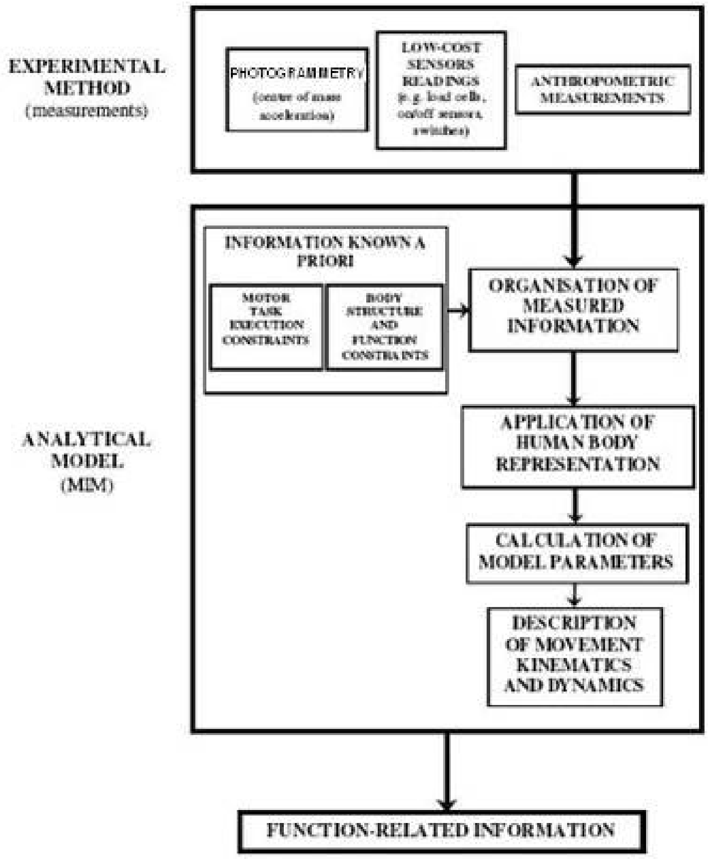

Figure 3.7. Schematic representation of the structure of a Minimum Measured-Input Model.

On the basis of the above description, it should now be clearer how a MMIM allows minimising the number of measured variables. In the next paragraph the model proposed in the present thesis for the analysis of the upper part of the body is reported.

3.3.2 The mechanical and the mathematical model

When referring to a “model”, it should be kept in mind that this actually consists of two components, i.e. the mechanical model and the mathematical model. The mechanical model is the simplified representation of the structure of the system under investigation. The mathematical model associated with the physical model consists in the set of equations that describe the behaviour of this latter model. Mechanical models consisting in segmental representations of the musculoskeletal system are considered in the present thesis. The associated mathematical models are systems of differential equations that describe the kinematics and dynamics of the segmental model. In particular, in the present work these equations establish a link among the kinematics of motion (displacements, velocities, and accelerations).

First the model proposed in based on the three steps reported in the previous chapter. Thus, in response of the first step, level walking has been preferred to analyse the individual's motor ability. Then, the experimental method used for the identification of the biomechanical variable is based on stereophotogrammetric system. However, the stereophotogrammetry could be replaced by tri-axial accelerometers. This equipment could allow for the analysis of walking patterns in a space wider than that of a laboratory, and in more natural conditions. In some recent studies, the accelerometers have successfully been used for gait analysis. In addition, they can be perfectly integrated with the environment, and thus made invisible for the test subject.

This instrumentation is coupled with an instrumented mat, which allows to identify the contact between each foot and the ground, representing a very suitable tool for the achievement of the objectives of this thesis. In fact, they can be extremely accurate and immediately available for processing. Moreover, it is possible to extend this model and the experimental protocol and analysis techniques to other locomotor acts such as running and race walking, in order to comprehend the strategy of locomotion in all of its possible configurations.

The mechanical model of the upper part of the body was based on the mechanics of the rigid bodies. Relevant critical choices hat to be made concerning the segmentation of the system being analysed in to parts each of which was to be modelled as a rigid body. The upper portion of the body was segmented in to the following five parts, as reported in Figure 3.8:

1. The head, defined by two sections: one from the posterior surface of the neck through the first cervical vertebrae, the other tangential to the mandible;

2. and 3. Right and left upper limbs, defined by sections parallel to the sagittal plane, passing through the glenohumeral joints.

3. The upper torso, defined by the above sections at the first cervical vertebrae and shoulder level, and by a transverse section conducted through the body of the fourth lumbar vertebrae.

4. The pelvis, defined by the section through the fourth lumbar vertebrae and by two sections through the hip joints that would leave the hip bone intact.

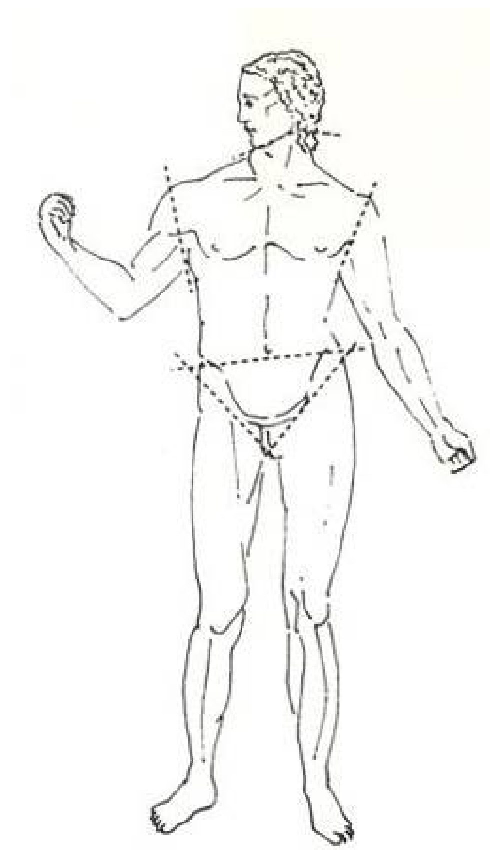

Figure 3.8. Upper body segmentation.

The constraints of rigidity were imposed on these five body segments. This was based on the following principal assumptions that were considered to be normally satisfied during walking.

1. The distance between the centroids of the fourth lumbar vertebrae and first cervical vertebrae is constant.
2. The articulation of the shoulder girdle do not move.
3. The torsional movement of the upper torso about its longitudinal axis has no relvant dynamic effects. The whole of the upper torso segment can be considerate to rotate about its longitudinal axis rigidly with the shoulder girdle.
4. The position of the vertebral column below the fourth lumbar vertebrae is rigid with the pelvis.
5. The cervical vertebral column is rigid with the upper torso.

In addition, for the purpose of the human movement analysis the body can be considered as an open kinematic chain composed of n rigid bodies. In particular, in the context of the present thesis it is of interest an open kinematic chain in which:

1. The rigid bodies composing the chain can be modelled as simple links, i.e. these bodies are substantially mono-dimensional.
2. Adjacent links are joined at their respective ends by means of ideal, frictionless hinges.
3. A rotatory actuator is applied at each joint, with its axis coinciding with the axis of the hinge.
4. The most distal end of the kinematic chain is stationary.
5. Such an open kinematic chain is schematically depicted in Figure 3.9 for the case n=3 and for a planar motion.

As the matter of fact the upper part of the body was represented with the least possible number of degree of freedom. Focusing the attention on the locomotor tasks the minimisation of the input measures could be done in two different approaches. On one hand it is possible to simplify the experimental protocol using the stereophotogrammetric system. This could be done using the minimum number of markers placed onto the skin for identifying the movements of the upper part of the body. On the other hand it is possible to use an alternative instrumentation more comfortable and portable as goniometer, accelerometer or gyrostar.

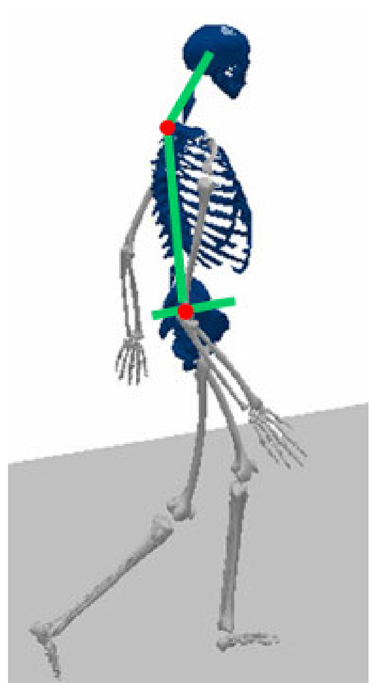

Figure 3.9. Upper part of the body proposed model

As seen previously this equipment is easy to use and does not require fixed frames of reference. However, methodological problems may be one reason for the limited use of accelerometers in gait analysis and must be solved in order to test validity and reliability of the method. For this reason, it was decided to use the stereophotogrammetric system to analyse both the displacement and the acceleration measurements (described in Chapter 5).

The upper body model proposed moves using the kinematics of three point located at the head, upper trunk (at shoulder level), and pelvis levels. The kinematics of these points along the three axes were observed using the trajectories of the centroid of the body section located at the relevant level. These points were computed using the following markers (Figure 3.10):

- Head: the midpoints between the markers placed on to the back and the front of the head using an elastic band.
- Upper trunk: the midpoints between two markers placed in to the spinal process of the seventh cervical vertebrae and the spinal process of the jugular notch.
- Pelvis: the centroid of the markers placed on to the spinal processes of the four superior iliac spines.

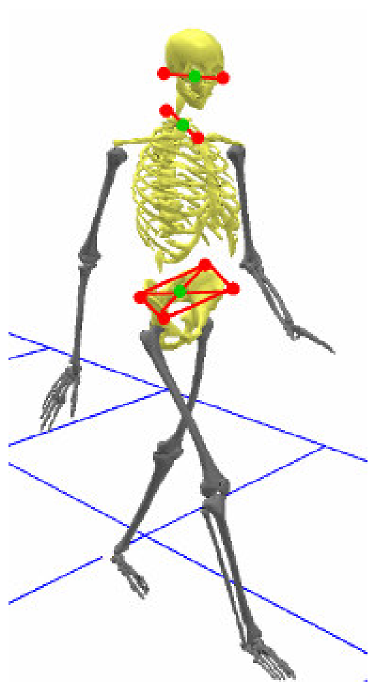

Figure 3.10 Upper body marker placement. Real marker placed on to the skin at the pelvis and shoulder levels and fixed on an elastic band at the head level (red dots). The centroids of the relevant upper body segments (green dots) were used to analyse the movement of the head, upper trunk, and pelvis.

The trajectories of the head, upper trunk, and pelvis were represented with respect to the global reference system (i.e. the laboratory reference system). Once the mechanical model has been defined, the trajectories of the above-mentioned points has been reconstructed and submitted to the mathematical model.

In steady state walking the kinematic functions associated to body landmarks are of a periodic nature with respect to an observer that translate at a constant velocity equal to the mean velocity of progression of the subject. Although it is impossible for these kinematics function to assume identical values at the ends of one gait cycle, the differences between these values can be vary small, often negligible with respect to the relative experimental error. In any case that difference, i.e. the eventual aperiodic component of the kinematic function, does not carry information within the aims of the present study. After these considerations the displacement, with respect to the stated observer, the velocity and the accelerations of the body landmarks during the walk trials on which this study was based were hypothesised to be strictly periodic with a period equal to the gait cycle. The determination of the gait cycle is clearly reported in the next chapter.

As previously reported, for the purpose of this thesis, the movement of the upper part of the body is defined by the trajectories of three “virtual” markers which describe the movement of the head (HE), upper trunk (at shoulder level, SH), and pelvis (PE). The trajectories of each body segments were recalculated with respect to new system of reference and deprived of their aperiodic component. Thus, prior to submitting these coordinates to harmonic analysis, they were recalculated with respect to a new laboratory reference system rotated about the Z axis (vertical) in order that the new X (anterior posterior) axis coincided with the mean direction of progression during any one particular test. The angle γ, by which this rotation was carried out, was obtained as:

$$\gamma = \tan^{-1}\left(\frac{Y_{PE}(M) - Y_{PE}(1)}{X_{PE}(M) - X_{PE}(1)}\right)$$

The rotation of the reference frame was carried out using the trajectory of the centroid of the pelvis because the body's centre of mass is located close to the level of the pelvis. This coordinate transformation made the direct comparison of coordinate functions relative to the different tests possible.

The coordinate functions in the new laboratory set of axes will be referred to as

$$X_S, Y_S, Z_S \qquad i = 1..M \qquad S = HE, SH, PE$$

These functions were thereafter deprived of their aperiodic component. With reference, for instance, to the *X'* coordinate the following was done:

$$X^*(i) = X(i)' - \left(\frac{K_x T(i-1)}{M-1}\right)$$

where

$$K_x = \frac{X'(M) - X'(1)}{T}$$

and T is equal to the stride duration, i.e. the walking cycle period.

Having done this, the functions $X^*_i(i)$, $Y^*_i(i)$, and $Z^*_i(i)$ could be considered periodic and could therefore be submitted to harmonic regression.

This computation allow to define a new system of reference in which the antero-posterior axis is aligned with the vector representing the subject's mean velocity of progression. The vertical axis pointed upwards and the medio-lateral axis pointed to the right of the subject. In order to analyse the antero-posterior oscillation the system of reference moves consistently with the subject's mean speed of progression.

CHAPTER 4 Assessment of level walking aperiodicity [1]

4.1 Introduction

When performing gait analysis, subjects are normally asked to walk at a constant speed of progression (at steady-state). The resulting estimated kinematic and kinetic quantities are assumed to be periodic, and, as such, are described with reference to a single walking cycle (Cappozzo, 1981; Perry, 1992). This cycle is commonly defined by the interval of time (T) that starts at the initial contact of one foot and ends at the following contact of the same foot (Perry, 1992; Winter, 1984). It is evident that the biological phenomenon that is dealt with is assumed to be periodic for the sake of simplicity, but, strictly speaking, it can only approximate periodicity and it may be referred to as pseudo-periodic (Winter, 1984).

The cyclic nature of gait data patterns emerges from gait initiation and is ultimately clearly identifiable only during steady-state pace. Since such pace is reached after negotiating some steps (Miller and Verstraete, 1999) short walking pathways, as found in many gait laboratories, might be one of the causes that contribute to the pseudo-periodicity in the recorded data. This might in turn reflect into an undesired augmented variability in the data, hiding the valuable information obtainable from variability analysis (Hausdorff, 2005; Schaafsma et al., 2003; Sheridan et al., 2003).

When talking about gait periodicity, reference should be made to the mechanical state of the locomotor system and to its reiteration after a given interval of time. Relevant state variables can be derived from the quantities normally measured in the gait analysis laboratory, such as joint angles (Forner-Cordero et al., 2006) or mechanical energies (Miller and Verstraete, 1999). Starting from a reference instant of time, the system mechanical state variation, determined in any subsequent instant of time, provides a measure of

[1] These results are published as a full paper in *Journal of Neuroengineering and Rehabilitation* (Pecoraro et al., 2006), and preliminary results have personally been presented at the *SIAMOC 2005* (Pecoraro et al., 2005).

the aperiodicity of the phenomenon as observed in that interval of time. When this variation reaches a minimum value, aperiodicity is at a minimum and, therefore, the corresponding interval of time may be considered as the best estimate of the pseudo-period ($\hat{T}$) and the relevant mechanical state variation as a measure of pseudo-periodicity or aperiodicity. The normally used foot-floor contacts represent a very partial description of the locomotor system mechanical state and, as such, may be not fully adequate for the determination of the walking pseudo-period.

The above considerations lead to the formulation of the following questions, which this paper aims at providing an answer:

1. Given a gait stride, perceived by the walking subject as performed at steady-state, how far is it from being a cycle of a periodic phenomenon and is it associated with a pseudo-periodic or an aperiodic gait?
2. Does a limited walking pathway length cause an increase in gait pseudo-periodicity?
3. As far as the above listed issues are concerned, is there a difference between the pseudo-periodic characteristics of the movements of the lower part and those of the upper part of the body?
4. How valid is the foot-floor contact method for determining the duration (pseudo-period) of the walking cycle?

In principle, the hypothesis of periodicity should be verified and quantitatively assessed by comparing relevant quantities recorded during a series of consecutive strides. However, this is hardly ever possible when using stereophotogrammetry and dynamometry, since the measurement volume normally does not host more than three consecutive steps. A method for the quantification of the discrepancy between periodicity and pseudo-periodicity or aperiodicity of walking through the observation of a single gait stride will be proposed in this paper. The information provided by this method is expected to be useful for two reasons: from a heuristic point of view, it allows an insight into a possible methodological, external, cause of the variability of gait strides (Chau et al., 2005), and, from a practical standpoint, it allows for a control of the consistency of the observed gait stride with the hypothesis of steady state.

4.2 Materials and methods

4.2.1 Subjects and protocol

Ten young healthy subjects (6 males, 23±5 years, 62±12 kg, 1.68±0.08 m) volunteered for the study and signed an informed written consent. Subjects participated in two sets of experiments each characterised by a different walking pathway length. They were asked to walk along the linear pathways at three different self-selected speeds of progression: slow (SS, "walk at a slow speed"), natural (NS, "walk naturally") and fast (FS, "walk as fast as you can"). In all cases the subjects were explicitly asked to reach and maintain a constant speed of progression. Three trials were performed for each condition.

4.2.2 Instrumentation

A purposely built instrumented mat was used to measure the beginning (t_b) and end (t_e) of a stride determined by the consecutive contacts of the same foot with the mat, and relevant stride duration (T) was then computed as $T = t_e - t_b$. Adhesive 5 mm wide copper stripes were attached parallel to each other at a 3 mm distance along a 4 m length linoleum mat. Alternative stripes were connected to an electric circuit so that, when short circuited, a signal was generated. Two independent circuits (Figure 4.1) were constructed for right and left foot. Subjects wore custom designed socks, the bottom part of which was covered with conductive material.

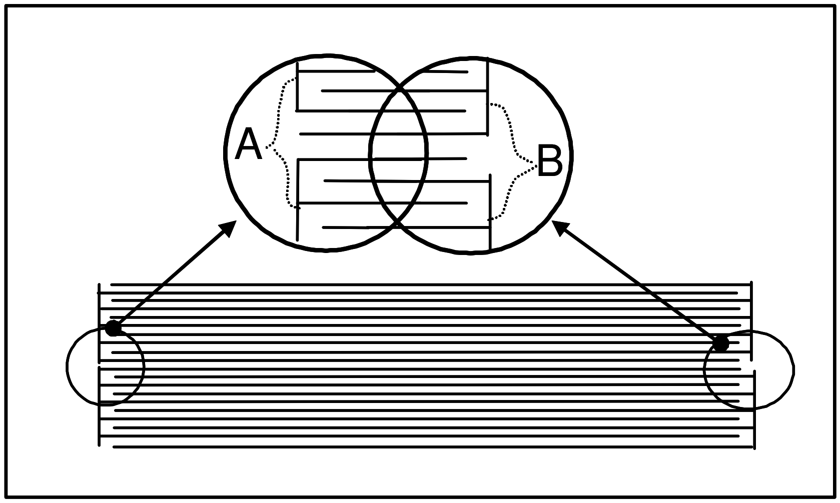

Figure 4.1. Schematic representation of the foot-switch mat.

Using this mat the subject should walk unconstrained and thus fulfilling the basic requirement of not influencing the subject's way of walking.

The accuracy of the mat was assessed by comparing its data to those simultaneously acquired with a strain gauge force plate (Bertec Corporation, Ohio, USA, sample frequency = 120 samples/s) while a subject stepped on it. An example of this procedure is depicted in Figure 4.2. The first sample at which the vertical force was greater than its mean value plus two standard deviations recorded for 1 s while the force plate was unloaded, was chosen as indicator of the foot contact.

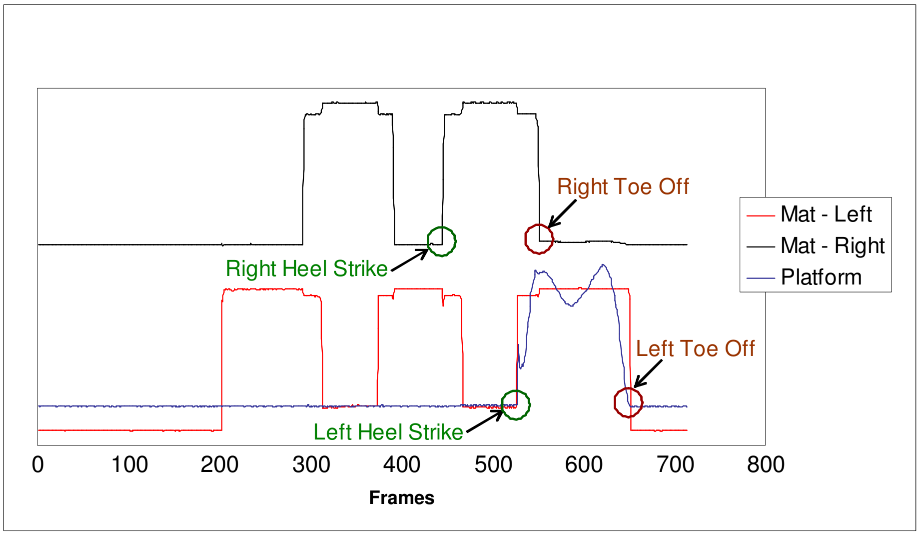

Figure 4.2. Example of the vertical component of the force acquired using the platform (blue line) compared with the signal acquired using the mat (green line).

The differences found between the time events detected with the mat and with the force plate were computed for ten different trials, and were always lower than 0.025 s.

A nine-camera VICON® system (Oxford Metrics, Oxford, UK) was used to reconstruct the 3D positions, relative to the stereophotogrammetric set of axes, of 19 markers placed on the body of the subjects. The markers were placed on the head (three markers attached on an elastic band), trunk (spinal process of the seventh cervical vertebra, acromion processes), pelvis (anterior superior iliac spines, midpoint between the posterior superior iliac spines), and lower limbs (greater trochanters, femoral lateral epicondyles, lateral malleoli, calcanei, and second metatarsal heads). From now on, the cluster composed by all the above listed markers will be referred to as whole body (WB) cluster. Two sub-clusters will also be considered: the lower body (LB) cluster, including the 13 markers located on pelvis and lower limbs, and the upper body (UB) cluster, including the 9 markers located on head, trunk and pelvis. While defining the latter cluster, it was decided not to include upper limb markers because of the low sensitivity of the overall gait pattern to the movement of the upper limbs, which, for this reason, may tend to be more aperiodic than that of the rest of the body. In addition, most gait analysis protocols do not include these segments.

Stereophotogrammetric and mat data were simultaneously collected at a sampling frequency of 120 samples/s.

4.2.3 Experiments

As mentioned previously, two sets of experiments were performed. The first set aimed at placing the subjects in the best condition for reaching steady-state walking and at properly assessing periodicity by observing more than a single stride. The second set aimed at simulating a standard laboratory situation where only a single stride per limb fits in the measurement volume and the walking pathway length is limited.

The first set of experiments was performed exploiting the entire length of a 20x8 m laboratory such that subjects were able to walk for at least twelve consecutive strides and the stereophotogrammetry measurement volume hosted two strides per limb (among the fifth, sixth and seventh stride). This pathway allowed the subjects to reach what they perceived to be a steady-state walking pace (Miller and Verstraete, 1999).

The second set of experiments used the same protocol as described above, but was carried out along an 8-m pathway and within a stereophotogrammetric measurement volume that hosted only three consecutive steps.

4.2.4 Data analysis

Through a rigid transformation, 3-D marker position data were represented relative to a laboratory set of axes, the X axis of which was aligned with the analysed subject mean speed of progression, and the Y axis was vertical. This data was filtered through a low-pass fourth-order Butterworth filter with a cut off frequency of 8 Hz (Winter, 1990) and was used to describe the variations of the mechanical state of the subjects' whole body, and of its upper and lower parts.

Each cluster was considered as an ensemble of particles with equal mass and was represented, in each sampled instant of time during movement and relative to the laboratory frame, by the global position vector (${}^{g}\mathbf{p}$) of its centre of mass and by the orientation matrix (${}^{g}\mathbf{R}$) of an arbitrarily chosen set of local axes. To this purpose, the singular value decomposition technique was used (Spoor and Veldpaus, 1980). The position vectors of the markers in the local frame is referred to as ${}^{l}\mathbf{p}$. Using this information, energy-like quantities were calculated and used to describe the instantaneous "mechanical state" variation of each cluster and, in turn, of each related body system. Such variations were calculated relative to the reference instant of time t_b.

The vertical coordinate $h(\mathrm{t})$ of the marker cluster centre of mass was considered to represent a gravitational potential energy-like quantity G(t). Its variation was calculated as:

$$\Delta G(t) = h(t) - h(t_b). \tag{4.1}$$

The first derivative of the centre of mass position vector was estimated via a three-point central difference differentiation method. The modulus of the instantaneous velocity thus obtained ($v(\mathrm{t})$) was used to calculate a linear kinetic energy-like quantity K(t). Its variation was given by:

$$\Delta K(t) = v^2(t) - v^2(t_b). \tag{4.2}$$

The instantaneous angular velocity ($\omega(\mathrm{t})$) of the cluster was computed from the orientation matrix ${}^{g}\mathbf{R}$ (Berme et al., 1990). The modulus of $\omega(\mathrm{t})$ was used to calculate a rotational kinetic energy-like quantity R(t). Its variation was given by:

$$\Delta R(t) = \omega^2(t) - \omega^2(t_b). \tag{4.3}$$

Besides height and velocities variations, during movement the clusters may undergo a variation in orientation and a deformation, both of which were described by elastic potential energy-like quantities. The orientation variation of a cluster between time t_b and time t may be thought to correspond to a rotation of the local set of axes about the corresponding finite helical axis against an elastic torsional constraint. From the orientation matrices of the cluster at times t_b and t, the relevant rotation vector ($\boldsymbol{\theta}$(t)) was calculated (Cappozzo et al., 1997). The following torsional elastic potential energy-like quantity was, thus, determined:

$$\Delta T(t) = \|\boldsymbol{\theta}(t)\|^2. \qquad (4.4)$$

Similarly, the variation of the markers local position vectors between time t_b and time t allowed for the calculation of another elastic potential energy-like quantity associated with marker cluster deformation:

$$\Delta D(t) = \frac{\sum_{i=1}^{N} \left\| {}^{l}\mathbf{p}_i(t) - {}^{l}\mathbf{p}_i(t_b) \right\|^2}{N}, \qquad (4.5)$$

where N is the number of markers of the relevant cluster.

A measure of the system mechanical state variation, in any observed interval of time, could be obtained through the sum of the absolute values of the above-defined energy-like quantities. However, since such quantities have arbitrary dimensions, their values are incomparable and should hence be normalised. The maximum amplitude of one (arbitrarily chosen) of the energy-like quantities could be considered as a reference normalisation factor for the others. In such way, the variation of the mechanical state of the system can be calculated according to the following weighed sum:

$$\Delta E(t) = \frac{|\Delta G(t)|}{k_G} + \frac{|\Delta K(t)|}{k_K} + \frac{|\Delta R(t)|}{k_R} + \frac{|\Delta T(t)|}{k_T} + \frac{|\Delta D(t)|}{k_D}, \qquad (4.6)$$

where k_G, k_K, k_R, k_T, and k_D are weighing constants. These constants, for the *i*-th trial, are arbitrarily calculated considering (for example) the maximum amplitude of gravitational potential energy-like quantity as the reference normalisation factor (i.e. setting $k_G = 1$):

$$k_{G_i} = \frac{\max|\Delta G_i(t)|}{\max|\Delta G_i(t)|} = 1 \qquad (4.7a)$$

$$k_{K_i} = \frac{\max|\Delta K_i(t)|}{\max|\Delta G_i(t)|} \tag{4.7b}$$

$$k_{R_i} = \frac{\max|\Delta R_i(t)|}{\max|\Delta G_i(t)|} \tag{4.7c}$$

$$k_{T_i} = \frac{\max|\Delta T_i(t)|}{\max|\Delta G_i(t)|} \tag{4.7d}$$

$$k_{D_i} = \frac{\max|\Delta D_i(t)|}{\max|\Delta G_i(t)|} \tag{4.7e}$$

However, if different trials are to be compared, a fixed reference value of the constants should be chosen for all of them. Since no reference values were available to this purpose, previously (unpublished) available kinematic gait data, recorded at natural speed from a similarly aged group of 15 healthy subjects adopting the same instrumentation and marker set as the ones in the present study, were used. The values of the constants were computed as in (4.7a-4.7e) for each trial, and their mean values (k_G = 1.00, k_K = 0.04, k_R = 0.27, k_T = 0.05, k_D = 0.70) were used in the rest of the study.

The variation of the state of the system during a gait cycle, starting from a foot contact (t_b) and normalised with respect to its maximum value, was assessed by means of the index:

$$J(t) = \frac{\Delta E(t)}{\max \Delta E(t)} \times 100. \tag{4.8}$$

The sought aperiodicity index was computed as:

$$J_{min} = \min (J(t)). \tag{4.9}$$

The larger the J_{min} value, the further the analysed gait is from a periodic process. The time instant for which J(t) = J_{min} is proposed as an estimate of the end of the period ($\hat{t}_e$), which can be used to determine the pseudo-period $\hat{T} = \hat{t}_e - t_b$. The value assumed by J(t) at t_e, i.e. that measured by the mat at the end of the stride, will be referred to as J_e.

To assess the sensitivity of the index J_{min} to the values of the constants, a set of 100 different combinations of values was generated by randomly varying them in the ranges defined by their corresponding mean values plus or minus one standard deviation, computed over the above described 75 trials (k_G=1.00, k_K=0.04±0.03, k_R=0.27±0.15, k_T=0.05±0.03, k_D=0.70±0.19).

As previously mentioned, gait aperiodicity depends on step length, cadence, and width, which can differently affect cluster kinematics: step length variation can be expected to mostly affect h(t), and partly v(t) and $^{l}\mathbf{p}$(t); step cadence variation mostly affects v(t); step width variation mostly affects h(t) and $^{l}\mathbf{p}$(t). Thus, it can be hypothesised that within the same stride, the quantities ΔG(t) and ΔD(t) are the most sensitive to changes in step length and width, and the index ΔK(t) to changes in step length and cadence. Moreover, the two terms ΔR(t) and ΔT(t) are expected to be negligible when walking straight. In such case, the equations (4.6), (4.8) and (4.9) can be replaced by the following:

$$\Delta\hat{E}(t) = \frac{|\Delta G(t)|}{k_G} + \frac{|\Delta K(t)|}{k_K} + \frac{|\Delta D(t)|}{k_D}, \tag{4.6a}$$

$$\hat{J}(t) = \frac{\Delta\hat{E}(t)}{\max \Delta\hat{E}(t)} \times 100, \tag{4.8a}$$

$$\hat{J}_{min} = \min(\hat{J}(t)). \tag{4.9a}$$

The above described hypotheses were verified by means of *ad-hoc* constrained tests. One subject (male, 23 years, 1.70 m, 70 kg) was asked to follow the auditory input of a metronome to modulate step cadence (C), and the visual input of markers on the floor to control step length (L) and width (W) while walking along the 8-m pathway. L, C, and W were first kept unconstrained, and then made to vary, one at a time, from step to step (ΔL = 0.4 m, ΔC = 1 step/s, ΔW = 0.2 m).

4.2.5 Statistical analysis

The coefficient of determination (R^2) was used, for both mat-measured period (T) and pseudo-period ($\hat{T}$), to assess the equivalence of the duration of the first and second stride of the same trial. A two-way ANOVA analysis was used to assess the effects of two between group factors: speed (three levels: SS, NS, and FS) and pathway length (two levels: short, SP, and long, LP). When significant differences ($p<0.05$) were found, a post-hoc analysis was performed using an unpaired samples two-tailed t-test with Bonferroni correction (significance level: $p = 0.017$). Finally, a two-tailed t-test for paired samples ($p = 0.05$) was used to compare the results obtained for the three clusters of markers and to assess the differences between T and $\hat{T}$.

4.3 Results

The first three steps of the analysis consisted in the validation of the proposed method in terms of: robustness of the index J_{min} to the variation of the constants k; sensitivity of the energy-like indices to the gait characteristics; suitability of the method to detect periodicity by observing changes between subsequent strides.

When varying the five constants in the computed ranges, the index J_{min} varied by less than 10% of its initial value and the corresponding $\hat{T}$ remained unaltered.

Figure 4.3 illustrates, for a representative LP trial, an example of the variations of the WB indices $\Delta G(t)$, $\Delta K(t)$, $\Delta R(t)$, $\Delta T(t)$, $\Delta D(t)$, and $J(t)$ from their values at t_b. The sensitivity of these indices to stride parameter variations is illustrated by the data reported in Table 4.1: as expected, J_{min} was sensitive to variations in step length, cadence, and width. When length and cadence varied, the contribution, at instant $\hat{t}_e$, of ΔG and ΔK to the overall J_{min} reached 71% and 61%, respectively.

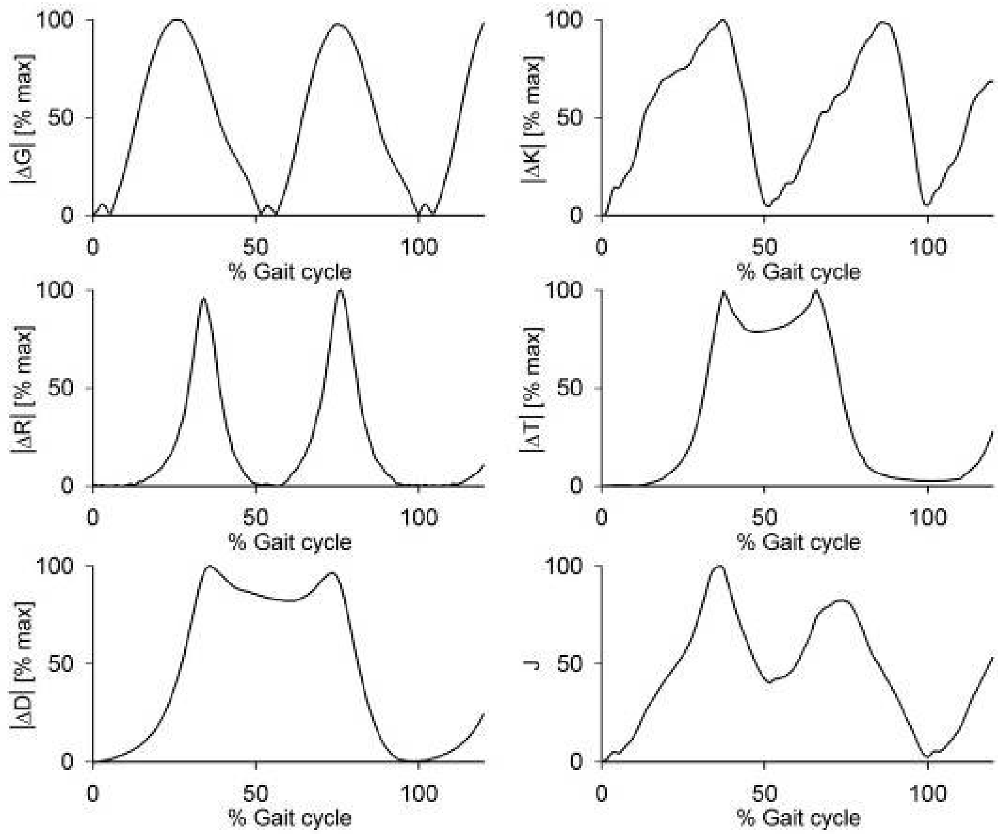

Figure 4.3. Example of the time patterns of the "energy-like" indices used to describe the mechanical state of the system during one gait trial. The data along the abscissa are normalised with respect to the duration of the first stride, determined as $\hat{T} = \hat{t}_e - t_b$**.**

Gait Factor	J_{min} (%)	ΔG ($\%J_{min}$)	ΔK ($\%J_{min}$)	ΔR ($\%J_{min}$)	ΔT ($\%J_{min}$)	ΔD ($\%J_{min}$)
None	10	81	10	1	5	3
Length	29	71	5	3	3	19
Cadence	21	13	61	1	5	20
Width	37	46	36	1	4	13

Table 4.1. Results of the trials performed by one subject in controlled experimental conditions (the gait factors, namely step length, cadence, and width, varied from step to step: ΔL=0.4m, ΔC=1step/s, ΔW=0.2m, respectively). The percentage contributions of the variation of the gravitational potential (ΔG), linear (ΔK) and rotational (ΔR) kinetic, torsional (ΔT) and deformation (ΔD) elastic potential energy-like quantities to the total value of J_{min}, as computed at $\hat{t}_e$, are shown.

The results in Table 4.1 show that, as hypothesised, the two terms ΔR(t) and ΔT(t) can be neglected when walking straight. These indices, in fact, contributed to the overall J_{min} by no more than 5%. As a result, from now onward, the index $\hat{J}_{min}$ will be used for the assessment of periodicity.

An example of the $\hat{J}(t)$ time patterns obtained for the WB cluster between subsequent strides is reported in Figure 4.4. In particular, the data obtained during the constrained tests of a representative subject when asked to walk at steady-state (Figure 4.4a) and when asked to vary progression speed freely between the two strides (Figure 4.4b) is illustrated. In the reported figure, during steady-state, the $\hat{J}_{min}$ values were similar in the two subsequent strides (8.7% and 7.8%, respectively, computed within the relevant T) and were lower than those obtained when the subject was accelerating during the first stride (32.2% and 10.3%, respectively).

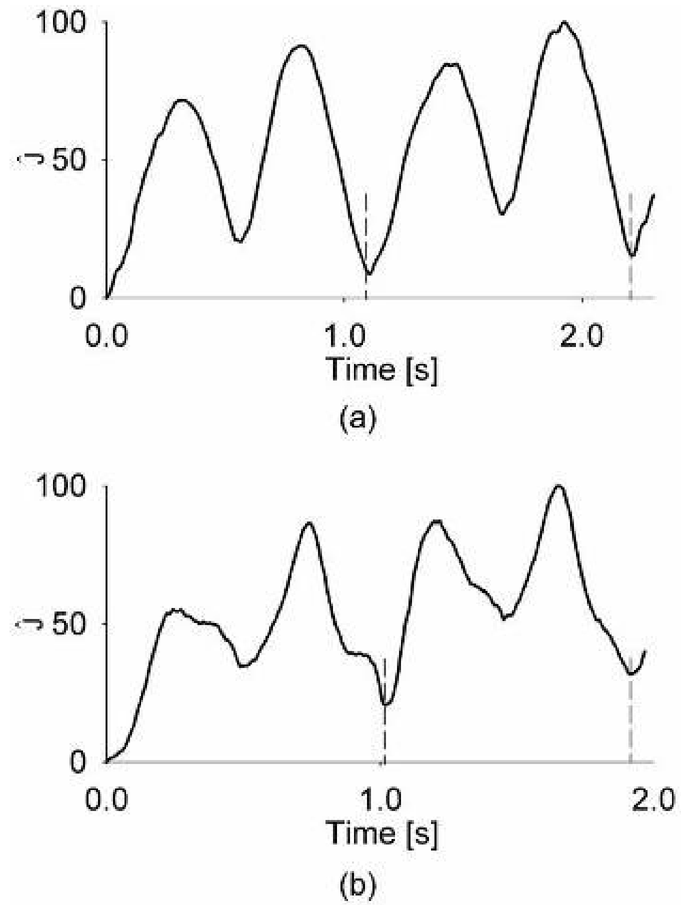

Figure 4.4. The time patterns of $\hat{J}(t)$ are reported for two different trials, one at steady-state (a) and one during which the subject was deliberately accelerating (b). The vertical dashed lines indicate the t_e values recorded from the mat.

Due to the longer stride length at the faster speed, it was not possible to collect reliable data for two consecutive strides in the FS experiments performed by the subjects along the LP. For this reason, only the SS and NS trials were included in this first part of the analysis. These trials actually resulted to be pseudo-periodic, since:

a) consecutive strides had the same duration, as shown by the high coefficient of determination between the duration of the first (T_1) and of the second (T_2) stride ($R^2 = 0.97$ for T_1 and T_2 and $R^2 = 0.90$ for $\hat{T}_1$ and $\hat{T}_2$), and

b) the two time curves of $\hat{J}(t)$ obtained in T1 and T2 (and resampled to 100 samples) were highly correlated (Pearson correlation coefficient r = 0.92±0.13) and very similar to each other (RMS = 9±8%). The same stands for the curves computed using $\hat{T}_1$ and $\hat{T}_2$ (r = 0.92±0.12 and RMS = 10±7%).

The values of $\hat{J}_{min}$ obtained for this sub-set of experiments (WB: 9±9%; LB: 7±6%; UB: 16±15%) can be used to set a threshold between pseudo-periodicity and aperiodicity for the three clusters of markers.

All the 90 LP trials were then compared with the SP trials for the three speeds of execution of the task. Mean speeds of progression, calculated as the product between stride length and stride frequency, did not significantly change between SP and LP trials (SS: 0.93±0.22 ms^{-1} vs 0.84±0.16 ms^{-1}, NS: 1.16±0.18 ms^{-1} vs 1.20±0.25 ms^{-1}, and FS: 2.21±0.14 ms^{-1} vs 1.91±0.52 ms^{-1}).

As reported in Table 4.2, the results of the ANOVA showed that the two factors, speed and pathway length, when considered separately, affected $\hat{J}_{min}$ of the UB cluster only: relevant $\hat{J}_{min}$ values increased with increasing speed and shortening of the pathway length (Table 4.3).

Factor	**$\hat{J}_{min}$ – WB**		**$\hat{J}_{min}$ – LB**		**$\hat{J}_{min}$ – UB**	
	F	**p**	**F**	**p**	**F**	**p**
Speed	3.262	0.062	2.526	0.108	7.795	0.004
Pathway Length	2.457	0.151	1.432	0.262	10.862	0.009
Speed × Pathway Length	2.914	0.080	1.192	0.327	2.684	0.095

Table 4.2. Results of the ANOVA performed on the $\hat{J}_{min}$ values obtained for the three clusters of markers.

The results of the comparison among the three body clusters are highlighted in Table 4.3, where the mean (standard deviation) values of $\hat{J}_{min}$ computed in all the experimental conditions are shown. Whereas no significant differences were found between the WB and LB, the UB almost always showed the highest $\hat{J}_{min}$ values. The only exception was relative to walking at NS along LP in which case $\hat{J}_{min}$ was not significantly different between WB and UB.

Trial type	$\hat{J}_{min}$ – WB (%)		$\hat{J}_{min}$ – LB (%)		$\hat{J}_{min}$ – UB (%)	
	LP	SP	LP	SP	LP	SP
SS	9 (7)	11 (2)	7 (5)	8 (2)	14 (10)*^	20 (5)*^
NS	9 (6)	12 (2)	7 (5)	9 (1)	10 (8)^	20 (5)§*^
FS	10 (8)	15 (4)	8 (6)	10 (3)	15 (11)°*^	27 (7)°§*^

Table 4.3. Mean values (standard deviation) obtained for the two sets of experiments (long pathway, LP, and short pathway, SP) in the different trial types: slow (SS), natural (NS) and fast (FS) walking speeds. The values of the index $\hat{J}_{min}$ are reported for the whole body (WB), lower body (LB) and upper body (UB) marker clusters. (°significant differences between FS and NS; §significant differences between LP and SP; *significant differences between UB and WB; ^significant differences between UB and LB).

Table 4.4 shows the results of the comparison between the stride durations, once estimated ($\hat{T}$) using the marker clusters and once measured with the mat (T) using the heel strike. The differences between the two quantities were, on average, less than or equal to 3% of T for WB and LB and less than or equal to 7% of T for UB. Significant differences were observed for all clusters at both SS and NS, except for WB and LB when walking along LP.

Trial type	WB		LB		UB	
	LP	SP	LP	SP	LP	SP
SS	1 (1)	2 (1)	1 (1)	1 (1)	3 (1)	3 (1)
NS	2 (1)	2 (1)	1 (1)	2 (1)	4 (1)	3 (1)
FS	2 (1)	3 (2)	2 (1)	3 (2)	7 (9)	4 (5)

Table 4.4. Mean values (standard deviation) of the differences between the stride duration values measured with the mat (T) and those estimated with the index $\hat{J}_{min}$ ($\hat{T}$), expressed as a percentage of T. Results are reported for the two sets of experiments (long pathway, LP, and short pathway, SP). The values in bold indicate the experimental conditions in which the differences between T and $\hat{T}$ were significant.

Table 4.5 shows the results of the ANOVA performed on the $\hat{J}_e$ values: for all clusters, only the speed factor caused a significant change in $\hat{J}_e$, with the highest values found at FS along SP (Table 4.6); for the UB cluster, $\hat{J}_e$ was not significantly different at FS and SS.

Factor	$\hat{J}_e$ – WB		$\hat{J}_e$ – LB		$\hat{J}_e$ – UB	
	F	p	F	p	F	p
Speed	14.624	0.000	21.453	0.000	11.086	0.001
Pathway Length	1.947	0.196	1.002	0.343	3.462	0.096
Speed × Pathway Length	0.801	0.464	0.489	0.621	0.394	0.680

Table 4.5. Results of the ANOVA performed on the $\hat{J}_e$ values obtained for the three clusters of markers.

Trial type	$\hat{J}_e$ – WB (%) LP	$\hat{J}_e$ – WB (%) SP	$\hat{J}_e$ – LB (%) LP	$\hat{J}_e$ – LB (%) SP	$\hat{J}_e$ – UB (%) LP	$\hat{J}_e$ – UB (%) SP
SS	11 (7)	13 (3)	9 (5)	10 (2)	20 (13)*^	25 (5)*^
NS	11 (6)	15 (3)	8 (4)	11 (2)	16 (9)*^	23 (5)*^
FS	14 (11)	21 (6)†°	13 (10)	16 (4)†°	23 (14)*^	35 (9)°*^

Table 4.6. Mean values (standard deviation) obtained for the two sets of experiments (long pathway, LP and short pathway, SP) in the different trial types: slow (SS), natural (NS) and fast (FS) walking speeds. The values of the index $\hat{J}_e$ are reported for the whole body (WB), lower body (LB) and upper body (UB) marker clusters. (†significant differences between FS and SS; °significant differences between FS and NS; *significant differences between UB and WB; ^significant differences between UB and LB).

The use of t_e instead of $\hat{t}_e$ led to an increase in the estimate of gait pseudo-periodicity: the values of $\hat{J}(t)$ at t_e ($\hat{J}_e$, Table 4.6) were significantly higher than those at $\hat{t}_e$ ($\hat{J}_{min}$, Table 4.3) for all clusters in all experimental conditions.

4.4 Discussion

The objectives of this study were: 1) to gather information concerning the periodicity of walking cycles and to set a threshold between pseudo-periodic and aperiodic walking; 2) to describe the effects of a limited walking pathway on gait pseudo-periodicity; 3) to assess differences in the movements of the lower and of the upper part of the body; 4) to assess the validity of the foot-floor contact method for determining the duration (pseudo-period) of the walking cycle.

To achieve the above listed objectives, a mechanical energy-like index computed from the kinematic data recorded during one stride only has been devised. This method was validated performing *ad-hoc* experiments which allowed for the comparison of two consecutive strides recorded in a pathway which certainly allowed the subjects to reach the steady-state condition. The results indicated that the proposed index is suitable for the measure of gait aperiodicity using one stride only.

The first two objectives were reached by assessing gait pseudo-periodicity. It was shown that, if considering the whole body cluster, a value of 18% (mean + one standard deviation) of the global variation of the mechanical energy-like index can be considered as a threshold of physiological pseudo-periodicity of young, healthy adult gait. Values below this threshold, in fact, were found when subjects were asked to walk along the 20-m pathway at NS and SS. Gait periodicity seemed reduced when subjects were asked to walk along the 8-m pathway and this was most evident at their maximal speed. These differences, however, were significant only for the upper part of the body.

The third objective of this study required the assessment of the periodicity of the different parts of the human body. In almost all the experimental conditions, the upper part of the body showed higher aperiodicity than the lower part. This behaviour can be explained by the lower number of functional constraints that trunk and head movements have to comply with during gait. Lower limbs, in fact, are responsible for forward progression and must hence act in a quite regular and constrained fashion, whereas head and trunk can, theoretically, freely behave while being “carried” by the lower part of the body (Winter, 1991).

Finally, the fourth objective of the paper was tackled and the validity of considering the foot-floor contact events as markers of the period of gait cycles was assessed. The error (mean + one standard deviation) that can be made in estimating the gait cycle duration for the whole body from the heel contacts, is, on average, less than 3% of the period in long pathway condition at all gait speeds. This error can increase up to 5% while walking at fast speed along the short pathway, and can lead to an increase of the pseudo-periodicity value from 19% to 27%. The different behaviour of lower and upper body described in the above paragraph was confirmed by the differences found between the periods estimated for the two clusters along the 20-m pathway: whereas the period estimated for the lower body was the same as that measured from the foot-floor contacts, noticeable differences were recorded for the upper part of the body. This proves that special attention should be dedicated when the foot-floor contact method is used for detecting the period of the whole and lower body along short pathways, and it should never be considered valid for the upper body.

4.5 Conclusions

This study showed that young, healthy adult human gait is pseudo-periodic, and this is more marked for the upper part of the body. A control of aperiodicity should always be performed if trials are conducted in common gait laboratories. If any instrument is available for the detection of the beginning of a stride, for example a force plate, then the proposed index could be used to accurately estimate stride duration.

CHAPTER 5 Intrinsic and extrinsic patterns of level walking in older women [2]

5.1 Introduction

As reported in Chapter 2 a detailed analysis of upper body movements provides information for the assessment of ontogenetic [Assaiante and Amblard 1993; Bril and Ledebt, 1998; Baumberger et al., 2004; Gill et al., 2001), pathological (Pozzo et al., 1991; Hirasaki et al., 1993; Mamoto et al., 2002), and deteriorative phenomena (Winter, 1991; Gill et al., 2001; Hirasaki et al., 1993; McGibbon and Krebs 2001; Menz et al., 2003; Paquette et al., 2006). It has been suggested that the harmonic analysis of the linear displacement of points located at head, shoulder, and pelvis levels may be useful to identify two human movement patterns: an intrinsic and an extrinsic patterns. The intrinsic pattern has been referred to as a characteristic of a specific population, whereas it has been suggested that the extrinsic pattern is a characteristic of an individual or of a sub-group of individuals in a given population performing the locomotor act (Pecoraro et al., 2004; Cappozzo, 1981).

The aim of this study was to assess short- and long- term reliability of the harmonic analysis to identify both the intrinsic and the extrinsic patterns in healthy women aged 75-85. The hypothesis to be tested was that, for this selected population sample, the intrinsic pattern was both intra- and inter-subject repeatable, while the extrinsic pattern was only intra-subject repeatable.

[2] These results are published as a full paper in *Gait and Posture* (Pecoraro et al., 2006b), and personally presented at the *1st joint ESMAC-GCMAS 2006* (Pecoraro et al., 2006c).

5.2 Materials and methods

5.2.1 Subjects and protocol

Participants were recruited through an advert on a local newspaper in the city of Glasgow. They were fully independent community-dwelling older women and were selected as "medically stable" participants for exercise studies using the criteria proposed by Greig et al. (1994). The Ethics Committee of the University of Strathclyde (protocol number 13:01/02) approved the study.

Eleven volunteers (age: 77±2 years; stature: 1.54±0.05 m; body mass: 63.5±8.1 kg) were asked to walk 10 laps at self-selected comfortable speed on an oval shaped 20-m walkway circuit, rectilinear for 8 m on each side. A stereophotogrammetric system (6-camera VICON 612®) was used to reconstruct the trajectories of 10 markers (anterior and posterior superior iliac spines, jugular notch, C7 spine process, front and back of the head, and heels). Stereophotogrammetric data were acquired at 120 frames/second for one stride performed in each lap in the mid-portion of the rectilinear part of the walkway circuit. To assess both short- and long-term repeatability, the same protocol was repeated after 6 and 12 weeks which led to the analysis of 30 strides per subject. Participants were asked to maintain their habitual lifestyle during the period of the study.

In order to assess stereophotogrammetric errors a spot check was carried prior to each experimental session using the protocol described in Della Croce and Cappozzo (2000). Soft tissue artefacts were deemed negligible at head and upper trunk level while no relevant clue was available at pelvis level.

5.2.2 Temporo-spatial parameters

The identification of the beginning (t_b) of the relevant stride was detected using the velocity of the appropriate heel marker derived from the relevant coordinates as suggested by Mickelborough and colleagues (2000).

To verify the validity of the method proposed the authors have compared the results obtained using the markers' velocity with the kinetic data extracted from a force platform. This method shows an high inter-subject reliability in determining timing on foot contact events during walking the protocol based on foot marker kinematics. In addition, in our laboratory group this method was tested by comparing even time detected using the velocity of the marker placed on to the heel and the event time assessed using the instrumented mat (Figure 5.1). For the purpose of this analysis, ten young healthy subjects were asked to walk twenty times at their self selected comfortable speed. In Figure 5.2 are shown the differences between the heel strike events determined using the mat and the heel marker velocity. These differences are depicted both as a the error to detect the heel strike and the error in terms of gait period.

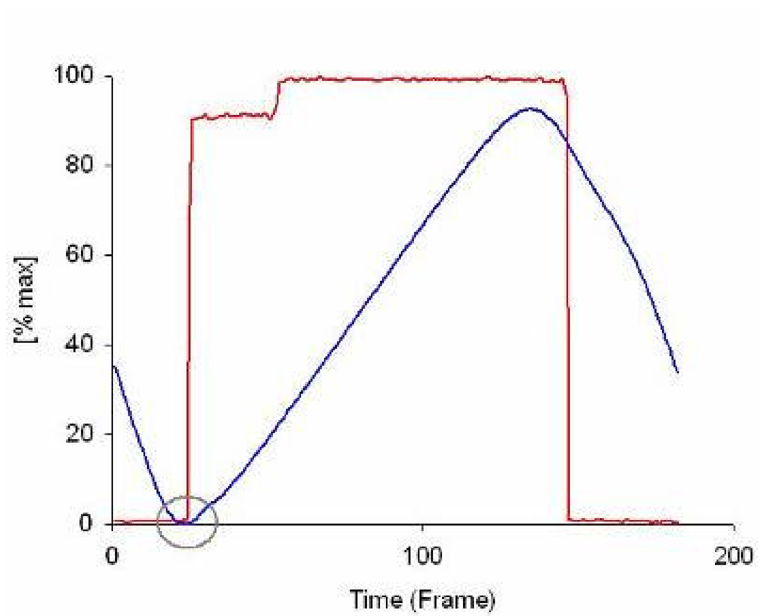

Figure 5.1. Example of the heel marker velocity (blue line) compared with the signal acquired using the mat (red line).

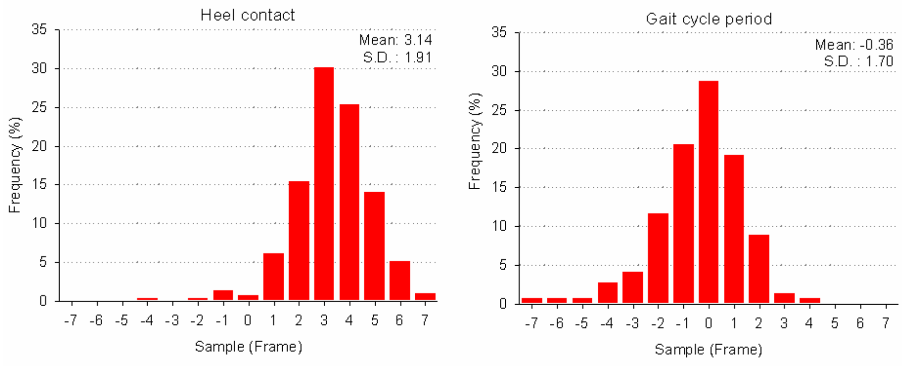

Figure 5.2. Differences between the instant of time detected using the mat and using the kinematics data. Positive values indicate that the data determined with the method is shorter with respect to the data determined with the instrumented mat method.

The reported results show that the difference in sample times between the proposed method and the instrumented mat is lower then 0.05s for the detection of the heel strike instant. In addition, the differences between the periods determined using the two methods are lower than 0.016 s. Note that each sample represents 0.008 (1/120) s.

The following temporo-spatial parameters were computed for each stride: stride period ($T=t_e-t_b$); stride frequency (SF) computed as 1/T, stride length (SL) computed as the antero-posterior displacement of the C7 marker between two sequential heel strikes of the right leg. Walking speed (WS) values were obtained as the product between SL and SF.

5.2.3 Harmonic analysis

The mechanical and mathematical model was used to describe the movement of the upper part of the body and to recalculate the trajectories of the points representing the movement of the head, upper trunk, and pelvis, with respect to new system of reference which has been deprived of the aperiodic component as reported in Chapter 3. Hence, the trajectories of the head, upper trunk and pelvis along the antero-posterior, medio-lateral and vertical directions could be considered periodic and could therefore be submitted to harmonic regression.

These points were computed as the midpoints between the head markers (HE), C7 and the jugular notch (SH), and as the centroid of the four iliac spines (PE), respectively. The HE, SH, and PE trajectories were represented with respect to a reference system having the antero-posterior (AP) axis aligned with the vector representing the subject's mean velocity of progression, and moving consistently with it. The vertical axis (V) pointed upwards and the medio-lateral axis (ML) to the right of the subject. The harmonic coefficients of the relevant coordinate time functions were calculated via discrete Fourier transform.

For each coordinate, the mean value and the Fourier coefficients – the amplitude and phase for each harmonic – were estimated and, thus, represented in the following analytical form:

$$\begin{cases} X^*(t) = a_{xo} + \sum_{j=1}^{N_x} a_{xj} \sin\left(j\omega_o t + \varphi_{xj}\right) & 0 \le t \le T \\ Y^*(t) = a_{yo} + \sum_{j=1}^{N_y} a_{yj} \sin\left(j\omega_o t + \varphi_{yj}\right) & 0 \le t \le T \\ Z^*(t) = a_{zo} + \sum_{j=1}^{N_z} a_{zj} \sin\left(j\omega_o t + \varphi_{zj}\right) & 0 \le t \le T \end{cases} \tag{5.1}$$

where a_{xo}, a_{yo}, and a_{zo} are the mean values; a_{xj}, a_{yj}, a_{zj}. are the amplitudes values and φ_{xj}, φ_{yj}, and φ_{zj} are the phases values of the *j*-th harmonic, respectively. In addition the fundamental frequency of the harmonic analysis is set to:

$$\omega_0 = \frac{2\pi}{T} \tag{5.2}$$

The relative power (RP_h) of each harmonic representing the coordinates in the AP (RP_{AP}), ML (RP_{ML}), and V (RP_V) directions at the three upper body levels, was computed using the following equation:

$$RP_h = \frac{A_h^2}{\sum_{h=1}^{H} A_h^2} \cdot 100, \tag{5.3}$$

where A_h is the amplitude of the h-th harmonic and H is the total number of the analyzed harmonics. The denominator of the equation represents the total power of the signal.

Amplitudes and phases of the even harmonics in the AP and V directions and of the odd harmonics in the ML direction, described the intrinsic pattern. The extrinsic pattern was described using the odd harmonics in the AP and V directions and the even harmonics in the ML direction.

5.2.4 Statistical analysis

Repeatability of the temporo-spatial parameters was assessed using the coefficient of variation (CV), defined as the ratio between the standard deviation and the mean value (Hausdorff et al., 1997; Terrier et al., 2003). Within session coefficient of variation (CV_W) was computed as $CV_W=SD_1/ME_1$, where ME_1 is the mean value of a given parameter computed over the trials performed by each subject in the first experimental session and SD_1 is the relevant standard deviation. Between sessions coefficient of variation (CV_B) was computed as $CV_B=SD_B/ME_B$, where ME_B is the mean value of a given parameter computed over the three experimental sessions and SD_B is the relevant standard deviation. A CV value lower than 0.1 indicates excellent repeatability (Menz et al., 2004).

The reliability of the amplitudes and phases of the harmonics was assessed using the standard error of measurement (SEM) (Hunter et al., 2000; Domholdt et al., 2000; Drouin et al., 2004). The SEM value, expressed in the same metric unit as the measurement, was computed as $SEM = SD\sqrt{1 - ICC}$, where SD is the within-subject standard deviation, and ICC is the intraclass correlation coefficient (Shrout and Fleiss, 1979). Within session SEM (SEM_W) was assessed for the data recorded in the first experimental session using the ICC (2,1) model:

$$\mathrm{ICC}(2,1)=\frac{\mathrm{BMS}-\mathrm{EMS}}{\mathrm{BMS}+(\mathrm{n}-1)\cdot\mathrm{EMS}+\mathrm{k}\left(\frac{\mathrm{RMS}-\mathrm{EMS}}{\mathrm{n}}\right)}, \tag{5.4}$$

where BMS is the between subjects mean square, EMS is the error mean square, RMS is the within subject mean square, n is the number of subjects, and k is the number of tests performed by each subject. Between sessions SEM (SEM_B) was assessed using the ICC(2,3) model computed over the three experimental sessions:

$$\mathrm{ICC}(2,3)=\frac{\mathrm{BMS}-\mathrm{EMS}}{\mathrm{BMS}+\frac{\mathrm{RMS}-\mathrm{EMS}}{\mathrm{n}}}. \tag{5.5}$$

Repeatability of the intrinsic and extrinsic patterns was investigated using the "coefficient of multiple correlation" (CMC) of the relevant harmonics. The CMC is a measure of the overall waveform similarity of a group of curves, and its magnitude is close to 1 if the waveforms are similar, and close to 0 otherwise. A CMC value higher than 0.7 indicates excellent repeatability (Kadaba et al., 1989; Queen et al., 2006). Intra-subject repeatability was assessed by first computing the CMC of each variable for each subject, and then averaging these values over the eleven subjects. Inter-subject repeatability was assessed computing the CMC considering all the trials performed by all the subjects at the same time. Intra- and inter-subject CMCs were then assessed both within- (CMC_W) and between sessions (CMC_B), according to the equations provided by Kadaba et al. (1989).

A two-tailed t-test for paired samples (significance level p=0.01) was used to assess the differences of the amplitudes and phases among the three investigated upper body levels.

5.3 Results

5.3.1 Temporo-spatial parameters

The repeatability of the temporo-spatial parameters was excellent both within and between sessions (SL: 1.39±0.04 m, CV_W: 0.03±0.01, CV_B:

0.04±0.02; SF: 0.93±0.04 s^{-1}, CV_W: 0.02±0.01, CV_B: 0.03±0.02; WS: 1.31±0.04 ms^{-1}, CV_W: 0.03±0.01, CV_B: 0.06±0.02).

5.3.2 Harmonic analysis

In all directions and at all body levels, the ratio between the sum of the power of the first four harmonics and the total power was higher than 99%. In all analysed strides, the amplitude of the harmonics of order higher than four was lower than 0.7 mm. Since the spot check showed that the stereophotogrammetric system had an accuracy in the order of 1.5 mm, the latter harmonics were associated with the experimental error. The third and fourth harmonics had amplitudes always lower than 2 mm. Since this figure alone did not provide evidence that these harmonics were reliable, the relevant phases were also analysed. Their values, as observed within subjects, were scattered in intervals close to 2π rad. Both observations led to the association of these harmonics to experimental errors. Identical considerations applied to the second harmonic of the ML coordinates at all upper body levels. From now onwards, the first two harmonics of the V and AP coordinates and the first harmonic for the ML coordinate will be taken into consideration as carrying significant information.

Table 5.1 shows, for all the body levels, the relative power of the significant harmonics of the trajectories recorded along the different directions.

Harmonic	Body Level	Relative Power (%)		
		AP	ML	V
First	HE	56.0±31.2	100	10.3±8.2
	SH	37.3±28.8	100	11.2±8.6
	PE	29.3±15.3	100	9.1±8.4
Second	HE	44.0±31.2	-	89.7±8.2
	SH	62.7±28.8	-	88.8±8.6
	PE	70.7±15.3	-	90.9±8.4

Table 5.1. Average (± standard deviation) relative power (RP) of the harmonics of the head (HE), shoulder (SH), and pelvis (PE) trajectories, along the AP, ML, and V directions. The values are reported as a percentage of the total power of the correspondent signal.

The average amplitudes of the first harmonic are reported in Table 5.2, together with the corresponding within and between sessions standard errors of measurement.

	Level	AP			ML			V		
		AV	SEM_W	SEM_B	AV	SEM_W	SEM_B	AV	SEM_W	SEM_B
A (mm)	HE	6	1.05	1.20	22	1.01	1.17	6	0.26	0.41
	SH	5	0.77	0.97	19	0.61	0.85	6	0.27	0.42
	PE	6	1.18	1.14	15	0.43	1.50	9	0.44	1.20
φ (rad)	HE	-0.5	0.11	0.41	3.0	0.05	0.08	-1.7	0.03	0.08
	SH	0.2	0.11	0.32	3.1	0.06	0.10	-2.8	0.23	0.12
	PE	-0.1	0.18	0.19	2.9	0.09	0.15	-2.8	0.11	0.36

Table 5.2. Average (AV) values of the first harmonic amplitudes (A) and phases (φ) together with the standard errors of measurement (SEM_B and SEM_W).

In the V direction, the SEM_W and SEM_B of the first harmonic amplitudes increased going from head to upper trunk and to pelvis level, despite no appreciable changes in the average amplitudes. Higher SEM_W and SEM_B were found for the ML amplitudes, but they were still lower than 10% of the relative average values. Higher ratios (up to about 20%) were found in the AP directions. The SEM_W and SEM_B obtained for the phases in the V and the ML directions were similar at all body levels, and were lower than those found in the AP direction, where the PE phases were the most reliable.

The average amplitudes of the second harmonic are reported in Table 5.3, together with the corresponding within and between sessions standard errors of measurement. Along the AP direction, a significant increase ($p<0.001$) in the amplitudes and in the phases ($p<0.001$) were obtained going from HE to SH and PE level. The ratios between the SEM_W and SEM_B and the relative average values of the amplitudes were lower than 6%, except for the HE in the AP direction (10%). The SEM_W and SEM_B obtained for the phases in the V direction were similar at all body levels, and were lower than those found in the AP directions.

	Level	**AP**			**V**		
		AV	**SEM_W**	**SEM_B**	**AV**	**SEM_W**	**SEM_B**
A (mm)	HE	5	0.52	0.39	18	0.15	0.65
	SH	7	0.40	0.36	19	0.18	0.64
	PE	12	0.44	0.58	20	0.17	0.74
φ (rad)	HE	-0.8	0.34	0.18	-2.1	0.03	0.06
	SH	-0.4	0.06	0.11	-2.1	0.03	0.06
	PE	-0.1	0.06	0.06	-2.2	0.06	0.06

Table 5.3. Average (AV) values of the second harmonic amplitudes (A) and phases (φ) together with the standard errors of measurement (SEM_B and SEM_W).

According to the hypothesis of the work and to the results of the power spectrum analysis, the intrinsic pattern was represented by the AP and V second harmonics and by the ML first harmonic and the extrinsic pattern was represented by the AP and V first harmonics.

The intra- and the inter-subject CMCs of the intrinsic pattern, both within and between sessions, are shown in Table 5.4.

	Level	AP		ML		V	
		CMC_W	**CMC_B**	**CMC_W**	**CMC_B**	**CMC_W**	**CMC_B**
Intra-subject	HE	0.90±0.08	0.87±0.12	0.98±0.01	0.97±0.01	0.99±0.00	0.99±0.00
	SH	0.97±0.01	0.96±0.02	0.98±0.02	0.97±0.02	0.99±0.00	0.99±0.00
	PE	0.99±0.01	0.98±0.01	0.96±0.02	0.95±0.03	0.99±0.00	0.99±0.00
Inter-subject	HE	0.83	0.82	0.93	0.93	0.97	0.97
	SH	0.90	0.90	0.93	0.93	0.97	0.97
	PE	0.96	0.96	0.91	0.91	0.97	0.97

Table 5.4. Intra- (average ± standard deviation) and inter-subject coefficients of multiple correlation for the intrinsic pattern for both within (CMC_W) and between sessions (CMC_B).

The intra- and the inter-subject CMCs of the extrinsic pattern are shown in Table 5.5. The values are reported for the analyses within and between sessions.

Figure 5.3 shows an example of graphically summarizing the two characteristic patterns (intrinsic and extrinsic) of the investigated group for the purpose of comparing them: the radius of the plotted polar sectors were obtained from the mean (± standard deviation) amplitudes of the corresponding harmonics and their orientation (angle with respect to the horizontal axis) were obtained from the average (± standard deviation) phases.

	Level	AP		V	
		CMC_W	CMC_B	CMC_W	CMC_B
Intra-subject	HE	0.79±0.10	0.70±0.15	0.89±0.12	0.86±0.20
	SH	0.80±0.15	0.72±0.17	0.90±0.12	0.88±0.16
	PE	0.85±0.06	0.83±0.11	0.89±0.09	0.81±0.17
Inter-subject	HE	0.18	0.20	0.79	0.79
	SH	0.29	0.30	0.79	0.79
	PE	0.66	0.61	0.70	0.70

Table 5.5. Intra- and inter-subject coefficients of multiple correlation for the extrinsic pattern for both within (CMC_W) and between sessions (CMC_B).

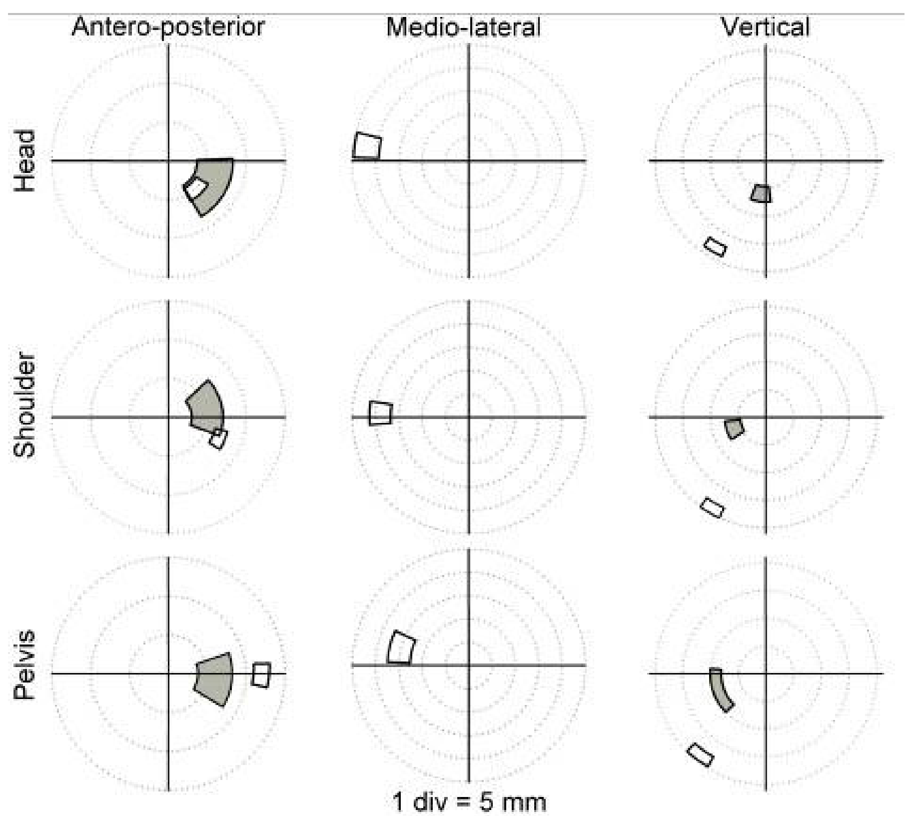

Figure 5.3. Polar diagram representing the intra-subject average and standard deviation of the amplitudes and the phases of the intrinsic (white sectors) and extrinsic (grey sectors) pattern at HE, SH, and PE levels along the AP, ML, and V directions.

5.4 Discussion

This study proved the suitability of the harmonic analysis for describing the upper body kinematics and for detecting the intrinsic and extrinsic patterns of gait in older women, which reveal fundamental mechanisms governing their walking.

The global strategy that the older women adopted to walk was consistent either within or between sessions, as demonstrated by the excellent repeatability of the temporo-spatial parameters. Our results confirm that the magnitude of the stride-to-stride fluctuations in stride length and step timing are unaltered in healthy older adults (Gabell and Nayak, 1984; Hausdorff, 2005).

The reliability analysis of the investigated harmonics gave excellent results. The Fourier coefficients of the first harmonic were highly reliable both within and between sessions, at all body levels, and the second harmonic was even more reliable than the first one. Therefore, we could appropriately proceed with the repeatability analysis of both the extrinsic and intrinsic patterns.

As hypothesized, the intrinsic pattern, which is characteristic of the whole investigated population of older women, showed excellent intra- and inter-subject repeatability, as confirmed both within and between sessions, at all levels, and along all directions. The detection of the intrinsic pattern allowed to highlight a clear oscillatory pattern of the upper body during walking in the older women (Figure 5.3): the oscillations of the pelvis were wider and the peak occurred earlier than those of the upper trunk and the head. These results are in line with the observations carried out in other populations of both young and older individuals using different methods of analysis (Menz et al., 2003; Pozzo et al., 1990). The compass movement of the lower limbs ruled the oscillations that characterize the V and ML displacements of the upper body during each stride: the V oscillations were biphasic, with the second harmonic having the highest relative power content; the ML oscillations were monophasic, with the relative power content of the second harmonic being negligible (Table 5.1). These results in the older women are consistent with the observations in young individuals of Cappozzo (1984), with the exception of the lack of a second harmonic in the ML direction. This lack, which indicates a greater gait symmetry than in young subjects, might be a unique feature of the older women's gait. Further investigations are needed to verify this hypothesis. In the AP direction, the head and trunk relative power content was distributed between the first and the second harmonics (Table 5.1), in contrast with the young individuals of Cappozzo (1984), in which the second harmonic was predominant. Interestingly, these results support the suggestion of Menz and colleagues (2003) that a small ratio between the amplitude of even and odd harmonics, measured through accelerometers, would relate to a higher risk of falls in older individuals.

The extrinsic pattern, which is characteristic of each individual within the population of older women performing the locomotor act, was excellently intra-subject repeatable, as shown by the values of both CMC_W and CMC_B. As hypothesized, the inter-subject repeatability of the extrinsic pattern was smaller in magnitude, thus indicating that upper body movements are unique to each individual. The inter-subject repeatability analysis also revealed differences along the different directions and between upper body segments. The lowest CMC_W and CMC_B values were found in the AP direction, and the head behaviour was less repeatable than the upper trunk and pelvis ones. These results agree with those recently reported by Kavanagh and colleagues (2005), who showed that the accelerations of the head and upper trunk are less repeatable in AP than in V direction, with the head having the lowest repeatability values.

5.5 Conclusions

In conclusion, this study has given an insight into some of the mechanisms of coordination ruling the movements of the upper body in order to control head stability in older individuals. One of the strengths of the harmonic analysis is the simplicity and clarity to quantitatively express laws governing human walking, which makes it a promising tool for motor ability assessment.

CHAPTER 6 Age related changes in walking stability [3]

6.1 Introduction

As reported in Chapter 2, a locomotor act can be seen as the superimposition of two different movement patterns: an intrinsic pattern characterised by perfect symmetry with respect to the anatomical planes, and an extrinsic pattern representing the deviations from the above mentioned symmetry. These patterns are reliable defined (Chapter 5) through the harmonic analysis of the linear displacement components of points defined at head, shoulder, and pelvis level during walking (Cappozzo, 1981; Pecoraro et al., 2004). This method can be used to describe the age-related differences on the mechanisms of coordination ruling the movements of the upper body in order to control head stability in older individuals. In addition the acceleration patterns can be observed to analyse the stability during walking as previously proposed by other authors (Menz et al., 2003; Kavanagh et al., 2005). Focusing the attention on the aging process, many studies suggested that age-related changes in the overall walking patterns can be interpreted as associated with the search for a safer gait (Woollacott et al., 1997; Prakash and Stern, 1973; Sudarsky, 1990; Sudarsky and Ronthal, 1992). Older individuals walk with conservatively lower pelvis accelerations than observed in younger people, however head accelerations are higher, thus suggesting a decline in the ability to control the movement of the upper body (Winter, 1991; Kavanagh et al., 2004). An insight in the age-related differences in the coordination of the upper part of the body segments is reported in Chapters 2, 3 and 5.

The aim of this study was, therefore, to examine the effect of normal ageing on the pattern of head, trunk, and pelvis mechanics during walking at different speeds of progression. The hypothesis to be tested is that the young and healthy elderly subjects would exhibit similar patterns of head accelerations, reflecting the essential need to maintain head control during

[3] Preliminary results have been published in abstract form in Pecoraro et al. (2004) and personally presented at the *SIAMOC 2003*.

walking. However, as anticipated in Chapter 2 the ageing process, which is generally associated with changes in lower limb gait dynamics, may impact on the dynamics of the mechanics of the lower trunk (i.e. pelvis).

6.2 Materials and methods

6.2.1 Subjects and protocol

Twenty young (age: 24±4 years; height: 1.66±0.05 m; weight: 57.7±7.1 kg) and twenty elderly healthy women (72±4 years, 1.54±0.06 m, 64.5±7.9 kg) volunteered for the study and signed an informed written consent. Subjects were asked to walk along the linear pathways at two different self-selected speeds of progression: natural (NS, “walk naturally”) and fast (FS, “walk as fast as you can”). In all cases the subjects were explicitly asked to reach and maintain a constant speed of progression. Five trials were performed for each condition.

6.2.2 Instrumentation

The stride period (T) was determined using the method described in Chapter 4. The beginning (t_b) of each stride was measured using an instrumented mat; the relevant end of the stride cycle was then computed as the time instant for which the index J reached the minimum value (= J_{min}) is used to determine the end ($\hat{t}_e$) of the pseudo-period $\hat{T} = \hat{t}_e - t_b$.

A stereophotogrammetric system (9-camera VICON 612®) was used to reconstruct the trajectories of 8 markers used to define the model proposed in chapter 3: anterior and posterior superior iliac spines, jugular notch, C7 spine process, front and back of the head. Stereophotogrammetric data and mat input signals were acquired at 120 frames/second.

6.3 Data analysis

6.3.1 Harmonic analysis

The harmonic analysis of the trajectories of the points at the head, shoulder, and pelvis levels was used to identify the intrinsic and the extrinsic patterns of upper body movements as described in Chapter 5. In addition to the analysis performed in Section 5.2.3 the velocity and the accelerations of the upper body segments were obtained using the following equations:

$$\begin{cases} X'(t) = \sum_{j=1}^{N_x} j\omega_0 a_{xj} \cos\left(j\omega_o t + \varphi_{xj}\right) & 0 \le t \le T \\ Y'(t) = \sum_{j=1}^{N_y} j\omega_0 a_{yj} \cos\left(j\omega_o t + \varphi_{yj}\right) & 0 \le t \le T \\ Z'(t) = \sum_{j=1}^{N_z} j\omega_0 a_{zj} \cos\left(j\omega_o t + \varphi_{zj}\right) & 0 \le t \le T \end{cases} \tag{6.1}$$

$$\begin{cases} X''(t) = -\sum_{j=1}^{N_x} \left(j\omega_0\right)^2 a_{xj} \sin\left(j\omega_o t + \varphi_{xj}\right) & 0 \le t \le T \\ Y''(t) = -\sum_{j=1}^{N_y} \left(j\omega_0\right)^2 a_{yj} \sin\left(j\omega_o t + \varphi_{yj}\right) & 0 \le t \le T \\ Z''(t) = -\sum_{j=1}^{N_z} \left(j\omega_0\right)^2 a_{zj} \sin\left(j\omega_o t + \varphi_{zj}\right) & 0 \le t \le T \end{cases} \tag{6.2}$$

The estimation of the displacement (Equation 5.1), velocity (Equation 6.1), and acceleration (Equation 6.1) and the relative harmonic analysis were thus obtained. It is evident that, in this way, only the acceleration associated with the relatively large displacements could be assessed as is the case with the low frequency vibrations.

The validity of the acceleration obtained using the double differentiation of the trajectories of the markers was assessed by comparing its data to those simultaneously acquired with a tri-axial accelerometer. To compared the acceleration obtained using the double differentiation of the trajectories of the markers and the acceleration directly measured using a tri-axial accelerometer

A solid state capacitive readout triaxial linear accelerometer (Analog Devices ADXL202E) with a marker fixed on top of it was used to measured the linear accelerations the trajectory respectively, along the antero-posterior, medio-lateral, and vertical directions. The accelerometer was secured on to an elastic band worn by the subject in a way that the three axes were aligned close to the three anatomical axes. In addition to calibrate the accelerometer before each testing session, it was placed with each of the orthogonal axes vertically, first pointing up, then down, which enabled the device to be statically calibrated to estimate the ±1 g values. Special care was taken in order to place the instrument centrally on the lumbar spine at the level of the third lumbar vertebrae where the opposing rotations of the thorax and pelvis most efficiently neutralise each other (Moe-Nillsen, 1998). The accelerometer was connected to a battery driven portable A/D converted which was transmitted to a portable PC via a bluetooth connection. Acceleration signals were acquired using purpose-written software and saved on computer for subsequent analyses sampled at 100 frames per seconds. Thus, in order to compare the two measures the trajectories of the marker were resampled at 100 frames per second. Data obtained from the two systems simultaneously were collected during sixty level walking trials performed by one subject (male, 28 years, 1.63 m, 60 kg) walked at him self selected comfortable walking speed.

The linear accelerations in the global system was derived from those in the sensor (local) coordinate system. This was done projected from the sensor coordinate system into the global system based on the sensor orientation output multiplying the orientation vector of the accelerometer by the linear local accelerations. The marker accelerations were then obtained using the double differentiations of the relevant harmonics (i.e. the first two harmonics along all directions), and the accelerations measured with the accelerometer were reconstructed using the same method, as reported for the trajectories in Chapter 5.

An example of the acceleration patterns obtained using the accelerometer and using the stereophotogrammetric system within one stride is reported in Figure 6.1.

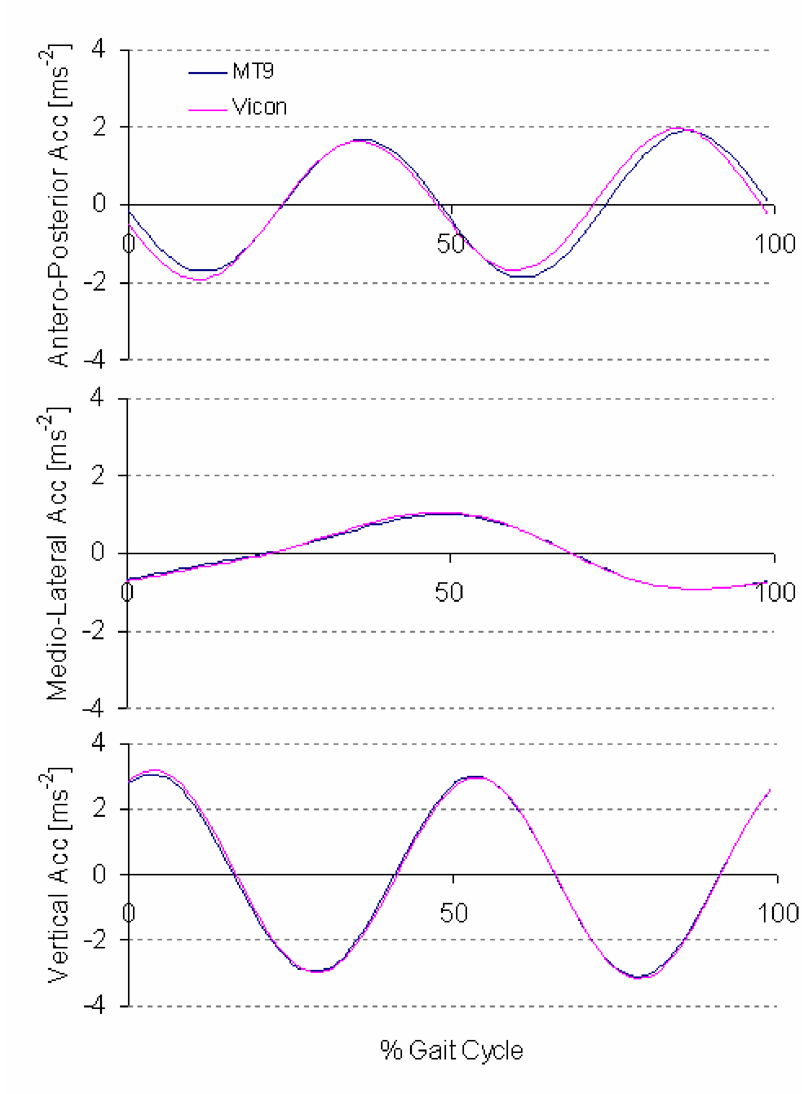

Figure 6.1. Example of the time patterns of the acceleration assessed using the double differentiation of the marker trajectories (Vicon, pink line) and measured using the accelerometer (MT9, blue line) during one gait trial. The data along the abscissa are normalised with respect to the duration of the walking cycle.

In the reported figure the acceleration RMS values were similar using the two methods (MT9: 1.31 ms^{-2}, Vicon: 1.35 ms^{-2}; MT9: 0.80 ms^{-2}, Vicon: 0.64 ms^{-2}; MT9: 2.11 ms^{-2}, Vicon: 2.07 ms^{-2} computed along the AP, ML and V directions, respectively). The time curves of the acceleration patterns obtained using the stereophotogrammetric system and the accelerometer resulted highly correlated as reported by the Pearson correlation coefficients (AP: r = 0.88±0.04; ML: r = 0.84±0.07; V: r = 0.98±0.01) and very similar to each other as shown by the high coefficients of determination between the two acceleration RMS (AP: R^2 = 0,73; ML: R^2 = 0.35; V: R^2 = 0.92).

From the acceleration data the root mean square was derived (aRMS). The aRMS is a measure of dispersion of the data relative to zero, as opposed to the standard deviation, which is a measure of dispersion relative to the mean. However, as the acceleration signals were transformed to give a mean equal to zero, the RMS in this study is synonymous to the standard deviation. This value was used to provide an indication of the average magnitude of accelerations in each orthogonal plane during a complete walking trial.

In addition three coefficients of attenuation were defined as follow:

- Trunk-to-Head attenuation: $\text{Tr-to-He}=\dfrac{aRMS_{SH}-aRMS_{HE}}{aRMS_{SH}}$
- Pelvis-to-Trunk attenuation: $\text{Pe-to-Tr}=\dfrac{aRMS_{PE}-aRMS_{SH}}{aRMS_{PE}}$
- Pelvis-to-Head attenuation: $\text{Pe-to-He}=\dfrac{aRMS_{PE}-aRMS_{HE}}{aRMS_{PE}}$

where $aRMS_{HE}$, $aRMS_{SH}$, and $aRMS_{PE}$ are the acceleration RMS at the head, shoulder and pelvis levels.

6.3.2 Statistical analysis

A two-way ANOVA analysis was used to assess the effects of two between group factors: speed (two levels: NS, and FS) and age (two levels: young group, YG, and elderly group, EG). When significant differences ($p<0.05$) were found, a post-hoc analysis was performed using an unpaired samples two-tailed t-test with Bonferroni correction (significance level: $p = 0.025$).

6.4 Results

The young group walked at higher speed of progression both at comfortable and fast speeds ($p<0.01$). The increased in walking speed are accomplished by increases in both cadence ($p<0.05$) and step length ($p<0.05$). No differences were found for the stride width at both comfortable and fast speed (Table 6.1).

	YG		EG	
	CS	FS	CS	FS
SL [m]	1.39 (0.15)^	1.61 (0.15)^	1.14 (0.17)	1.25 (0.15)
SF [s^{-1}]	0.93 (0.12)^	1.47 (0.18)^	0.85 (0.09)	1.30 (0.18)
WS [ms^{-1}]	1.30 (0.30)^	2.32 (0.21)^	0.97 (0.17)	1.59 (0.17)
Fn	0.20 (0.10)^	0.63 (0.11)^	0.12 (0.04)	0.31 (0.06)
SW	0.18 (0.04)	0.18 (0.03)	0.20 (0.03)	0.19 (0.03)

Table 6.1. Mean (standard deviation) of the temporo-spatial parameters for both groups performed at comfortable and fast speed (^ differences between young and elderly group).

In all directions and at all body levels, the ratio between the sum of the power of the first four harmonics and the total power was higher than 99%. In all analysed strides, the amplitude of the harmonics of order higher than four was lower than 0.7 mm. Since the spot check showed that the stereophotogrammetric system had an accuracy in the order of 1.5 mm, the latter harmonics were associated with the experimental error. The third and fourth harmonics had amplitudes always lower than 2 mm. Since this figure alone did not provide evidence that these harmonics were reliable, the relevant phases were also analysed. Their values, as observed within subjects, were scattered in intervals close to 2π rad. Both observations led to the association of these harmonics to experimental errors. From now onwards, the first two harmonics in all directions will be taken into consideration as carrying significant information.

Table 6.2 shows the results of the ANOVA performed on the parameters computed in the antero-posterior direction. According to the ANOVA analysis, speed per se affected all the parameters but the first harmonic amplitude at shoulder level and the ratio between the harmonics at pelvis level. Age influenced the first harmonic at HE and SH level, and the second harmonic at PE level. Finally, the interaction between the two was significant only at shoulder level, for the first harmonic and the harmonic ratio.

AP		**Age**		**Speed**		**Age x Speed**	
		F	**p**	**F**	**p**	**F**	**p**
HE	I	2.0	0.16	2.2	0.14	0.6	0.45
	II	6.4	*0.01*	17.3	*0.00*	0.6	0.45
	Ratio	0.3	0.56	11.7	*0.00*	4.1	0.05
	RMS	0.0	0.91	55.3	*0.00*	0.0	0.97
SH	I	4.6	*0.03*	0.0	0.93	6.2	*0.02*
	II	0.3	0.59	41.3	*0.00*	0.5	0.47
	Ratio	0.1	0.76	16.9	*0.00*	6.0	*0.02*
	RMS	20.3	*0.00*	35.6	*0.00*	2.6	0.11
PE	I	1.8	0.19	5.3	*0.02*	2.0	0.17
	II	4.6	*0.04*	57.9	*0.00*	1.3	0.26
	Ratio	6.1	*0.02*	2.6	0.11	2.3	0.13
	RMS	43.3	*0.00*	159.9	*0.00*	2.0	0.16

Table 6.2. Results of the ANOVA performed along the antero-posterior direction obtained for the harmonic amplitudes, harmonic ratio and RMS of the acceleration pattern.

Table 6.3 shows, for all the body levels, the amplitude of the informative harmonics and the harmonic ratio of the trajectories recorded along the antero-posterior direction for both groups. The second harmonic amplitude decreased with increasing speed, whereas the RMS increased with speed, at all levels.

	AP	YG		EG	
		CS	FS	CS	FS
HE	I [mm]	6.3 (2.5)	5.9 (1.8)	7.5 (2.4)	6.3 (2.1)
	II [mm]	4.1 (1.2)*	2.3 (0.9)^	4.7 (2.0)*	3.5 (1.7)
	Ratio	0.7 (0.4)*	0.4 (0.2)	0.7 (0.2)	0.6 (0.2)
	RMS [ms^{-2}]	0.42 (0.12)*	0.65 (0.21)	0.41 (0.14)*	0.71 (0.25)
SH	I [mm]	7.1 (3.5)	9.0 (5.5)^	7.4 (2.3)	5.3 (1.5)
	II [mm]	8.1 (1.8)*	4.2 (2.3)	7.4 (2.6)*	4.3 (2.4)
	Ratio	1.4 (0.7)*	0.6 (0.4)	1.0 (0.4)	0.9 (0.4)
	RMS [ms^{-2}]	0.82 (0.37)*^	1.14 (0.42)^	0.61 (0.20)*	0.80 (0.29)
PE	I [mm]	7.3 (2.1)	6.8 (2.8)	8.8 (1.9)*	6.8 (2.5)
	II [mm]	15.0 (2.9)*^	9.7 (2.3)	13.0 (2.7)*	9.0 (2.3)
	Ratio	2.2 (0.8)^	1.7 (0.9)	1.5 (0.4)	1.5 (0.7)
	RMS [ms^{-2}]	1.41 (0.23)*^	2.31 (0.49)^	1.06 (0.24)*	1.70 (0.43)

Table 6.3. Mean (standard deviation) of the harmonic amplitudes, harmonic ration and acceleration RMS performed along the antero-posterior direction (* differences between comfortable and fast speed; ^ differences between young and elderly group).

Higher amplitude of the second harmonic ($p<0.01$) and lower values of the RMS ($p<0.01$) were found at comfortable speed than at fast speed at all levels and for both groups. The young group exhibited higher harmonic ratio at comfortable speed than at fast speed ($p<0.01$) at the shoulder and head levels. The elderly group showed higher amplitude of the first harmonic at comfortable speed than at fast speed ($p<0.01$) at pelvis and shoulder levels. At pelvis level, at comfortable speed, YG exhibited higher oscillations of the second harmonic than EG ($p<0.01$), whereas no differences were found at fast speed. As opposite, at the head level, at fast speed YG exhibited lower oscillations of the second harmonic than EG ($p<0.05$). At the pelvis and shoulder levels, at both speeds, YG showed higher RMS accelerations than EG, whereas no differences were found at head level. A significant increase ($p<0.001$) in the amplitudes of the second harmonic and in the RMS acceleration were obtained going from head to shoulder and pelvis level at both speeds and for both groups.

Figure 6.2 shows the coefficients of attenuations computed for both groups and at both speeds.

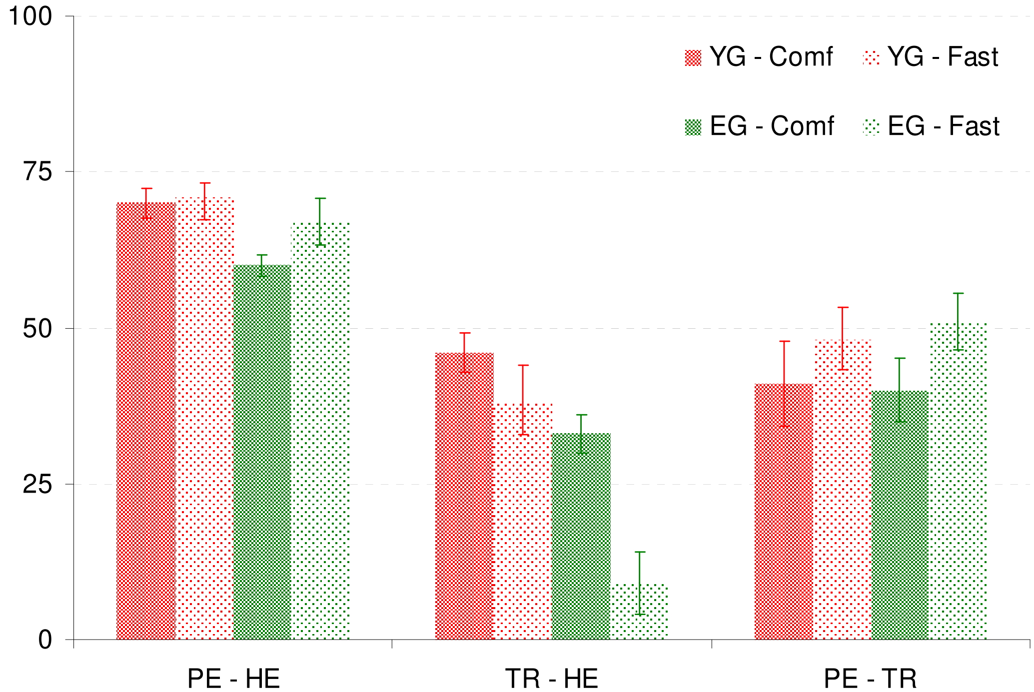

Figure 6.2. Coefficients of attenuation performed along the antero-posterior directions for the young and elderly groups at both comfortable and fast speed.

Both groups showed an attenuation of the RMS between pelvis and trunk and further attenuated between trunk and head, at both speeds. The group factor effected TR-to-HE ($F=15.8$, $p<0.05$) and PE-to-HE ($F=37.5$, $p<0.05$) coefficients whereas no effects where found for PE-to-TR coefficient. The speed factor and the interaction between speed and group did not affect the coefficients of attenuation. Higher attenuation were found for YG both from pelvis to head ($p<0.05$) and from trunk to head ($p<0.05$), at both speeds.

Table 6.4 shows the results of the ANOVA performed on the medio-lateral direction. As for the AP direction, speed influenced more parameters than age did. The interaction between two factors was significant only for the first harmonic (at all body levels) and for the acceleration RMS (SH and PE).

ML		**Age**		**Speed**		**Age x Speed**	
		F	**p**	**F**	**p**	**F**	**p**
HE	I	20.8	*0.00*	55.0	*0.00*	4.4	*0.04*
	II	3.6	0.06	0.0	0.89	0.1	0.82
	Ratio	3.7	0.06	26.4	*0.00*	0.9	0.34
	RMS	1.4	0.24	47.9	*0.00*	0.6	0.45
SH	I	20.5	*0.00*	66.0	*0.00*	5.5	*0.02*
	II	0.4	0.53	0.3	0.60	0.7	0.40
	Ratio	3.5	0.07	18.8	*0.00*	1.7	0.19
	RMS	9.3	*0.00*	121.8	*0.00*	10.3	*0.00*
PE	I	0.4	0.53	42.6	*0.00*	5.5	*0.02*
	II	0.4	0.52	0.0	0.88	2.0	0.16
	Ratio	0.0	0.95	13.2	*0.00*	0.6	0.42
	RMS	41.9	0.00	96.6	0.00	27.8	0.00

Table 6.4. Results of the ANOVA performed along the medio-lateral direction obtained for the harmonic amplitudes, harmonic ratio and RMS of the acceleration pattern.

Table 6.5 shows, for all the body levels, the amplitude of the informative harmonics and the harmonic ratio of the trajectories recorded along the medio-lateral direction for both groups. The first harmonic amplitude and the harmonic ratio decreased with increasing speed in both groups ($p<0.01$), and RMS increased with speed, at all levels. The first harmonic amplitude at HE and SH levels increased with age, and the RMS at HE and SH levels, at comfortable speed, increased with age, whereas at PE level, at fast speed, decreased with age.

ML		YG		EG	
		CS	FS	CS	FS
HE	I [mm]	15.4 (4.8)*^	9.4 (3.9)^	23.0 (6.2)*	12.2 (3.8)
	II [mm]	1.4 (0.4)	1.4 (0.5)	1.7 (0.5)	1.7 (0.8)
	Ratio	12 (4.7)*	7.1 (3.4)	15.4 (7.2)*	8.2 (3.0)
	RMS [ms^{-2}]	0.39 (0.11)*^	0.67 (0.23)	0.49 (0.12)*	0.65 (0.22)
SH	I [mm]	16.3 (5.4)*^	9.8 (3.3)^	24.2 (6.4)*	12.2 (3.3)
	II [mm]	1.3 (0.4)	1.4 (0.7)	1.5 (0.5)	1.4 (0.6)
	Ratio	13.6 (4.7)*	8.7 (7.5)	18.8 (9.6)*	9.6 (3.8)
	RMS [ms^{-2}]	0.40 (0.06)*^	0.70 (0.29)	0.50 (0.11)*	0.63 (0.19)
PE	I [mm]	18.6 (5.3)*	13.3 (5.1)	22.4 (7.0)*	11.2 (3.1)
	II [mm]	1.7 (0.4)	1.9 (0.6)	2.1 (1.2)	1.8 (0.7)
	Ratio	11.4 (3.5)*	8.2 (5.7)	12.4 (6.0)*	7.3 (3.8)
	RMS [ms^{-2}]	0.48 (0.18)*	0.94 (0.29)	0.48 (0.14)*^	0.63 (0.17)

Table 6.5. Mean (standard deviation) of the harmonic amplitudes, harmonic ration and acceleration RMS performed along the antero-posterior direction (* differences between comfortable and fast speed; ^ differences between young and elderly group).

A significant increase ($p<0.001$) was found for the young group, at both speeds, in the amplitudes of the first harmonic and in the acceleration RMS when going from head to shoulder and pelvis level. For both groups, at fast speed, the pelvis and shoulder accelerations were higher than that of the head. No differences were found between the trunk and the pelvis accelerations.

Figure 6.3 shows the coefficients of attenuation computed for both groups and at both speeds.

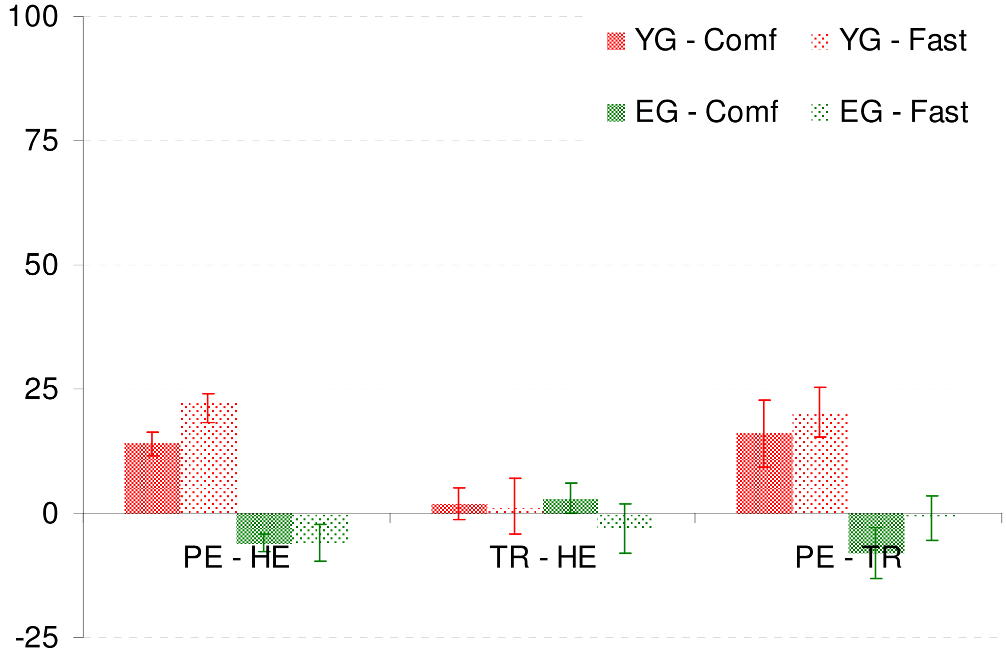

Figure 6.3. Coefficient of attenuation performed along the medio-lateral directions for the young and elderly groups at both comfortable and fast speed.

Both group and speed factors effected the PE-to-TR (group: F=18.2, p<0.01; speed: F=7.9, p<0.01), TR-to-HE (group: F=17.0, p<0.01; speed: F=8.2, p<0.01) and PE-to-HE (group: F=25.9, p<0.01; speed: F=9.8, p<0.01) coefficient of attenuation. Higher pelvis to trunk and trunk to head attenuations were observed for the young group than that found for the elderly group. In addition, these attenuations were higher at comfortable speed than at fast speed, for both young and elderly groups.

Table 6.6 shows the results of the ANOVA relevant to the vertical direction. The age factor affected all the parameters but the amplitude of the first harmonic at HE and PE level and the ratio between the two harmonics at SH level. The speed factor affected all the parameters but the second harmonic amplitude at PE and SH level. Finally, the interaction between the two factors was significant only for the RMS at PE and HE level.

V		Age		Speed		Age x Speed	
		F	p	F	p	F	p
HE	I	0.5	0.48	6.2	*0.02*	0.1	0.76
	II	8.8	*0.00*	4.7	*0.03*	0.9	0.34
	Ratio	6.7	*0.01*	7.1	*0.01*	1.1	0.29
	RMS	47.1	*0.00*	77.4	*0.00*	4.9	*0.03*
SH	I	3.2	0.08	9.2	*0.00*	1.0	0.31
	II	5.5	*0.02*	2.5	0.12	1.9	0.17
	Ratio	0.6	0.43	7.9	*0.01*	2.2	0.14
	RMS	38.4	*0.00*	90.0	*0.00*	3.4	0.07
PE	I	0.1	0.78	7.1	*0.01*	0.0	0.97
	II	7.5	*0.01*	0.5	0.49	0.6	0.43
	Ratio	5.6	*0.02*	5.3	*0.03*	1.9	0.18
	RMS	42.4	*0.00*	108.1	*0.00*	8.4	*0.01*

Table 6.6. Results of the ANOVA performed along the vertical direction obtained for the harmonic amplitudes, harmonic ratio and RMS of the acceleration pattern.

Table 6.7 shows, for all the body levels, the amplitude of the informative harmonics and the harmonic ratio of the trajectories recorded along the vertical direction for both groups.

	V	YG		EG	
		CS	**FS**	**CS**	**FS**
HE	I [mm]	4.3 (1.3)	5.5 (2.1)	4.7 (1.4)	5.7 (2.3)
	II [mm]	16.4 (4.0)^	13.1 (5.5)	12.2 (3.1)	10.9 (5.4)
	Ratio	4.1 (1.3)^	2.8 (1.8)	2.8 (1.2)	2.3 (1.6)
	RMS [ms^{-2}]	1.67 (0.86)*^	2.92 (0.78)^	1.01 (0.31)*	1.93 (0.66)
SH	I [mm]	4.3 (1.7)	5.9 (2.4)	4.0 (1.2)	4.8 (1.5)
	II [mm]	16.5 (4.2)^	13.1 (6.0)	12.2 (2.9)	12.0 (5.7)
	Ratio	4.4 (1.8)*	2.6 (1.6)	3.4 (1.7)	2.9 (1.9)
	RMS [ms^{-2}]	1.67 (0.85)*^	2.91 (0.78)^	1.01 (0.32)*	2.14 (0.73)
PE	I [mm]	4.3 (1.6)	5.5 (1.8)	4.4 (1.3)	5.6 (2.8)
	II [mm]	16.2 (4.4)^	14.4 (6.2)	12.0 (2.8)	12.1 (6.1)
	Ratio	4.3 (1.8)*^	2.9 (1.6)	2.9 (0.9)	2.6 (1.8)
	RMS [ms^{-2}]	1.66 (0.92)*^	3.20 (0.83)^	0.99 (0.32)*	2.17 (0.84)

Table 6.7. Mean (standard deviation) of the harmonic amplitudes, harmonic ration and acceleration RMS performed along the antero-posterior direction (* differences between comfortable and fast speed; ^ differences between young and elderly group).

At shoulder and head levels, the young group showed higher values of the second harmonic at comfortable speed than at fast speed ($p<0.01$), whereas no significant differences where found for the elderly group. Moreover, the young group exhibited higher harmonic ratio at comfortable speed than at fast speed at all levels. The RMS values were lower at comfortable than at fast speed for both group and at all levels. At comfortable speed, the young group showed higher amplitude of the second harmonic than the elderly group ($p<0.01$) at all levels. The RMS values were higher ($p<0.01$) for the young group than for the elderly group at all levels and at both speeds. All levels had similar amplitude of the two harmonic and of the harmonic ratio.

Figure 6.4 shows the coefficients of attenuation computed for both groups and at both speeds.

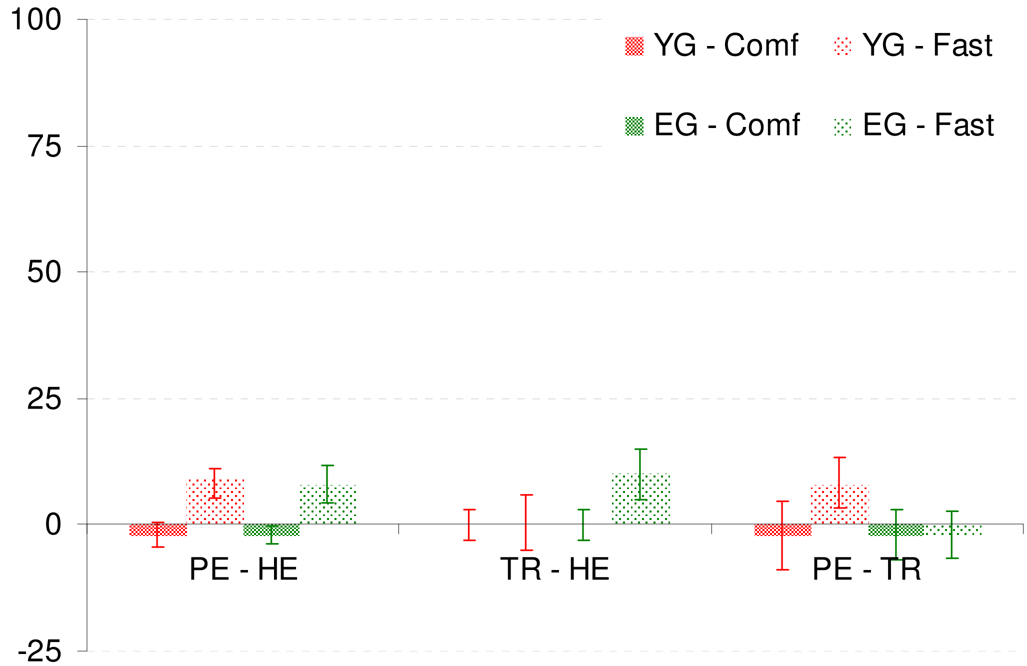

Figure 6.4. Coefficient of attenuation performed along the vertical directions for the young and elderly groups at both comfortable and fast speed.

The group factor effected only the PE-to-TR coefficient ($F=4.4$, $p<0.05$), whereas the speed factor effected the PE-to-TR ($F=26.4$, $p<0.01$), TR-to-HE ($F=33.2$, $p<0.01$) and PE-to-HE ($F=49.4$, $p<0.01$) coefficients. The interaction between the two factors effected only the PE-to-TR coefficient ($F=5.6$, $p<0.05$). Pelvis to trunk attenuation was higher for the young group than for the elderly group. In both groups these attenuations were higher at fast speed ($p<0.01$) than at comfortable speed.

6.5 Discussion

This study examined the effect of normal aging process on the coordinate mechanism among the upper part of the body segments during walking on a level surface. In particular the aim of this study was to determine whether the pattern and the structure of the head mechanics decline with age, and to what extent the trunk segment is involved in stabilising the head for young and elderly individuals.

Compared to the younger group, older subjects exhibited significantly reduced velocity, step length and increased stride period. These results are consistent with previous reports (Imms and Edholm, 1981; Cunningham et al., 1982; O'Brein et al., 1983; Hagemon and Blanke, 1986; Oberg et al., 1983; Lord et al., 1996; Bohannon, 1997; Eble et al., 1991; Hausdorff et al., 1997) and represent the characteristically 'cautious' gait pattern commonly observed in older people.

The harmonic analysis of the displacement of three points located at the head, shoulder, and pelvis level was used to define the intrinsic and extrinsic patterns as described in Chapter 1. Using this method the fraction between the harmonic describing the intrinsic pattern and the harmonic describing the extrinsic pattern was used to analyse the "smoothness" of the acceleration patterns (Menz et al., 2003b; Kavanagh et al., 2005). In addition, the acceleration was computed via a double differentiation of the relevant harmonics and its RMS was used as a magnitude of the acceleration at all body levels. These parameters provide a more direct indication of walking stability than temporo-spatial parameters of gait.

The results of the present study indicate that both accelerations and displacements at the head level were attenuated with respect to the trunk and to the pelvis levels. This finding is consistent with that reported in previous investigations (Kavanagh et al., 2005; Menz et al. 2003b; Winter 1991; Winter et al. 1990). In agreement with the literature (Winter 1991; Menz et al., 2003b; Kavanagh et al., 2005) the magnitude of the accelerations increased with increasing speed, at all body levels. This velocity dependence is problematic when investigating the effects of ageing on gait-related accelerations, as it is difficult to determine whether differences in acceleration patterns are due to gait velocity, or the ageing process. The result reported in this study are in agreement with the general view that the walking speed declines with age

(Himann et al., 1988; Imms and Edholm, 1981; Lord et al., 1996; Dobbs et al., 1993; Bohannon et al., 1997M Winter et al., 1990). Clearly, walking speed needs to be taken into account when comparing acceleration RMS patterns between subjects who may be walking at different speeds. For example, since the young subjects walk at different speed than the elderly subjects, it would not be appropriate to directly compare their acceleration RMS values as a measure of head stability, as the younger subject, presumably with better head stability, experiences larger head accelerations than the older subject simply because they are walking more quickly. However, when studying the pattern of the acceleration RMS two different factors influenced the magnitude of the acceleration: the square of the stride frequency (as reported in the materials and method section) and the amplitude of the relevant harmonics of the displacement that composed the acceleration signals. One of the results of the present work is that an increasing in walking speed and thus in stride frequency is accompanied by a reduction of the amplitude of the intrinsic harmonic. This behaviour suggests that, when an individual is asked to perform the locomotor act at maximum speed, since the stride frequency is higher than at his/her comfortable speed, he/she reduces the magnitude of the trajectory oscillations in order to attenuate the magnitude of the accelerations recorded at the upper body segments. Due to this behaviour the acceleration RMS of the upper body segments is lower than that if the trajectory oscillations were similar at comfortable and fast speeds. This reduction permits to preserve the systems (i.e. visual and vestibular) located in the head by excessive accelerations.

During walking the elderly group had the acceleration RMS at the pelvis level generally smaller than that found in the young group. It can be attributed to the reduction in walking speed, and thus it is possible that older subjects adopt this slower speed to keep the magnitude of the upper body accelerations at a tolerable level. However, when analysing the intrinsic harmonic it revealed that the elderly subjects had either lower pelvis or higher head displacement with respect to the young subjects, especially in the antero-posterior and medio-lateral directions. In addition, the lower pelvis accelerations recorded for the elderly group coupled with the magnitude of the head accelerations which did not differ between the two groups at both speeds, suggesting that the elderly subjects may have some difficulty in attenuating the acceleration at the head level during walking. This results is particularly highlighted by the coefficients of attenuation: the attenuation between the pelvis and the trunk and between the trunk and the head revealed the difficulty for the elderly subject to attenuate the head accelerations with respect to the young subjects in particular along the medio-lateral direction.

Despite their impaired physiological capabilities the elderly subjects exhibited the same degree of smoothness in their trajectories patterns, as evident by the lack of differences between the two groups with regard to the harmonic ratios. Previous studies have reported that older people with balance problems have smaller harmonic ratios than those without balance difficulties (Menz et al., 2003b; Yack and Berger, 1993). In this study the harmonic ratio is computed differently than in those papers: in particular, dividing the amplitude of the intrinsic harmonic by the amplitude of the extrinsic harmonic. The harmonic ratio proposed in literature is defined using the same criteria but it is calculated by dividing the sum of the amplitudes of the even harmonics by the sum of the amplitudes of the odd harmonics (as opposite along the medio-lateral direction the harmonic ratio is calculated as the sum of the amplitudes of the odd harmonics divided by the sum of the amplitudes of the even harmonics). Since, in this study, as reported there is a tendency of the intrinsic harmonic amplitude to be higher at comfortable speed than at fast speed in both groups whereas no speed related differences where found for the extrinsic pattern there is a tendency for the harmonic ratio to be higher at comfortable speed than at fast speed. Given that this variable is a useful indicator of overall gait stability (Menz et al., 2003b; Yack and Berger, 1993), it can be concluded that both groups at fast speed reduced their smoothness of the trajectories of the upper body segments.

6.6 Conclusions

In conclusion, this study has given an insight into some of the mechanisms of coordination ruling the movements of the upper body in order to control head stability in young and elderly subjects. The reported results has shown that elderly subjects modify their basic gait pattern to stabilise their head with respect to the base of support identified by the pelvis. This is particularly evident in transverse plane where the trunk is able to reduce both the accelerations and the displacement of the head with respect to those found at the pelvis level. Thus, in elderly subjects without any particular deficit and with optimum physiological abilities, reduced walking speed, stride length and stride frequency are accompanied by a reduction in the pelvis oscillations and accelerations is helpful to maintain the stability of the head both at comfortable and at fast speed.

CHAPTER 7 Conclusions

The general objective of this thesis was to assess the locomotor ability of an individual through a detailed analysis of the upper part of the body mechanics during walking on a level surface. For this purpose the strategy adopted by young and elderly subjects has been observed in terms of: maintenance of walking stability and symmetry and simplicity of the body movements. With reference to the hypotheses and to the aims of this thesis, the following conclusions can be reported:

- a detailed analysis of the upper body mechanics can be used to provide useful information to assess the individual's locomotor ability;
- the analysis of level walking pseudo-periodicity showed that young healthy adult human gait is pseudo-periodic, and this is more marked for the upper part of the body. Moreover, it has been proven that a control of aperiodicity should always be performed in common gait laboratories;
- the harmonic analysis proved to be a reliable tool for identifying the intrinsic and extrinsic patterns of gait in older women. The intrinsic pattern, which is characteristic of the whole investigated population of older women, exhibited excellent intra- and inter-subject repeatability (coefficient of multiple correlation, CMC, ranging between 0.82 and 0.99). The extrinsic pattern, which is characteristic of each individual within the population of older women performing the locomotor act, was excellently intra-subject repeatable (CMC ranging between 0.70 and 0.90). The inter-subject repeatability of the extrinsic pattern was smaller in magnitude, thus indicating that upper body movements are unique to each individual;
- the comparison between the elderly and the young groups highlighted the different motor strategies adopted by the two groups

in controlling the overall stability of the upper body. As expected, walking speed was reduced for the elderly subjects and was associated with lower oscillations at pelvis level, especially in the antero-posterior and medio-lateral directions. On the contrary, no significant differences were found between the two groups at head level, suggesting that both groups were able to control the stability of this segment. This hypothesis was strengthened by the results of the accelerations analysis. In both groups, lower accelerations were recorded at head level than at pelvis level; whereas no differences were found between the two groups at head level, the elderly subjects expressed lower accelerations at pelvis level.

In conclusion, the results of this thesis showed that the harmonic analysis of the movement at the head, shoulder, and pelvis levels reliably describes upper body kinematics in elderly individuals. This analysis provides elements for detecting the intrinsic and extrinsic patterns of level walking. The detection of these patterns can provide information about the motor strategy adopted by a subject and can allow for the identification of physical impairments and for the description of their role in determining mobility limitations in older subjects.

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Appendices

A. Effects of speed training using a treadmill with body weight unloading on temporo-spatial parameters of gait in healthy older women [4]

A.1. Introduction

A slowing of gait, caused mainly by a prolongation of the double support phase, is commonly reported in older individuals, which has been related to an increased risk of falls (Winter, 1991). Few training interventions have been specifically designed to improving gait. Walking on a treadmill with an apparatus of Body Weight Unloading (BWU) is an effective tool for the rehabilitation of patients with neurological and orthopaedic diseases (Gazzani et al., 1999). This tool might also be safely used for healthy older individuals to carry out specific training programmes designed to increase walking speed. The higher speed of movement, which is achieved on the treadmill with BWU without increasing the energy cost (Thomas et al., 2003), would represent a stimulus to the neuromuscular system to respond with an increase in walking speed also in the open field. The aim of this study was to investigate the adaptations in temporo-spatial characteristics of gait of older women on the ground following a 12-week speed training program on a treadmill with BWU.

A.2. Materials and methods

With ethics committee approval, 22 volunteers were randomly assigned to exercise group (EG, N=11, 79.6±3.7 years; mean ± S.D) and non-intervention control group (CG, N=11, 77.6±2.3 years). They were selected according to the exclusion criteria to define "medically stable" older participants for exercise studies, as proposed by Greig and colleagues (1990). During the first 6 weeks,

[4] These results are published, in an abstract form, in Thomas, Pecoraro et al., (2005) and have personally been presented at the SIAMOC 2005.

the EG performed walking interval-training on the treadmill with 40% of BWU at the speed corresponding to heart rate at Ventilatory Threshold (4 sets of 5 min with 1 min interval in between). BWU was then progressively reduced to 10% during the last 6 weeks of intervention. Before, in the middle and after training, gait analysis of each participants was performed using a six camera VICON 612® system while walking on an oval shaped 20-m walkway circuit (rectilinear for 8-m on each side) for 10 laps at their self-selected walking speeds (slow, comfortable and fast). Two strides were captured from the mid-portion of the rectilinear part of the walkway circuit during each lap. Stride duration (T) was computed as the elapsed time between sequential heel strikes of the same leg. The stride frequency (SF) was taken as 1/T and the stride length (SL) was computed as the antero-posterior displacement between sequential heel strikes of the same leg. Speed values (v) were computed as the product of SL and SF. In addition, gait analysis was carried out during a 6-m maximal speed walking test. Statistical comparisons of the parameters between groups (EG and CG) at three stages (before, in the middle and after training) were carried out by ANOVA for repeated measures, followed by Student's t-tests with Bonferroni adjustment where appropriate. Statistical significance was set at p=0.05.

A.3. Results and discussion

Following the 12-week intervention, the fast walking speed of the EG increased by 7% while no significant changes were measured in the CG (Figure A.1).

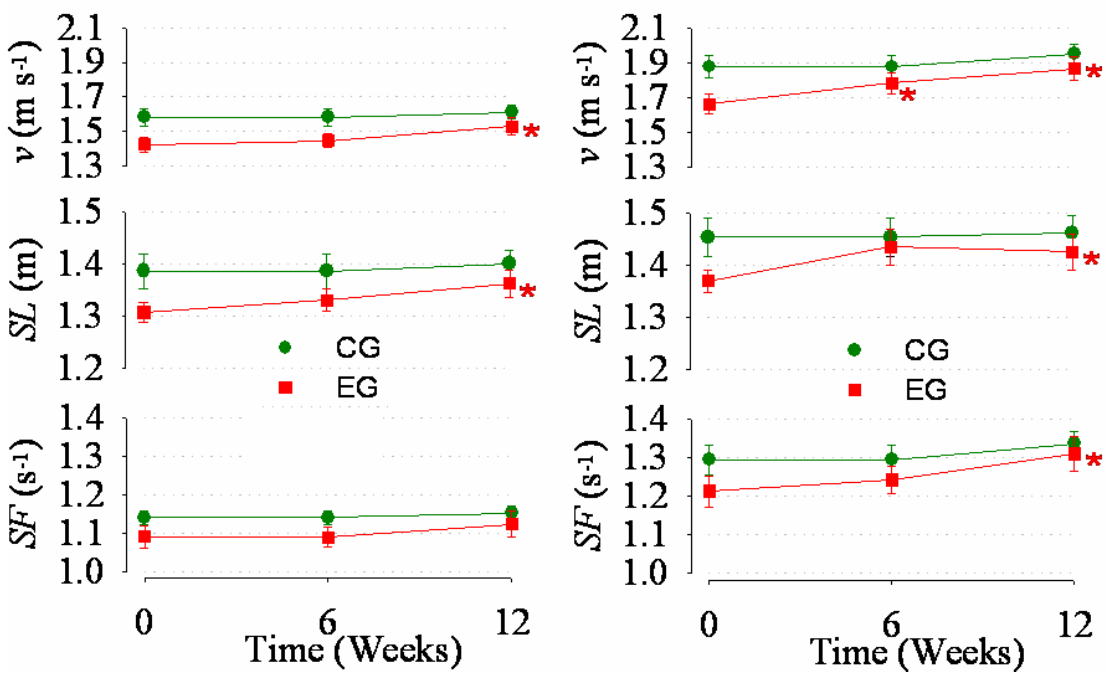

Figure A.1. Mean (± S.E.M.) *significantly different from week 0.

No significant changes occurred at slow and comfortable speeds. Maximal walking speed of EG significantly increased after 6 and 12 weeks by 7% and 12%, respectively, with no significant changes in the CG (Figure A.1). The increase in fast and maximal walking speed in the EG was accompanied by a significant increase in SL, which could be attributed to the specificity of the training programme and indicate an increased postural stability.

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B. A Neurofuzzy inference system based on biomechanical features for the evaluation of the effects of physical training [5]

B.1. Introduction

Individual task-related motor strategies change in relation to many modifications of the musculoskeletal system where a lack of physical activity can play a significant role. This is mainly true for elderly subjects, where the modifications may result in deteriorated walking pattern strategies such as reduced walking speed and stride length and increased step width (Winter et al, 1990). Similar limitations may also be pointed out in other motor acts that are considered important in daily living activities such as rising from a chair (Alexander at al, 1991). These limitations may result in a loss of functionality typical of the ageing process (Buchner et al, 1993) or of a disease.

Training programs entail important changes in chair-rise performance (Alexander et al, 2001), contributing to postpone the onset of functional limitations. The relevant improvement may be associated with changes in the execution of movements and in the related motor pattern. Evaluating these changes is an important task while assessing the benefits of a training process or of a rehabilitative treatment. This task requires finding an indicator of the expected enhanced motor performance.

The quantitative evaluation of physical training is usually provided by scores, based on specifically devised exercises involving a single joint. Conversely, the use of complex motor acts, as the Sit-to-Stand (STS) test, may enhance this evaluation by globally assessing the individual motor capacity using an appropriate subset of features. These features may be selected from mechanical quantities either measured or estimated in the movement analysis laboratory during the execution of a motor act. STS tasks have been already proposed for the evaluation of rehabilitative treatments (Kolbe-Alexander et

[5] These results are published, in an abstract form, in Vannozzi G., Pecoraro F., et al., (2003) and submitted for publication as a full paper in *Computer methods in biomechanics and biomedical engineering* (Vannozzi G., Pecoraro F., et al. 2007) .

al, 2006; Taaffe et al, 1999). The conclusions of these studies were mainly based on the analysis of one biomechanical feature at a time or, less frequently, using multivariate analyses based on features subjectively selected by the analyst. However, when a synthetic descriptive score is sought and when a lack of knowledge subsists about the biomechanical features to select, a global classification index that takes into account the appropriate feature set is desirable. This endeavour entails a non-trivial problem to be tackled, where the most convenient analytical method is searched for extracting useful information from motion data (Chau, 2001).

The goal of this work was the design, implementation and testing of a inference system, specifically devised for evaluating whether a subject benefits of a given lower limb physical training. Such a classifier receives as input a set of biomechanical features extracted from force-plate measurements carried out during the execution of STS trials. It allows for the evaluation of the training of a group of participants: the physical training is hypothesised successful when the sample group can be clustered in a way that participants can be separated from control subjects. Moreover, the inference system is designed to carry out the classification task when new participants are recruited.

B.2. Materials

The database used in this study included a sample of 31 female subjects (age 60-85 years old) divided into an experimental (EG) and a control group (CG). This division was based on an index of maximum muscular power obtained during an exercise of leg extension (Bassey et al, 1990) that mainly involved the action of quadriceps, the most stressed muscle during the foreseen training duration. Once the resulting power score was obtained, each subject was assigned to one of the two groups, following the goal of obtaining two homogeneous groups. In figure 1, such subdivision step is represented:

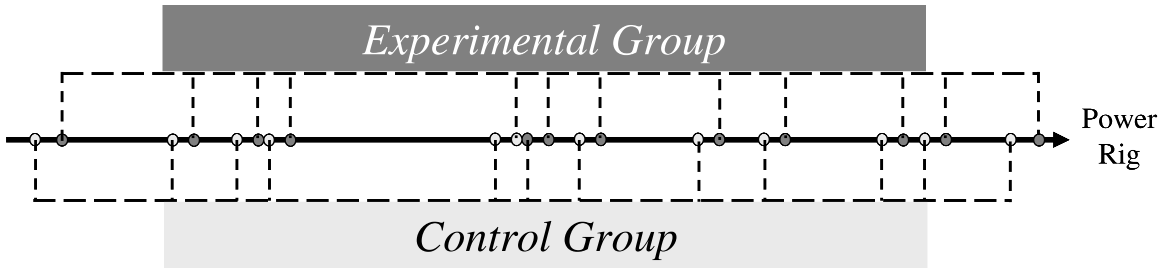


Figure B.1. Procedure of subdivision of the whole sample group into an experimental and a control subgroup by means of the power index

A set of quantitative evaluations of STS was carried out on the two groups. Each STS trial was executed at the maximum velocity and using a modular seat adjustable to subject's height. The experimental apparatus was composed of a six-component force-plate (Kistler 9281B) and a purposely implemented software for data acquisition (National Instrument LABVIEW).

Only the experimental group followed a heavy-resistance strength training cycle of 12 weeks, purposely designed for the lower limbs on cybex weight training machines. Such training aimed at increasing the muscular power of the lower limb. The training programme comprised sequences of dynamic exercises including concentric and eccentric muscle contraction on resistance machines. The training intensity was set at 75% of 1RM. A typical training session lasted about 75 minutes and included double leg press, double/single leg extension, double calf rise, double leg curl, and double seated leg press. At the end of the training period, both experimental and control groups underwent a further STS acquisition following the same protocol.

B.3. Methods

B.3.1 Data acquisition, modelling and feature selection

Ground reaction forces and moments, recorded from the forceplate, were obtained from both pre- and post-training STS acquisitions. When looking for differences due to physical training, however, no appreciable differences due to training were detected using raw data alone.

This circumstance suggested to consider the application of models of the musculoskeletal system, referred to as Minimum Measured Input Models (Cappozzo, 2002), that included the invariant aspects of both the modelled system and the motor task, enhancing the possibility to interpret data. The adopted model was the Telescopic Inverted Pendulum, TIP, model (Papa and Cappozzo, 1999), that describes human locomotor acts as characterised by a rotation about the base of support of a massless link, accompanied by a shortening or elongation of the musculoskeletal structure as a whole. This model takes into consideration only the kinematics of, and the dynamic actions on, the centre of mass (CM) of the body portion involved in the movement. Dealing with the STS test, a two TIP model was proposed that split the entire task in two phases: TIP1 which represents the task from the beginning until the seat off time instant, TIP2 which represents the last part of the motor act until the end. The orientation in space of the TIP link is controlled by a rotational actuator acting on the sagittal plane (SA), while movements on the frontal plane are neglected because of the particular task analysed. The elevation of the CM has taken into account by a linear actuator (LA), which exerts a lengthening on the link itself.

The input variables to the model are the coordinates of the CM of the body portion in motion as computed from the measured ground reaction forces. The model parameters are the mass and the initial geometry of the modelled body portion. The outputs of the model are the mechanical variables associated with the actuators. Temporal patterns yielded from the model were typical of the TIP applied to the STS (Papa and Cappozzo, 2000). While ground reaction forces were found to be unable to highlight differences between pre- and post-training, TIP mechanics was shown to be more powerful in this respect. Therefore, the use of the TIP model gave a stronger chance to evidence the presence of an increased level of individual physical performance. This finding definitely confirm the use of the TIP model for this application and its inclusion in the process of knowledge discovery. Figure 2 shows, for instance, the time behaviour of SA power during the STS execution for both pre- and post-training trials.

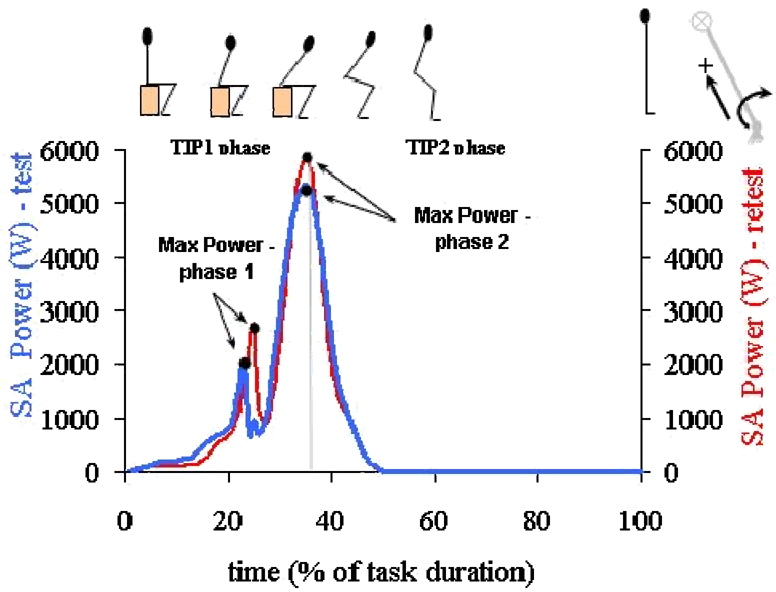


Figure B.2. Effect of training on SA Power

From these set of curves, the feature selection step was carried out in the following fashion: from time characteristics of both LA and SA mechanics, maxima of the two TIP1 and TIP2 phases were calculated and included in the database. In Table 1, the selected parameters are listed:

	Max Displacement (d, φ)		Max Velocity (v, ω)		Max Force / Couple (f, c)		Max Power (p_L, p_S)	
Actuator	LA	SA	LA	SA	LA	SA	LA	SA
TIP1	*d1*	*φ1*	*v1*	*ω1*	*f1*	*c1*	p_L1	p_S1
TIP2	*d2*	*φ2*	*v2*	*ω2*	*f2*	*c2*	p_L2	p_S2

Table B.1. List of selected features

This step of the process allowed to obtain a reduced set of parameters to consider for further analysis. Data obtained are ready to enter the Data Mining phase.

B.3.2 The Neurofuzzy inference system

Data mining technique selection was driven by the analysis of the characteristics of the problem (the evaluation of training using parameters related to motor capacity) and of the sampled population (elderly people). First, a general absence of an a-priori knowledge about the correspondence between each TIP parameters and its physiological or functional meaning was observed. This lack of knowledge called for the use of a heuristic neural approach that allows to learn from the existing data. Furthermore, the presence of uncontrollable factors, as environmental changes and personal lifestyle differences, were noticed. These could have masked the information about the experimental and control groups in a way that each of them could not be fully considered as a crisp set, but should be better regarded as fuzzy set to account for non-statistical uncertainties in data (Bezdek, 1992). This circumstance led to the adoption of a fuzzy inference system, that allow to deal with computational manipulations of imprecise concepts (Mamdani and Assilian, 1975) or with applications in the field of evaluation and diagnosis as for the diplegic gait (O'Malley et al, 1997), usually difficult to be quantitatively described in the real world. Fuzzy inference is the process of formulating a mapping from a given input to an output using fuzzy logic, that provides a basis from which patterns can be distinguished and relevant decisions be made. The rules used by the inference system, when the expert's knowledge lacks, need to be calculated by an artificial neural network. Since this was the case, a NeuroFuzzy Inference System (NFIS) was devised to carry out the classification task, merging the benefits of the use of fuzzy logic with the possibility to learn from data (Lin and Lee, 1996). In figure 3, the NFIS flowchart was reported.

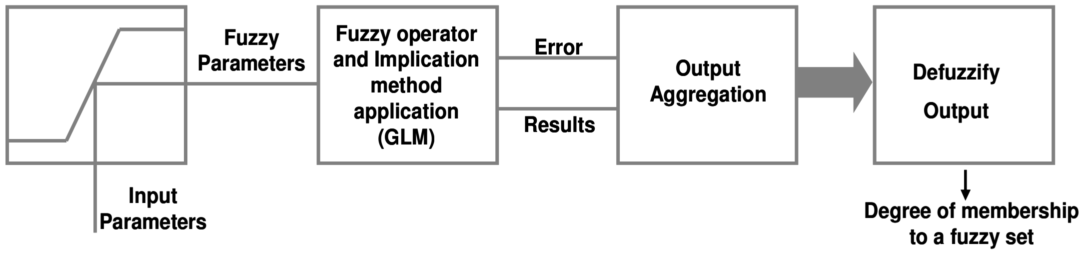
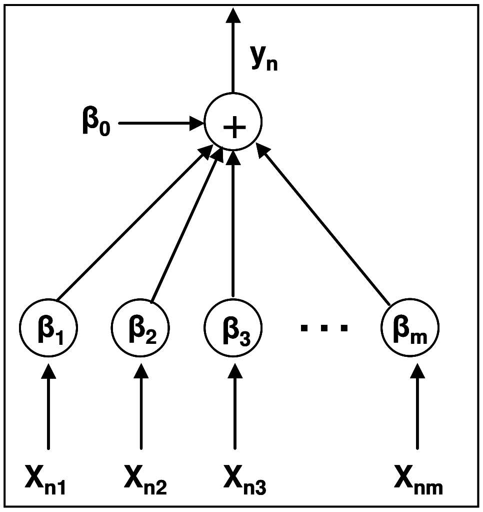


Figure B.3. FIS applied to fuzzy clustering. Classical FIS steps are: *Fuzzify Input*, doing the transformation [min, max] → [0, 1] to obtain standard input data; *Fuzzy operator application*, to merge fuzzy input data and *Fuzzy methods application*, that determines the outputs of each rule; *Output Aggregation*, output values, calculated for each rule, are aggregated; *Defuzzify Output*, using the inverse operation of the fuzzify input step, the aggregated output is normalised, giving as a result, the membership percentage of the analysed pattern to the group.

The first NFIS phase aimed at converting crisp numerical values into fuzzy sets expressed with a qualifying linguistic set (always a 0-1 interval). This entailed the use of a *membership function*, chosen as linear to minimise the loss of information. The further NFIS phase included both the “fuzzy operator application”, that merged all the fuzzified input values into a single truth value for the antecedent of each rule, and the “application of the implication method”, that concerned the transmission of the mentioned antecedent truth value to the consequent of each rule.

The core of the process concerned the definition of the rules (the knowledge) and the following subject evaluation using that knowledge. To this purpose, a feed-forward neural network without hidden layers was adopted that corresponded to a linear regression carried out using the Generalized Linear Model, GLM (Bishop, 1995). The associated network map was reported in figure 4a.



a)

$$y_n = \beta_0 + \sum_{i=1}^{m} \beta_i x_{ni} + \varepsilon_n$$

b)

$$\varepsilon = \sum_{n=1}^{N} (y_n - \sum_{i=0}^{m} \beta_i x_{ni})^2$$

c)

Figure B.4. (a) A neural model for the multiple regression (GLM); (b) GLM equation; (c) Cost function to be minimised in order to find β_i coefficients.

The GLM equation model (figure 4b) was implemented as a supervised neuronal model, where independent and dependent variables represented respectively the network inputs and output, and the β_i coefficients corresponded to the weights of the network. Output of the GLM was the membership degree of each pattern to “trained” (T) and “not trained” (NT) fuzzy sets. In addition, an error ε_n was produced representing the error associated to the n^{th} instance, while coefficients β_i were estimated by minimising the objective function reported in figure 4c. The GLM allowed for data partitioning by defining a *n-1*-dimensional plane in the *n*-dimensional space (or simply a line in 2D). Its training technique included a supervised algorithm having weight vector adjustments that were obtained by minimising the error ε committed while linearly subdividing the cloud of data into two groups. This minimisation task was carried out using the Iterative Reweighted Least square algorithm (IRLs, Bjorck, 1996) that, for each biomechanical feature, calculated a separating line in the relevant *two*-dimensional plane (post-training vs pre-training).

Since NFIS classifications were based on the testing of the rules, each resulting fuzzy set was properly combined into a single fuzzy output using an *aggregation method*. Whereas a lack of knowledge about the relationships between parameters was observed, the aggregation entailed a global approach based on the calculation of the centroid of all the output fuzzy sets. In this work, the weighted average of few resulting data points was used, discarding the rules with weakest discriminating power (i.e. those corresponding to higher ε_n). Each r^{th} rule had a specific incidence to the final result, expressed by the relevant partial result $\{o_{r1}, o_{r2}, \ldots, o_{rn}\}$ and error ε_r. The final aggregation was carried out using the following expression:

$$\{o_1, o_2, \ldots, o_n\} = \sum_r f(\{o_{r1}, o_{r2}, \ldots, o_{rn}\}, \varepsilon_r)$$

Finally, the aggregate fuzzy set was the input for the "defuzzification" phase, which processes data using an inverse procedure of the first fuzzify phase, while the output was a single membership percentage value.

The core of the classifier was the GLM module that accomplished the task of generating the fuzzy rules and that included, also, the evaluation of the validation set. The implemented GLM module (Figure 5) received as input the fuzzy parameters and fed them into a sub-module that managed the subdivision of the entire group into a training group, that participated to the network training (generating the actual weights) and a validation group. After the determination of the validation set, the GLM managed the training of the network for each feature under analysis. The choice of the GLM met the requirements of efficacy, simplicity and reduction of data processing time. Each couple were processed and a partial subdivision of the validation set was obtained together with the weight that this function had in the total calculation given by the output error of the IRLs. As output, a percentage of membership of each subject to each of the group was obtained.

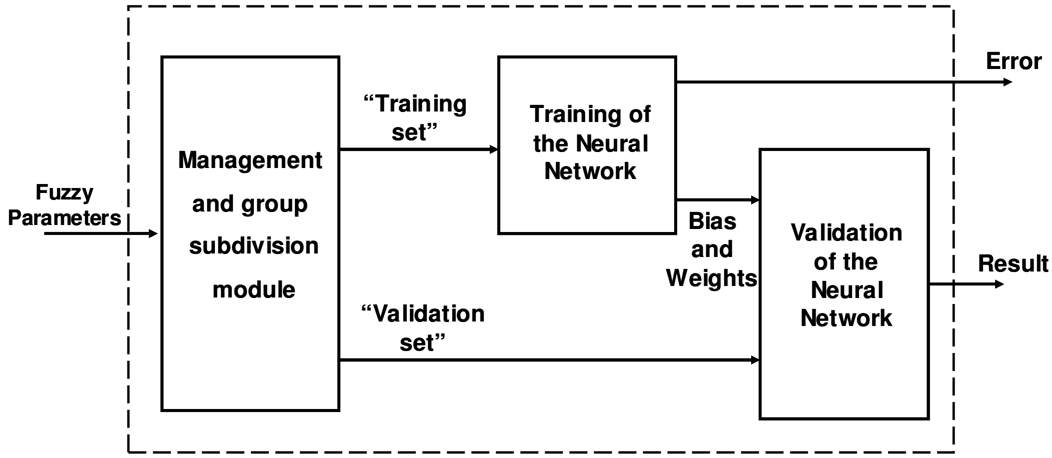


Figure B.5. Diagram representing the phases of the GLM application

Whereas the goal of clustering is to automatically find natural groupings in the data (Duda and Hart, 1973), each group having crisp boundaries and each data point assigned into one group only, fuzzy clustering allowed each point to simultaneously have partial memberships to both the T and NT fuzzy groups. This result was obtained by imposing all the sample to participate to both training and validation set, obtaining a subject positioning in a two-dimensional plane.

Furthermore, the problem of classifying new subjects was faced. Because of the limited amount of subjects available, the generalization ability of the classifier was evaluated using a k-fold cross-validation procedure (Ripley, 1996). A number of k equal to 4 subjects was used as validation set, equally formed of both EG and CG participants, while the remaining 27 subjects participated to the NFIS training. All possible combinations were imposed, so that all participants entered the validation set.

Classifier performance was assessed in both cases by the measures of accuracy (*Acc*), sensitivity (*Sens*) and specificity (*Spec*) as proposed by Begg and Kamruzzaman (2005). The following expression were used:

$$Acc = \frac{TP + TN}{TP + FP + TN + FN} * 100\% \text{ , } Sens = \frac{TP}{TP + FN} * 100\% \text{ , } Spec = \frac{TN}{TN + FP} * 100\% \text{ ,}$$

where TP and TN were the number of true positive and true negative, respectively, indicating the number of subject correctly classified, while FP and FN were the number of misclassifications. The *Acc* index indicates the overall classifier accuracy, *Sens* defines the ability of the classifier to recognize a trained subject and *Spec* indicates the classifier's ability not to generate a false decision (control subject).

B.4. Results

Clustering based on the differences between test and retest fuzzy values was carried out considering each subject as participating in both NFIS training and validation. For each of them, relevant percentages of membership to the T fuzzy set ($T_{\%}$) were pointed out. Percentages to the NT fuzzy group ($NT_{\%}$) were easily obtained using the following obvious relationship: $NT_{\%} = 100\% - T_{\%}$. In figure 6, $T_{\%}$ percentages pertaining to the whole sample are shown.

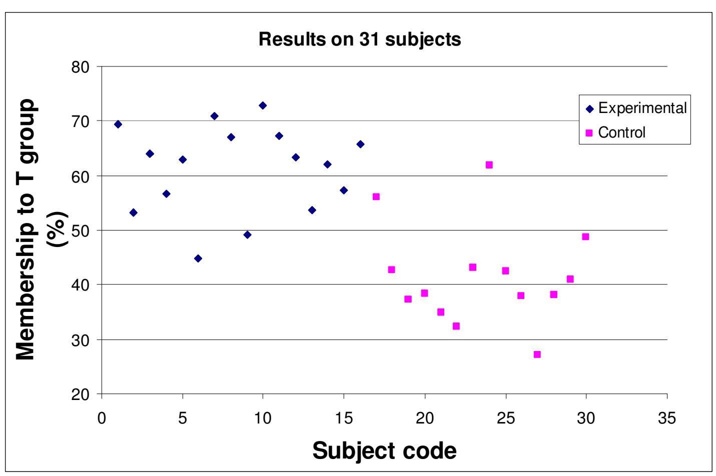


Figure B.6. Results of fuzzy clustering (31 subjects)

The features that mostly determined the group placement, i.e. having the higher weight coefficients while aggregating the outputs, were those belonging to the second phase of the motor task, particularly those related to forces and power (*f2*, *c2*, p_L2 and p_S2), the linear velocity v2 and both displacements d2 and φ2.

In table 2, $T_{\%}$ means and standard deviations were reported for both EG and CG. A one-tailed T-test was performed on both resulting $T_{\%}$ and $NT_{\%}$, confirming a very high statistically significant difference ($p<0.001$) between the two fuzzy sets.

Group of origin	$T_{\%}$ mean (std. dev.)	Misclassification rate (%)
Experimental	61.26 (8.07)	12.5
Control	41.32 (8.77)	13.3

Table B.2. Performance on fuzzy clustering: membership to T fuzzy set

Subjects belonging to both T and NT fuzzy sets showed misclassification rates of 12.5% and 13.3%, respectively, corresponding to only two misclassifications for each group. Accuracy, sensitivity and specificity of the classifier reached similar high percentages (*Acc*=87.1%; *Sens*=87.5%; *Spec*=86.7%).

The classical neural approach, with 4 of the 31 subjects used as validation set, confirmed the amount of subjects to be not yet suitable for the classification task. Relevant $T_{\%}$ and $NT_{\%}$ (mean ± st.dev.) were 53.8±7.6 and 48.9±8.2, respectively. A one-tailed T-test, performed on them, confirmed that the differences were statistically not significant ($p>0.05$). In this circumstance, misclassification rates were higher than 25% and the classifier performance worsened (*Acc*=71.0%; *Sens*=73.3%; *Spec*=68.8%).

B.5. Discussion

The NFIS presented in this work was implemented as a tool for the evaluation of the benefits of a physical training on lower limbs. Such evaluations are usually carried out using a single index, generally related to power expressed by the subject during the execution of specifically devised tasks. The classifier was conceived to allow for an automated evaluation, based on a set of biomechanical features obtained during the act of rising from a chair, deemed to be related to individual motor capacity. The classifier exploited the differences in the individual performance level while comparing biomechanical features obtained both before and after a given period of treatment. The physical training was judged as effective for a participant when, after the period of intensive training, a higher membership percentage to the T fuzzy set than to the NT was obtained.

The study showed that detecting any changes in the physical training level of a widely aged group of elders was a difficult task, that became even harder because of the presence of uncontrollable environmental factors that might affect the subject performance level. In this context, whereas the analysis of a single feature was not able to correctly cluster the two groups, the algorithm devised in this work aimed at overcoming this limitation.

The benefits of the physical training were, first, statistically evaluated by including each subject in both NFIS training and validation sets. This clustering exercise entailed the definition of a linear separating function, as output of the GLM, that generated the two T and NT fuzzy sets. Whereas most of the experimental and of the control group subjects fell into T and NT, respectively, the classification was considered successful. It is worth to note that similar results were previously not remarkable, if ground reaction forces alone were analysed and, also, if a single feature is considered as performance indicator.

Biomechanical features that mostly influenced the fuzzy output aggregation were mainly those referred to the phase that follows the seat-off time instant, particularly those referred to forces and powers. Since the training was given specifically to the lower limbs, this result was expected. Deepening into the meaning of the features involved, it is useful to recall the TIP model that uses a link to connect the individual centre of mass to the ground (corner higher right of Figure 1). Angular mechanics in the last STS phase is associated to the link rotation along the sagittal plane, typically executing a backward movement while elevating the trunk toward the upright posture. Linear mechanics in the same phase is associated to the link elongation that corresponds to the elevation of the centre of mass while the subject is reaching the upright stand. A difference in this last mechanics recalls the existing difference between old and young people that showed a different strategy while executing the STS task. Particularly, for instance for the displacement, the difference is in a more rotated trunk during the first phase before the seat-off. This circumstance entails that, at the beginning of the second phase, the link is less elongated than for a young subject and, therefore, d_2 is higher. For the same reason, the $\varphi 2$ feature weighed significantly while aggregating fuzzy outputs. The set of features highlighted accorded partially to those evidenced previously (Alexander et al, 2001) as denoting the effects of the physical training.

The percentages obtained while doing the clustering exercise showed statistically significant differences between the two groups. Coherently, subject belonging to the experimental group presented higher membership percentages to the T fuzzy group, while subjects from the control group showed higher $NT_{\%}$. However, this last assignment was not so clear-cut, but showed to be strongly influenced by different factors related to environment, personal habits and general subject activity that increased parameter value fluctuations. Thus, the clustering task was more reliable in correctly assigning subjects belonging to the experimental group, while control subjects were still hard to be assigned to the right group. This finding suggested that, while the experimental group was followed carefully in its training program, subjects of the control group were more heterogeneous in their performance level and no information was available about the aforementioned environmental factors that significantly influenced their placement in the two-dimensional plane.

The clustering performance exhibited values around 87% for *Acc*, *Spec* and *Sens* indexes that are appreciably high when compared to those available in the literature (Begg et al, 2003), indicating also that the aggregation phase highlighted the peculiar contribution of each feature to the output membership percentage.

Results obtained with the neural classification showed a weaker performance, probably due to the limited number of subjects available. Although the k-fold cross-validation procedure was included, relevant results showed no significant pre-post differences (p>0.05), even if not far from the significance threshold. Furthermore, $T_{\%}$ and $NT_{\%}$ means and standard deviations differed slightly. The same situation was depicted when calculating the classifier performance parameters (*Acc*, *Sens* and *Spec*), found to be lower than 73%. If compared to those obtained in the clustering exercise, these results evidenced the worsening of $T_{\%}$s for the experimental group, while similar values persisted for the control group. Improvements in the classification ability may be expected, whereas an increased amount of subjects is made available to train the NFIS and, in this respect, relevant $T_{\%}$ percentages may be expected to lead to significant differences among groups.

B.6. Conclusions

An evaluation process that includes: the execution of an appropriate motor task, a musculoskeletal model for its interpretation, and a heuristic technique for the extraction of information may be considered as a suitable tool for a global description of the subject's capacity to move.

The undertaken way of analysing slightly different groups using the NFIS was shown to be promising. From a technical standpoint, the application was encouraging, even considering all the limitations related to the task and the points of vagueness associated to the knowledge about the level of individual performance. Moreover, the NFIS allowed to overcome the lack of knowledge about the importance of each biomechanical feature while synthesising a global score for evaluating the effectiveness of a specific training program on the lower limbs.

Future perspective in this work will be the investigation of NFIS modifications aimed at making it more robust against environmental and seasonal fluctuations of the individual physical level.

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