



DESIGN OF A GAIT ASSISTANT EXOSKELETON FOR INDIVIDUALS WITH LOWER LIMB IMPAIRMENTS

Camilo Andrés Acosta-Márquez ✓

A thesis submitted in partial fulfilment of
the requirements of the University of
Abertay Dundee for the degree of
Doctor of Philosophy
October 2006

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IMPAIRMENTS**

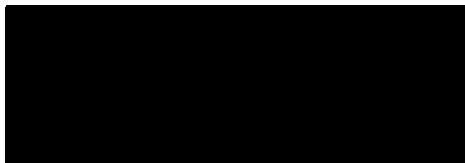
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Doctor of Philosophy

October 2006

I certify that this thesis is a true and accurate version of the thesis
approved by the external examiners.

Signed



Date *14 Nov 2006*

Declaration

I hereby declare that while registered as a candidate for the degree for which this thesis is presented I have not been a candidate for any other award. I further declare that, except where stated, the work in this thesis is original and was performed by myself.

Signed (Camilo A. Acosta-Márquez) October 24, 2006

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DEDICATION

Dedicated to my parents, without their massive support and encouragement this project would have never occurred. You will always be sadly missed in my heart.

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LIST OF ABBREVIATIONS

SCI	: Spinal Cord Injury
FES	: Functional Electro Stimulation
KAFO	: Knee Ankle Foot
HKAFO	: Hip Knee Ankle Foot
HGO	: Hip Guidance Orthosis
RGO	: Reciprocating Gait Orthosis
DoF	: Degree of Freedom
CNS	: Central Nervous System
AL	: Articulated Locomotion
MBD	: Mineral Bone Density
CoG	: Centre of Gravity
SC	: Spinal Cord
NACIS	: National Acute Spinal Cord Injury Study
3D	: Three Dimensional
2D	: Two Dimensional
BISAM	: Biologically Inspired Walking Machine
ZMP	: Zero Moment Point
PC	: Personal Computer
WABIAN	: Waseda Bipedal Humanoid

INSERM	: Institut National de la Santé et de la Recherche Médicale
INRIA	: Institut National de Recherche en Informatique et Automatique
DARPA	: Defense Advanced Research Projects Agency
DoD	: Department of Defense
BLEEX	: Berkeley Lower Extremity Exoskeleton
MRC	: Model Reference Controller
FSC	: Finite State Controller
DAQ	: Digital Acquisition
I/O	: Input / Output
ROM	: Range Of Motion
NEAT	: New and Emerging Application Technologies
NeXOS	: NEAT Exoskeleton
TEM	: Therapeutic Exercise Machine
KA	: Knee Arthroplasty
CPM	: Continuous Passive Motion

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SUMMARY

The goal of the research was to investigate the design of a gait aid that could help individuals with lower limb impairments to support the rehabilitation of the walking function in individuals with certain types and classes of lower limb impairments.

After reviewing different types of lower limb impairments, it was established that in many cases this is a non reversible condition that renders^s muscles in the legs useless either by physically impairing them or by severing the control mechanisms that allow them to be flexed and relaxed at will. This review process was important in validating the concept of the gait aid as at present a cure for this condition is not available.

The research used the analysis of human gait and bipedal robotic gait to develop the concept of an exoskeleton based on a telescopic knee, which could be integrated with an appropriate control system in which the user provides a control input via the crutches. This unique approach to restoring the walking function can be seen as one of the most representative contributions of the project, alongside the methodology developed to analyse the movement profiles and the comparison in between the generated trajectories. The results of the research are a full analysis of the proposed system together with a $\frac{1}{4}$ scale model of a single leg incorporating a telescopic knee.

The initial assumption that an exoskeleton could replace wheelchairs as the main stream aid for mobility in this population also had to be completely revised based on the user centred approach and changed into a different one in which a wheelchair contains the exoskeleton that can detach itself and allow a short stroll nearby the wheelchair.

The wheelchair would then contain the bulk of the energy storage and anything else that is not absolutely necessary for the exoskeleton in order to minimise the weight of the structure. The development of the full size exoskeleton was declined in favour of a therapeutic approach whereby the exoskeleton design procedure is used to investigate an application for a robotic therapeutic rehabilitation device for the lower limbs. This research have been used to extend that initial concept to that of a full-scale machine to be used for the rehabilitation of the lower limbs while employing a control strategy integrated within a telehealth environment.

Finally twelve papers have been produced¹ taking into account both projects put together, in fact although the dissertation only discusses that part of the research pertinent to the exoskeleton and how it led to the NeXOS project, it is in many ways easier to understand the research efforts of the past six and a half years as just one big project.

¹ The complete text of the papers can be found on Appendix C and digitally in the accompanying DVD-ROM.

Chapter 1

INTRODUCTION

1.1 Introduction to Research

An inability to achieve a normal gait is a reality faced by groups of individuals who experience certain types and classes of lower limb disability. These groups can broadly be categorised as those suffering from either a nervous based disease or a muscular based disease. Though most of the illnesses that express these conditions are in fact a mix of both, as illustrated in Figs. 1.1 and 1.2.

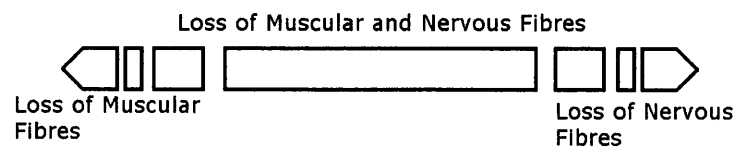


Figure 1.1 : *Spectrum of lower limb disease.*

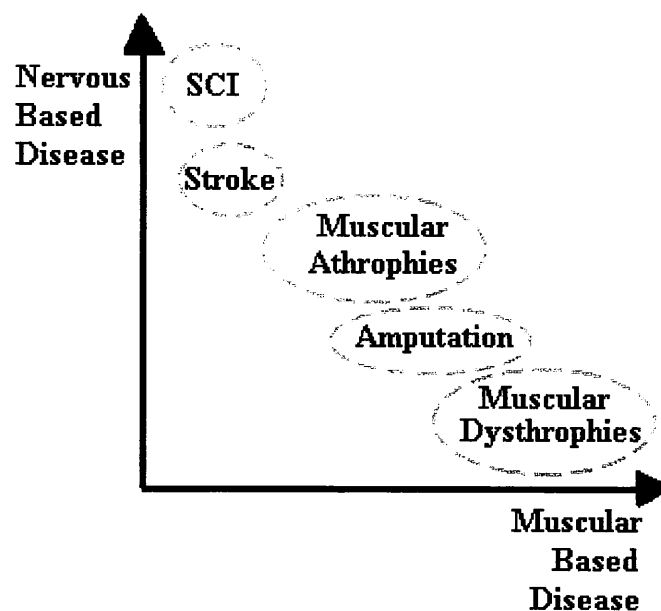


Figure 1.2 : *Nervous Based vs. Muscular Based disease mapping¹.*

¹ SCI stands for Spinal Cord Injury

Chronic and acute diseases alike lead patients to rehabilitation centres where they receive appropriate treatment which tends to focus around the issue of enabling them to gain or regain mobility and independence. Therefore, rehabilitation tasks tend to concentrate on giving an individual the ability to live independently. Under such circumstances, the inability to walk may well seem to be a minor problem when compared to an inability to feed or dress oneself, or to control the sexual function. Losing the ability to walk does not however mean losing the ability to move. Depending on the nature of the disease, many patients regain mobility through the use of various locomotion aids, primarily wheelchairs. Nonetheless it is only human to keep alive the hope of regaining the ability to walk, and with the current technology available it is possible to investigate alternatives to mobility aids that can facilitate the rehabilitation of the walking function.

Spinal Cord Injury (SCI) patients already can (under certain circumstances) receive electro stimulation therapy. This allows them to move with partial control their legs, and depending on the type of equipment, achieve different types of movement including limited gait mobility, usually with the aid of crutches.

A deeper analysis of the issues surrounding both electrical wheelchairs and Functional Electro Stimulation (FES), has the potential to provide some clarity as to the requirements for a different type of rehabilitation tool. Such a tool could provide the advantages of wheelchairs and FES, while minimising their disadvantages. This rehabilitation tool could be a bipedal walking machine in the form of an exoskeleton that in theory provides support while assisting the gait of an individual with lower limb impairments (the target population is made up of individuals with SCI and muscular dystrophies and atrophies).

1.2 Outline of thesis

This thesis is organised into a total of nine chapters. The remainder of this first chapter discusses current practice in support for and rehabilitation of the walking function. Subsequent chapters develop the research carried out to investigate the feasibility of the design and development of a tool for gait assistance, in the form of an exoskeleton for the rehabilitation of the walking function.

In this context, Chapters two and three provide the required insight into the issues surrounding human gait and the pathologies associated with the inability to achieve it. Chapter four provides a thorough and comprehensive compilation of state of the art and previous developments in the field of walking machines. Together with chapter one, these four chapters provide the basis for the analysis carried out allowing the determination of potential configurations of the exoskeleton so that it could become a useful long term aid for individuals with lower limb impairments.

Chapters five and six chart this analysis with a systematic and meticulous modelling of the alternatives and the implementation of different simulations. Chapter six is devoted to the physical implementation of the model designed to test a methodology for synthesising gait patterns that complied with the predefined goals as well as to help in refining the criteria required for such synthesis. Chapter seven explains the results of the tests carried out for this purpose.

Finally, chapter eight describes the practical implementation of the methodologies developed in a spin-off project funded by the Department of Health. The conclusions in Chapter nine rounds off the journey undertaken by the ideas explored in this thesis.

1.3 Current Practice in Mobility Aids

1.3.1 Wheelchairs as Mobility Aids

Wheelchairs are by far the most widely used mobility aid among people who experience lower limb impairments that lead to an absence of the walking function. Those suffering from lower limb disabilities largely retain their mobility through the use of a wheelchair. Their simplicity of operation and high range of mobility make them the obvious alternative in many instances (see Figure 1.3).



Figure 1.3: *Powered and unpowered wheelchairs*

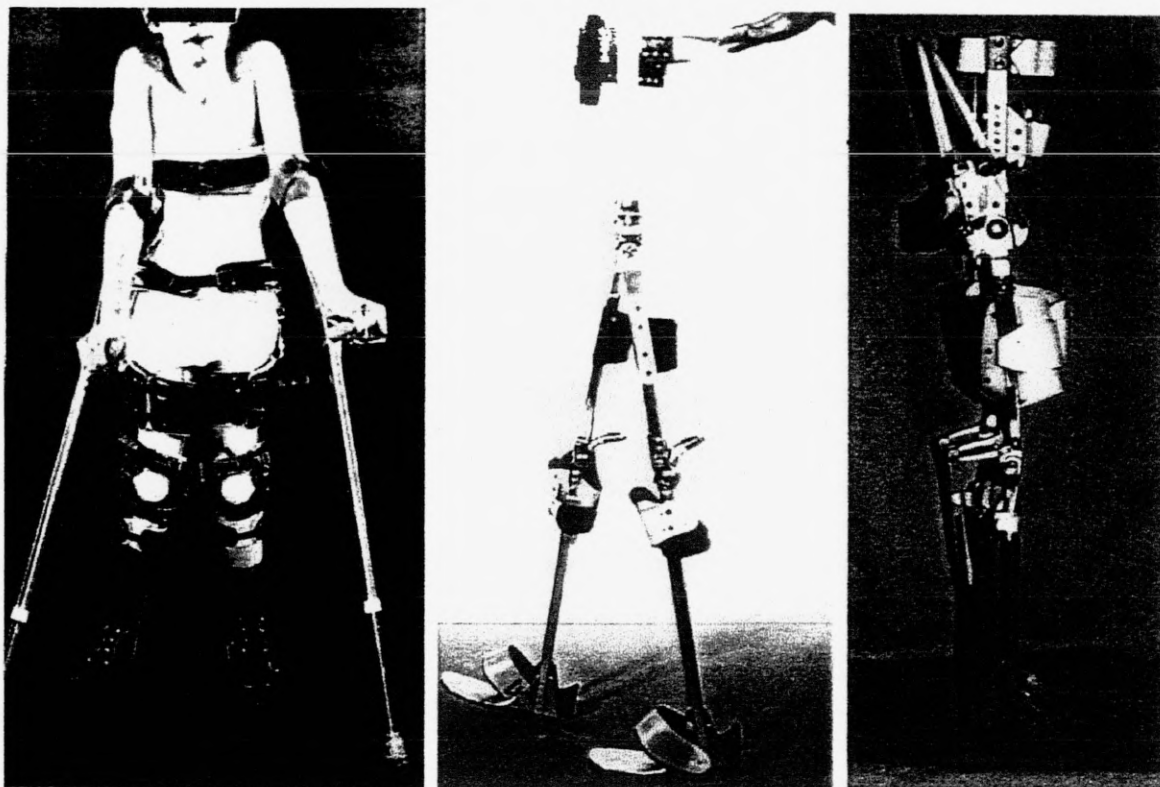
Although some powered wheelchairs can climb stairs or have elevating seats, their utility is in many cases limited by simple obstacles which do not normally present a problem to the average individual. As a result, buildings must be designed so as to satisfy the requirements of wheelchair users by the inclusion of features such as ramps.

1.3.2 Orthoses in the Rehabilitation of the Walking Function

There are however alternatives to wheelchairs to support mobility. One alternative for the rehabilitation of the walking function is that of a passive orthosis, which is basically a pair of long leg braces attached to the user. The braces have different joints depending on the type of system employed, namely Knee-Ankle-Foot orthosis (KAFO), Hip-Knee-Ankle-

Foot orthosis (HKAFO), Hip-Guidance orthosis (HGO) or Reciprocating Gait orthosis (RGO). The joints are mechanically locked or unlocked to allow movement (see Fig. 1.4).

Early studies into the long term usage patterns of KAFO's revealed limited success. Although restoration of the gait could have been achieved, use was practically confined to a clinical or domestic environment ^[1]. The poor functional achievements of long braces paved the way for the development of the Hip-Knee-Ankle-Foot orthosis (HKAFO) in the late 1970's and early 1980's.



(a) Hip-Knee-Ankle-Foot orthosis

(b) Hip-Guidance orthosis

(c) Reciprocating Gait orthosis

Figure 1.4: *Different orthotic systems* ^[23].

HKAFO's made walking a feasible option, within limitations, for SCI patients with higher level lesions resulting in their retaining more mobility than those with lower level lesions.

Despite this, the HKAFO did not completely fulfil the requirements for daily use. The reciprocating nature of the mechanism prompted the entrance to the market of the Reciprocating Gait orthosis, which proved to have a better public response. However in a follow-up study carried out in the UK ^[2], it was found that most of the users had stopped using the orthosis for daily activities within two years of use, and felt that the wheelchair was more appropriate for moving around. They also claimed that the Reciprocating Gait orthosis was uncomfortable and the difficulties of donning and removing it were the main reported reasons for the cessation of use.

The results for the HGO or ParaWalker orthosis ^[1] were more encouraging. Although ParaWalker users reported that the preferred activity for the orthosis was exercising, the majority were still using it 1 or 2 times per week after 3 years without the need for assistance.

Most of the problems encountered in using these types of orthoses in paraplegia lie in the fact that this type of mechanism is not designed to replace the wheelchair. However, these passive orthoses can be incorporated into daily activities if compatibility with the wheelchair can be achieved. Most of the design improvements in passive orthoses over the years have none-the-less focused on improving the efficiency of walking. This is an important aspect, but not the most important one from a user's point of view, where comfort is a key factor ^[1].

SCI patients can also benefit from the fact that the nature of the illness is one that compromises the functionality of the Central Nervous System (CNS) while the skeletal-muscular apparatus is left relatively intact (although rendered useless). The muscles in fact

deteriorate rather quickly due to the lack of transport of the vital neurotrophic elements that maintain the normal volume of the muscles. Therefore the muscles affected by the lesion will suffer a progressive process of atrophy. Similarly, the mineral bone density progressively decreases and in many cases the patients are left with an early type of osteoporosis ^{[3][4]}.

1.3.3 Electro Stimulation in the Rehabilitation of the Walking Function

When properly stimulated, muscles respond positively and walking can be restored by the electrical activation of the paralysed or spastic muscle in individuals with paraplegia. The first modern trials on Functional Electrical Stimulation (FES) were performed on paraplegic subjects some 40 years ago ^[5]. Nevertheless FES has some technical limitations, including a lack of selectivity and of user feedback on energy consumption. Thus, simple movements performed effortlessly by normal healthy subjects become a significant challenge to FES users.

During FES studies, intramuscular electrodes are typically connected percutaneously to a microprocessor-controlled stimulator. Transcutaneous stimulation of the trunk and posterior thigh muscles is used to supplement the intramuscular stimulation. A depolarising electric field is applied either through the skin or through the implanted electrode to the nerves that activate the muscles required to perform the movement. Transcutaneous stimulation achieves stronger contractions but lacks selectivity. No more than 8 electrodes (each connected to an equal number of stimulation channels) are typically used for the trunk and the posterior thigh muscles in transcutaneous procedures ^[6]. In commercially available units however, the number of channels varies from 4 to 32

for skin surface type electrodes, that do not require implantation, and are limited to Spinal Cord Injury subjects with lesions below T4 ^[7] and a mild spasticity.

The percutaneous intramuscular electrode is the usual choice. Once implanted, the electrode remains in the body for as long as it produces active contraction. An average survival rate of the intramuscular electrode is 70% after 1 year and in many cases electrodes have been known to survive well over 5 years ^[6]. Although this is the type of electrode that requires the least surgery, it is nonetheless a cumbersome procedure requiring the use of a special needle to determine the motor point that gives the maximum desired muscle response. Once the most suitable point is determined, the electrode is implanted and clamped in the vicinity of the nerve by means of an anchor at its tip ^[8].

Also, because the muscles have undergone a prolonged period of disuse, the quality of the fibre type composition is usually compromised and is mainly made up of atrophied fibres ^[9]. When electrically stimulated these produce less than normal force and fatigue quickly overcomes users. Conditioning of the paralysed muscle is therefore a prerequisite to initiating artificial walking.

Selectivity is a big challenge in FES systems, and as many as 48 channels could be required for a functional system. Not only is placing the electrodes cumbersome and time consuming, but they can easily fail during the first weeks after implantation. The electrodes have to be monitored constantly as the only way to know if they are working properly is to activate them one by one and to assess the strength of the muscle contraction and electrical impedance. There is not a natural feedback loop in the system, so improving the selectivity by adding more electrodes increases the complexity of placing and monitoring the electrodes.

In Europe, the ‘Stand Up And Walk’ program is the most advanced in its class. The objective is to implant electrodes and control the artificial contraction of the muscles via wireless communication in six selected patients in six European countries (France, Italy, Netherlands, Germany, UK & Denmark) for evaluation. The patients have control over three simple commands; accelerate, slow down and change direction ^[10], see Figure 1.5.

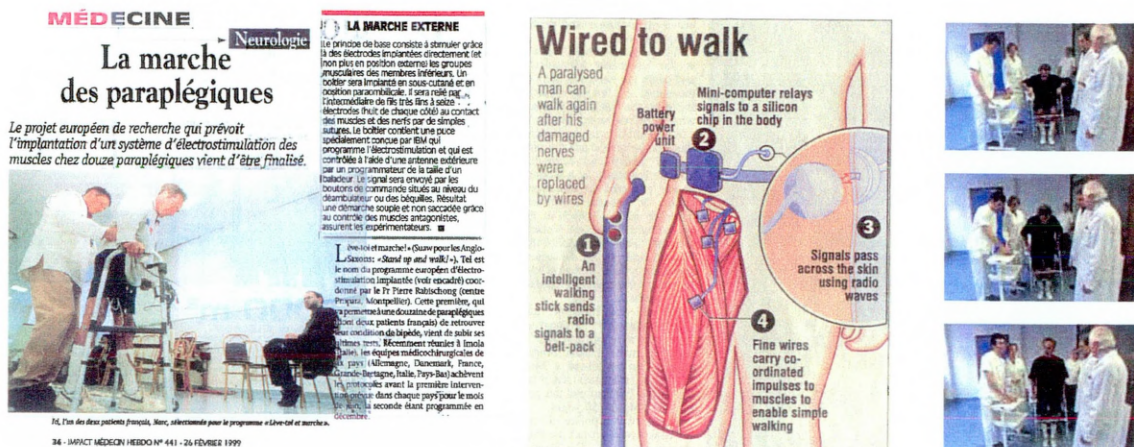


Figure 1.5: The ‘Stand Up And Walk’ project ^[10].

The more fibres that can be stimulated individually, the finer the movement that can be performed. Therefore resolution in this sense, is a key design parameter, since it is necessary to connect fine wires that will co-ordinate the impulses transmitted to the muscles directly inside the leg. The dangers of such procedures are plain to see in projects like the so-called ‘Bionic Man’ in the 1980’s. In this particular project surgeons in the USA selected a Vietnam War veteran (Steve Winter) to undergo surgery and inserted more than 180 fine wires to be connected to a remote control that stored a number of leg movements, including gait patterns.

For over three years Steve Winter was paraded around the US, taken to numerous conferences and hailed as the proof of mankind’s defeat of illness, until the wires started to deteriorate and corrode inside his body. Because of the unusually high number of fibres

implanted, it became virtually impossible for the doctors who pioneered the procedure to remove them without compromising the integrity of the leg, i.e. the risk of damaging the leg to the point of amputation was just too high ^[11]. The project was abandoned and Steve Winter now has to periodically undergo surgery to remove the wires that are liable to cause infections due to corrosion, see Figure 1.6

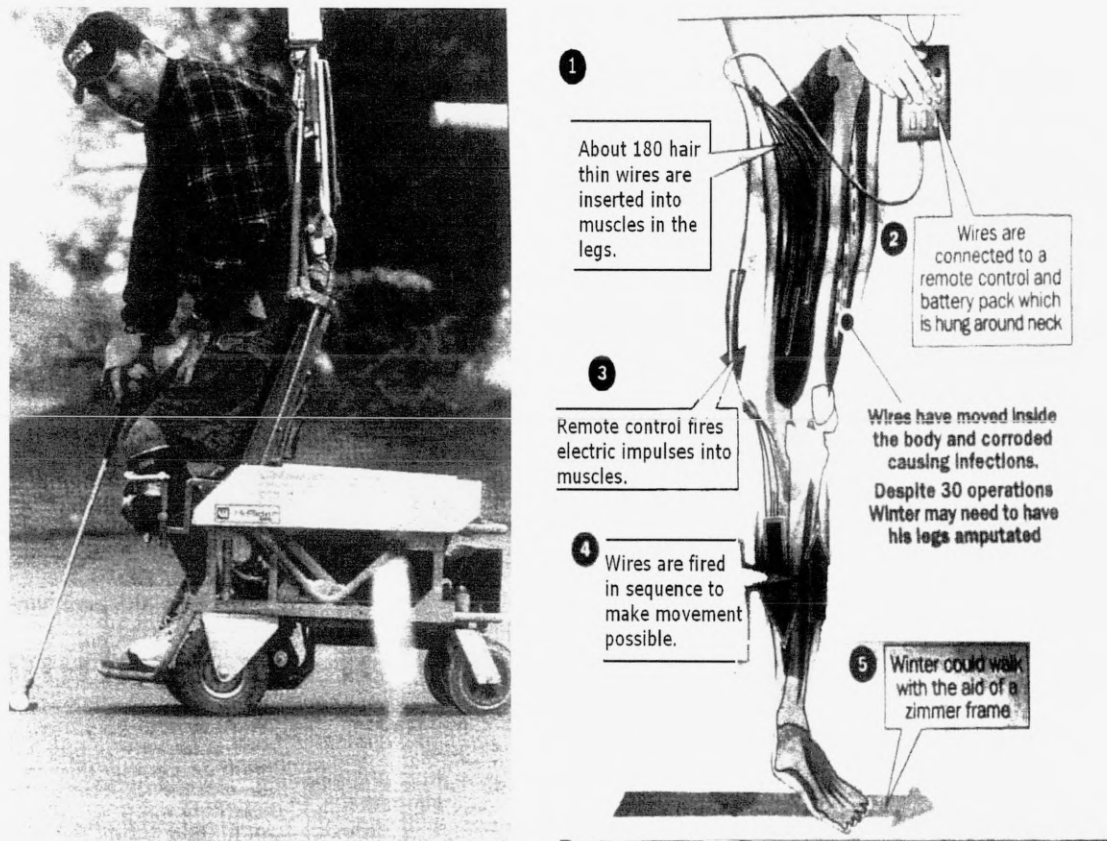


Figure 1.6: Steve Winter - The failed "Bionic Man" system, was hailed in the 80's as a cure for paraplegia^[11].

Despite all the technical difficulties, synthesis of artificial gaits can be achieved, although movements tend to be jerky and lacking in smoothness ^[6]. Artificial walking requires excessive metabolic energy ^[6] as a result of the lack of feedback. In addition the poor quality of the muscles causes over-exercise of the muscular packages and subjects tire quickly.

1.3.4 Implications for the Restoration of the Walking Function

Functional Electro Stimulation extends the idea of restoring mobility to that of restoring bipedal locomotion.

For individuals with lower limb disabilities it is important to consider that one of the goals is not just restoring the ability to go from point A to point B, but to enable them to walk from point A to point B. Bearing in mind the idea of restoring bipedal locomotion, other alternatives can be considered. For example, a powered exoskeleton based on electrical, pneumatic or hydraulic actuators may be an alternative approach to achieving mobility. Because they can be controlled more optimally and less heuristically, and because several parameters can be monitored and a proper control strategy that takes into account the energy consumption implemented, such actuators could provide the required movement of the lower limbs, satisfying the premise of restoring bipedal locomotion.

From interviews with SCI patients and videos ^[12], it is clear that although rehabilitation of the walking function is not a matter of life and death, it can restore independence and build confidence while providing marginal benefits like increase in Mineral Bone Density and overall fitness.

1.4 Walking Machines as a Rehabilitation Tool

1.4.1 Bipedal Walking Machines

The ability to move is a vital life function for all creatures in the animal world. In animals locomotion is usually achieved by means of articulated limbs, referred to in the literature as Articulated Locomotion (AL) ^[13]. In insects there are usually more limbs involved in the process of movement than in terrestrial mammals.

This has an empirical explanation in the variation in energy requirements due to the size constraints between groups of animals. The bigger the animal, the less number of actuators it can afford to power. There is also a computational power issue because the musculo-skeletal systems of most mammals boast a few hundreds Degrees of Freedom (DoF). Thus humans have an operational capability of several hundreds of muscles for locomotion-manipulation activities, or the equivalent of around 350 DoF ^[14].

There are many reasons why animals did not evolve wheels or tracks. Fundamentally, when dealing with obstacles, legs perform far better than anything else. Although from our everyday experience this is easy to appreciate, a more formal explanation is important in order to highlight the fact that a robot designed as a tool for rehabilitating the walking function has to be a legged machine. In Active Bipedal Walking machines control becomes a critical factor of operation ^[13]. According to Kato ^[13] the following considerations must be taken into account :

1. Control of limbs is of synchronic nature.
2. Control of positioning and acceleration for the limbs is required.
3. Stability can be achieved by controlling the position of the Centre of Gravity (CoG).
4. Control must be able to adapt to the environment.
5. As with any autonomous vehicle, optimum control of appropriate energy requirements is a must.

Passive bipedal walkers extend the consideration of optimum control of the appropriate energy requirements. The spring-loaded legs can theoretically walk over level ground with no energy cost at all i.e. to keep walking with no energy input at all ^[86]. More relevant

perhaps, is the realisation that the more basic the walker is, the simpler the control strategy thus lowering the requirement for on board intelligence. A more thorough and extended discussion on the dynamics of bipedal walking can be found in Chapter 2.

The use of legs requires a very high degree of automation that is usually achieved through a learning process. In humans it is considered that normal walking patterns require approximately 7 years to be fully developed ^[13]. The evidence from the evolutionary record ^[15] indicates that the original, and still primary, function of the brain is not to think and plan, but to act and react which, at a very fundamental level, can be seen as controlling the body in pursue of a goal-seeking strategy by trial-and-error.

In human movements the motion control system may be considered at three levels ^[13]:

- (a) Brainstem-spinal System
- (b) Limbic System
- (c) Neocortex System

Of these, level (a) could be seen as a servo-system that derives the necessary selection of walking patterns in response to instructions from the (b) and (c) levels. Level (b) would then be a feedback system for control of the Centre of Gravity (CoG), in order to maintain balance during walking. Level (c) makes decisions under walking conditions, such as speed and direction, and provides overall control of the lower levels.

Research into Artificial Intelligence led to a growing number of projects that attempted to build walking machines ^[16]. With the ability to easily implement on-board microprocessor based controllers in small vehicles allowing for the rapid expansion of this field of research ^[17]. Without computer control it would be very difficult to coordinate the

movement of the many joints required in order to achieve the different motion profiles that generate the diverse type of gaits.

1.4.2 Considerations for Medical Exoskeletons as Walking Machines

If a Bipedal Walking Robot can be designed in such a way that it can accommodate the legs of a person, then it could be argued that the legs of the person could be made to move by the robot. Furthermore, if the type of gait expected to be implemented is bound by criteria such as safety, simplicity and optimal (low consumption) energy consumption of the device, it could constitute a significant tool in the rehabilitation of the walking function.

It is important when considering exoskeletons to highlight the fact that orthoses and FES systems all make extensive use of bracing and balancing aids such as crutches and standing frames. Rehabilitation alternatives have been explored around external aids but not designed as exoskeletons ^[16].

The concept of an exoskeleton does not have to benefit only SCI patients. The possible applications of such a device can be filtered down to the whole spectrum of lower limb impairments provided the users retain significant mobility above the waist. For example, those with a severing of the spinal cord exhibiting lesions below T4 with mild signs of spasticity could be considered. Younger candidates being more likely to retain the arm strength necessary to help themselves to retain balance while walking with crutches.

Individuals with severe atrophies (lower limb problems) could also benefit due to the mild spasticity that they may exhibit.

The majority of these diseases are muscular atrophies or dystrophies such as Duchenne's and Becker's muscular dystrophy, facio-scapulo-humeral dystrophy, and some other conditions like congenital myopathy, spinal muscular atrophy of the Kugelberg-Welander type, chronic polyomyositis as well as some difficult cases of myasthenia gravis ^[14]. In this same category are those suffering from amputations well above the knee. With this type of amputation the whole musculo-skeletal apparatus is lost and a simple prosthetic device may not suffice for the requirements of the user.

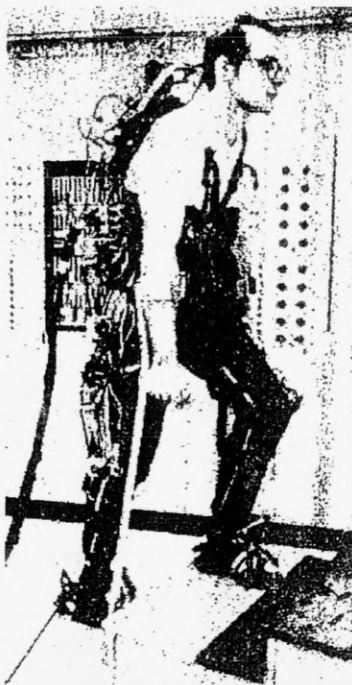
Exoskeletons aimed at rehabilitation have been designed and tested in the past and Chapter 3 develops this subject thoroughly. They include those developed at the Tokushima Hospital in Japan and at the Mihailo Pupin Institute in Belgrade ^[14] ^[5] in the early 1970's, both of which enabled paraplegic patients to walk supported by the exoskeleton and crutches or a four-legged-wheeled aluminium walker. A further type of active orthosis developed at the Mihailo Pupin Institute was a semi-soft structure for more comfort and less decubital risk ^[5]. The target population for this system was the dystrophic group of patients encompassing similar diseases characterised by the lack of muscular strength.

Also in the 1970's other exoskeletons were developed for research purposes such as the Multitask Exoskeleton from the University of Wisconsin ^[18], see Figure 1.7 (a), and an active orthosis in St. Petersburg ^[19], see Figure 1.7 (b). Another active orthosis was also developed in Torino in 1999 as a rehabilitation device for paraplegic users. The device has a pneumatically powered reciprocating knee as the only active joint ^[20], see Figure 1.7 (c).

As indicated above, the 1970's saw the Mihailo Pupin Institute take centre stage as far as developments in medical exoskeletons were concerned, the kinematic walker being one of

the earliest successful designs. This used two main pneumatic actuators for the hip and the knee ^[14]. Because of the reduced number of DoF's, only movement in the sagittal plane was achieved. After trials conducted in healthy subjects, results provided the basis for moving a man of medium size and also proved that an impaired individual could adapt to this type of assistive machine. More DoF's were then added and the exoskeleton was tested.

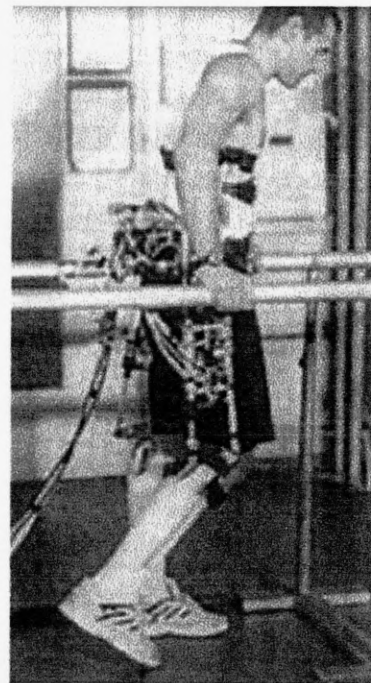
The main considerations deal with the locations of the attachments to the body of the subject because of the possibility of developing local wounds or decubitous in places of higher pressure during extended use, see Figure 1.8. Fully paraplegic individuals could walk with the assistance of two people and a rolling aid.



(a) *Uni. of Wisconsin* ^[18]



(b) *Uni. of St Petersburg* ^[19]

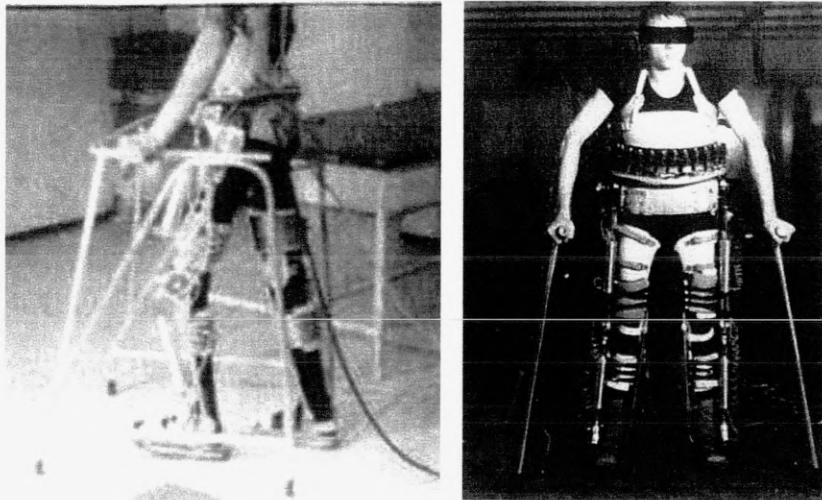


(c) *University of Torino* ^[20]

Figure 1.7: *Examples of active orthosis*

The main drawback of the design was the weight, which increased enormously because of the use of industrial pneumatic valves. The next design replaced these with a series of miniature pneumatic valves and used light alloy and plastic materials to reduce the weight

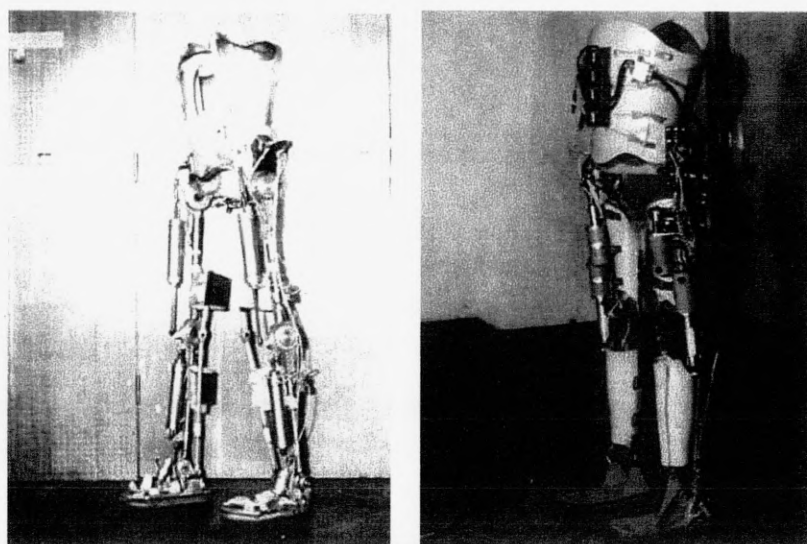
of the complete structure. This lighter exoskeleton allowed patients to walk with the aid of crutches alone, passing through doors and turning, see Figure 1.9 (a). Because of the drawbacks of compressed air generation, further progress on pneumatically driven autonomous walking aid was not practical and this type of exoskeleton was solely confined to controlled environments like clinics and laboratories. The development of an electrically driven exoskeleton was therefore considered as an alternative, see Figure 1.9 (b).



(a) Patient with light support^[14]

(b) Patient with crutches^[14]

Figure 1.8 : Patients undergoing trials for pneumatic exoskeletons from the Mihailo Pupin Institute.



(a) Pneumatic Exoskeleton^[14]

(b) Electrical Exoskeleton^[14]

Figure 1.9: Second generation exoskeletons from the Mihailo Pupin Institute.

1.5 Clinical Gait Analysis Considerations

In order to obtain a better understanding of human gait, considerations of Clinical Gait Analysis are useful. A Clinical Gait Analysis Report gives information on the main rotation angles, net moments and power consumption for the principal joints for pitch, yaw and roll together with the ground reaction forces.

1.5.1 Energy Expenditure Considerations

Patients wear markers around the knees, hip and ankles and an array of infrared cameras tracks the markers while in motion and a computer assembles the information from the different cameras into a continuous sequence. Of special interest are the values for the power consumption at the hip, knee and ankle ^[21].

For the ankle power curve of Figure 1.10, Phases I and II show the double support phase of gait when both feet are touching the ground (see Figs 1.11(a) and (b)), and Phases III and IV indicate the single support phase when only one foot is in contact with the ground (see Figs 1.11(c) and (d)). In Phase I the swinging foot lands on the ground, Fig. 1.11(a) and at the end of Phase I the stance foot is raised and the swinging leg becomes the stance leg.

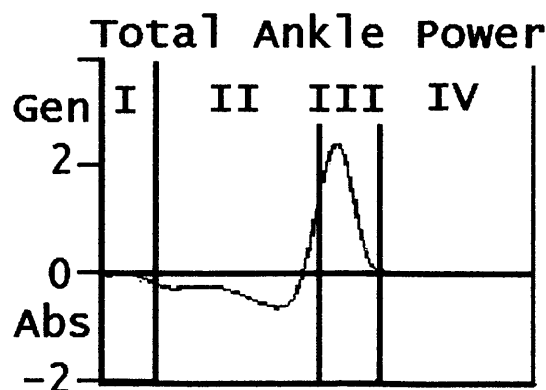


Figure 1.10 : Ankle power curve from Clinical Gait Analysis ^[21].

(courtesy of Dundee Limb Fitting Centre)

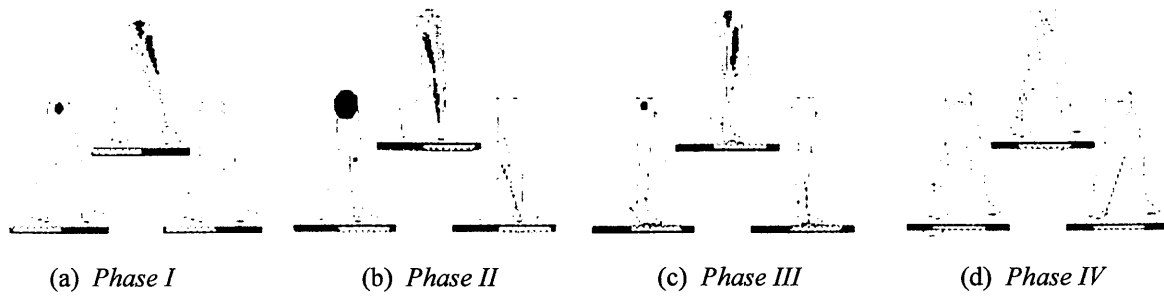


Figure 1.11 : Simulation of human gait using the GaitLab simulation software

■ muscle actuators, ■ joint loading and ■ ground reaction forces

During Phase II, the stance leg absorbs all the impact of the falling body and the swinging leg and accumulates energy for thrusting the body forward at the end of the phase. Phase III captures all the swinging of the leg as it flies from the back of the body to the front of the body, effectively preparing the body to advance. The swinging leg touches the ground at the end of Phase IV and the stance leg starts to rise during Phase IV. By the end of Phase III the stance foot is not touching the ground and becomes the swinging one.

Ankle power consumption is a key factor in the human gait because (along with the knee) it is the joint with the highest instantaneous power consumption and therefore the point at which more energy is released. As there is little that can be done to change the consumption pattern in the hip and the knee, the ankle joint is consequently the one to moderate. As a result it is important to understand why the ankle power curve has the form shown.

Firstly, the power curve for the ankle is obtained from the sagittal components of force and velocity (see Fig. 1.10), for that reason any consideration is valid for a biped robot walking in this plane. Secondly, this energy release is mainly due to the thrust required to bring the centre of gravity of the body in front of the stance leg in order to rotate the knee and raise the swinging leg to allow the forward displacement of this swinging leg.

From these considerations, it is clear that a method of enabling the movement of the centre of gravity in a slower fashion is required, and that the ankle is required only for lifting off the swinging leg. Therefore, if the knee can accomplish the movement of the centre of gravity, it will minimise the power consumption of the ankle while limiting it to support duties. As the knee is only raising the swinging leg, a telescopic joint can be employed and through a careful choice of gait strategy, the contribution of the ankle to the gait can be effectively limited to providing double support and easing the balance control strategy.

One serious problem is that of balance, in order to simplify the task the user will be required to employ crutches to provide extra support in the single leg support phases of a step, effectively using a static balance strategy whose possible major impact on the gait synthesis will be that of producing slow patterns. If an increment of walking speed is required, statically unstable states could be introduced with the aid of the crutches so that the dynamic effects will shorten the double support phase and reduce the effort of transferring the body forward ^[22].

1.6 Design Considerations

Clearly the link between Bipedal Walking Machines and Rehabilitation of the Walking Function has to be the target application of this technology to produce purpose designed legged machines that will act as gait assistance devices for individuals with certain classes of lower limb impairments.

In general terms for most orthotic devices, from the user's point of view there are usually 5 basic design requirements ^[1] for the successful implementation of such devices in the long term. These are:

- Function
- Comfort
- Cosmetic Appearance
- Trustworthiness
- Cost/Benefit

The long term acceptance of a gait assistant of this type, and of any orthotic or rehabilitation device, lies within its functioning during other activities. So either the set of activities involving the use of the gait aid are constrained to a very few specific ones, or the global functional requirements must incorporate daily life activities ^[1].

When extended use in public is required, acceptance becomes as influential as the technical side of the design and that drives it in a way that meets the expectations of the user. Two aspects require consideration. The first one directly related to the aesthetic appearance of the exoskeleton (wheelchairs have become a part of our everyday life, and may be regarded as socially accepted in the main). The second aspect worthy of consideration is *dynamic cosmesis* ^[1]. This is a term coined to signify the idea that the walking pattern will resemble normal gait patterns as much as possible.

Since it is widely accepted that aided gait in lower limb impairment is usually accompanied by the use of walking aids such as crutches, the gait pattern may not be expected to be exactly the same as that of an average individual ^[23]. Nonetheless the design of the exoskeleton, and the gait patterns it produces, must address this issue.

Safety issues such as system failures should not cause physical injury to the users so that they can use the exoskeleton without any fear that it will cause any harm. Likewise,

temporary non-use due to failure of the system (lack of reliability) may result in the user losing confidence and abandoning the exoskeleton.

Ultimately, a device of this type must take into account individual needs of the user in order to maximize the independence that they used to enjoy prior to the onset of their specific impairments (user centred design approach). If long-term use is to be achieved, almost no need for assistance when donning and doffing into and out of the exoskeleton, should be at the top of the requirements.

1.7 Project Objectives

The key objectives of the project, can now be summarised as follows:

- To investigate the feasibility of the design and development of a novel gait assistant exoskeleton for the rehabilitation of the walking function in individuals with lower limb impairments (primarily Spinal Cord Injury) as a replacement for an autonomous wheelchair and an alternative to Functional Electro Stimulation.
- To determine successful configurations of the exoskeleton so it could become a useful long term aid for individuals with lower limb impairments.
- To develop and test a methodology for synthesising gait patterns that comply with the previous objectives as well as to define the criteria required for such synthesis.
- To construct a representative model of the exoskeleton as a test bed to carry out the required experiments to achieve the previous objectives.

Chapter 2

LOWER LIMB DISEASE

v

2.1 Introduction

Nervous or muscular based diseases (such as spinal cord injuries and muscular atrophies) are the main causes of lower limb impairments. Spinal cord injuries and muscular atrophies and dystrophies render certain groups of individuals completely unable to walk.

2.2 Spinal Cord Injury

Spinal Cord Injuries are typically characterised by the loss of nerve fibres, usually leaving the skeletal-muscular structures unharmed and, as such, are life threatening conditions recognised by many early civilizations ^[24]. In the context of this thesis, it is acute traumatic Spinal Cord Injury (SCI) and its associated lower limb impairment that is most commonly associated with the use of wheelchairs.

Table 2.1: *Causes of SCI recorded at the Duke of Cornwall Spinal Centre (1993-95)*

Admission Related Cause of Injury	Percentage of Admissions
Domestic and Industrial Accidents	37.0
Road Traffic Accidents	36.0
Sports Injuries	20.5
Self harm and Criminal Assault	6.5

Since such conditions tend to be associated with sports injuries and road accidents, the epidemiology of SCI is generally an illness affecting young males, mostly occurring amongst persons aged 16 – 30 ^[25]. Complete injuries are more common amongst younger individuals and men, than older adults and women. Due to the advances in emergency services most of the SCI patients are incomplete, and therefore the majority suffer from spasticity issues. Table 2.1 shows the typical

distribution of causes for SCI at the Duke of Cornwall Spinal Treatment Centre in England for the period 1993 to 1995.

Over the past 50 years, neuroscience techniques have resulted in a better understanding of the neuropathophysiology of the primary and secondary events that cause progressive loss of neural tissue. A number of experimental models and studies have been developed to simulate acute clinical spinal cord compression and to explain the further damage caused by haemorrhagic necrotic tissue. The primary injury as a rule involves blunt cord compression, usually as a result of the dislocation of spinal vertebral segments or displaced bone fragments ^[25].

Other studies ^{[26] [27]} have dealt with the processes resulting from cells undergoing necrosis as they release chemicals that injure the surrounding tissue producing an inflammatory response furthering the damage created by the initial compression. Improved early management of SCI patients has led to a decrease in mortality, an increase in life expectancy and a more rapid rehabilitation and re-entry into society.

2.2.1 The Central Nervous System

The Central Nervous System (CNS) is divided in two main parts, the brain and the Spinal Cord (SC). The spinal cord is protected by the vertebral column and is surrounded by different protective layers and a plasma-like liquid called the Cerebrospinal Fluid and communicates with the body through 31 pairs of nerves (the spinal nerves). Each spinal nerve emerges in two parts : a dorsal root and a ventral root (see Figure 2.1).

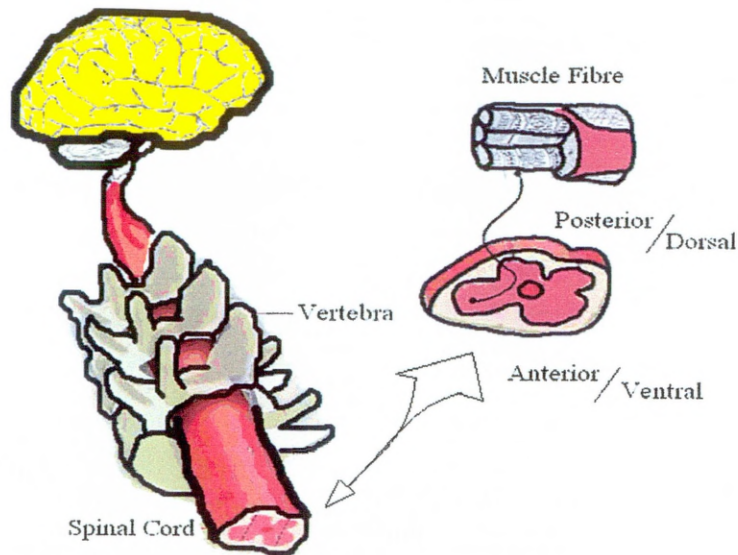


Figure 2.1 : Transverse section of the SC, the posterior is involved with the sensory activity, the anterior area contains the motor neurons that control the contraction of the muscle fibres

This morphological division has a physiological impact as the dorsal roots contain all the efferent fibres involved with sensory activity while the ventral roots contain all the afferent fibres involved with motor and autonomic activity (see Figure 2.1). Hence the place where the cord compression incident occurs determines to a great extent the degree and level of injury.

2.2.2 Neuropathology of the Central Nervous System

In the immediate aftermath of a blunt cord compression incident, the spinal cord undergoes a sequential progression of pathologic changes as set out below ^[28].

Pathologic Changes	<ul style="list-style-type: none"> - Haemorrhage - Edema - Neuronal necrosis - Axonal fragmentation - Demyelisation
--------------------	--

A persistent compression of the material surrounding the Spinal Cord (the Cerebrospinal Fluid) leads to a distension (erythrocyte distension) of the capillaries irrigating the area where the bodies of the neurons and most of the dendrites are present (the grey matter) ^[29]. After 10 to 25, minutes the distension turns into a series of small haemorrhages within the capillary areas feeding the compressed section. Within 1 hour after injury, lack of blood irrigation affects the metabolism of some lower motor neurones (the anterior ventral horn cells) ^[30], these cells then start to undergo an irreversible process of cell death (ischemia).

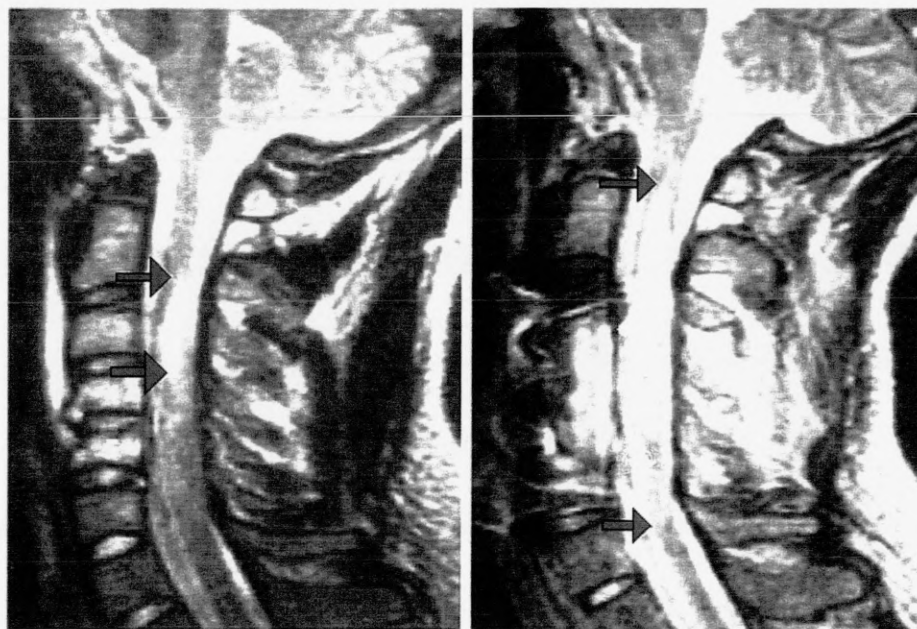
The haemorrhaging also causes necrosis of the region as they disturb the intracellular concentration of sodium-potassium. This precipitates a toxic poisoning process (cytotoxic edema), intracellular acidosis and increased calcium permeability through the cell membrane. As these structures present high metabolic requirements, the resulting imbalance of oxygen and glucose causes the death of the cells by lack of oxygen (hypoxic-ischemia) ^[25]. This is known as the secondary mechanism of injury.

As the intracellular levels of calcium increase, a cascade of calcium-mediated events are responsible for releasing toxins ^[31] that will damage and destroy cell membranes by dilation (vasospasm) ^[25]. The increase in intracellular calcium also releases different excitatory neurotransmitters, amongst them glutamate.

Glutamate is thought to be the most harmful mediator of neuronal cell death due to its excitotoxicity role in hypoxic-ischemic neuronal and glial death ^[32]. The release of glutamate under these uncontrolled circumstances will trigger the release of more neurotransmitters, which are themselves toxic substances and are key to the process of

synapsis. Usually after the synapsis, the neurotransmitters are degraded or metabolised. However, the increasing volume of neurotransmitters released into the system creates more excitatory synapsis, releasing more toxic material and the regulatory mechanisms become overloaded and finally collapse.

The Central Nervous System disposes of dead cells in a cleansing process called phagocytosis, other secondary mechanisms of injury include the inflammatory response whereby the amount of phagocytic cells at the injury site correlates with quantitative damage ^[25].



(a) 6 hours after injury

(b) 3 months after injury

Figure 2.2 : Magnetic Resonance Images of a 21 year old with cervical trauma ^[25]. The arrows indicate the extension of the lesion, changing from 2cm to 6 cm in three months.

As the number of dead cells increases, so does the number of cleaning cells disposing of the remains of neurons and glial cells. The congestion in the zone then leads to an inflammatory process resulting in further damage to the surrounding tissue. Figure 2.2 illustrates this process 6 hours after injury and 3 months later.

2.2.3 Therapeutic Research into Spinal Cord Injury

Current therapeutic research into SCI has focused on two areas of pharmacotherapy. The first aims to minimise the effects of the secondary mechanisms of injury, so the acute stage treatments aim towards diminishing the inflammatory response and the excitotoxicity processes ^[33]. In the subacute and chronic stage, the initiation of neuroregenerative therapies with neurotrophic substances, possibly in combination with transplantation of stem cells ^{[34][35][36]}, is under consideration.

Methylprednisole was the first drug proven to alter the neurologic outcome of SCI. Although motor improvement is minimal, the National Acute Spinal Cord Injury Study 2 (NACIS 2) in the United States of America suggested the future potential for this type of treatment ^[25].

Transplantation techniques aimed at replacing neuronal tissue include stem cells, foetal tissue and peripheral nerves. Foetal tissue transplants combined with neurotrophic therapy have proved to completely reverse neuronal atrophy in rats ^[34].

After SCI, there is little re-growth across the injury area. The affected neurons undergo a process of dieback or involution of the axon proximal to the injury. Neutralisation of the factors that block axonal regeneration will prevent dieback. Tests in rats of the monoclonal antibody IN-1 has led to pronounced axonal regeneration in their damaged spinal cords ^[37].

So, although at present there is significant research into identifying new therapies for rehabilitating SCI, there is currently no effective treatment that allows regeneration of the affected CNS tissue.

2.3 Muscular Dystrophies and Atrophies

Muscular Dystrophies leave the Central Nervous System essentially intact but render the directly affected muscular tissue virtually useless. A more precise definition of Muscular Dystrophy is given by Emery as:

“A group of inherited disorders which are characterized by a progressive muscle wasting and weakness, in which the muscle histology has certain distinctive features (muscle fibre necrosis, phagocytosis, etc..) and where there is no clinical or laboratory evidence of central or peripheral nervous system involvement or myotonia” ^[38].

2.3.1 Pathology of Muscular Dystrophies and Atrophies

As can be seen from Table 2.2, muscular dystrophies encompass a number of disorders that vary considerably in their onset, severity and distribution of muscle involvement.

Although the basic biochemical defect is not yet known for most of these diseases ^[38], pathological studies carried out on individuals suffering from muscular dystrophies all show a similar result whereby the membrane known as the sarcoplasm that binds every muscle fibre (see Fig. 2.2) disappears, leaving the muscle fibre to undergo necrosis ^[38].

Table 2.2: *Types of muscular dystrophies*

Type	Groupings		
X-linked dystrophies	- Proximal	- Duchenne	
		- Becker	
		- Others	
	- Quadriceps	- Quadriceps Myopathy	
	- Scapuloperoneal		
Autosomal recessive dystrophies	- Proximal	- Congenital Forms	- Rapidly Progressive
			- Slowly Progressive
		- Childhood Forms	
	- Quadriceps	- Adult Forms	- Scapulohumeral
		- Quadriceps Myopathy	- Pelvifemoral
Autosomal dominant dystrophies	- Facioscapulohumeral	- Usual form	
		- With Coat's Disease	
	- Scapuloperoneal		
	- Proximal	- Dominant limb girdle dystrophy	
		- Hereditary Myopathy limited to females	
		- Hereditary Myopathy limited to males	
	- Distal	- Childhood form	
		- Adult form	
	- Ocular	- Ocular form	
		- Oculopharyngeal forms	

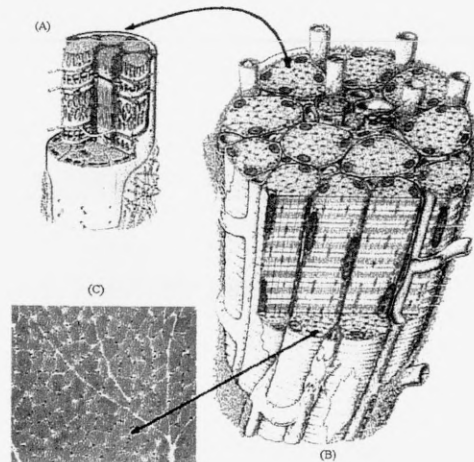


Figure 2.3 : *Effects of muscular dystrophy* ^[38]

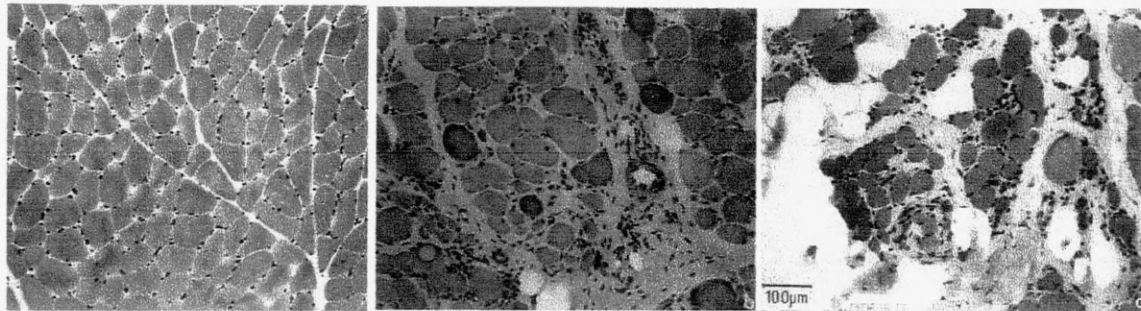
(A) *Diagrammatic representation of a single muscle fibre with the packages of myofibrils*

(B) *A small fascicle bundling a small number of muscle fibres*

(C) *Transverse section of a healthy muscle.*

These exposed fibres are removed from the body by the process of phagocytosis. Normally an affected fibre will regenerate, but in the case of muscular dystrophies this is not the case. As a muscle fibre contains the elements that are contractile (myofibrils), and therefore produce the movement, when the muscle fibres are lost, so are the myofibrils and hence the ability to contract a muscle, see Figure 2.3.

Depending on the muscle and the number of muscle fibres affected, the sufferer experiences a sensation of weakness and when a sufficient percentage of fibres is lost, so is the ability to perform movement. In cases like Duchenne's Dystrophy, the muscle tissue is replaced by connective and fat tissue, in a process called pseudohypertrophy, see Figure 2.4 ^[38].



(a) *Healthy tissue* (b) *Early case of Dystrophy* (c) *Advanced case of Dystrophy*

Figure 2.4 : *Transversal view of muscular tissue for Duchenne's Dystrophy - note the lack of muscular fibres and the proliferation of fat tissue in (c) ^[38].*

The muscle tissue also requires certain chemicals that are transported in the nervous system. When a person exercises their musculature, the muscles experience a process of hypertrophy and these chemicals are known as trophic substances. When the muscles do not exercise and stop receiving these trophic substances, they undergo an atrophic process. There are three main classes of atrophies :

- Spinal muscular atrophies
- Progressive muscular atrophies
- Peripheral neuropathies.

In most of these atrophies, there is a degeneration of the lower motor neurons with no evidence of pyramidal tract or peripheral nerve involvement. The pyramidal tract is a subsystem of the motor system which, from the cerebral cortex passing through the

Spinal Cord, instructs the execution of almost all the voluntary and complex learned movements of the body (except the eyelids). Every movement understood as a sequence of changes in the contractions of different muscle fibres has to go through the lower motor neuron.

These cells receive multiple excitatory requests from the pyramidal and extra pyramidal tracts and from the sensitive areas involved with reflexes. Each lower motor neuron innervates a group of muscle fibres, referred to as a motor unit. When a motor neuron is excited, the muscular fibres it controls will contract, if the motor neuron is inhibited the fibres will relax ^[39]. Thus in cases of muscular atrophies there is no evidence of spasticity or hyper-reflexia, the problems are those of paralysis, hypotonia and areflexia.

2.4 Summary

Lower limb disease as explained in this chapter is a non reversible condition that render^s the muscles in the legs useless either by physically impairing them or by severing the control mechanisms that allow them to be flexed and relaxed at will. At present although there are promising research developments a cure is not available to rehabilitate the walking function. Therefore a mechanical implementation is a viable alternative until a definitive cure or treatment becomes available.

Non pathological human gait is the subject of the next chapter, where clinical gait analysis and light studies form the basis for understanding human gait and its implication for the design of the exoskeleton

Chapter 3

HUMAN GAIT

3.1 Introduction

Human bipedal locomotion and its understanding through rigorous analysis, has for many years enabled clinicians and health specialists to identify normal and pathological conditions in human locomotion.

Locomotion is the process by which humans (and animals alike) move themselves from one location to another. It includes; starting, stopping, changes in speed and direction and compensation for changes in slope. Walking is a non-specific term whose interpretation is that of a cyclical pattern of body movements repeating themselves over and over, step after step. It is usually defined as the rhythmic displacement of body parts that maintain the individual in a constant forward progression ^[6] (see Figure 3.1). In research terms, gait analysis generally refers to the analysis of the rhythmic movement of the body in the sagittal plane while walking, excluding starting and stopping (see Figure 3.2).

Human walking differs from that of the majority of mammals in that most animals are quadrupedal. When walking slowly, quadrupeds tend to coordinate their legs so that three of their feet are always in contact with the ground. Balance is a key issue here, because with humans stability is lost as we are bipedal. Although coordinating two legs seems simplest, in fact it demands a more complex control strategy ^[6].

Learning to master erect bipedal locomotion takes up the first years of our lives. It is only when our peripheral nervous system completes its process of myelination that we are able to attempt to stand up on our own, roughly at about our first year after being born, ^[39]. So if walking is a learned activity, it is logical to argue that personal peculiarities can be

detected alongside a basic common pattern. Not only do individuals walk differently due to a range of factors such as anatomical differences, but everyone has their own idiosyncratic way of walking and it is not a simple task to find an average person.

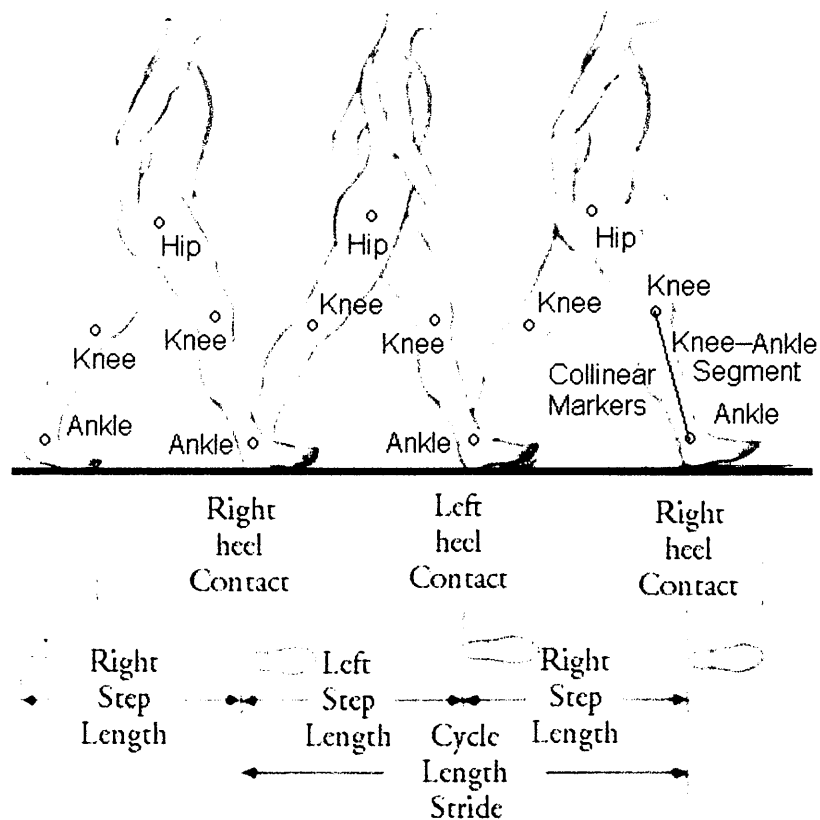


Figure 3.1 : Definitions of walking cycle dimensions, progression refers only to sagittal plane ^[6]

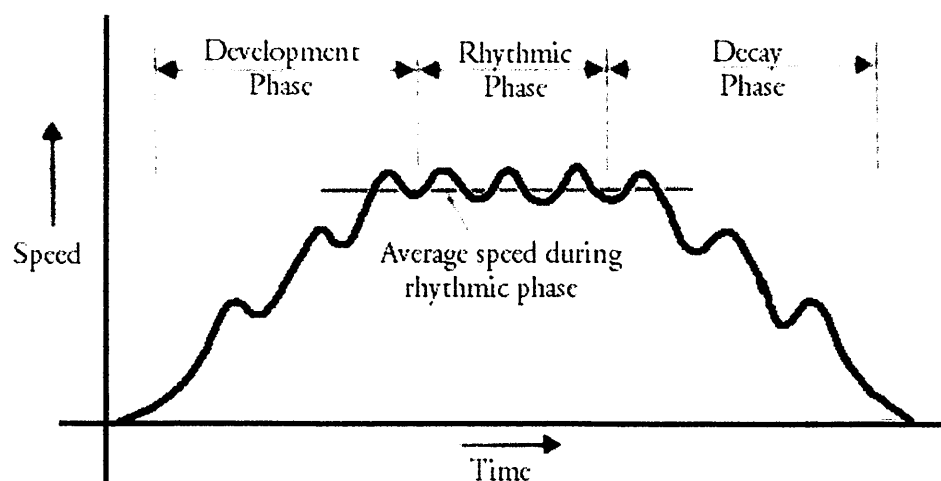


Figure 3.2 : Accelerographic analysis of trunk speed during complete act of human walking, from standing start to stop. Rhythmic phase shows where clinical gait Analysis focuses its scrutiny ^[6]

However, our anatomy imposes a common set of similarities in the way that each individual walks. All of us must oscillate our legs, and as we do it our bodies rise and fall with each step. Superimposed onto the movements parallel to the plane of progression (sagittal plane) are other small movement occurring in planes close to the coronal and transverse planes of the body. These are not really taken into account as most of the analysis centres around the sagittal plane.

3.2 Gait Analysis

Human locomotion expresses itself in many different ways, depending on many different factors. Gait, for instance, is influenced by the characteristics of the terrain, the motivation (a slow paced calm stroll while talking, the purposeful stride to keep an appointment, the carefully selected movements of a soldier on parade), and of course more technical factors such as age, sex, build and height ^[40].

Lettre ^[41] has described human locomotion in relation to three very distinct stages :

- Development stage - from rest to constant velocity.
- Rhythmic stage - constant average velocity.
- Decay stage - from constant velocity to rest.

In order to simplify the study of human gait, most research ^[6] has focused on the rhythmic stage also known as free speed-walking. The basic gait can be divided into different movements which can be individually kinematically analysed. Observations ^{[6] [40]} show a remarkable consistency between individuals and this has led to the identification of events

regularly occurring in a certain interval of time while walking. This succession of events defines a gait cycle.

3.3 Gait Cycle

The gait cycle refers to the cyclic pattern of bodily movements that repeat themselves over and over, step after step when walking. This process of locomotion basically produces the movement of the body by supporting all of the weight first on one leg and then on the other. As the body passes over the supporting leg, the other is swinging forward in preparation for its next support phase, see Fig. 3.3.

During walking, one full *gait cycle* or *stride time* signifies the time interval between two consecutive heel strikes of the same foot with the ground. The full gait cycle is divided into two major parts :

- Stance phase
- Swing phase.

The stance phase is defined as the percentage of the cycle when the foot (of the stance leg) is in contact with the ground (while the other leg, the swinging leg, is moving forwards).

The swing phase is defined by the time when the foot is in the air.

Typically, initial foot contact is at the heel. As soon as the foot strikes, both feet are in contact with the ground, this initial *foot strike (heel contact)* is designated as 0% (the second foot strike then being 100%). About 12% into the cycle, the swinging leg is lifted with the toe being the last part to leave contact. This event is also known as *toe off*. After half the cycle, the swinging leg lands and first contact is again at the heel.

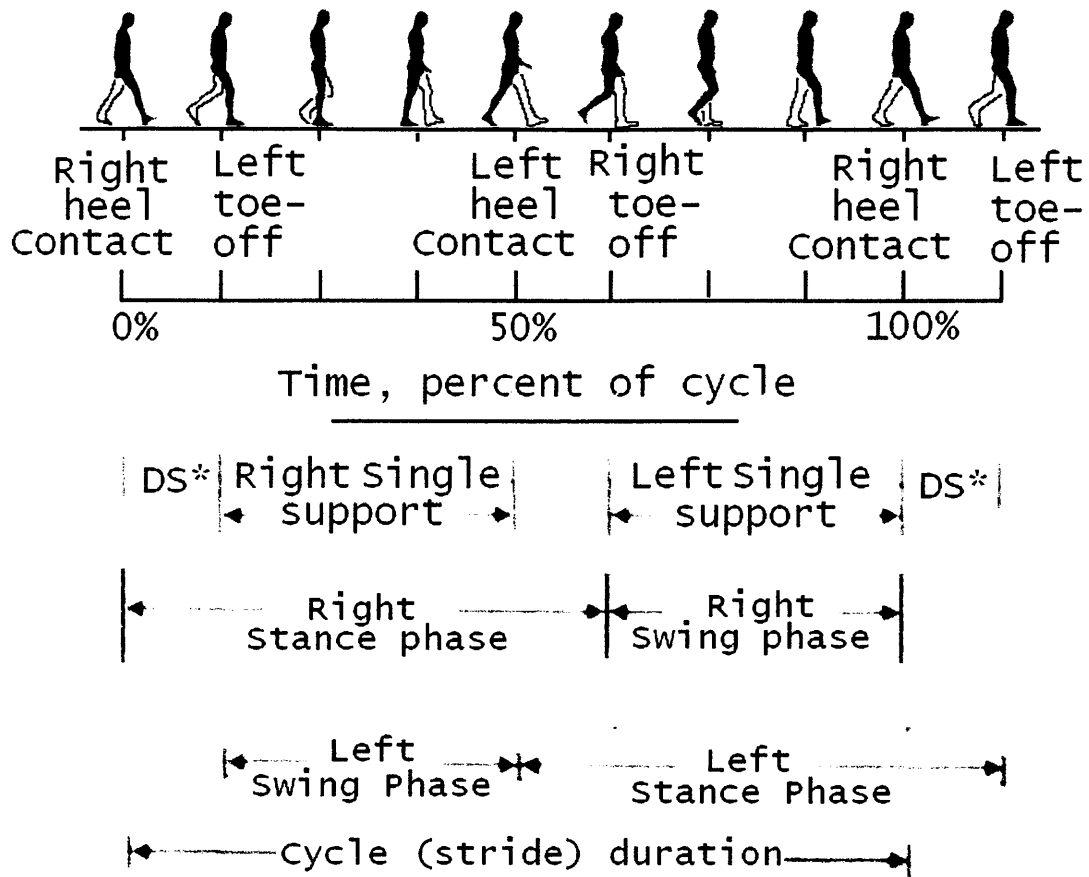


Figure 3.3 : Time Dimensions of walking cycle ^[6]

The stance phase for each foot starts with the heel strike (double support) and finishes with the toe off (double support). This lasts around 62% of the cycle whilst the swinging phase takes up the remaining 38% ^{[6][40]}.

After *toe off* the leg is in swing and the weight-bearing limb is in single limb stance. As the body passes over the fixed foot, the centre of mass rises to its peak elevation, while the forward and vertical velocities decrease. At this point, at around 30% of the cycle, forward shear then reverses to aft shear (potential energy is at its maximum and kinetic energy is at its minimum), the centre of mass falls, and vertical velocity increase ^[6]. Once the peak in elevation of the centre of mass is achieved, it falls until it is stopped by means of the opposite leg foot strike. This is why walking can be understood as a cyclical controlled fall.

Opposite foot strike occurs at around 50% of the cycle and then the whole process repeats itself. Each foot thus spends part of the walking cycle in contact with the walking surface and the rest of the time on the air. During the period of contact with the walking surface the foot remains stationary providing support for the leg, during this stance phase the heel of the foot strikes first and the toe strikes last. If gait symmetry is assumed then:

$$\text{Stance Phase Time} = \frac{\text{Cycle Time}}{2} + \text{Double Support Time} \quad \text{Eq. 3.1}$$

and therefore:

$$\text{Swing Phase Time} = \frac{\text{Cycle Time}}{2} - \text{Double Support Time} \quad \text{Eq. 3.2}$$

By analysing the gait cycle it is possible to establish a general pattern common to the majority of people exhibiting normal gaits. The body is supported through a dynamic, continuous balance of ground reaction forces and during this dynamic balance, each foot moves from one support position to a next support position. For the purposes of human gait analysis, this movement will be taken to be in the direction of progression i.e. in the sagittal plane.

This cycle has been expressed diagrammatically by Perry ^[20] as shown in Fig. 3.4. Figure 3.5 shows the various conditions and actions associated with human gait. These form the basic requirements for any form of bipedal walking implementation, thus any form of bipedal implementation requires to specify body motions (depending on the anatomical geometry of the biped).

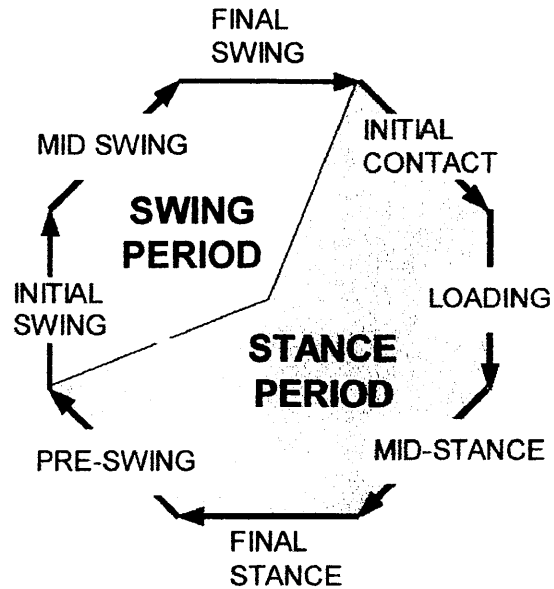


Figure 3.4 : Perry's gait nomenclature for subdivisions of the walking cycle ^[20]

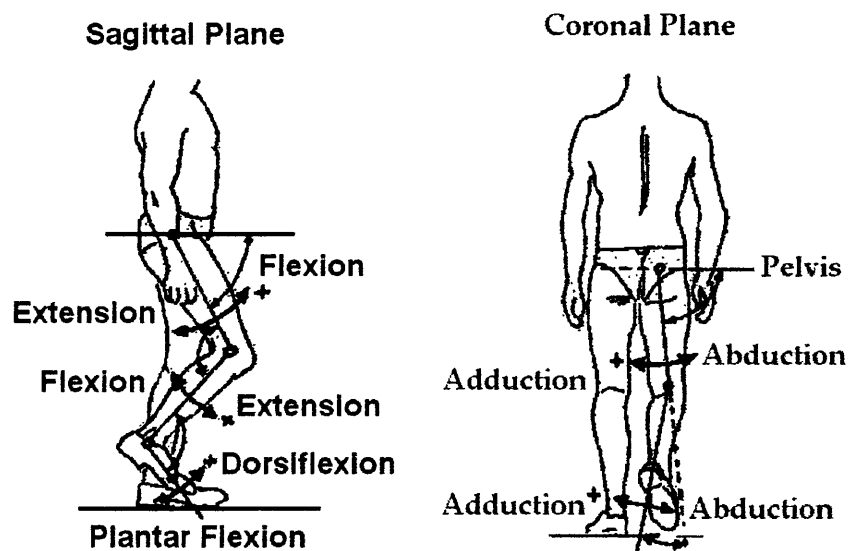


Figure 3.5 : Human gait definitions for different types of basic movements ^[6]

Because of the inherent instability of the system (as at any given time a maximum of only two points of support are available) at each step the body speeds up and slows down. The stance foot starts its support function ahead of the body and then it passes under it finishing at the back. As the body passes over, the stance leg rises and swings until it touches the ground again.

3.4 Segmental Analysis

Analysis of the process of walking takes into account the displacements of the entire body through space. However, this displacement is achieved by angular displacements of the individual segments of the body about the axes of the joints. Therefore, useful tools for describing human locomotion are the curves described by the angular and vertical displacements of the hip, knee and joint.

Kinematically speaking, the relevant information is contained within the joints as they are the points of attachment to which the segments are linked. As these segments lack independence of movement, the effective centres of mass may be located outside the segments. The targets for translational measurements are thus located as closely as can be estimated to the joint centres. This task is difficult to achieve with total precision because human walking is in reality a three-dimensional motion (See Fig. 3.6), To simplify its analysis it is useful to use only the sagittal plane ^[6].

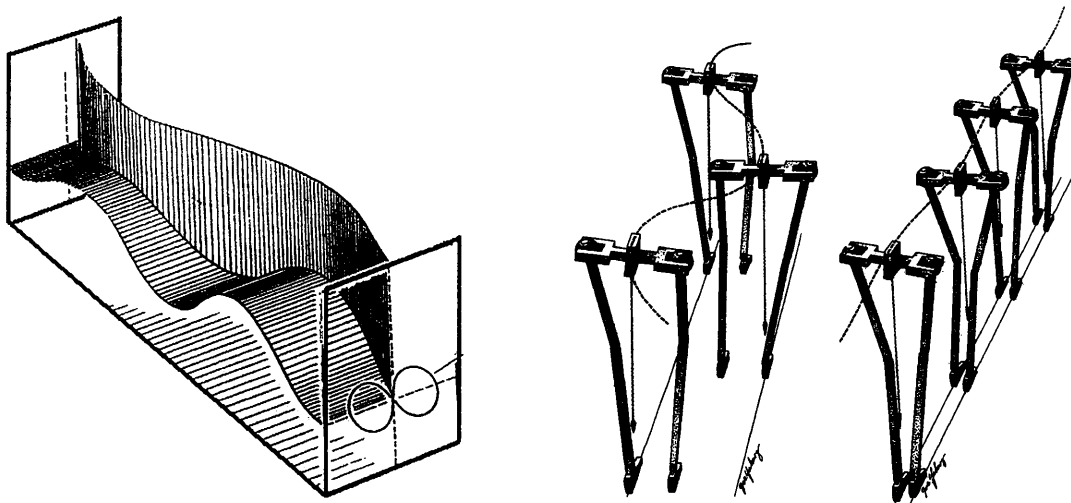


Figure 3.6 : *Three-dimensional representation of the movement of the middle point in between hips around the sagittal and coronal planes for human gait* ^[6]

The implication is that in three-dimensional bodies, the joint centres are not located on the surface of the body and clearly are not accessible as targets for measurements. During a typical clinical gait analysis procedure, markers are located on the surface of the patient indicating interior points. A three-dimensional optical system then records the movement of the targets in all three dimensions and errors can be removed by projecting onto a single plane for two-dimensional analysis. This may be either the sagittal or the coronal plane, see Fig. 3.7 (a). In engineering terms, the coronal plane corresponds to the frontal view or projection and the sagittal plane to the side view, a third, top, view is known as the transverse plane ^[6].

In reality, systems like that by Viacom ^[16] limit the information acquired to a few possible motions and although a loss of accuracy and precision will be experienced, there is an overall gain because the system does not heavily rely on precise placing of the markers, which theoretically should be located on two perpendicular planes. Most of the time this is almost impossible to achieve, so it is preferable to design the system to be more robust, despite it being less accurate.

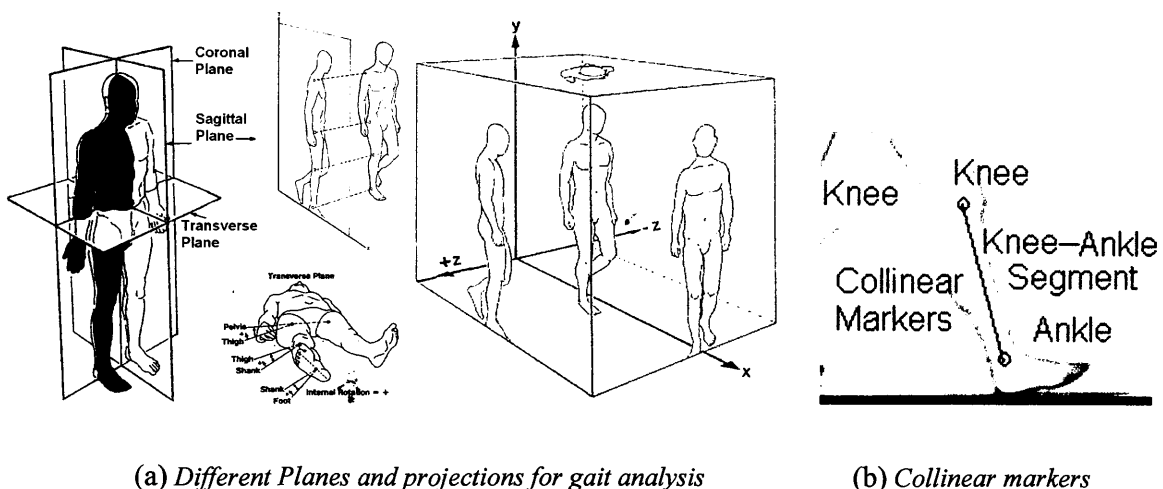


Figure 3.7 : Three-dimensional projections onto a single plane for two-dimensional analysis ^[6]

As normal human walking is achieved by the action of muscles producing rotations of the different segments of the leg, these angular changes or joint rotations are the principal subject of measurement during clinical gait analysis along with the ground reaction forces. The segments are obviously those enclosed by two collinear markers, see Fig. 3.7 (b), and the angular relationship between these two segments is represented in a plane perpendicular to the axis of rotation.

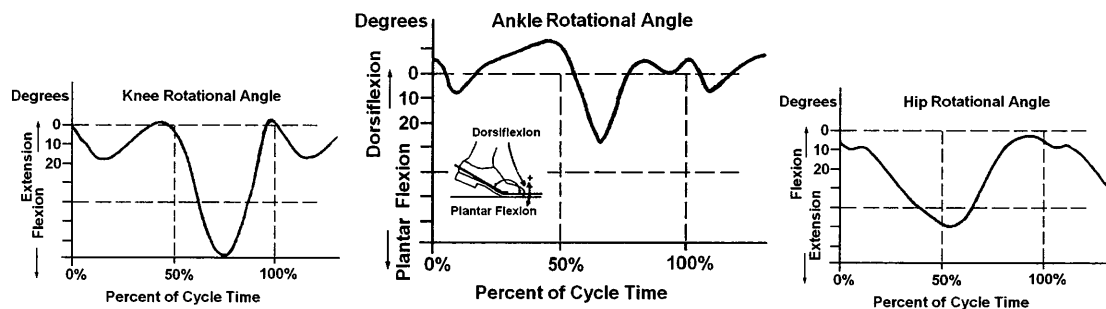


Figure 3.8 : Typical rotational joint values (in degrees) for an average gait of 110 steps/min[†] [6]

Figure 3.8 depicts the typical values for normal gait for the full cycle for the rotational joint values of the hip, knee and ankle. Unless otherwise stated, the values chosen have been taken from individuals walking at about 110 steps/min, which corresponds to a normal average pace.

The hip rotation is enclosed by the angle formed by the pelvis and the thigh. Basically, the hip extends during most of the stance phase, it starts to flex around the time of the heel impact of the other leg and reaches its maximum value at around 85% of the cycle, rebounding a little bit upon heel impact. Knee flexion maintains the hip in a nearly maximally flexed position during the first 10% of the walking cycle. The range of hip

[†] Approximately 85 metres per minute.

flexion-extension correlates with the stride length; flexion and extension values increase or decrease to accommodate changes in the stride length.

The knee rotation is contained in the angle formed by the thigh and the shin. Rotation starts with initial heel contact in preparation for maximum swing phase at about 15% of the cycle. This initial flexion is closely related to the walking speed, increasing as walking speed increases. The knee then starts to straighten, reaching maximum extension just before the middle of the cycle. The knee then starts to flex, reaching complete flexion at around 70-75 % of the cycle. It then straightens up again, reaching maximum extension just before heel contact.

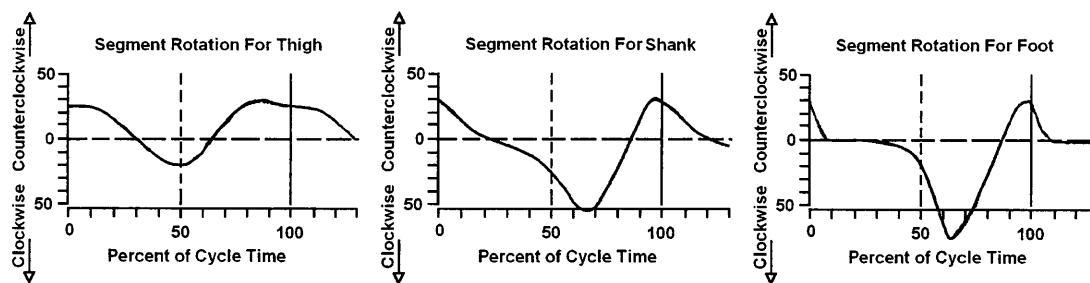


Figure 3.9 : Typical rotational segment values (in degrees) for an average gait of 110 steps/min^[6]

Segment rotations are usually described in terms of the way the segments incline either clockwise or counter-clockwise relative to the joint they pivot around. For example, the thigh at heel contact is at nearly its maximum counter-clockwise position and is then set to rotate clockwise throughout most of the stance phase. Midway through the cycle it reaches its maximum clockwise position and by the time of heel contact the swinging leg starts its counter-clockwise rotation (swing phase). By the time of toe-off, the thigh is approximately vertical and continues to rotate counter-clockwise until it reaches its maximum position of rotation at around 90% of the cycle.

The foot curve of Fig. 3.9 shows how when approaching heel contact the foot is held at around 20° and is then lowered to 0° , at which point it is parallel to the ground. The foot remains flat on the ground until the middle of the stance phase, when the heel lifts off the walking surface, with the toe being the final element to lift off. A clockwise rotation is sustained for the remaining part of the phase. Throughout the swing phase, the foot then rotates counter-clockwise in preparation for heel contact. Similar analyses can be made for the other segments, the shin basically rotates clockwise during the stance phase and counter-clockwise during the swing phase.

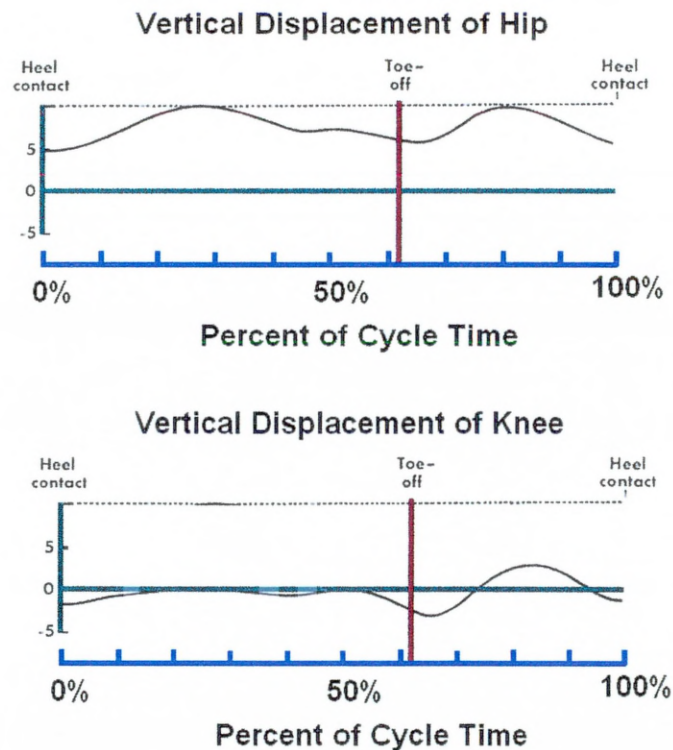


Figure 3.10 : Typical vertical displacements (in cm) for an average gait of 110 steps/min ^[6]

Vertical displacements for the hip and knee are shown in Fig. 3.10 from which there seems to be a marked separation of the stance and the swing phase. For the hip, the maximum point of swing elevation occurs at midswing, towards the end of a stance phase, dropping to its minimum height at heel contact.

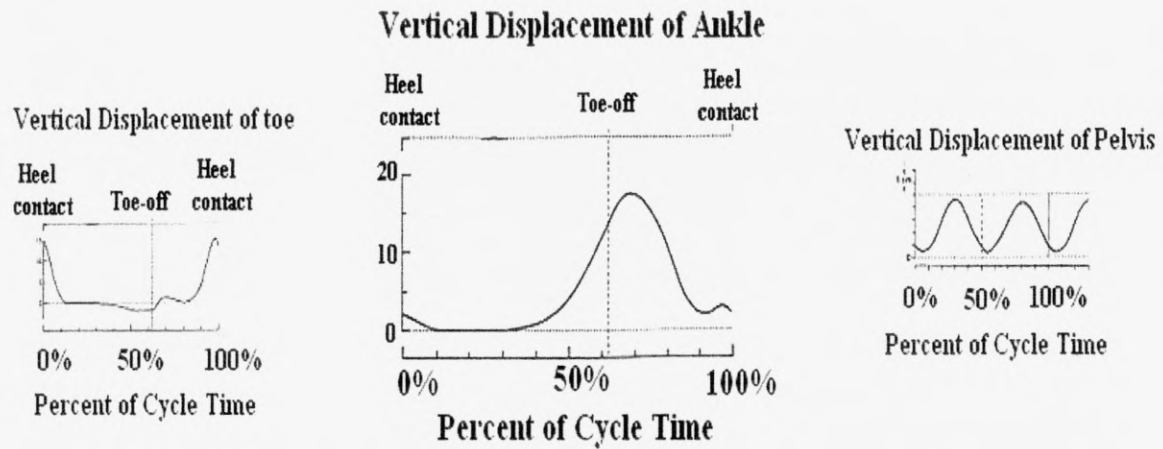


Figure 3.11 : Typical vertical displacements (in cm) for an average gait of 110 steps/min ^[6]

The knee is at its lowest elevation at heel contact and *toe off* and reaches its maximum value when the hip reaches its maximum swing phase flexion position. Referring to Fig 3.11, at heel contact the ankle is close to its lowest elevation, it drops slightly 2 or 3 cm as the foot is lowered onto the ground, just after the midpoint of the stance phase. The heel then rises slightly, raising the ankle joint and then continues to rise reaching maximum elevation at the time the knee is fully flexed. It then descends until the foot is flat on the ground, there is a small local maximum due to the way the foot lands on the ground with the heel touching first.

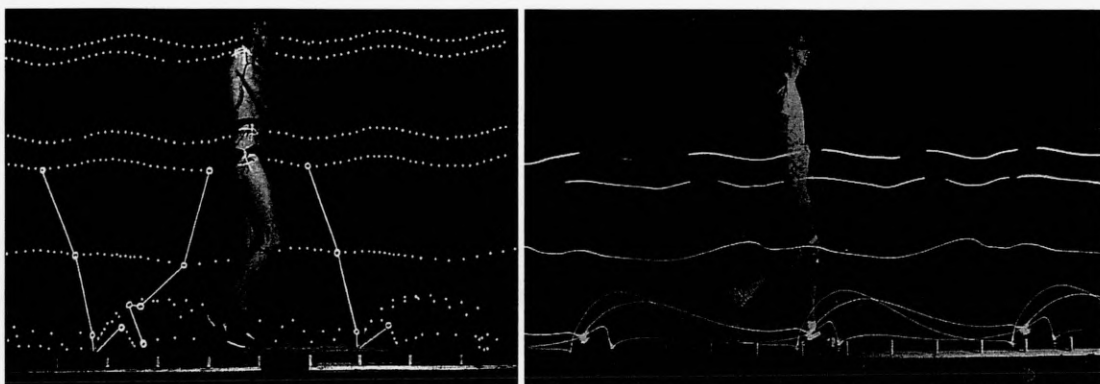


Figure 3.12 : Light studies for normal gait subjects walking at 110 steps/min ^[6].

3.5 Light Studies

Besides direct measurement of the values of rotation and vertical and horizontal displacement, there is another technique which is more intuitive and does not require the in-depth analysis associated with direct measurement. When using interrupted (see Figure 3.12) or continuous light (see Figure 3.13) it is possible to see all the relations of the rotations and displacements of the joints and segments all at once. Figures 3.12 to 3.15 show results obtained by this approach ^[6].

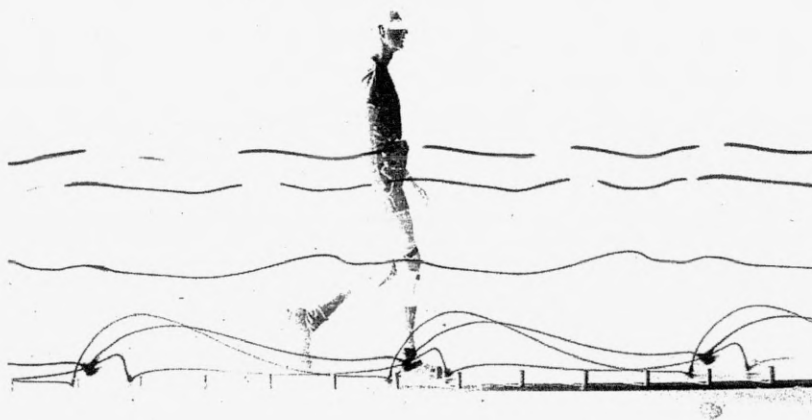


Figure 3.13 : *Continuous light studies for normal gait subject walking at 110 steps/min* ^{[6]†}

In particular, this technique of analysis is very useful in synthesising patterns that resemble the most human like gait. Thus, direct measurement using light studies is very useful when implementing a control strategy because it provides the criteria for controlling and maintaining the trajectories within the specified boundaries of human like gaits.



Figure 3.14: *Detail of ankle and foot tracks for continuous light studies for a normal gait subject walking at 110 steps/min* ^[6]

† Negative image.

In contrast to segmental analysis, light studies allow a direct comparison of a robot leg movement with human gait and are therefore very useful as a design tool. This is especially true with an exoskeleton configuration where the movement of the exoskeleton can be directly superimposed onto the human leg movement and the definition of geometries and gait patterns is simplified overall.



Figure 3.15: *Interrupted light studies for normal gait subject walking at 110 steps/min* ^[6]

3.5.1 Muybridge Technique

Eadweard Muybridge pioneered the technique of photographing the human body in motion in the late 19th century, placing several cameras that were activated in sequence to produce results such as that shown in Fig. 3.16 ^[42]. This series of studies can be digitised and the trajectories tracked just as in clinical light studies.

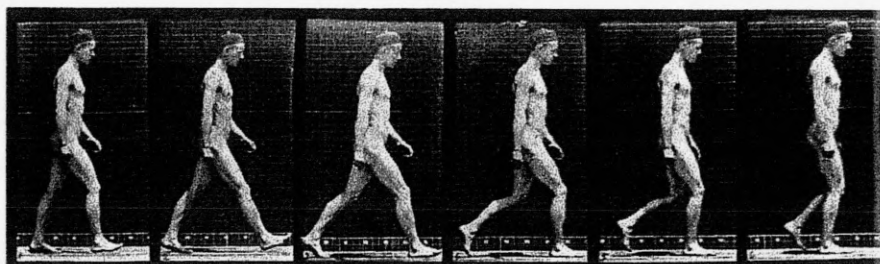


Figure 3.16: *Muybridge study on human motion showing a man walking on a straight line trajectory at a constant normal pace of 110 steps/min. This plate was first published in 1887* ^[42].

With the validation of this software marking and tracking technique, which will be called from here onwards the Muybridge Technique (MT), it is possible to further the concept and mark digitally any movement recorded on a video camera. Figure 3.17 shows an example of simple light studies carried out in the laboratory using the MT.

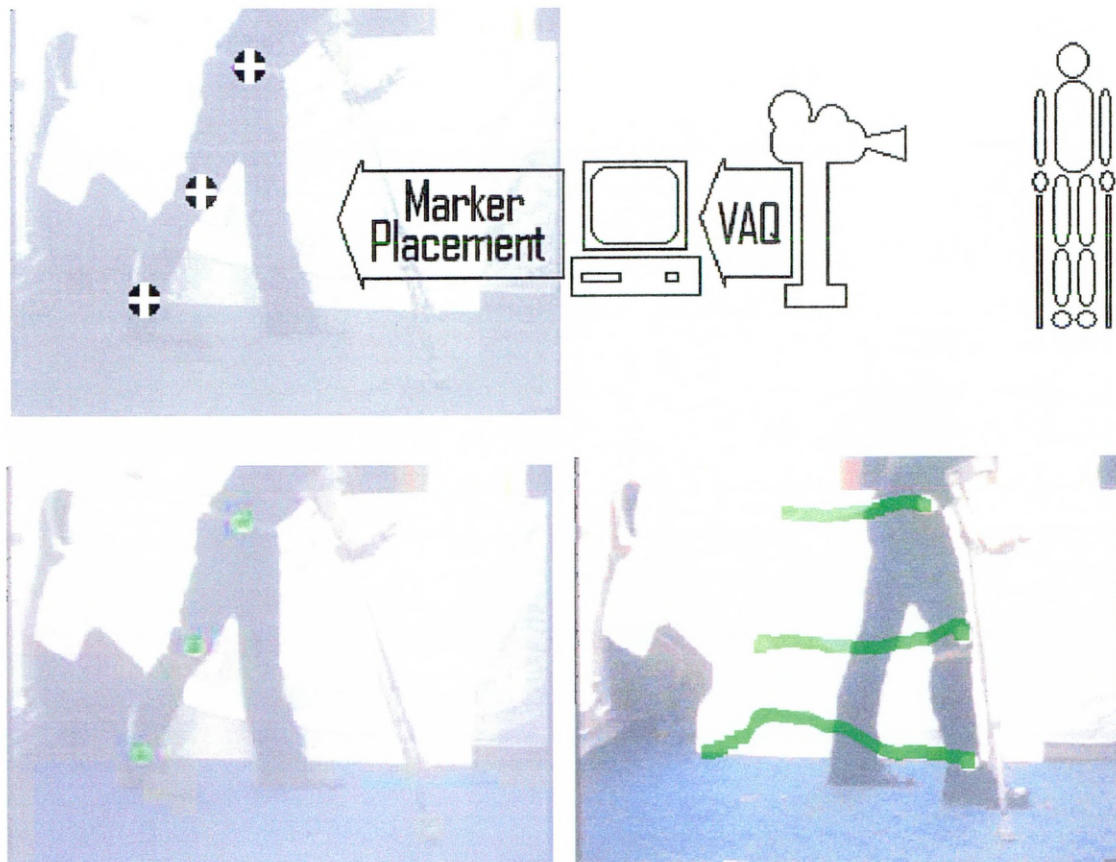


Figure 3.17 : *Light studies generated in the laboratory at the University of Abertay for normal gait subject walking with crutches at 30 steps /min. Software marker placement and digital visualisation.*

The videos are digitised with a generic Video Acquisition System (VAQ), then the strip of frames is used to follow the markers placed in different positions, figure 3.17 shows the markers located at the hip, knee and ankle level. Each selected frame is made into bit map file and then the markers are added up in succession in order to obtain a continuous description of the markers trajectories. The frames are then put back together in a generic video editing software package and made into an audio visual file.

The videos contain the movement of the segments and the tracking curves of the digital markers identify the pattern of movement for gaits such as that aided by crutches.

3.6 Energy Expenditure

With the kinematic information available it is possible to obtain an estimate of the energy levels for the different segments. The energy level of any segment has three components; a translational kinetic energy component, a rotational kinetic energy component and a potential energy component.

If, for every segment the mass, location of centre of mass and length are known, then it is possible to calculate the energy values for the different segments. There are tables available that specify these coefficients, known as Fischer's coefficients. They are based on measurements made with cadavers and were first published in 1886 ^[6]. The curves presented in the chapter make use of the velocity information derived from the forward motion analysis on the sagittal plane. Figure 3.18 represents the different energy levels for the leg. Unsurprisingly, the leg energy characteristic is minimum during the stance phase and achieves its maximum value during the swing phase. Most of the energy increase taking place during the double support period.

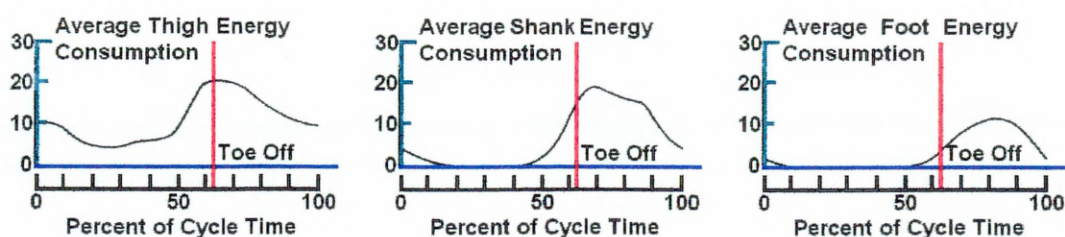


Figure 3.18 : Average energy levels in Joules for an individual walking at 110 steps/min ^[6]

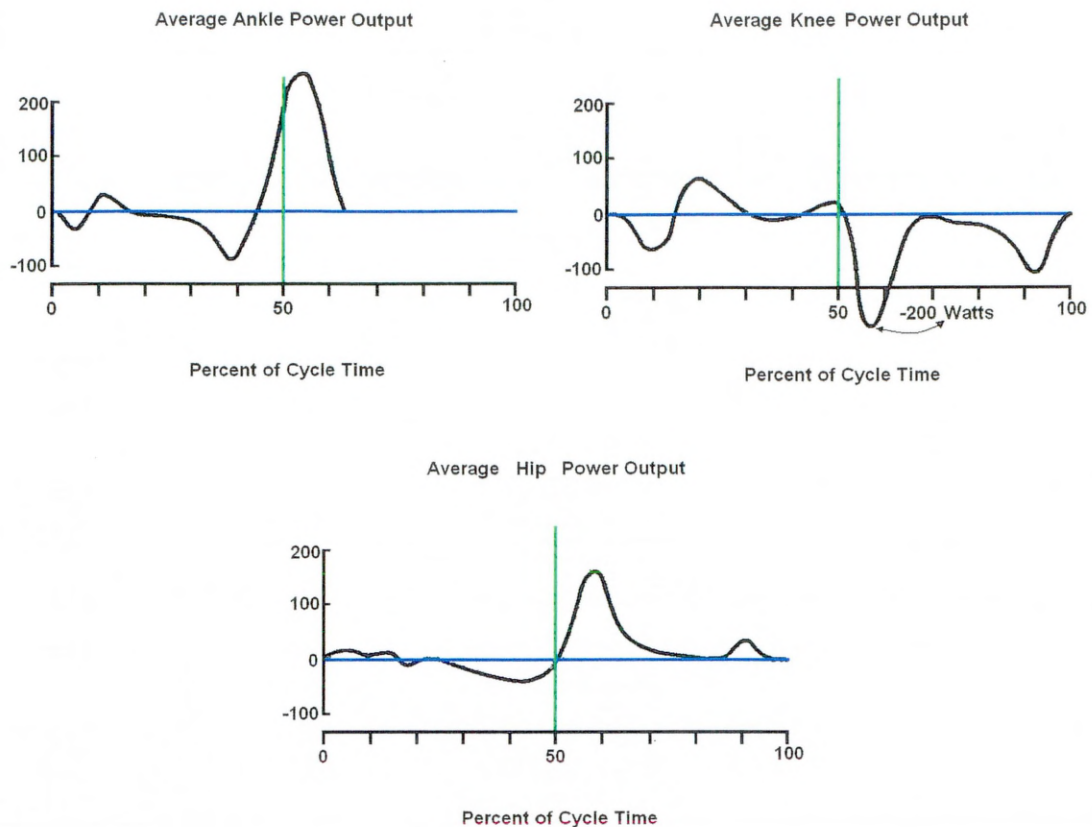


Figure 3.19 : Joint power curves for a normal individual walking at 110 steps/min (power output in Watts)^[6]

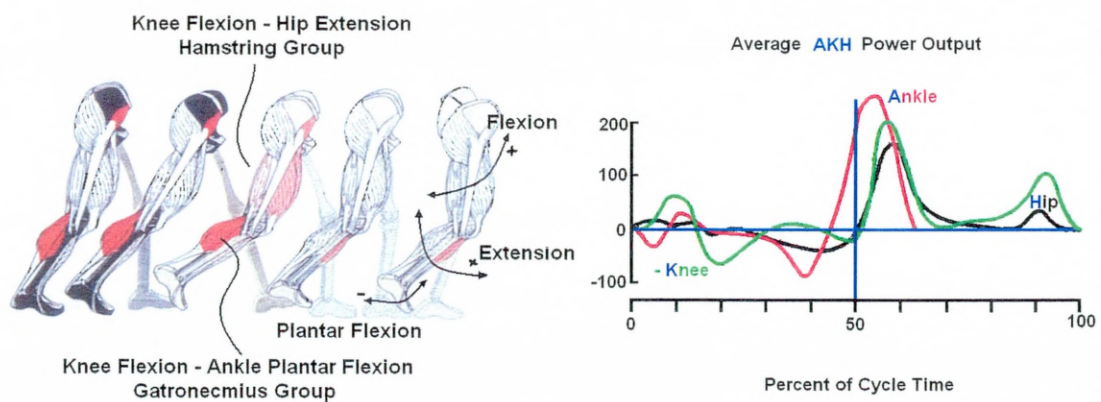


Figure 3.20 : Joint power output curves for a normal individual walking at 110 steps/min. Ankle contribution is greater when gastronecmius rotates the ankle into plantar flexion for toe off^[6].

The power requirement can be derived from these energy curves and compared with the power output developed by the joint moments and the angular velocities derived from the angular rotations of the segments at the hip, knee and ankle and shown in Figs. 3.19 and

3.20. At each joint the peak is reached during the double support period when preparing for the leg take off, the muscles acting on the knee are storing energy whereas the others are absorbing energy. At the knee there is a flexion movement produced by the hamstring and the gastronemius. While developing this flexion moment, they also develop extension moments at the hip (the gastronemius) and plantar flexion moments in the ankle (hamstrings), see Fig. 3.20. The knee stores energy as specified by Inman ^[6] in Eq. 3.3.

$$\frac{\text{Work Output}}{\text{Work Input}} = \text{Energy Stored} \approx 0.3 \quad \text{Eq. 3.3}$$

Here, the work output and the work input are calculated from the curves shown in Figure 3.19

If for the leg joints, all the powers are combined, a total power output is obtained, which can be compared to the power requirement for the leg as derived by Inman ^[6] from the energy levels at the thigh, shank and foot, see Fig. 3.21.

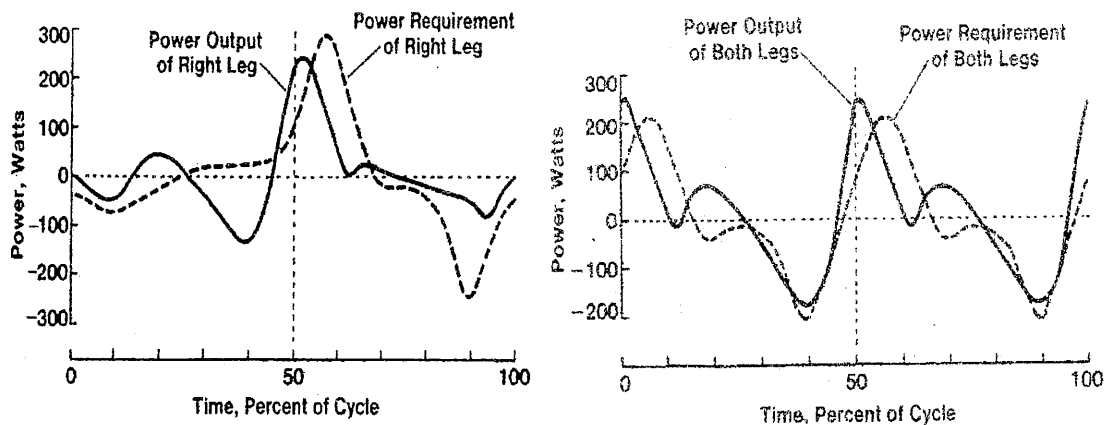


Figure 3.21 : Power Output v Power Requirement for a normal individual walking at 110 steps/min for one leg and for both legs ^[6].

The curves are very similar and the differences can be put down to inaccuracies in the calculation of the potential energy stored in the springs of the leg and perhaps in other parts of the body, which indicates that energy is actively transferred between the leg and other parts of the body.

3.7 Summary

Human gait is a very complex cyclical process whereby most of the hundreds of degrees of freedom available in the body are used to maintain the balance when the body is supported on either one or two legs. Bipedal walking is unsurprisingly extremely difficult to achieve if the biped in question aims to mimic a human walking.

A more realistic approach must be to simplify the balance issue, which is the critical one in terms of the added complexity versus the benefit that can be obtained from a more comprehensive approach where the assistive method employs only two legs to maintain the balance. Since walking is defined in terms of walking in the sagittal plane, constraining the movement to this plane also reduces the complexity of the problem. This simplification still leaves plenty of room to tackle delicate issues such as the definition of the number of Degrees of Freedom, the exact geometry of the exoskeleton and generation of trajectories.

Understanding how machines walk and the different geometrical configurations for mechanisms developed for walking machines is also important in defining the geometrical configuration of the exoskeleton. The next chapter elaborates on the subject of walking machines with special attention to medical exoskeletons and bipedal walking machines.

Chapter 4

MACHINE GAIT

4.1 Introduction

Professor Ichiro Kato, a pioneer in autonomous bipedal walking machines once said that the history of engineering is one of the enlargement and replacement of human capabilities. Mankind has invented a variety of vehicles that extend the locomotion function of human and animal legs through the utilisation of rotary motion. However, three main themes can be identified for the practical purposes of walking machines; rough terrain transport, bipeds (orthotics and humanoid) and climbing machines for hazardous environments such as nuclear power stations.

Rough terrain realisations are interesting to analyse since they deal with issues that are of relevance for the project. Since many individuals with lower limb disabilities use wheelchairs which as vehicles based around wheels or caterpillars, are bound to be limited by the environment in which they exist. Specifically, wheels are not as effective without appropriate roads being prepared for them.

This chapter will first present some of the legged machines that have been documented and researched in order to establish the common points of interest in design terms with direct implications for the design of the exoskeleton.

4.2 Rough Terrain Transport Applications

The earliest papers found in the 20th century were written some 40 years ago ^[43] ^[44] and dealt mainly with the same issues that are relevant today. Designers at that time realised that nature, having spent so much time in perfecting its designs, has populated our planet with a vast array of walking creatures. In the animal world movement is generally made

possible by means of articulated legs, hence locomotion in living creatures is sometimes referred to as articulated locomotion.

4.2.1 The W.H. Allen Quadruped

In the 20th century perhaps the earliest documented attempt for building an articulated walking machine was by A.C. Hutchinson in 1940. The project was developed at W.H. Allen & Co. Ltd. and consisted of a scale model for a very large armoured vehicle in the 1000 Ton Class. At the time military theorists like B. H. Liddell Hart ^[43] argued that all warfare was bound to degenerate into trenches and mud following the 1917 pattern and that the tank had been devalued by anti-tank guns and terrain.

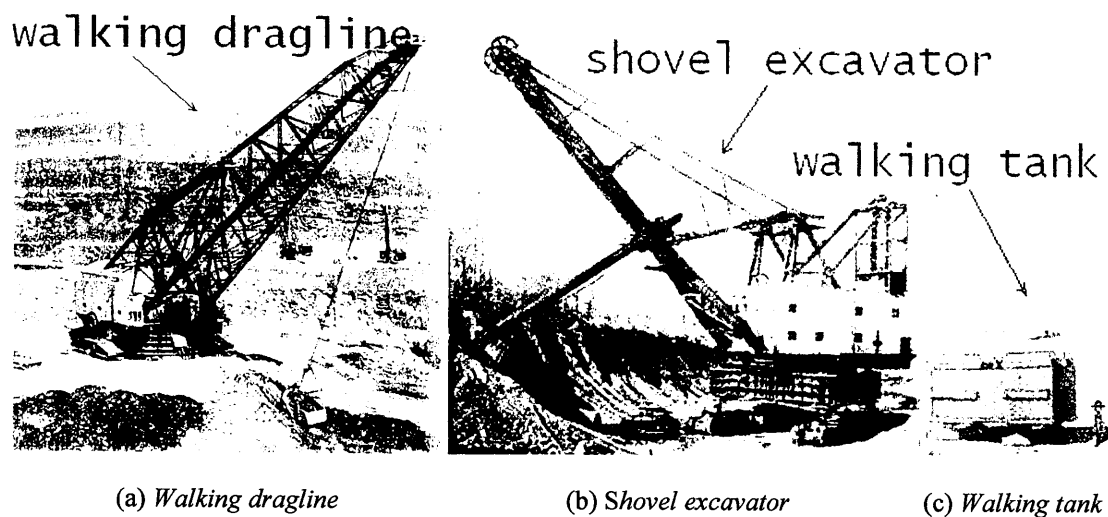


Figure 4.1 : *Walking machines* ^[45] ; (a) *A walking dragline that rises itself off the ground with its own feet;*
(b) *A pre-World War II 1000 ton shovel excavator;* (c) *A 12 ton walking tank .*

Following an idea for a walking fighting machine, the design brief stated that legs would be better than tracks and the designers opted for four legs and a quadruped crawl gait with a rolling thigh joint and a telescopic leg ^[43]. Hutchinson claimed to have had the idea after realising how much more effective farm horses were in the mud than tractors. The control

mechanism consisted of a feedback loop per leg, with the four legs each being controlled by the hand or the foot of the driver. Figure 4.1 shows a sketch of the proposed Mark I version of a 12 Ton armoured vehicle along with examples of dragline excavators.

4.2.2 The Land Locomotion Laboratory

Nearly ten years later a group associated with the University of Michigan and the US Army Tank-Automotive Centre (Shingley) initiated studies investigating walking vehicles ^[44] as a result of the interest of the US Army in rough-ground transport. After World War II, the group performed several studies in the laboratory including several types of levered vehicle capable of leaping, galloping, bouncing, and running strategies. A periodical, the Terramechanics Journal, was also developed in the mid 1960's ^[45].

Liston and Shingley ^[46], established the requirements for a practical walking machine as :

- The machine must have a uniform velocity while the feet are in contact with the ground.
- The stride must be long in relation to the physical dimensions of the walking machine to achieve adequate speeds.
- The height and length of the stride must be controllable by the operator.
- The height of the step should be large compared with the dimensions of the machine.
- The feet should have a high stride to return-time ratio, i.e. it must move slowly.
- A mechanism integral to the legs must be provided for steering the vehicle.
- The vehicle must be capable of moving either in the forward or reverse directions.
- The inertia forces and torques must be balanced.

- The energy lost in lifting the foot should be recovered when lowering the foot.
- The height of the body of the machine above the ground should be controllable by the operator.

A number of different machines such as those shown in Fig. 4.2 were built to try out a range of diverse mechanical solutions such as non-circular gears for reducing the accelerations of the legs and different numbers of legs. These were unsuccessfully tested along with some machines such as Spartaco Moritori's *Golden Horse* (see Fig. 4.3(a)) which fulfilled all of the listed criteria ^[46]. Ironically, the Land Locomotion Laboratory evaluated various designs from other inventors and rejected the Scruggs Walking Machine of Fig. 4.3(b) that would be the foundation of the design of the successful Komatsu underwater robot some 20 years later.

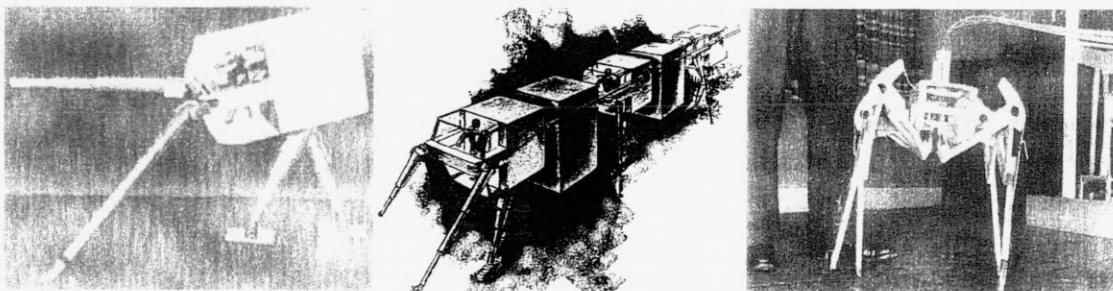
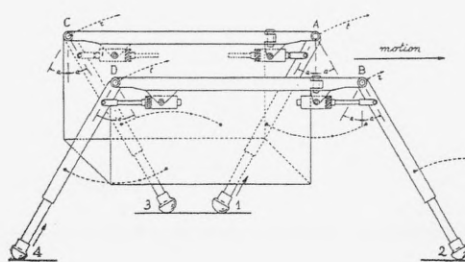
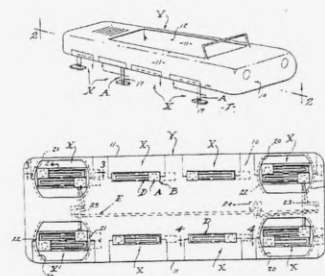


Figure 4.2 : Land Locomotion Laboratory Automated Anthropomorphic Machines ^[45].



(a) The 'Golden Horse'



(b) Scruggs' Walking Machine

Figure 4.3: Walking machines ^[45].

The Land Locomotion Laboratory worked very closely with General Electric in establishing the feasibility of a walking machine based on force feedback. The design study concluded that such a walking machine was feasible, and that the single most critical task to be undertaken by the operator would be that of maintaining static balance ^[46]. It also concluded that the major locomotion role would continue to be filled by vehicles based on wheels.

4.2.3 The Walking Truck

In the late 1960's the Land Locomotion Laboratory cooperated with General Electric on the Handyman project. This used manipulators with force feedback from the arms of the operator to provide force amplification.



Figure 4.4 : *Early concepts for the Walking Truck* ^[43].

The idea was to build a bipedal walking machine, however, the final chosen configuration was not a bipedal one but a quadrupedal one and the device was known as the “*Walking Truck*”. This was capable of carrying a 230 kg load and had a gait speed of 5 km/h. The vehicle was over 3m long and weighed about 1400 kg. Hydraulic actuators were chosen for driving the vehicle, and a 90 hp petrol engine provided the power for the pumps ^[47], see Figures 4.4 and 4.5.

The vehicle performed reasonably well but proved extremely demanding for the driver who had to control it with both hands and legs. The hand controllers for the front legs had a force feedback with a ratio of 1 to 120 reflected by the joint servo loops. The driver's leg movements at both hip and knee were followed in a geometrically similar way by the machine's back legs. It also had to be equipped with large mirrors for controlling the posterior legs. The result of this poor control configuration relegated the project to the laboratory.

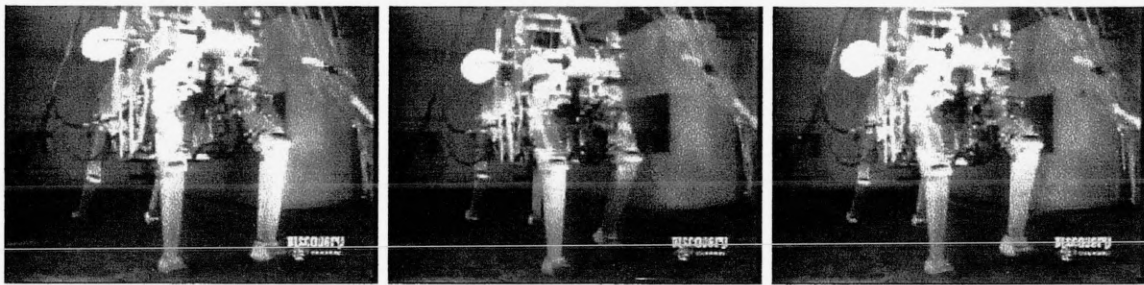


Figure 4.5: *Implementation of the Walking Truck (Image taken from the Discovery Channel)*

The project then evolved in the early 1970's into a version of a man-amplifier known as *Hardyman*. This was again a biped, but this time the exoskeleton was hydraulically powered, see Fig. 4.6. The master inner exoskeleton controlled the powered slave outer exoskeleton and the force amplification ratio was 25:1, i.e. if the operator lifted 250 kg it felt as if they were lifting 10 kg. The Achilles's Heel was the hydromechanical servosystem employed in the legs, the problem of balance could not be solved and *Hardyman* could never achieve effective unsupported walking^[45].

The real problem however was in controlling the interaction between the hydraulic servo joints, which sometimes resulted in a violently unstable behaviour of the machine. If an operator had been subjected to this, they could have been torn apart^[45].

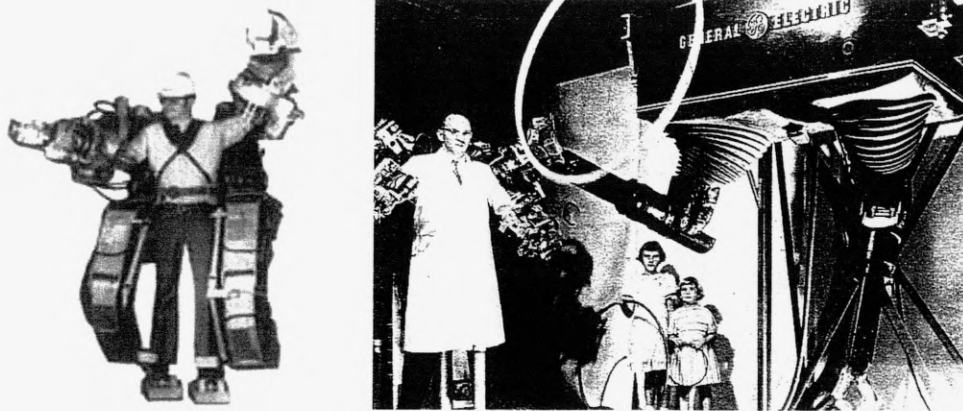


Figure 4.6: *Hardyman and operator* ^[45].

4.2.4 The Iron Mule Train

The last series of machines built without any electronic control was that known as the *Iron Mule Train* shown in Fig. 4.7(a) and developed by the Aerojet General Corporation. This was intended to work in tandem with others forming a train in which the first vehicle led the rest of the train. Static balance was again the key to the design ^[45].

4.2.5 The Phony Pony

The first attempt at introducing an electronic control strategy came with the *Phony Pony* project (see Fig. 4.7(b)) in the late 1970's. This used a finite state machine control strategy implemented by an electronic sequencer made of flip-flops.

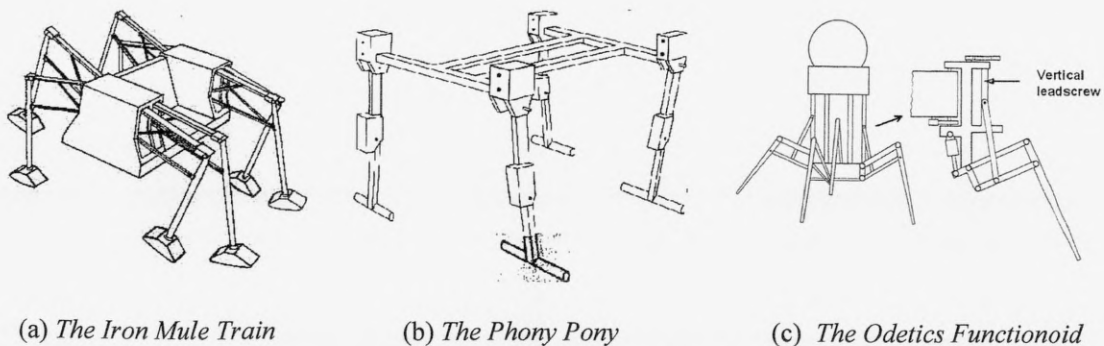


Figure 4.7: *Walking machines* ^[45].

As the electronics went through the different states, so the movement was achieved. It was also a quadruped with a passive load equalising the suspension mechanism ^[45].

4.2.6 The Odetics Functionoid

Other hexapods have been designed since, and one of the most successful designs is the Odetics Functionoid, created at Odetics Inc. of Anaheim and shown in Figs 4.7(c) and 4.8. This robot weighs 170 kg and a 24V, 25 A/h aircraft battery provides the self-contained power source with up to one hour of operation. An interesting feature of the design is the use of brakes on the motors and the storage of energy during the reverse stroke. The ability to maintain static balance no matter what the position, terrain configuration or pace was key to its design ^[47].

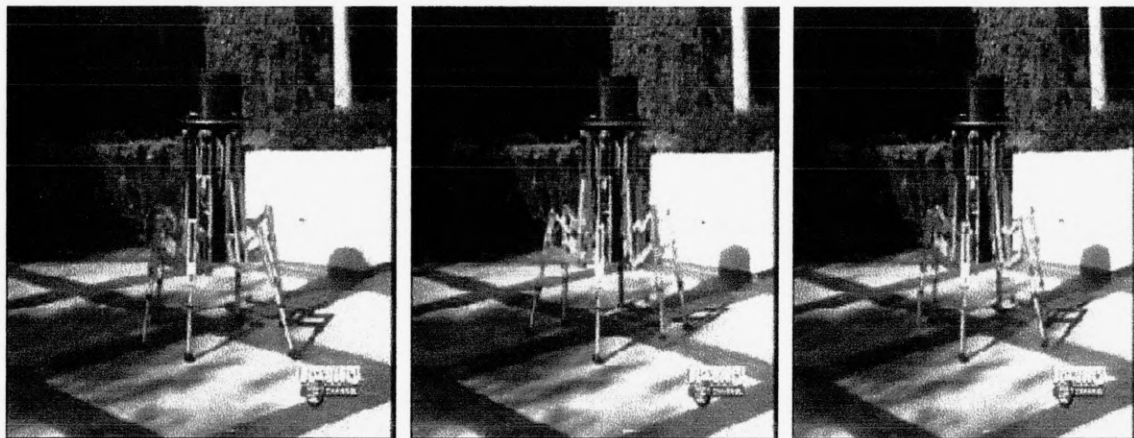


Figure 4.8: Odetics Functionoid *sequence for one step* (Image taken from the Discovery Channel).

4.2.7 The Adaptive Suspension Vehicle

One of the most expensive designs was the multimillion dollar *Adaptive Suspension Vehicle* from Ohio State University. The driver uses a six-axis joystick and computer assisted co-ordination of the legs. The selected gait pattern is aided by a scanner that searches for obstacles and checks ground configuration ^[47].

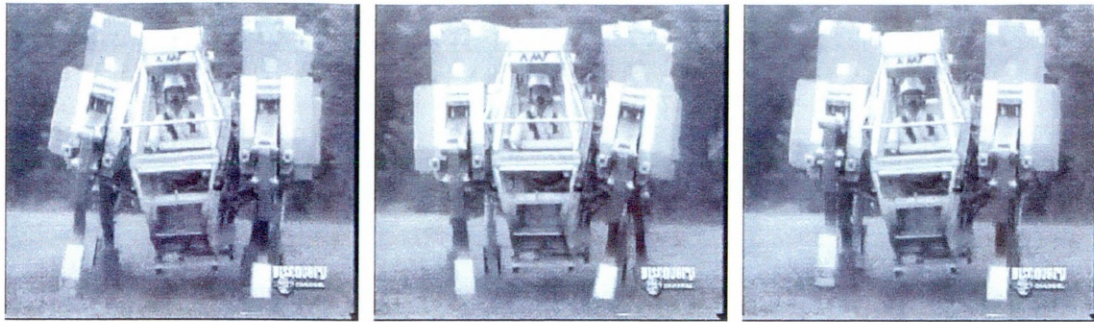


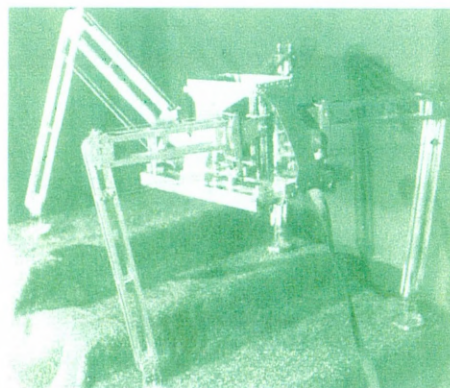
Figure 4.9 : Adaptive Suspension Vehicle *moving sideways* (Image taken from the Discovery Channel).

4.2.8 Sutherland's Hexapod

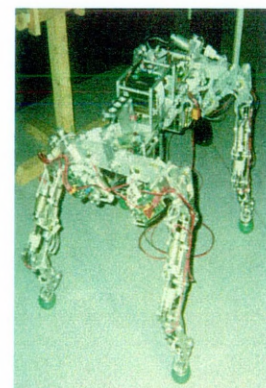
Designed at Carnegie-Mellon University, *Sutherland's Hexapod* (see Fig. 4.10(a)) was one of the first computer controlled man-carrying walking vehicles. The design of the leg mechanism was intended for control by a computer, which automatically coordinated the joint motions according to the navigation controls. Again it maintains static balance by means of a tripod configuration.



(a) Sutherlands Hexapod



(b) PV II



(c) BISAM

Figure 4.10 : *Walking machines* ^[45].

4.2.9 The PV II

Quadruped research at the Tokyo Institute of Technology saw a number of developments in walking machines in the early 1980's and the *PV II* of Fig. 4.10(b) was one of these designs. The leg design and gait strategy selects footholds while maintaining a constant gait in static balance by moving one leg at a time.

4.2.10 The BISAM

The Biologically InSpired Walking Machine (BISAM) of Fig. 4.10(c) was developed at the University of Karlsruhe and has been designed to perform sensor-based adaptation of motion. It has five active DoF's in the body along with the ability to rotate the shoulder and hip combined with reactive control that uses small modular units which take care of stability aspects, their overall output can be superimposed on the basic trajectory generation system ^[48]. The modular units implemented are :

EvenForceXY - Evenly distributes the forces at the feet touching the ground by shifting the central body in the xy-plane.

EvenForceZ - Evenly distributes the forces at the feet not through shifting the body, but by adapting the length of the legs in the z directions.

BodyHeight - Keeps the body at a specific ground level by calculating the representation of the body attitude from the average length of all the legs touching the ground.

BodyInclination - Inclinometers mounted on the BISAM's body keep the body parallel to the ground by rotating the body stretching and bending the legs

4.3 Bipedal Realisations

Most of the efforts dedicated to bipedal robots focus on integrating robots into normal day to day living, either as humanoid robots or as human strength enhancers. Because cities and the urban environments have been developed for two legged creatures, it makes sense that in order for the robot to be successfully integrated into our lives, they most certainly will need to walk. Problems are largely centred around achieving balance (see Fig. 4.11).



Figure 4.11 : Support ensure that this bipedal robot does not fall down. Balance is a key area of research in bipedal robots (Image taken from the Discovery Channel).

4.3.1 The Moscow State University semi-anthropomorphic biped

A project of semi-anthropomorphic nature was developed at the Institute for Mechanics of the Moscow State University by Lomonosov in 1988 ^[49]. Referring to Figs 4.12 and 4.13, the device consisted of telescopic legs and powered hips. Two dc servomotors drive each leg at the hip rotation and extension/retraction of the leg ^[14]

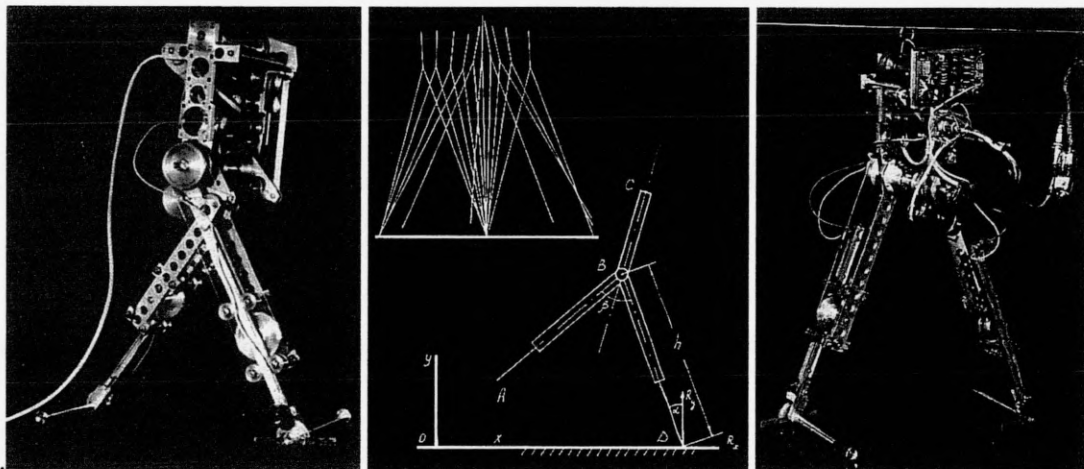


Figure 4.12: The Lomonosov Biped with telescopic knees ^[49].

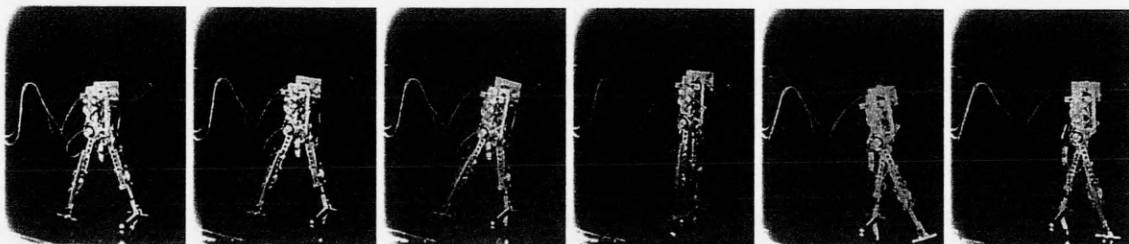


Figure 4.13: Sequence for a step of the Lomonosov Biped with telescopic knees ^[49].

This system is of particular relevance for the project as this biped, along with the BR-1 by Shih ^[50] developed at the National Taiwan Institute of Technology, shares the concept of telescopic legs that the proposed exoskeleton has at the core of its design.

Formalsky and Grishin explain that the robot control was designed to track the commanded path, the shin can move along the thigh over the rails, and in this way the overall length of the leg is controlled ^[49]. The shins were driven by a cable attached to a dc motor, the legs had passive feet that extended in the frontal plane thus avoiding tilting sideways, the robot could only perform a gait pattern in the sagittal plane.

The design imposed some restrictions as the two shins were linked together so that the superimposed step length of both legs remained constant, i.e. the extension of one leg caused the retraction of the other one. A dc drive maintains the longitudinal axis of the trunk in the middle of the angle formed by the two legs by swinging it during the single support phase ^[49].

The control strategy determined transition points ensuring that initial and final velocity in between transitions were matched. Times of transition were calculated to manage the overall velocity of the biped, force sensors defined the moment of the transition from single to double support.

The nominal walking regime for the biped was designed to comply with these requirements :

- The robot is not to fall forwards or backwards.

- The foot of the transferred leg should pass over the support foot without touching it.
- In walking the contact of the feet with the surface is assumed to be without slipping.

4.3.2 Taiwan Institute of Technology *Biped BR-1*

This biped was designed as an experimental test bed for the synthesis of walking patterns for ascending and descending stairs. Like the Lomonosov realisation, it featured variable length telescopic legs but balance was ensured by carefully controlling the centre of gravity (CoG) by displacing a weight along the y axis. This particular design relied on statically stable modes that maintain the CoG within its support region and by avoiding balance disturbance due to inertial effects from reciprocating legs by moving them slowly and by maintaining the Zero Moment Point, see Fig. 4.14.

The Zero Moment Point (ZMP) is the point on the ground around which the moments caused by the ground reaction forces are equal to zero ^[50], see also Fig. 4.14. It can be calculated by dividing the ankle torque by the total vertical ground reaction force ^[14]. When the ZMP exists within the length of the foot, preferably around the middle, the system is stable.

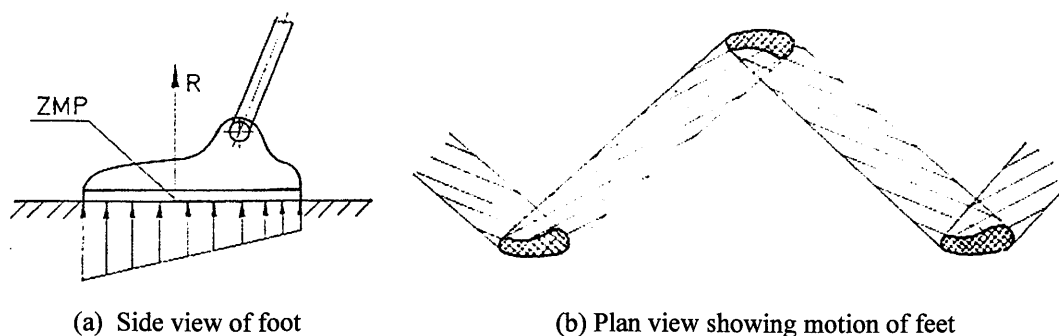


Figure 4.14: *Zero Moment Point concept and transition from single to double support for a step showing the dynamic stable triangular area where the CoG should be maintained* ^[50].

The BR-1 synthesised walking patterns while assuming the following conditions :

- The Biped is both left-right and forward-backward symmetric.
- The feet are rigid.
- The supporting foot is in flat contact with the surface.
- There is enough frictional force to prevent slippage.

The mathematical model is very similar to that used in the project and a full description and analysis can be found in Chapter 4 ^[50].

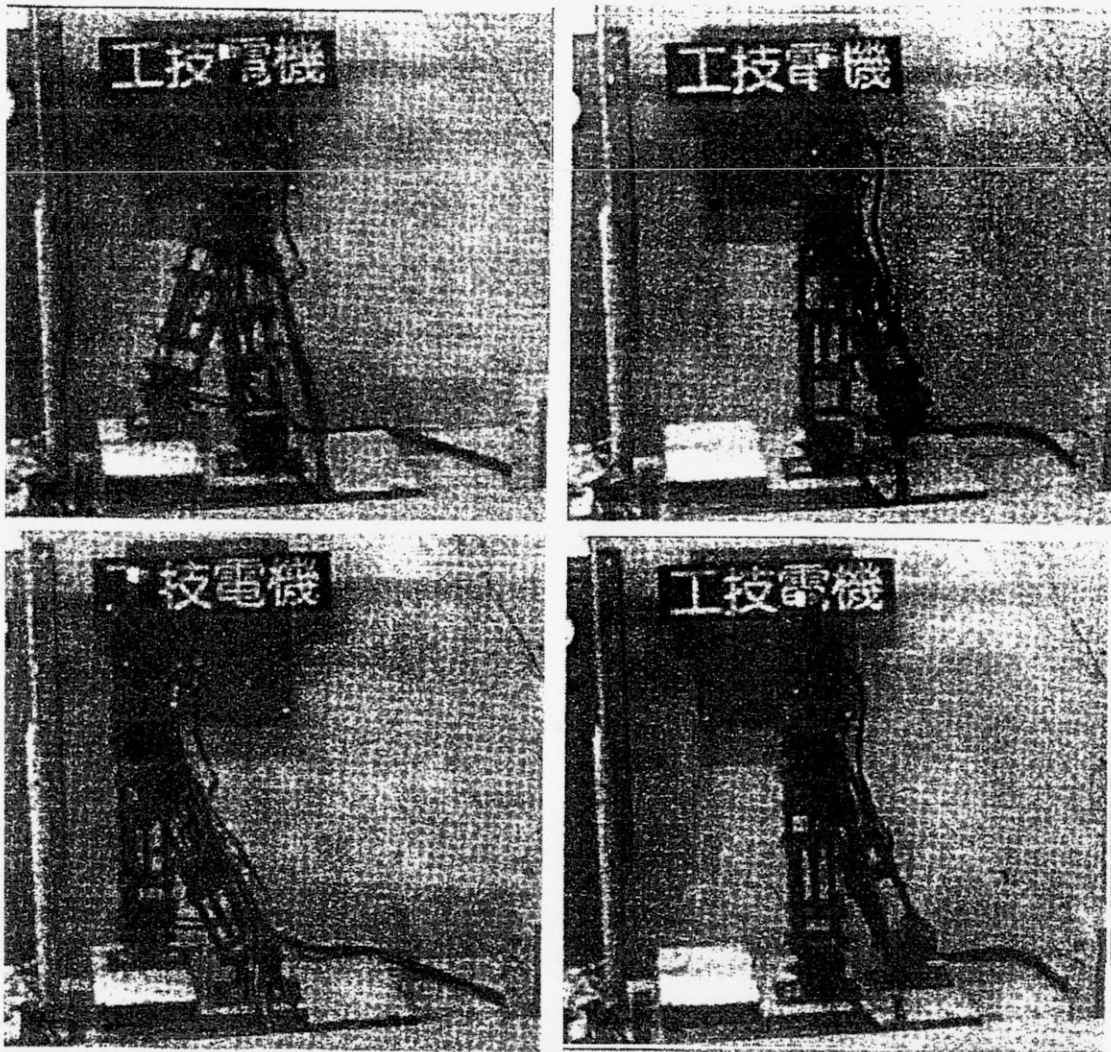
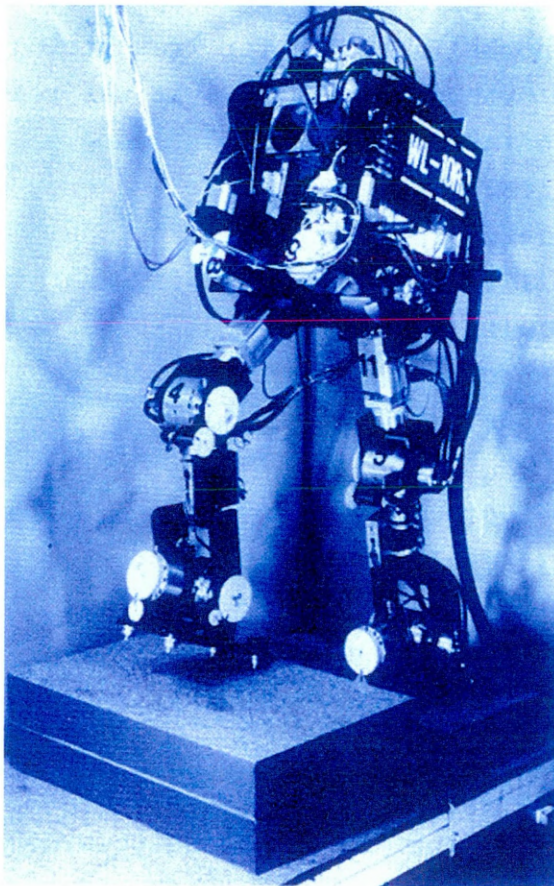
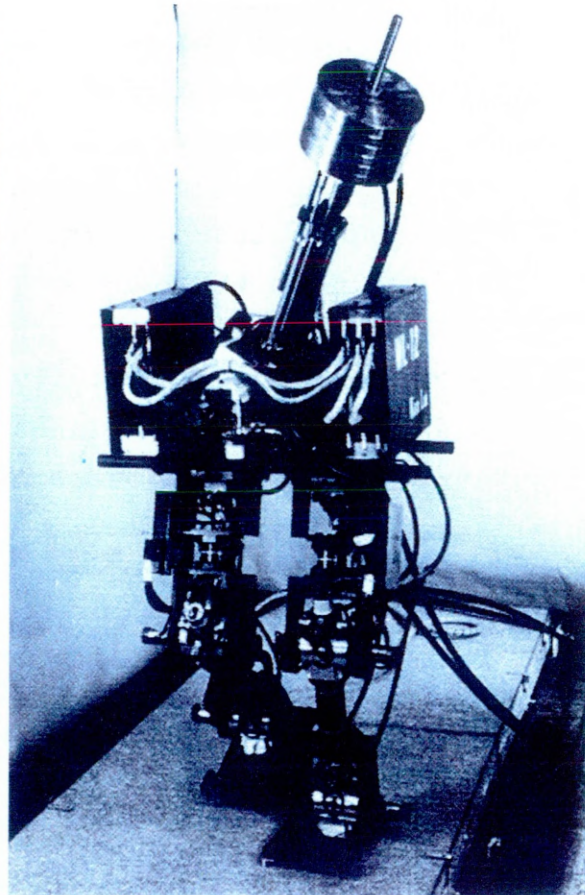


Figure 4.15: *Biped BR-1 with telescopic knees, sequence for step climbing* ^[50] (sequence is clockwise).

The imbalances caused by shifting the weight from side to side allowed for what Shih described as an “....*easy implementation of correction for imbalances*” ^[50]. Like Formalsky and Grishin, Shih had problems achieving vertical zero velocity of the landing foot immediately after contact. The solution implemented was to cover the foot with a shock absorber material. The swing foot was parallel to the surface of the ground and was maintained in that fixed orientation. Figure 4.15 shows *BR-1* in motion.



(a) The Waseda Leg refined WL-10RD



(b) The Wabot Waseda Leg 12 DoF WL-12

Figure 4.16 : Bipedal walking machines ^[47].

4.3.3 The WL-10RD

From the 1980's onwards several attempts to build working models of artificial bipedal walking devices have been made and in the Kato Robotics Laboratory at the University of

Waseda in Tokyo. One example of this bipedal robots is the *WL-10RD* (Waseda Leg – 10 Refined Dynamic) seen in Figs 4.16 (a) and 4.17, this bipedal machine was able to ascend and descend stairs and negotiate ramps of small inclination. On level ground it could walk with a pace of 2 to 5 seconds per step ^{[13][14]}.

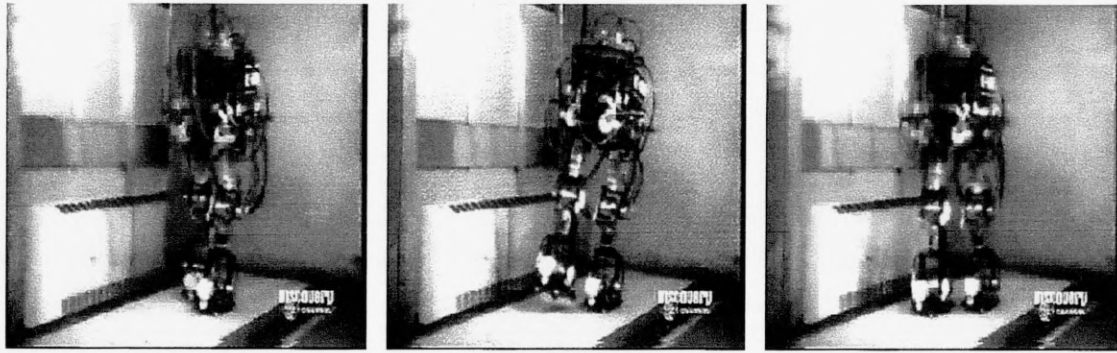


Figure 4.17: *Sequence for one step of the WL - 10RD (Images taken from the Discovery Channel).*

4.3.4 The Wabot WL-12

The WL-12 of Fig 4.16 (b) and 4.18 uses the concept of maintaining the Zero Moment Point around a region inside the foot outline and features complete microcomputer control with force sensors at the feet and at the joints and achieved dynamically stable conditions for transitions in between single and double support by moving the pendulum at the back for weight compensation ^[14].

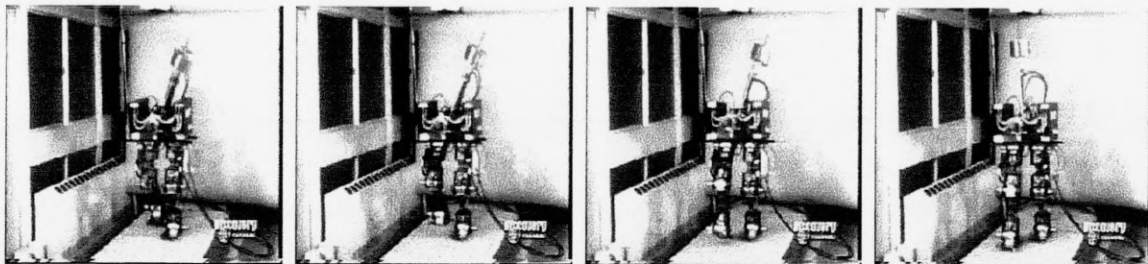
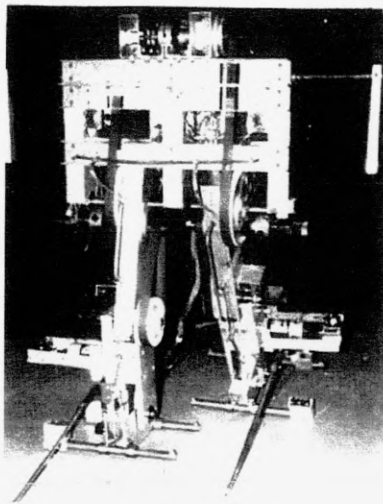


Figure 4.18 : *Sequence for one step of the Wabot WL-12 (Images taken from the Discovery Channel).*

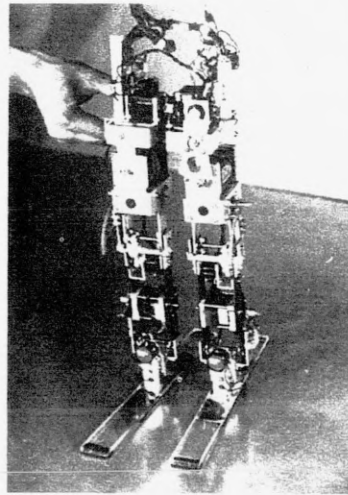
It was hydraulically powered with an upper body and a two-degrees-of-freedom waist. The emphasis in its construction was to realise a more human-like motion.

4.3.5 The Kenkyaku-2

The *Kenkyaku-2* of Fig 4.19 (a) is a seven link biped robot weighing 40 kg and with a walking period step of 1 sec. Extensions are attached to the feet to maintain lateral balance, so the robot only moves in the sagittal plane ^[14].



(a) Kenkyaku-2



(b) BIPER-4

Figure 4.19: Bipedal walking machines ^[6].

4.3.6 The BIPER-4

This seven-jointed robot shown in Fig 4.19 (b) was developed at the University of Tokyo. Miura and Shimonaya reported in 1984 ^[45] was capable of dynamically stable walking with the hip contributing to roll and ankle to the pitch, Each leg has an instrumented knee and a hip pitch joint connects both legs.

4.3.7 The BLR-G2

One of the most successful anthropomorphic biped robots of the 1980's was the *BLR-G2* from the University of Gifu and shown in Fig. 4.20. The sole senses the reaction forces and it performs flexible movements through a force control strategy. Three sets of strain gauges in the foot are used to determine pressure and ankle torque measurements.

Inclinometers and angular rate sensors measure pitch and roll angles, an ultrasonic speed sensor measures forward velocity and compares it with the extrapolated measure from the forward acceleration of the accelerometers ^[47].

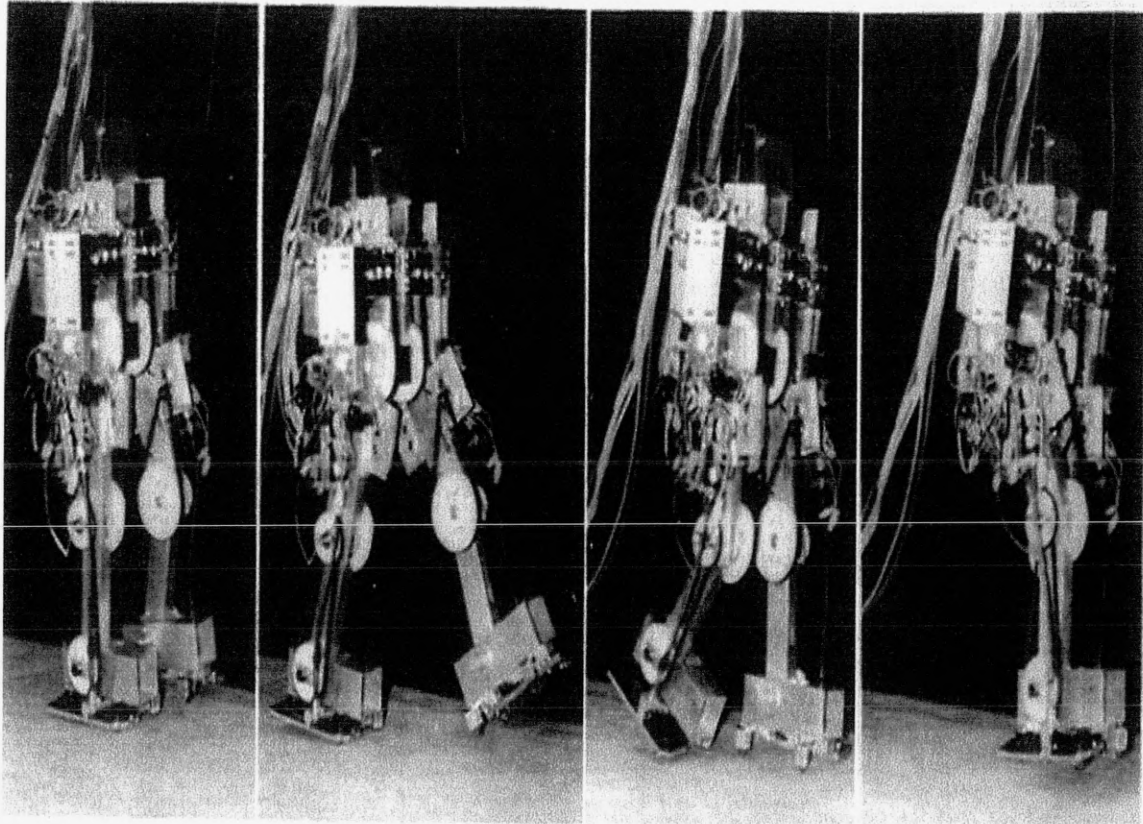


Figure 4.20: *Sequence for two steps of the BLR-G2* ^[47].

4.3.8 The MIT Spring Flamingo

Most robots built to date have integrated their visual or navigation systems with the locomotion control strategy. However the *Spring Flamingo* of Figure 4.21 relies on sensors at its feet and ankles to maintain balance in the sagittal plane, mimicking the way a blind person will walk. It also has a very human characteristic in that it uses tendons to drive the joints, leaving all the motors in a basket located above the hip ^[51].



Figure 4.21: *Sequence for one step of the Spring Flamingo (Images taken from the Discovery Channel).*

4.3.9 The MIT M2

A recent robot from the MIT Leg Laboratory is the *M2*. Its novelty comes from the fact that the robot is learning to move by itself. It still requires the assistance of students to prevent it from falling down. It is important to remember though that humans spend years learning to master standing still and walking before it becomes natural ^[51].



Figure 4.22: *M2 achieves a standing still posture (Images taken from the Discovery Channel).*

4.3.10 The WABIAN-2 LL

The WAseda Bipedal HumAnoid – 2 Lower Limb (WABIAN-2 LL) of Fig. 4.23 evolved from its predecessors the Wabot and the WL series of bipeds. The *WABIAN-2 LL* can walk with the knee stretched and can change the direction of its knee without moving its waist and foot ^[52]. WABIAN-2 LL is controlled by a PC which is mounted on its trunk, without any external support except for the power supply.

Each actuator system is equipped with an incremental encoder attached to the motor shaft, and a photo micro sensor attached to the joint shaft in order to detect the initial posture. Also, each ankle is equipped with a six-axis force/torque sensor which is used for measuring the floor reaction force and Zero Moment Point ^[52]. Figure 4.23 shows the WABIAN family of robots.

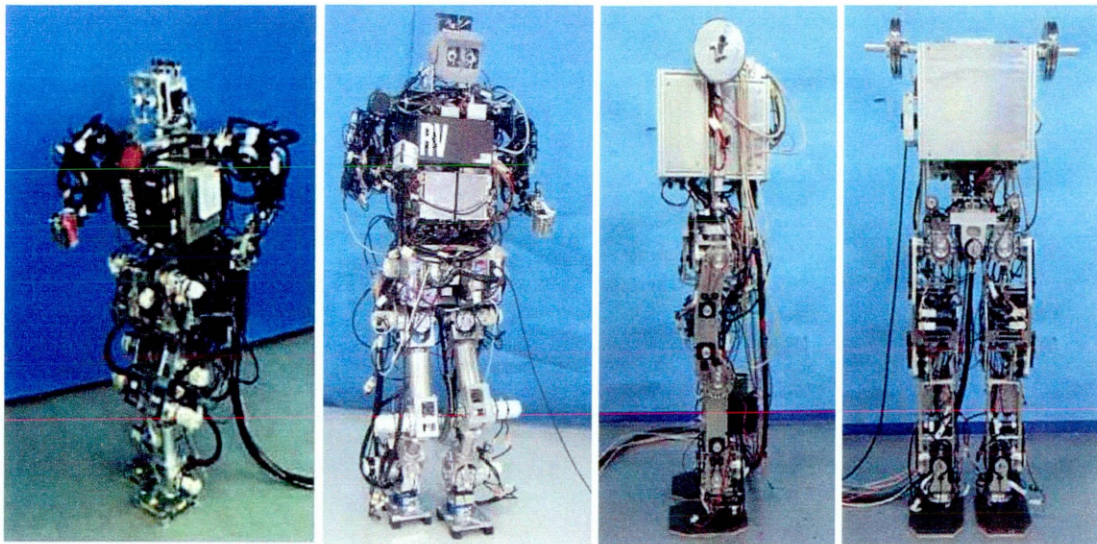


Figure 4.23: From left to right WABIAN, WABIAN RV, WABIAN-2 LL ^[52].

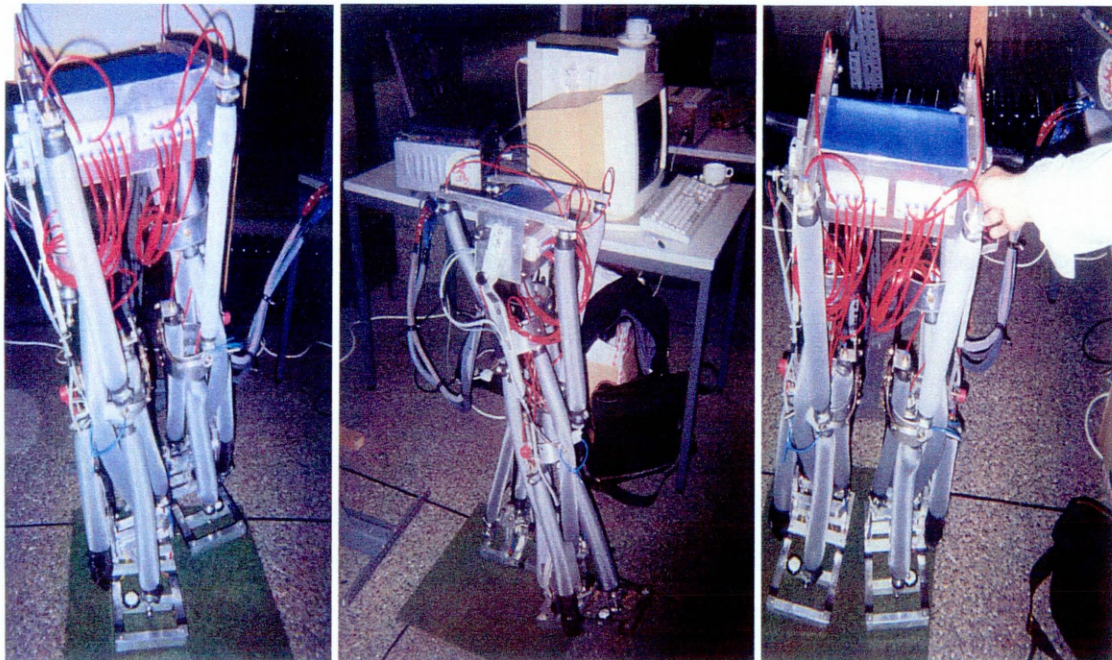


Figure 4.24: *The Salford Pneumatic Humanoid Robot project*
 (Image taken at the 4th CLAWAR Conference).

4.3.11 The Salford Pneumatic Biped

The *Salford Pneumatic Biped* of Fig. 4.24 is the lower section of the *Salford Pneumatic Humanoid Robot*. The main feature of the robot is that it employs pneumatic muscle actuators (pMAs) derived from the McKibben type ^[53]. The biped has 8 DoFs with each joint carrying a precision position sensor and being manipulated by two pMAs acting as an antagonistic pair.

Each foot has four contact switches at its corners as a means of establishing ground reaction information. The stiffness can be regulated independently and simultaneously for each joint, thus balancing the biped in the coronal and sagittal planes. Joint motion is achieved by producing antagonistic torques through cables and pulleys driven by the pMAs ^[53].

4.3.12 The Honda P3

While developments in Japan were initially confined to the universities, from the 1980's onwards a number of different projects have been developed funded privately by companies like Honda ^[54]. The Honda's *P3* shown in Figs 4.25 to 4.27 is perhaps the most advanced biped robot to date (and the most expensive at circa US\$ 100 million).

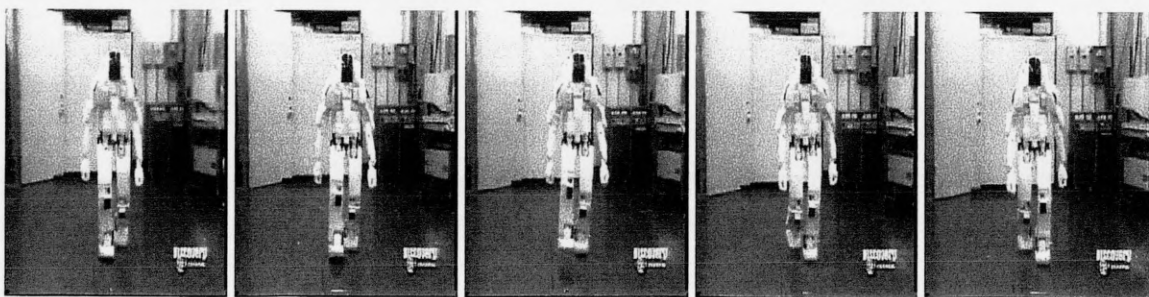


Figure 4.25: *Sequence for one step of the Honda P3 on level ground*
(Images taken from the Discovery Channel).

Like its predecessors it walks by displacing its weight from one leg to the other, achieving a true dynamic equilibrium performance. In conjunction with its advanced visual system, force sensors in every joint provide the information to enable the calculation of the movements of every actuator in order to perform different gait patterns, Including some that allow the robot to negotiate stairs. Like humans, the *Honda P3* can fall over. Should this situation arise, the robot will require assistance to be put back in a normal position, it cannot recover by itself.

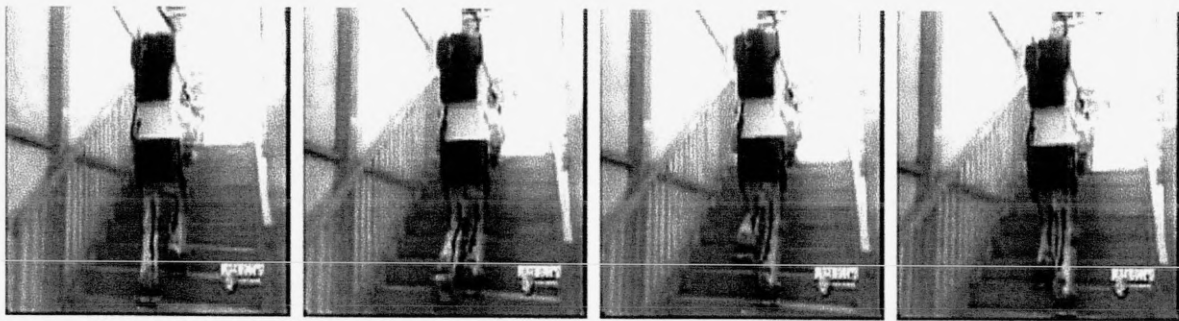


Figure 4.26: *Prototype of the Honda P3 climbing stairs (Images taken from the Discovery Channel).*

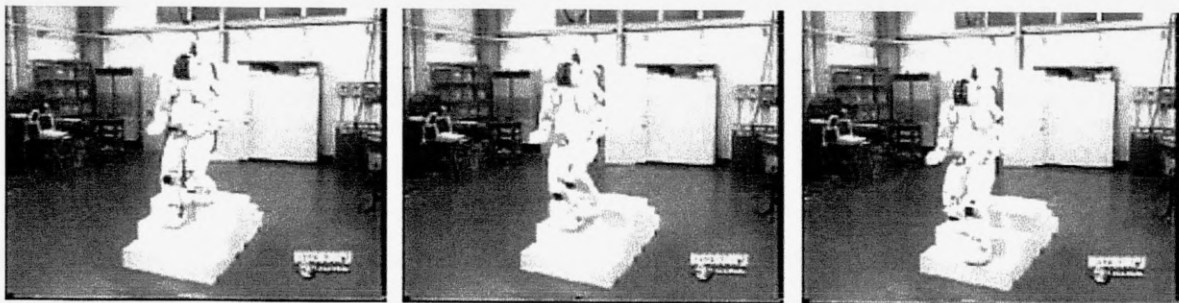
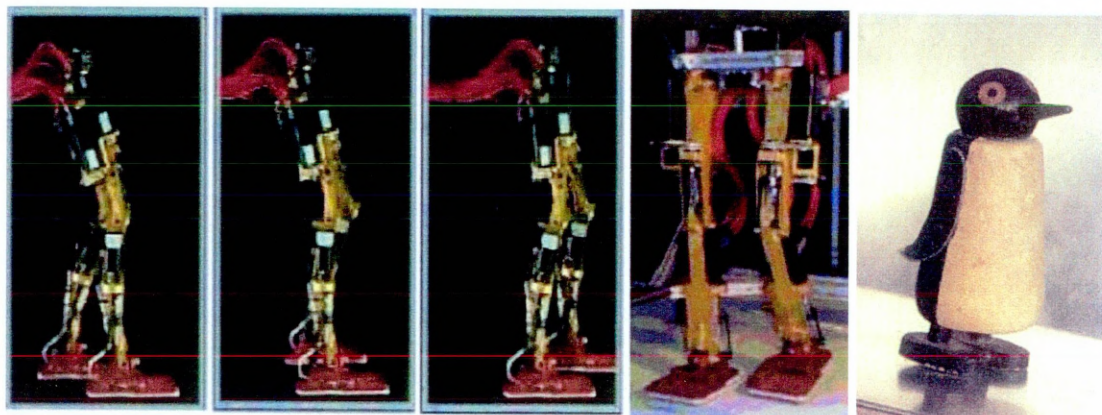


Figure 4.27: *The latest prototype of the P3 (Aibo) descending stairs (Images taken from the Discovery Channel).*

4.3.13 The BIP 2000

This French project from the Institute of Sciences (INSERM), the National institute of Applied Research (INRIA) and the Laboratoire Mecanique des Solides ^[55], took its first steps in March 2000 with the *BIP 1* version. The *BIP 2000* of Fig. 4.28 has doubled the number of DoFs to 15 and can maintain static balance on one foot while moving the other

leg. The limb dimensions and the mass properties of the lower extremities are similar to those in the human. In its first version, the system possessed only a trunk for carrying the electronics and computing systems. The robot does not have arms or head and it is not expected that the robot will carry its entire energy source. It will be able to walk in a human-like fashion on level ground or slightly inclined surfaces or ascend and descend stairs^[55].



(a) BIP 2000

(b) Wilson's Walkie

Figure 4.28: Sequence of movements for the BIP 2000^[56] and Wilson's Walkie toy^[86].

4.4 Passive Walking Robots

Passive walkers have been around for a long time, with toys in the market for more than 50 years, however the intricacies of the dynamics of this mechanisms have only become a matter of serious research only since the beginning of the 1980's, and as recently as 2005 saw the disclosure of three robots that walk using gravity and clever mechanisms with advanced control strategies.

4.4.1 Wilson's Walkie

In 1938 Jhon Wilson (a toy maker) developed a mechanism that walk^{ed} on two pivoted legs with a waddling motion that provided the ground clearance required by the swinging foot,

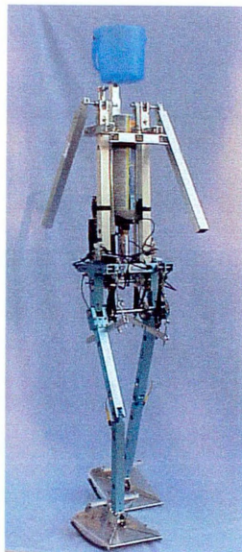
when placed on a gentle slope by the action of gravity. Figure 4.28 (b) shows the Wilson's "Walkie" mass produced version of his walking toy ^[86].

4.4.2 The Delft Biped

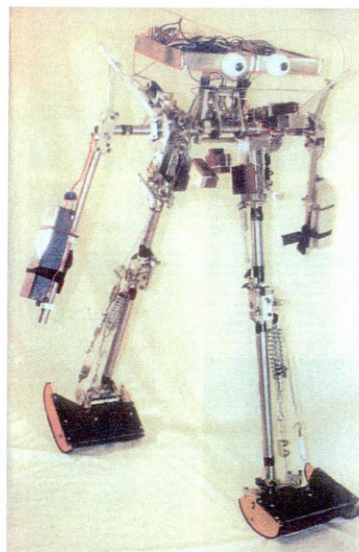
Denise is a level ground powered walking biped, that achieves dynamic balance using ankles that mechanically couple leaning with steering, i.e. while leaning, lateral foot placement played a major part in aiding to maintain the balance. In fact according to Martijn Wisse ^[87] *"the biped was not able to balance laterally without sufficient fore-aft swing leg actuation"*, implying the connection between sagittal and lateral balance in human walking ^[87].

4.4.3 The Cornell Walker

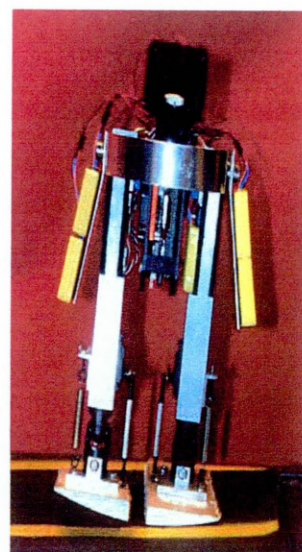
The Cornell Walker was developed in 2005, it has jointed legs a metre long, a compact torso and a flat slab-shaped head (see Figure 4.29 (b)). In terms of energy consumption is on a par with human walking ^[88] and about 15 times more efficient than Honda's Asimo ^[86].



(a) Delft's Denise



(b) Cornell's Walker



(c) MIT's Toddler

Figure 4.29: *Different Passive Walkers* ^[86].

4.4.4 The MIT Toddler

The MIT Toddler is a small robot (43 cm height) that has been programmed to learn to walk and maintain the posture by sensing the tilt of its body ^[86]. It is able to adapt to different surfaces and terrains (see Figure 4.29 (c)).

4.5 Powered Exoskeletons and Walking Chairs

4.5.1 Medical Exoskeletons

Early projects aimed at rehabilitation include those in the early 1970's at the Mihailo Pupin Institute in Belgrade, the University of Wisconsin and the Tokushima Hospital in Japan ^[14], already referred to briefly in Chapter 3. The Wisconsin Hydraulically Powered Exoskeleton (see Figure 4.30 and Figure 4.31 (a)), developed by Grundman, was under the command of the user who could select a number of tasks such as standing up from a seated posture, sitting down, walking with a 7.5 cm or 30 cm stride, stepping over an obstacle and stair climbing, all controlled through a set of switches ^[18].

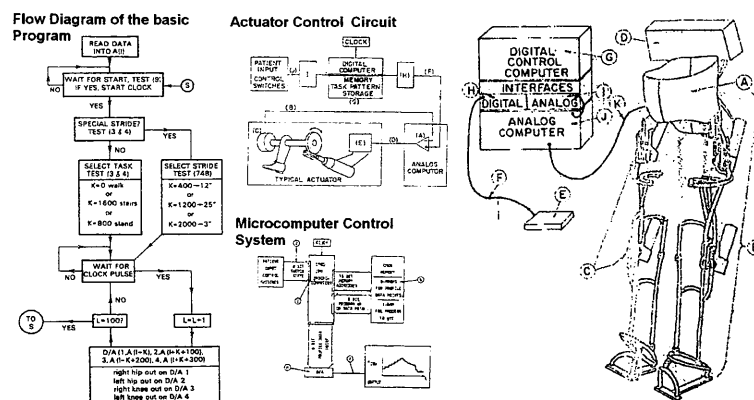
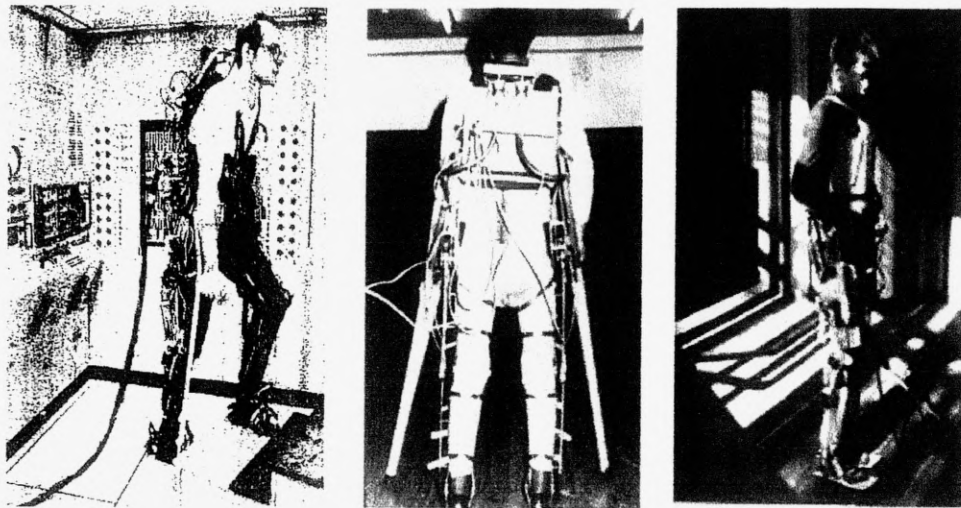


Figure 4.30 : Grundman's Control System Schematic for the Wisconsin hydraulically powered multi-task exoskeleton ^[18].

Walking started from the standing position with the right leg first swinging forward and ended when the stop push-button is depressed, after which the leg swing, moves forward to floor contact and the trailing leg ends action by closing the stride to a standing posture ^[18].



(a) *Wisconsin exoskeleton*^[18] (b) *Tokushima exoskeleton*^[14] (c) *CITO exoskeleton*^[14]

Figure 4.31 : 1970's exoskeletons.

The Wisconsin exoskeleton was hydraulically powered with the power unit worn as a backpack and hydraulic rotary actuators located at the hip and knee joints. It is first worn by a person who performs the desired tasks (teach & repeat) which are stored in memory and can then be recalled by reconstructing the voltage profiles that were recorded by the rotational transducers in each joint at intervals of 3.6 degrees.

At the Central Institute for Traumatology and Orthopaedy (CITO) in Moscow an active exoskeleton for the post-operational rehabilitation in children was successfully used during the late 1970's. This was electrically driven and was first reported to be used as a rehabilitation tool in 1976, see figure 4.31 (c) ^[14]. The Tokushima University in Japan developed an active exoskeleton for rehabilitation of paraplegics in 1973. This was electrically driven with DC servomotors actuating the hip and the knee (the active joints) and the ankle joint being passive, see figure 4.31 (b) ^[14].

An active orthosis has also been developed in Torino as a rehabilitation device for paraplegics, it shares the same principle as the Tokushima exoskeleton with a

pneumatically powered reciprocating knee and hip as the active joints, although in 2001 the Torino group reported at the 4th CLAWAR Conference that experiments had been carried out with only the knee being actuated [20]. The support structure is a commercial passive ARGO reciprocating gait orthosis that has been customised with different knee and hip joints to house the designed actuation schemes. The double acting pneumatic cylinders served the function of flexion and extension for the hip and the knee.

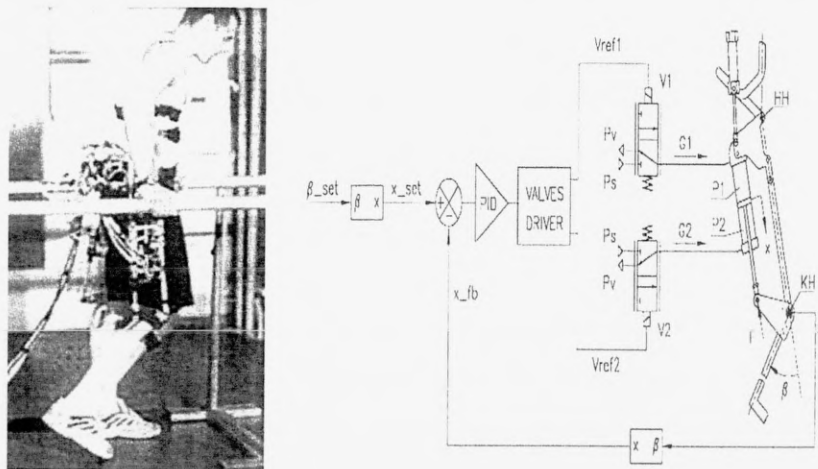


Figure 4.32 : Torino powered active ARGO gait orthosis with servo-system schematic layout [20].

Proportional valves were used and pressures ($P1$ and $P2$) were determined from the mass flow-rates ($G1$ and $G2$). The closed control loop generates the valve reference voltages (V_{ref1} and V_{ref2}) through the PID controller as a function of the position error (See Fig. 4.32) [20].

4.5.2 The Mihailo Pupin Institute

It could be argued that the pioneers on modern exoskeleton research are the members of the Mihailo Pupin Institute in Belgrade, where different types of active exoskeletons for rehabilitation that enabled paraplegic patients to walk supported by the exoskeleton and crutches were developed in the 1970's. A further type of active orthosis developed at the

Mihailo Pupin Institute was a semi-soft structure for more comfort and less decubital danger ^[14]. The target population for this system were SCI patients and the dystrophic group of patients encompassing diseases characterised by the lack of muscular strength ^[14].

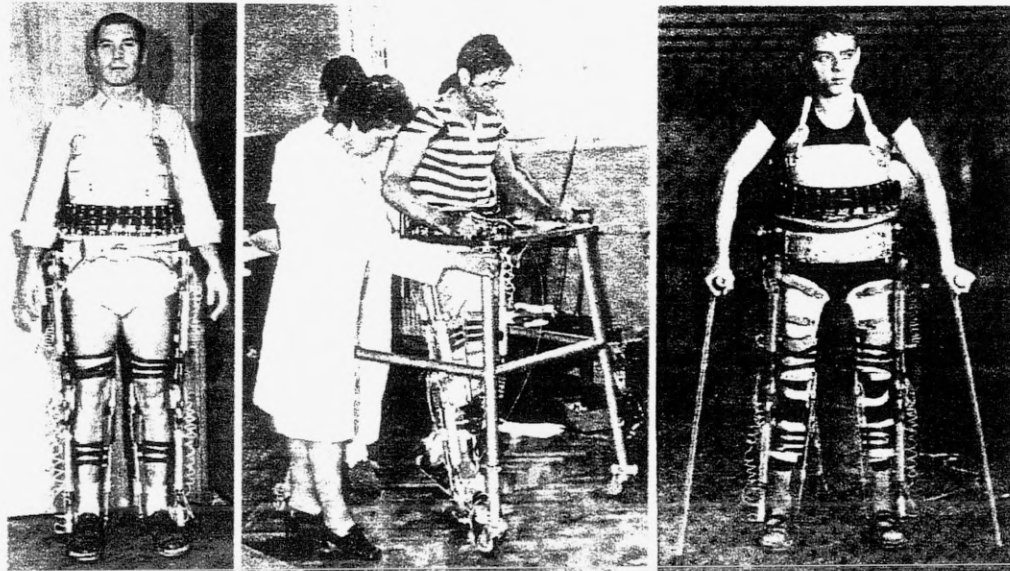
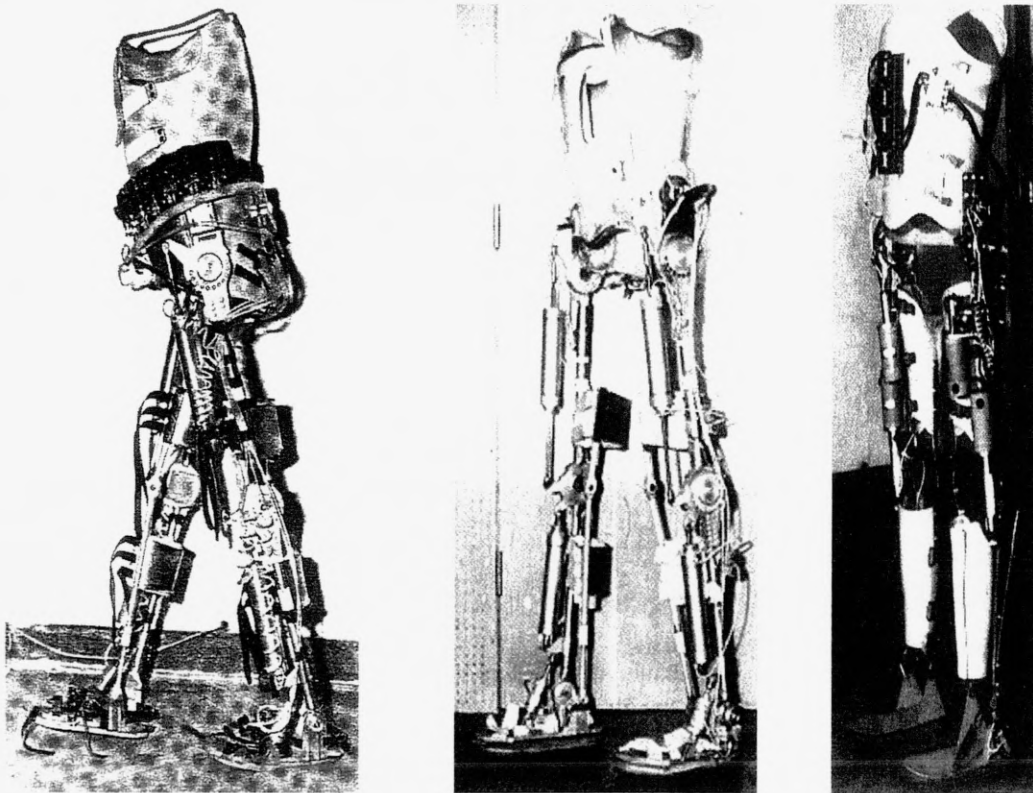


Figure 4.33 : Trials with the complete pneumatic exoskeleton at the Mihailo Pupin Institute and at the Orthopaedic Clinic in Belgrade 1972-1973 ^[5].

The Kinematic Walker

The kinematic walker is one of the Pupin Institute's earliest successful designs. It used two pneumatic actuators for the hip and the knee and produced a gait of the sliding foot type ^[5]. A simple electronic control system provided the necessary signals for triggering the pneumatic control valves.

Because of the reduced numbers of DoF, only movement in the sagittal plane was achieved. After trials conducted in healthy subjects, the conclusions provided the basis for moving a man of medium size and also proved that impaired individuals can adapt themselves to this type of assistant machine.



(a) Complete Pneumatic Exoskeleton (b) Light Pneumatic Exoskeleton (c) Electrical Exoskeleton

Figure 4.34 : *Different versions of the Complete Exoskeleton* ^[5].

Complete Pneumatic Exoskeleton

The next step consisted of a three DoF active exoskeleton to be tested on impaired subjects. One of the main considerations was the way of attaching the exoskeleton to the body of the subject because of the possibility of developing local wounds or decubitous (blisters on the skin) in places of higher pressure during extended use. So a model with a pelvic corset was implemented which used 14 solenoid valves for controlling 7 pneumatic actuators situated in the belt around the waist of the pelvic corset.

Control was sequential with a pulse generator governing the valves and the compressed air unit was outside the structure. Fully paraplegic individuals could walk with the mechanism with the assistance of two people and a rolling aid. Another version with a complete corset enclosing the chest was later developed. A force feedback system was added by means of

sensors in the soles measuring the reaction forces and correcting for the actuators to ensure protection from overturning. These two improvements allowed patients to walk with the aid of crutches alone^[14].

The main drawback of the design was the weight, which increased enormously because of the use of industrial solenoid valves. The next step was to replace these with a series of miniature valves. This, along with the use of light alloy and plastic materials, reduced the weight of the complete structure by 30%. This exoskeleton model increased patients agility and with a light support aid helped them to ease their way through doors and to turn corners.

Complete Electrical Exoskeleton

The next step was to change the power source because compressed air generation technology at the time limited this type of exoskeleton to the controlled environments in the institute like its own clinics and laboratories. To increase autonomy, it was therefore decided to develop an electrically driven exoskeleton. The result of a 2 year development programme proved to be about 25% heavier than the lightest pneumatic one. Control strategies changed from sequential event-driven ones to a finite state machine approach, again the power source was outside the exoskeleton^[5], see Figure 4.34 (c).

4.5.3 Augmentation Exoskeletons

Besides medical therapeutic uses, other applications have boosted the field of exoskeleton research. These type of exoskeletons are not designed for users with lower limb

impairments. Instead they are designed to enhance the capabilities of human beings so they can cover longer distances and carry higher loads.

Power Assisted Suits

A team of researchers from the University of Tsukuba led by Professor Sankai Yoshiyuki has developed a suit that supplements the power of the wearer's legs, see figure 4.35 (a). This suit has two main components: a metal frame that externally supports the legs and has a motor and sensors attached, and a regulator that is carried on the user's back. As the wearer tries to move, the sensors affixed to the surface of the skin pick up the low voltage electrical signals transmitted from the brain to the muscles and attempt to make the motor's action complement the motion of the wearer ^[57].

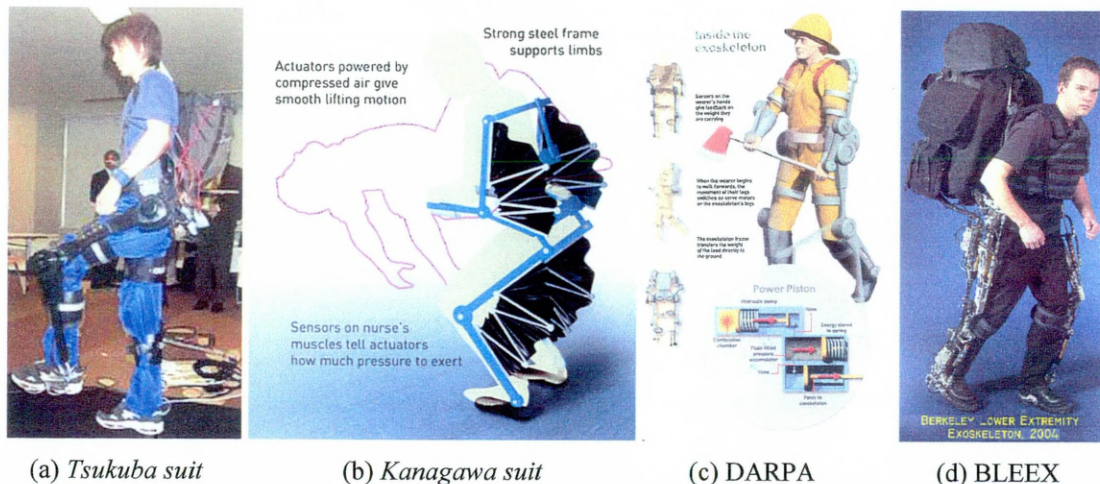


Figure 4.35 : From left to right (a) University of Tsukuba Power suit, (b) Kanagawa Institute of Technology Power Assist Suit, (c) DARPA's Petrol Powered Exoskeleton, (d) Berkeley Lower Extremity Exoskeleton.

The Kanagawa Institute of Technology Power Assist Suit's main role will be helping nurses and physiotherapists to lift patients on and off beds by giving wearers approximately twice their natural strength. When the wearer lifts someone up (see Fig. 4.35 (b)), the computer works out if their limbs and joints have enough artificial support so

the movements can be performed slowly thus creating low mechanical stress, inflating compressed-air actuators positioned in the arm, lower back, and knee areas ^[58].

The prototype suit weighs 18 kilograms and in tests, a nurse weighing 64 kilograms was able to pick up and carry a patient weighing 70 kilograms ^[58].

DARPA has been managing different projects aimed at developing an exoskeleton that lets troops jog effortlessly at 13 Km/h carrying a 70 Kg load for 12 hours before recharging the power unit ^[59]. The researchers have identified what they believe is the key to achieve such levels of performance in the energy storage ^[59], see Figure 4.35 (c). The core concept of the energy storage strategy involves using small combustion engines that provide the power output required for the target performance. The hydraulics necessary for heavy lifting are operated through a small purposefully designed combustion engine built into it. On a full tank the system should be able to run for as long as two hours ^[59].

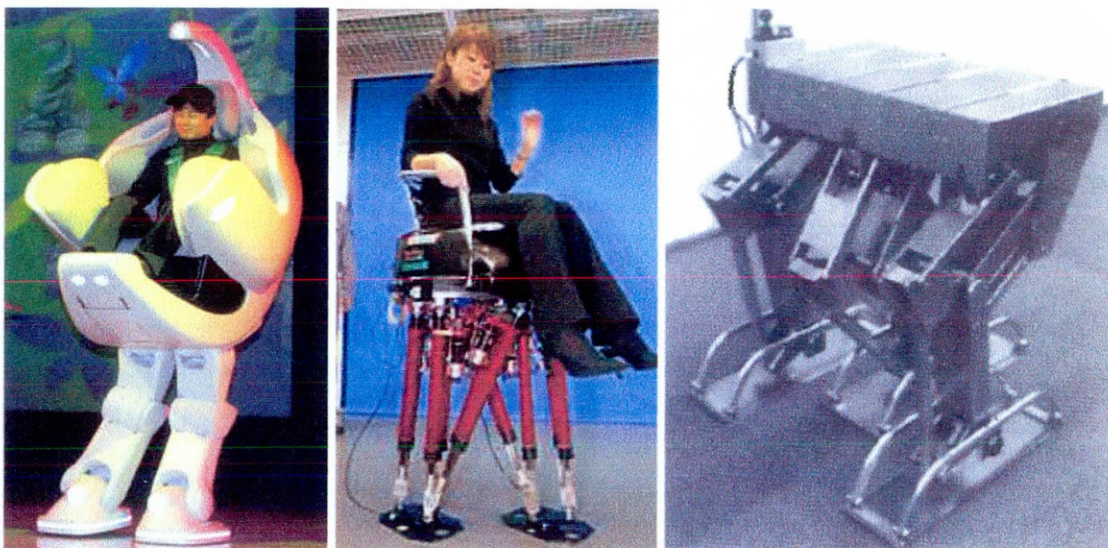
The key control feature is that the user needs no joystick, keyboard or buttons to operate it, leaving the hands free for other tasks. The control system ensures it moves in concert with the person wearing the exoskeleton ^[60]. One implementation is the Berkeley Lower Extremity Exoskeleton (BLEEX) (see figure 4.35 (d)). The device itself weighs 50 kg, with the control system ensuring that the centre of gravity is always within the pilot's footprint i.e. the machine takes its own weight, plus a 32 kilogram load within the backpack.

4.5.4 Walking Chairs

Besides the exoskeletons, there are a number of projects exploring the idea of walking from a different angle. For example the Toyota Bipedal Chair (see figure 4.36 (a)) and the

Waseda University in Tokyo, whose robot WL-16, is essentially an aluminium chair mounted on two sets of telescopic poles, see figure 4.36 (b).

WL-16 uses 12 actuators to move forwards, backwards and sideways while carrying an adult weighing up to 60 kilograms. The robot can adjust its posture and walk smoothly even if the person it is carrying shifts in the chair. At present it can only step up or down a few mm, but the team plans to make it capable of dealing with a normal flight of stairs ^[61].



(a) Toyota Bipedal Chair

(b) Waseda's WL-16 ^[61]

(c) Tokyo Institute of Technology WC II

Figure 4.36 : *Walking Chairs*.

Researchers at the Tokyo Institute of Technology and at the Shiabura Institute of Technology in Japan have also developed a Walking Chair as an alternative to a wheelchair, whereas the WL-16 is a biped WC I, II and III are tripeds driven by two servomotors. WC III was also aided by the arms of the user, see figure 4.36 (c).

4.6 Design Implications

After careful examination of the machines developed in the field, a number of considerations regarding the design of an exoskeleton for gait aid can be established. As a

research project the exoskeleton design can initially be limited to the sagittal plane. If the exoskeleton is to be taken forward as a commercial product it must include the Degrees of Freedom required for it to be able to turn and move around obstacles.

The choice of actuators is open to the most convenient and available. For the project to be taken forward it is important to consider the experience of the Mihailo Pupin project, which started with pneumatic actuators and finally opted for electrical actuators.

The DARPA military project opts for an internal (petrol) combustion engine as the main means for powering the exoskeletons. Human walking as explained in the previous chapter requires around 300 W. Due to the weight penalty, the DARPA exoskeletons require far more than that, probably around 2 kW. That is why they are willing to risk a design that will be used in the battlefield (in the case of the military application) and in environments surrounded by flames (in the case of the fire fighter application) with highly flammable fuel¹.

For the project to be taken forward as a commercial product, the autonomy of the exoskeleton must be taken into account. In the introduction it was mentioned that perhaps a solution whereby the exoskeleton could share the locomotion responsibilities with a wheelchair could prove to be more beneficial than the exoskeleton alone. Wheelchairs work perfectly well under certain circumstances and exoskeletons could be preferred under others.

¹ On a personal note it seems to me that the choice of application for such technology will be better placed in projects that involve some less hazardous conditions like the one at the core of this PhD, I would like to meet the brave soldiers and fire fighters that will strap themselves to those highly flammable exoskeletons.

A wheelchair-exoskeleton approach could also ease the weight burden imposed on the exoskeleton as some of its components would not have to be onboard, and the wheelchair could host some extra energy storage required by the exoskeleton. The wheelchair design is beyond the scope of the dissertation. Most of the bipedal research highlighted the balance problem where many gait patterns cannot ensure static balance even with both feet on the ground. It is therefore interesting to see what other non-bipedal robotic solutions have to offer. The quadrupeds and hexapods reviewed all relied on static balance by maintaining at least three points of support at any given time in a triangle. A stability criteria similar to ZMP was used where a point existed within the triangle such that the reaction forces caused this ZMP to exist.

The electronics governing such machines (the reviewed quadrupeds and hexapods) are simpler, and therefore in the eyes of the project far more robust. The exoskeleton nonetheless still is a bipedal mechanism, so the alternatives available are either crutches or support frames. The criteria for using one or the other should, in the opinion of the project, be left to the will of the user. The implication is that with a frame perhaps there will be more than four points of support. Consequently, the efforts in design could be focused on the use of crutches as, if it works for crutches, certainly it can work for frames, but if it works for frames it is difficult to infer the electronics of crutches.

Besides maintaining the balance, the other critical issue is that of generating the trajectories. The robots studied either have fixed trajectories or generate them according to the navigation plan. Leaving the robot to generate any possible trajectory and deciding on it may not render human-like trajectories and complete automation is not required since there will be an operator for the exoskeleton. What is required is enough intelligence to

translate simple basic commands into effective actions. Static balance seems to be the key, so the gait pattern with crutches has to take this into account. ZMP techniques have to be explored and a way to measure where the centre of gravity lies every time there is forward motion must be secured.

Involving the user in the operation of the robot not only gives more confidence to the wearer who feels in control of the gait assistant, but can increase the performance of the exoskeleton compared to the performance/cost ratio of an autonomous machine. There is no need for sophisticated terrain acquisition sensors, terrain mapping capabilities and complicated path planning. Ultimately, balance and foot placement could be left for the user to make the final decision resulting in a more practical approach. The next step necessitates the definition of the number of degrees of freedom (DoF) required to achieve a target type of gait and the geometrical configuration of the legged machine.

The medical exoskeletons developed in the 1970's provide an excellent case study into how not to design rehabilitation tools of this type if long term use is an important requirement. Although these exoskeletons were successful in replicating human gait and during clinical trials allowed certain SCI patients to briefly walk, the designs were mainly technology driven and not user centred, i.e. the Mihailo Pupin Institute pneumatic and electrical full exoskeletons required a team of assistants to strap the patients into the exoskeleton, so donning and doffing required assistance, which is an aspect of the design that is bound to affect the usage of the tool. The other key aspect that was also completely overlooked in the design of the medical exoskeletons previously examined is that none of the developed systems were designed to be incorporated into a wheelchair, all of them were designed to be used as stand alone equipment, thus automatically eliminating any

possibility for those exoskeletons to extend their autonomy beyond the confinement of their laboratories.

A more pragmatic approach based on the prospective user's needs will have to incorporate these two aspects, whereby the design requirements of the exoskeleton must incorporate a very simple system for donning and doffing and the possibility to be used in association with the user's wheelchair. Since the telescopic knee design provides advantages ^[16] in respect of the energy consumption at the ankle joint (it removes the need for the energy peak due to the push before toe off) and the movement of the leg is simplified (initialisation of the gait cycle with a telescopic knee immediately provides the toe clearance required to move the leg forward).

4.7 Summary

After careful examination of different walking machines and medical exoskeletons, constraining the exoskeleton design initially to the sagittal plane, three DoF and the use of a telescopic knee are clearly the basic set of parameters required for the definition of a model so that the implications of the design and its impact in the generation of trajectories can be modelled and analysed. If this basic setup is successful in recreating appropriate gait patterns, then the simplicity of the design is an advantage for an easy implementation of the donning and doffing tasks. The question of how to incorporate the exoskeleton onto the wheelchair's user could be more effectively answered if the wheelchair is the one designed around the exoskeleton requirements for donning and doffing and the transitions from sitting down to standing up. The next chapter is dedicated to the exploration of the different models developed following the criteria developed in this chapter.

Chapter 5

MODELLING & SIMULATION

5.1 Introduction

After taking into account the design implications resulting from the analysis of the human and machine gaits, the next step was to mathematically model the telescopic knee design and compare it with a human model. Three active degrees of freedom (DoF) were chosen to be the starting point required to achieve the target type of gait and the geometrical configuration of the legged machine. Although only two DoF are required to achieve bipedal locomotion ^[16] in the sagittal plane, one DoF per representative joint gives more flexibility in establishing different combinations of gait trajectories.

The telescopic design provides advantages ^[16] in respect of the energy consumption at the ankle joint since the complete movement of the leg is simplified as the telescopic knee immediately provides toe clearance when the gait cycle starts, thus removing the energy peak due to the push before toe off (see 3.2 for a more extensive discussion of the gait cycle). This semi-anthropomorphic geometry was the chosen design to be validated through the extensive modelling carried out for the project.

Table 5.1: *Subject anthropomorphic relationships*

Subject	Relative Anthropomorphic Measurements in cm			
ID number	Thigh	Shin	Sole to Ankle	Foot Length
01	39.5 cm	36.8 cm	10.0 cm	28.0 cm
02	59.0 cm	47.0 cm	13.0 cm	40.0 cm

5.2 Mathematical Definition of Kinematic Model

The human model was built around the anthropomorphic data collected from two members of the project, as set out in Table 5.1. The first 2D models were constructed

in MatLab with the actual measurements for the segments as collected from the first test subject.

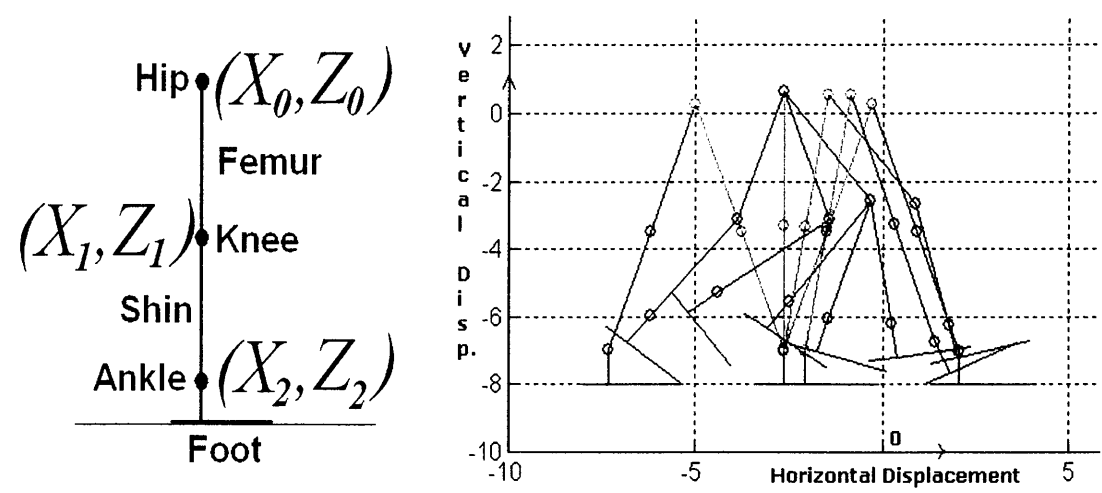


Figure 5.1 : Human Leg Model and simulation of a full step using MatLab.

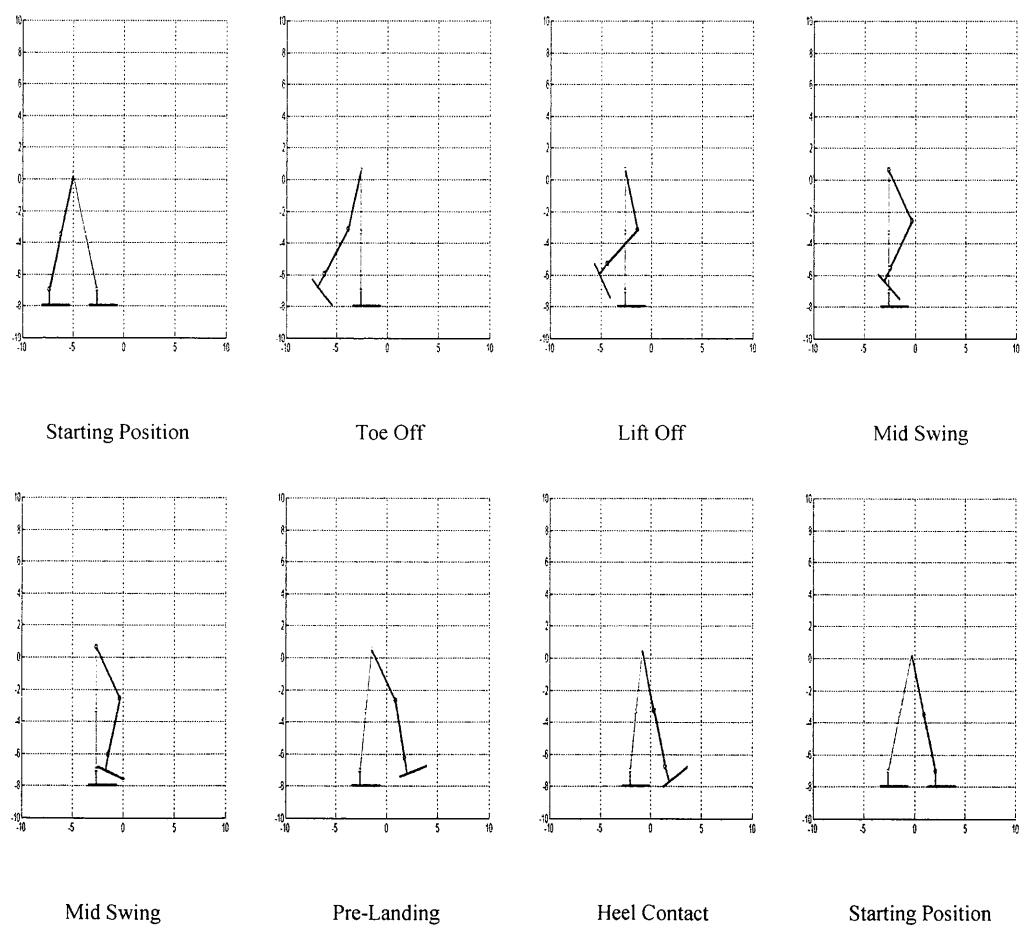


Figure 5.2 : Simulation of one step with Human Kinematical Model

The model had three segments connected through a 1 DoF rotary joint (hinge) representing the hip, knee and ankle. The data for joint rotation obtained from clinical gait analysis was fed into the model as a reference, and from that base different types of steps were obtained with different stride lengths and foot clearances (see Figs. 5.1 and 5.2). Different states were defined comparing the simulation to the gait analysis established in Chapter 3.

From these models it was possible to define the inverse and direct kinematics and hence the simulation of different trajectories, Figure 5.3 shows the geometrical model for the biped .

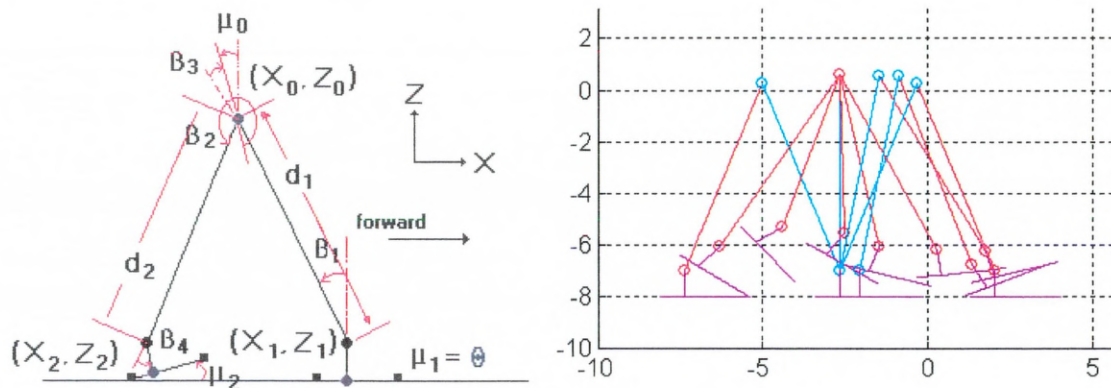


Figure 5.3 : Exoskeleton Kinematical Model and simulation of a full step using MatLab, this is the telescopic knee model (Figure 5.1 depicts a rotary knee model).

5.3 Direct and Inverse Kinematics

Consider the walking cycle to be the period of continuous motion when the biped is supported on one foot, followed by the impact of the heel at the end of the cycle. Provided that the friction between the sole and the ground is enough to prevent slippage, and therefore the stance foot remains in the same position throughout the

stance phase, the equations for direct $\mathbf{p} = \mathbf{h}(\mathbf{q})$ and inverse $\mathbf{q} = \mathbf{h}^{-1}(\mathbf{p})$, kinematics can be derived as follows (see Appendix A for a complete derivation) :

$$\mathbf{p} = \begin{bmatrix} x_0 \\ z_0 \\ \mu_0 \\ x_2 \\ z_2 \\ \mu_2 \end{bmatrix} = \begin{bmatrix} p_0 \\ p_1 \\ p_2 \\ p_3 \\ p_4 \\ p_5 \end{bmatrix} = \begin{bmatrix} x_1 - d_1 \cdot \sin(\mu_1 + \beta_1) \\ z_1 + d_1 \cdot \cos(\mu_1 + \beta_1) \\ \mu_1 + \beta_1 + \beta_3 \\ x_0 - d_2 \cdot \sin(\mu_0 + \beta_2) \\ z_0 - d_2 \cdot \cos(\mu_0 + \beta_2) \\ \mu_0 + \beta_2 + \beta_4 \end{bmatrix} \quad \text{Eq. 5.1}$$

$$\mathbf{q} = \begin{bmatrix} \beta_1 \\ d_1 \\ \beta_3 \\ \beta_2 \\ d_2 \\ \beta_4 \end{bmatrix} = \begin{bmatrix} q_0 \\ q_1 \\ q_2 \\ q_3 \\ q_4 \\ q_5 \end{bmatrix} \quad \text{Eq. 5.2}$$

The velocity transformations between \mathbf{p} -space and \mathbf{q} -space are defined by :

$$\dot{\mathbf{p}} = \mathbf{J} \cdot \dot{\mathbf{q}} \quad \text{Eq. 5.3}$$

where :

$$\mathbf{J} = \begin{bmatrix} \frac{\partial p}{\partial q_0} & \frac{\partial p}{\partial q_1} & \dots & \frac{\partial p}{\partial q_5} \end{bmatrix} = \begin{bmatrix} \frac{\partial p}{\partial \beta_1} & \frac{\partial p}{\partial d_1} & \frac{\partial p}{\partial \beta_3} & \frac{\partial p}{\partial \beta_2} & \frac{\partial p}{\partial d_2} & \frac{\partial p}{\partial \beta_4} \end{bmatrix} \quad \text{Eq. 5.4}$$

$$\mathbf{J} = \begin{bmatrix} \begin{pmatrix} z_1 - z_0 \\ x_0 - x_1 \\ 1 \end{pmatrix} & \begin{pmatrix} -\sin(\mu_1 + \beta_1) \\ \cos(\mu_1 + \beta_1) \\ 0 \end{pmatrix} & \begin{pmatrix} 0 \\ 0 \\ 1 \end{pmatrix} & \begin{pmatrix} 0 \\ 0 \\ 0 \end{pmatrix} & \begin{pmatrix} 0 \\ 0 \\ 0 \end{pmatrix} & \begin{pmatrix} 0 \\ 0 \\ 0 \end{pmatrix} \\ \begin{pmatrix} z_1 - z_2 \\ 2 \cdot x_0 - (x_1 + x_2) \\ 1 \end{pmatrix} & \begin{pmatrix} -\sin(\mu_1 + \beta_1) \\ \cos(\mu_1 + \beta_1) \\ 0 \end{pmatrix} & \begin{pmatrix} z_2 - z_0 \\ x_0 - x_2 \\ 1 \end{pmatrix} & \begin{pmatrix} z_2 - z_0 \\ x_0 - x_2 \\ 1 \end{pmatrix} & \begin{pmatrix} -\sin(\mu_0 + \beta_2) \\ -\cos(\mu_0 + \beta_2) \\ 0 \end{pmatrix} & \begin{pmatrix} 0 \\ 0 \\ 1 \end{pmatrix} \end{bmatrix}$$

Eq. 5.5

5.4 Definition of Gait Pattern

From the above equations, suitable trajectories were identified by implementing the movements of the structure in MatLab and visualising the movements in 2D. After several attempts to match the hip and knee values it became clear that the principle criteria to be used was that of matching the ankle, toe and heel values, effectively matching the movement of the feet soles on both models (the bipedal and the human) through the entire cycle, as seen in Figure 5.4.

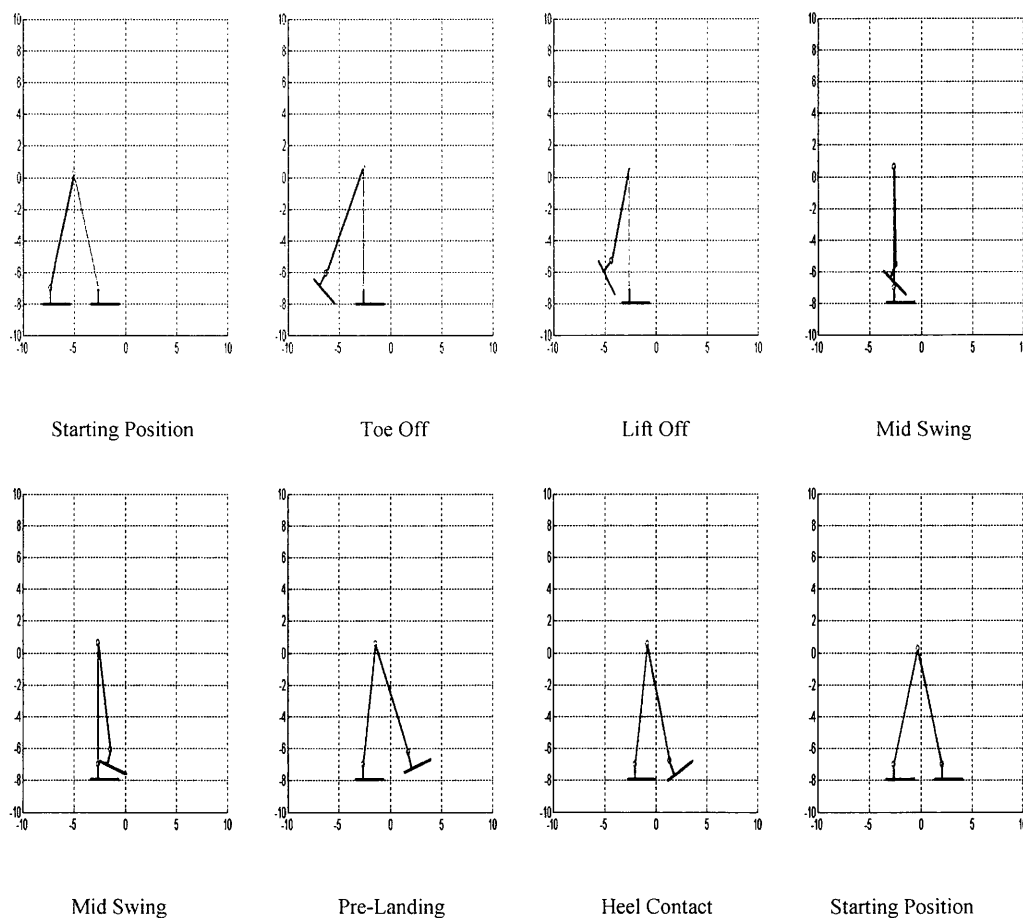


Figure 5.4 : *Simulation of one step with Exoskeleton Kinematical Model*

The procedure followed requires first establishing where both feet are at all times, this information is available from the human gait simulation, then the feet of the bipedal

model are matched to this positions and finally the required angles for the joints and length of the telescopic knee are calculated from the inverse kinematics. This set of values is the used in the direct kinematic configuration and the result is superimposed on the human gait simulation and the result can be seen in Figure 5.5.

The analysis of the results led to the conclusion that the implementation of the exoskeleton was feasible as it could reproduce human-like gait trajectories.

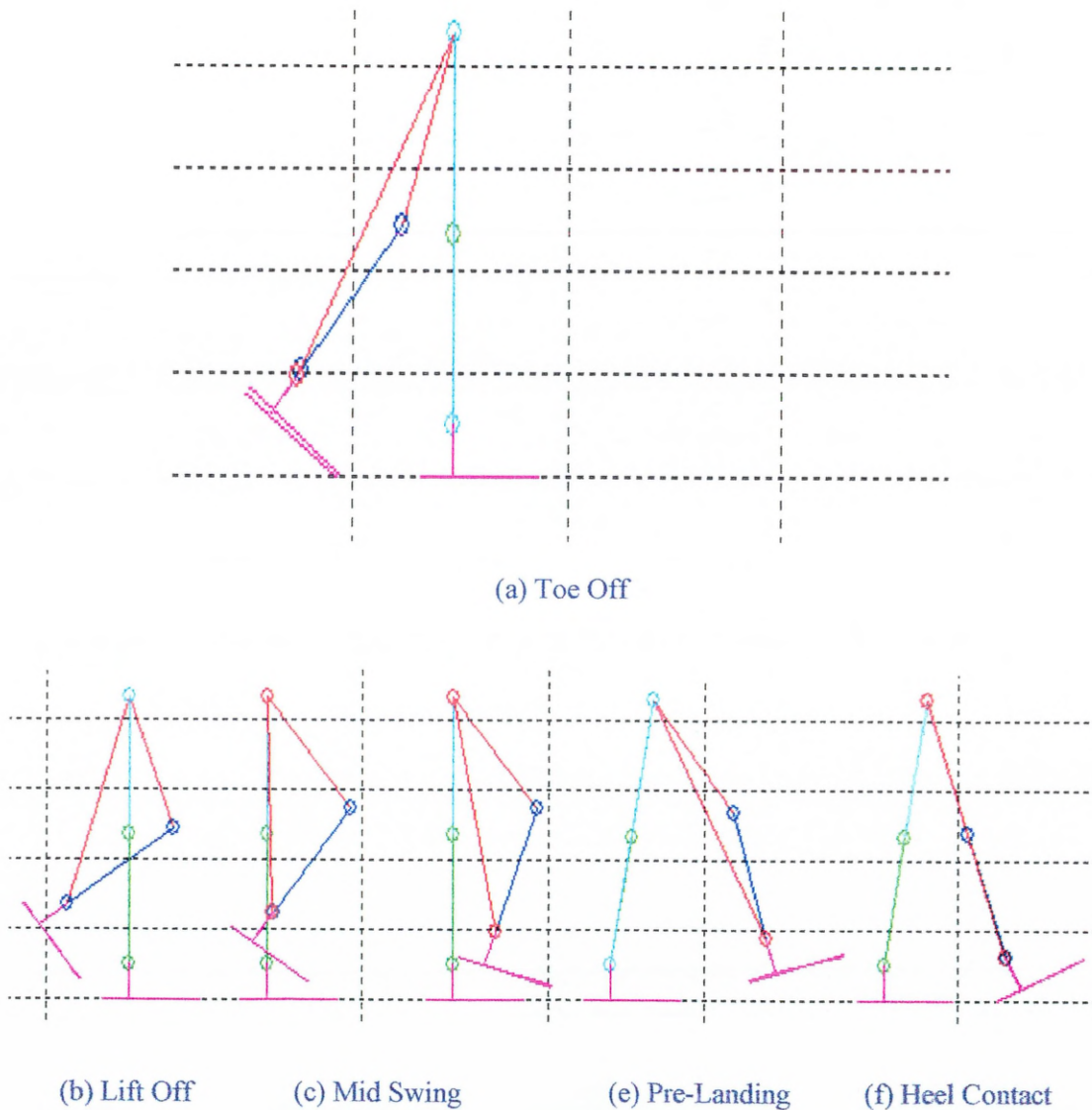


Figure 5.5 : Simulation of one step with Exoskeleton and Human Kinematical Model superimposed.

5.5 Crutch Assisted Gait

As for the balance problem, the simulations were performed assuming that static balance was maintained during the walking cycle. It was then decided that the user would make use of a pair of crutches to give them two extra support points. For this quadrupedal configuration, two main types of high speed gait patterns have been observed, the ipsilateral gait and the contralateral gait^[22].

The slower it is intended to move, the more stable the system. If it is required that the user goes faster, then it will become necessary to introduce unstable states^[22]. Results from the study by Zefran aimed at improving the crutch-assisted gait of FES patients demonstrated that forward motion provided by stimulation of the plantar flexors and the use of unstable states, make walking faster and smoother^[22].

The Zefran model (see Figure 5.6) incorporated the trunk to the extremities. The criteria for gait selection was to achieve smooth progression of the body and energy efficient walking^[22]. The premise is that energy consumption during FES assisted gait would be decreased if smooth continuous movement of the centre of the body (COB) could be achieved.

The participants in Zefran's study (more than 90 individuals trained in the use of the FES system) had control of the stimulation events throughout the whole cycle as the gait aid deployed used memorised functions accessible to the user through a panel of buttons located on the crutch handle. When the button is pressed, a flexion reflex is elicited and the leg is flexed and transferred forward^[22].

During the double support phase for the statically stable mode, the knee extensors of both legs are contracted (by stimulating some leg muscles ^[22]) and therefore providing the support, and maintaining a static posture. The user then transfers the body weight to the leading leg by positioning the crutches, and again the trailing leg is brought forward (see Figure 5.7).

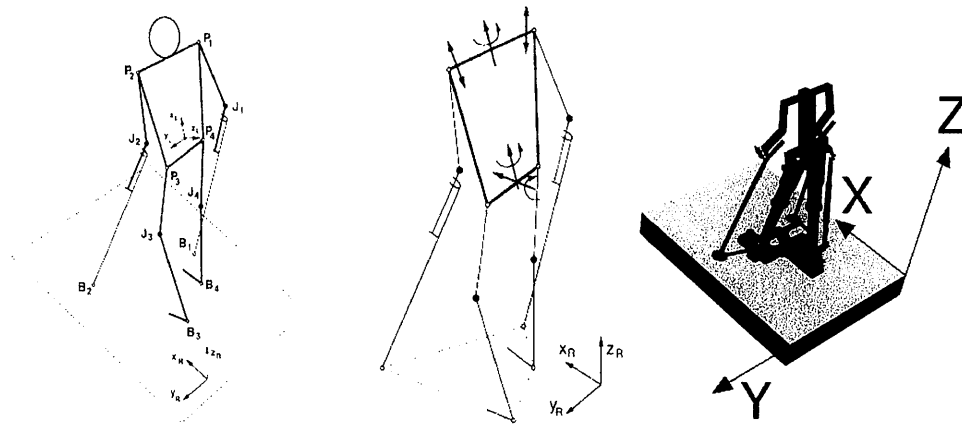


Figure 5.6 : Zefran's model with parallel linkage with flexible trunk ^[22] and Lazarus Model.

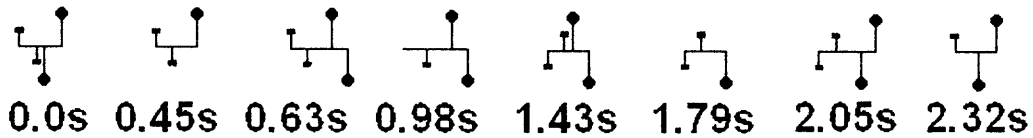


Figure 5.7 : Average step time for regular gait pattern for statically stable mode, three points of support with the Centre of Gravity lying within the area enclosed are guaranteed at any given time ^[22].

This movement keeps the leg on the ground for approximately 85% of the cycle (compared with about 60% of the cycle for typical individuals). The average step length was 18 cm and the time to complete a step was 1.38-1.43 s, the average velocity was 0.13m/s and swing of the crutch lasted about 0.2 s ^[22]. This regular gait pattern (as shown in Figure 5.7) maintains at least three support points at any given time thus becoming a statically stable gait. This is the same concept for the

implementation of that which the exoskeleton can perform and therefore can be used as a benchmark to evaluate the performance of the design.

Positioning the crutches can be done in a number of ways, besides the statically stable mode, ipsilateral and contralateral patterns can be used generating faster gait patterns, although not statically stable ones as seen in Figure 5.8.

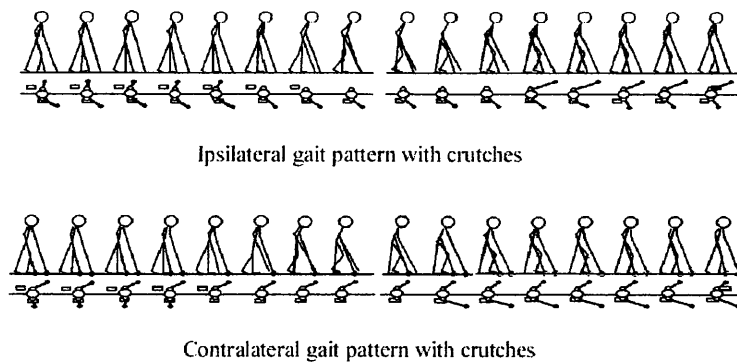


Figure 5.8 : *Unbalanced faster gait patterns with crutches (three points of support are not guaranteed at any given time).*

As Zefran's model included the trunk and the crutches, a new 3D model was developed (using Visual Nastran 3D Motion). With this model different sets of trajectories and walking cycles were simulated and analysed. The idea was to find a type of quadrupedal gait that would maximise the double support time over the step. Zefran's gait is a good reference and starting point in order to synthesise the desired gait pattern.

5.6 Full Walking Cycle

Table 5.2 describes an ipsilateral (better static balance) sequence for a full step with the aid of crutches. More detailed information on the values of the different parameters of the simulation can be found in Appendix A.

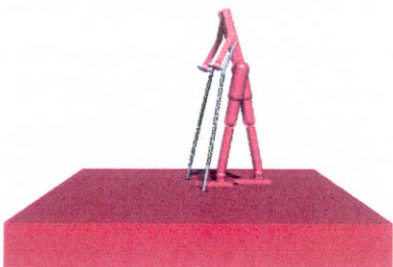
Table 5.2 *Walking Cycle for a Step with the aid of crutches(ipsilateral)*



Starting position

Stance leg in front of swing leg.

The starting position corresponds to the home position. This geometrical position corresponds to the most stable position with four points of support.



Swing Leg crutch's positioning

Both feet are flat on the ground. Right crutch is positioned ready for next step.

The crutch opposite to the stance leg is positioned around the point where the swing foot is expected to land, close to the toe but no further than half the length of the foot.



Stance Leg crutch's positioning

Both feet are flat on the ground. Left crutch is positioned ready for next step.

The second crutch is positioned parallel to the other crutch in preparation for the movement of the swinging leg. Body attitude has to be corrected to avoid tilting.



Preparation for Toe – Off

Swing foot is not flat on the ground, toe is still in contact with the ground.

The body attitude is corrected so the centre of gravity compensates for the moment generated by the lifting of the swinging leg. The toes maintain contact with the ground.



Toe – Off

Swing foot is about to lift off, but is still in contact with the ground.

Lift off is imminent, the triangle formed by the crutches and the stance leg ensures that on lift-off the weight of the swinging leg will not tilt the body backwards.



Lift – Off

Swing foot lifts-off.

The swinging foot is propelled upwards and towards the front, the moment generated by the weight of the leg pulling the body backwards is compensated by the three-point balance setup.



Mid-Swing

Swing foot is airborne and going through the swinging sequence

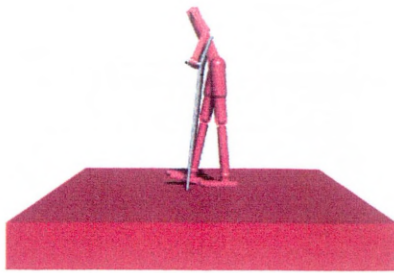
The swinging leg pivots around the hip maintaining the position with the knee bent. Balance is maintained as the body attitude remains the same.



Maximal Swing Position / Pre-Land

Swing leg achieves its maximum rotational value and prepares for landing.

The leg achieves the extreme point of the swing phase. The knee extends the leg forcing the foot downwards in preparation for landing.



Heel Strike

Swing leg achieves its maximum rotational value and prepares for landing.

The foot continues to travel downwards until the heel strikes the floor. The impact is soft allowing the foot to land gently.



Starting position

Swing leg achieves its maximum rotational value and prepares for landing.

Crutches can be relocated if they are not in the proper position. Once the support points are correctly positioned, then the whole sequence can be re-started.

The sequence starts with one leg in front of the other and the two crutches parallel to one another in a statically stable configuration, the aim is to advance the crutches to a position at which, when the swing leg lands, it will be symmetrical with the previous step. It is then just a matter of lifting and swinging the leg, maintaining the static balance all the way through. The simulation provides finally the information required to synthesise the gait trajectories.



Figure 5.9 : Exoskeleton complete Gait Pattern

With the information from the joints and the soles of the feet, it is possible to synthesise trajectories that match the movement of the feet, achieving human-like, gait patterns. Figures 5.9, 5.10 and 5.11 show the exoskeleton sequence for a step matching the trajectory of the soles described in Table 5.2. The sequence for walking thus follows a very simple but effective sequence. As the cylinders contract, the leg achieves toe clearance so the swing can be carried out. When the swinging leg achieves its maximal rotational value, the cylinders extend and the swinging foot lands, thus achieving the required transitions between single and double support.

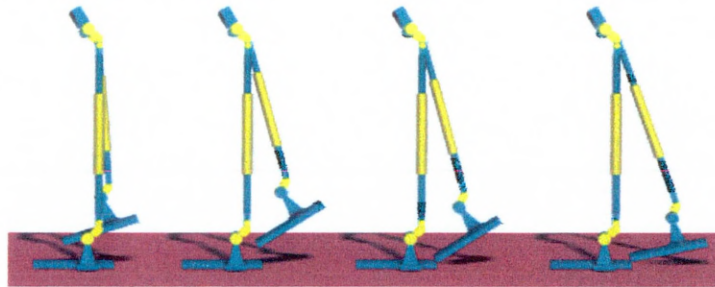


Figure 5.10 : *Exoskeleton Gait Pattern, soles matching human feet movement for Lift Off and Swing.*

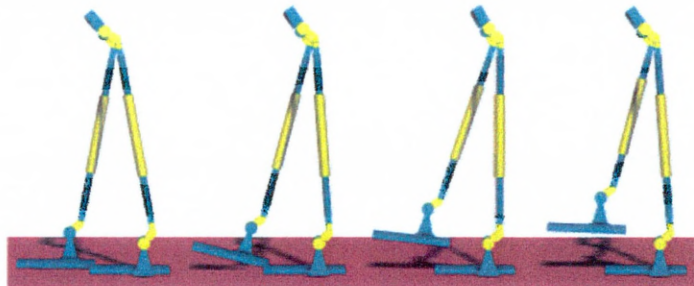


Figure 5.11 : *Exoskeleton Gait Pattern, soles matching human feet movement for Swing and Landing*

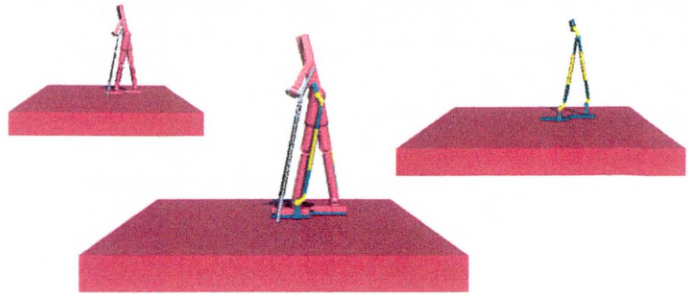
The next step was to match the trajectory of the soles of the feet as described in Table 5.2 with that of the soles of the exoskeleton. In the case of the simulation it was aimed to maintain a trajectory for the feet that would resemble as closely as possible to the normal trajectory of the feet when walking with crutches. Table 5.3 describes this situation for the complete sequence of a step.

Table 5.3 Exoskeleton Walking Cycle for a Step with the aid of crutches(ipsilateral)

Starting position

Stance leg is in front of swing leg. Leg actuators are fully extended.

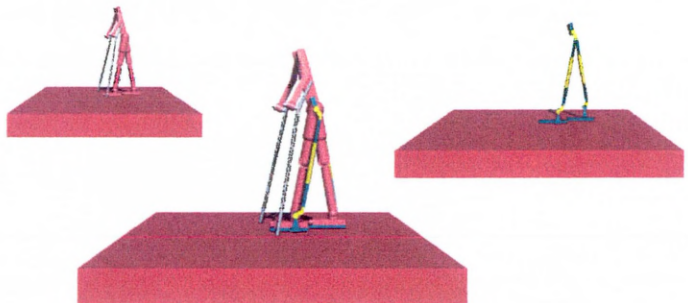
The starting position corresponds to the most stable position with four points of support. The actuators are fully extended and the tips of the crutches are near the toes.



Swing leg crutch's positioning

Both feet are flat on the ground. Right crutch is positioned ready for next step.

The crutch opposite to the stance leg is positioned around the point where the swing foot is expected to land, close to the toe but no further than half the length of the foot.



Stance leg crutch's positioning

Both feet are flat on the ground. Left crutch is positioned ready for next step.

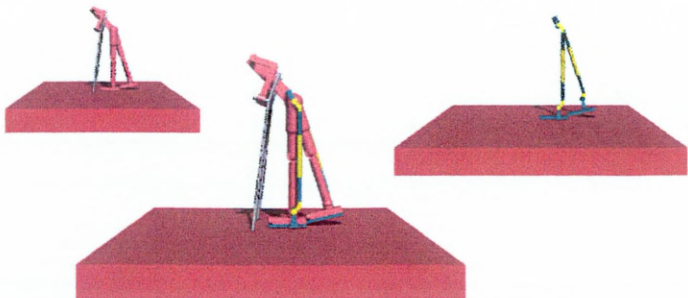
The second crutch is positioned parallel to the other crutch in preparation for the movement of the swinging leg. Body attitude has to be corrected to avoid tilting.



Preparation for Toe – off

Swing foot is not flat on the ground, toe still is in contact with the ground.

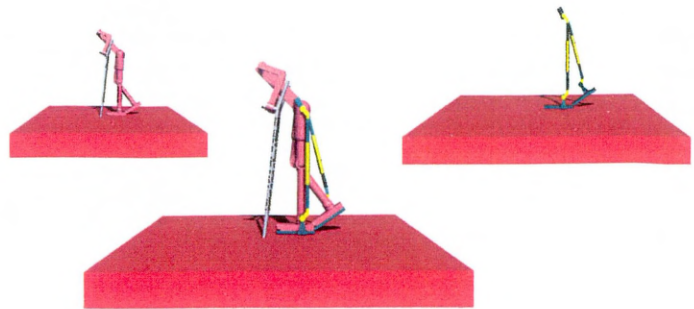
The body attitude is corrected so the centre of gravity compensates for the moment generated by the lifting of the swinging leg. The toes maintain contact with the ground.



Toe – off

Swing foot is about to lift off, but still is in contact with the ground. One actuator is fully extended and the other is fully retracted.

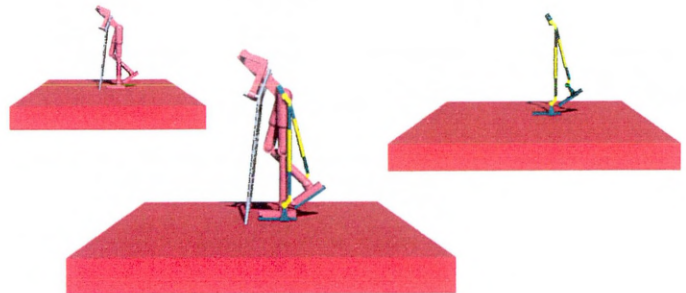
Lift off is imminent, balance is ensured by the remaining support points reaction forces.



Lift – off

Swing foot lifts off the maximum knee flexion value. Both actuators are fully retracted.

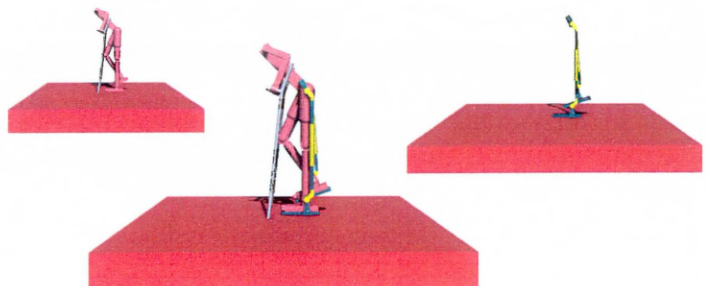
The swinging foot is propelled upwards and towards the front due to the contraction of the actuators bending the knee.



Mid-Swing

Swing foot is airborne going through the swinging sequence. Both actuators are fully retracted.

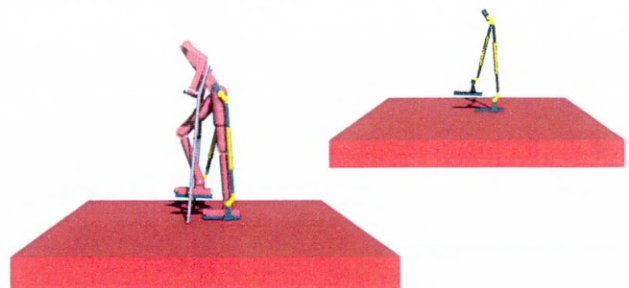
The swinging leg pivots around the hip and with the aid of the hip actuator. Balance is maintained as well as leg abduction.



Maximal Swing Position

Swing Leg achieves its maximum rotational value. Both actuators are fully retracted.

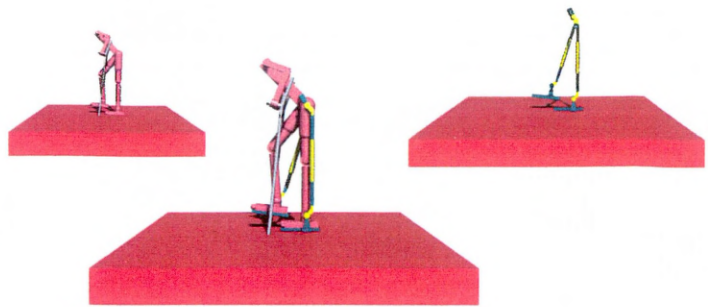
Maximum swing and pre-land are divided in two in the exoskeleton sequence, allowing maximum toe clearance during the swing phase by maintaining the leg abducted.



Pre-Land

After swing leg achieves its maximum rotational value it prepares for landing. One actuator is fully extended and the other is fully retracted.

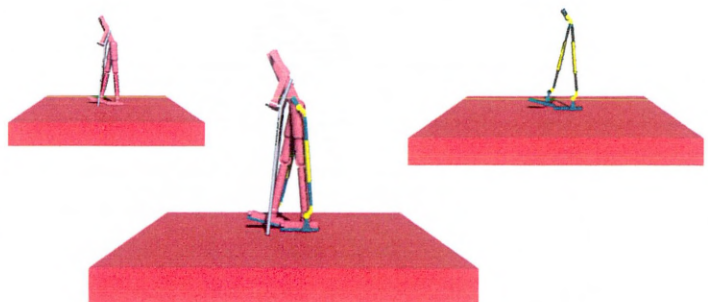
In pre-land, the cylinders start to extend and the position of the exoskeleton sole is matched.



Heel Strike

Swing leg achieves its maximum rotational value and prepares for landing. Both actuators extend fully.

The foot travels downwards until the heel strikes the floor, the impact is smooth allowing the foot to land gently.



Starting position

Swing leg achieves its maximum rotational value and prepares for landing.

Crutches can be relocated if they are not in the proper position. If the support points are correctly positioned then the whole sequence can be re-started.



The sequence of movements is nearly identical to that of Table 5.2 with a minor change in the maximum swing / pre-land as it was split into two movements, one places the heel on the ground (minimising the impact) and the second places the complete sole flat on the ground. It is also possible to avoid heel strike and place the sole directly flat onto the ground as toe clearance can be easily calculated. This sequence improves the toe clearance by swinging the leg while contracted¹.

¹ Both leg actuators are fully retracted to its maximum value of hip rotation for the length of the step.

5.7 Energy Analysis

The potential and kinetic energy of the biped system are described by equations 5.6 and 5.7 ^[50].

$$PE = m_0 g Z_0 + m_1 g Z_1 + m_2 g Z_2 \quad \text{Eq. 5.6}$$

$$KE = \frac{m_0}{2} \left(\left(x_0' \right)^2 + \left(z_0' \right)^2 \right) + \frac{m_1}{2} \left(\left(x_1' \right)^2 + \left(z_1' \right)^2 \right) + \frac{m_2}{2} \left(\left(x_2' \right)^2 + \left(z_2' \right)^2 \right) + \frac{I_0}{2} \left(\mu_0' \right)^2 + \frac{I_1}{2} \left(\mu_1' \right)^2 + \frac{I_2}{2} \left(\mu_2' \right)^2 \quad \text{Eq. 5.7}$$

This together with the energy analysis of human walking as seen in chapter 3 establishes a maximum power requirement for both legs in between 200 and 300 Watts. Seward ^[101] calculates human average power input of 320 Watts over an eight hour period, with an ability to provide a steady power output of 375 Watts for the first hour, followed by 80 to 150 Watts for a sustained period together with 27 kJ of energy which can be supplied as required bursts of up to 2.5 kW, overall this sums up to about 0.75 kWh ^[101]. Seward calculates for a complete electrical system based on a four hour duty cycle a battery in the range of 17 – 33 kg ^[101], although the testing system is a pneumatic one a complete electrical system is a better suited approach.

5.7.1 Energy Criteria for optimisation of Trajectories

The trajectories can be optimised and their efficiency calculated following a dynamic analysis. Starting with the dynamic equation :

$$\left(\text{diag}[m_0, m_0, I_0, m_2, m_2, I_2] \cdot p'' \right) + \left(\frac{\partial p}{\partial f} \right) \quad \text{Eq. 5.8}$$

And therefore the efficiency as derived by Shih ^[50] becomes :

$$\text{Efficiency} = \frac{\int_s^f (Mp'' + G)^T p' dt}{s} \quad \text{Eq. 5.9}$$

5.8 Profile Generation for a Quarter-Scale size model

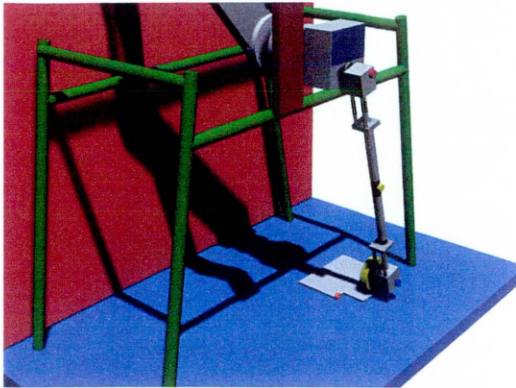
With the trajectories generated for a full size model, it was decided to build a quarter scale model in order to analyse the implementation of the gait pattern generation and evaluation of control strategies.

For this quarter scale model suitable trajectories were generated and simulated in preparation for a comparison with the recorded movements of the test rig. The preparation for the simulation led to the design of the mechanism. Pneumatics were chosen by the project because of the accessibility of components at Abertay and the knowledge gained from previous projects ^[16]. The leg height is estimated to be about 20 –25 cm with two possible actuators that could allow extra flexibility where by the two cylinders can be controlled independently.

The complete leg then can be mounted on a purpose design rig and the positions of each joint in space tracked for comparison purposes. The test rig was intended to emulate the behaviour of a leg undergoing the full sequence for a step. Therefore a hip (rotary actuator), a telescopic leg (linear actuator) and an ankle (rotary actuator) were required to be installed on a test bed suitable for performing the movements, and which could be constructed in the laboratory. For the type of movements and gait patterns expected to be simulated, a stepper motor with a gear box (40:1 to 50:1) mechanism and holding torques in the range of 500 – 600 mNm.

Table 5.4 explains the sequence simulated for the swinging leg in the test bed from the feet-sole movement in normal gait, for a full step, Appendix A provides more details.

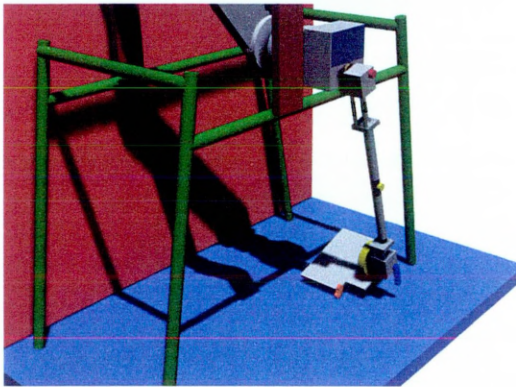
Table 5.4 *Full step sequence for swinging leg in Test Rig*



Starting position

Swing leg is in position. Both cylinders are fully extended. Markers are placed in position.

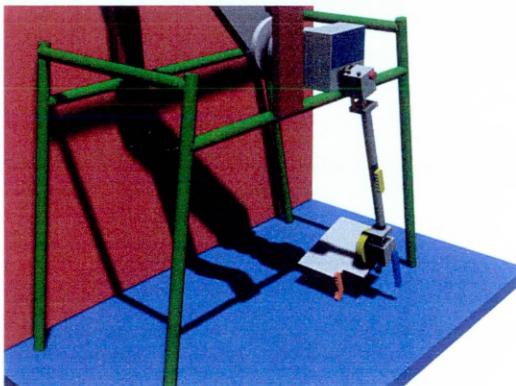
The leg is prepared for lift off, both cylinders are extended and the hip and ankle actuators maintain the leg and the foot at the right angle. Balance conditions are checked.



Toe – Off

Swing foot is lifting off, but still the extreme of the toe is in contact with the ground. One cylinder is fully extended and the other is fully retracted.

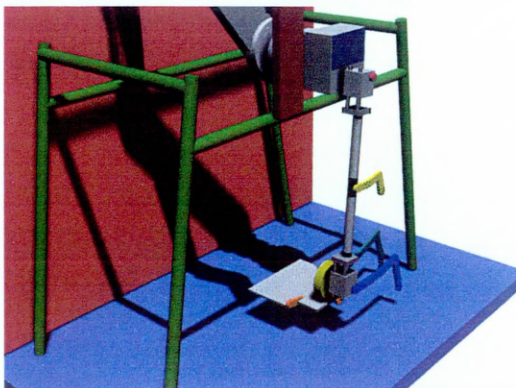
The leg is about to lift off, the ankle rotates and only the toe is in contact with the ground.



Lift – Off

Swing foot lifts-off the maximum knee flexion value. Both cylinders are fully retracted.

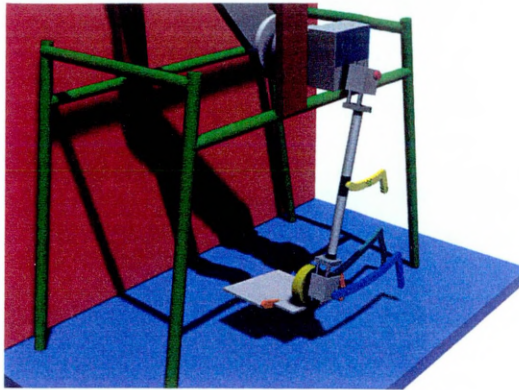
Lift-off occurs and the foot is propelled upwards and to the front. The ankle angle is maintained and toe clearance allows for the swing.



Mid-Swing

Swing foot is airborne going through the swinging sequence. Both cylinders are fully retracted.

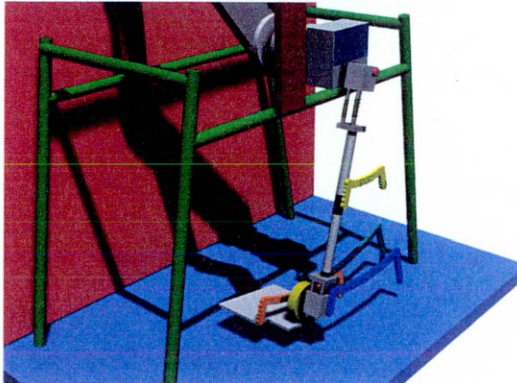
The hip rotates and swings the leg. The foot rotates as the ankle changes its value to increase toe clearance.



Maximum Swing Position

Swing leg achieves its maximum rotational value and prepares for landing. Both cylinders are fully retracted.

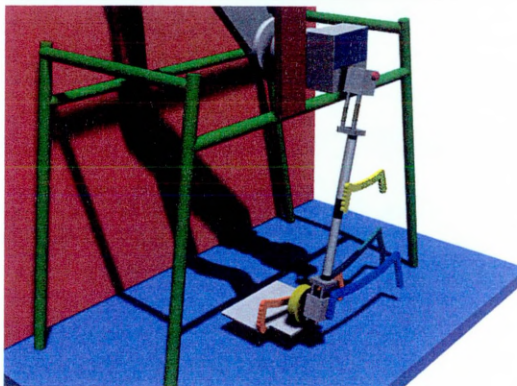
The hip achieves maximum value of rotation for the selected step length. The ankle positions the foot for heel strike.



Pre-Land

Swing leg achieves its maximal rotational value and prepares for landing. One cylinder is fully extended and the other is fully retracted.

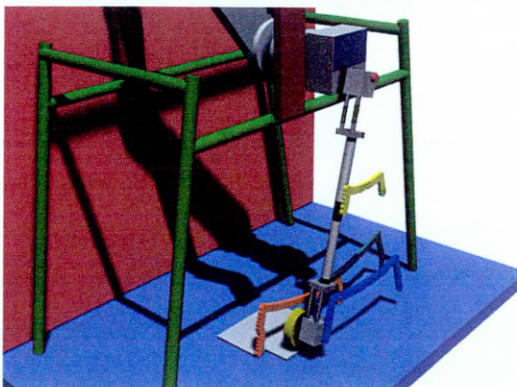
The cylinders start their extension sending the foot downwards in preparation for landing.



Heel Strike

Swing leg achieves its maximum rotational value and prepares for landing. Cylinders are almost fully extended.

The cylinders extend the leg almost completely until the heel border makes gentle contact with the ground.



Starting position

Swing leg achieves its maximum rotational value and prepares for landing.

The cylinders extend completely and the foot rotates, pivoting around the heel landing completely on the surface of the ground. Balance conditions are checked.

Figure 5.12 shows the model design that was finally chosen for the implementation shown in Table 5.4.

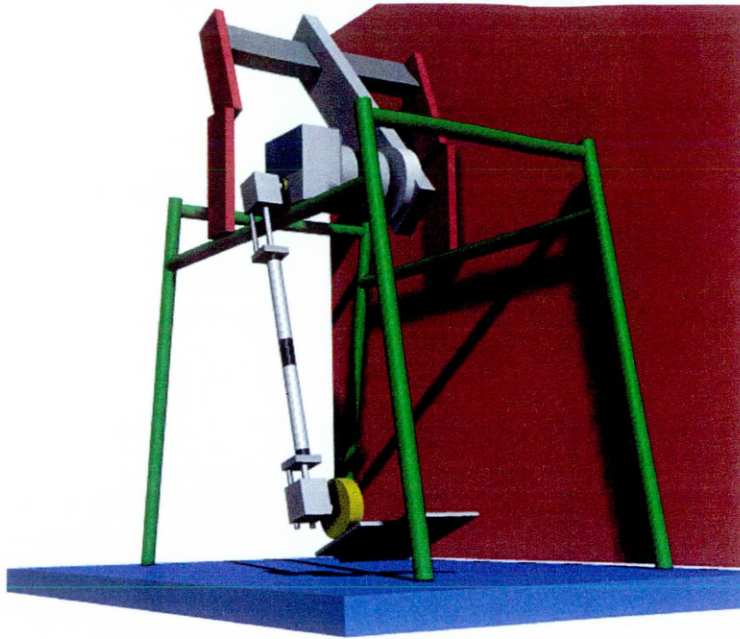


Figure 5.12 : *Test Rig Design Configuration for one leg with back to back actuators.*

Figure 5.13 shows the complete set of trajectories for one of the simulations used to compare the test rig with the gait studies (light and clinical). Chapter 7 provides a detailed explanation of the simulations and comparisons and analysis performed.

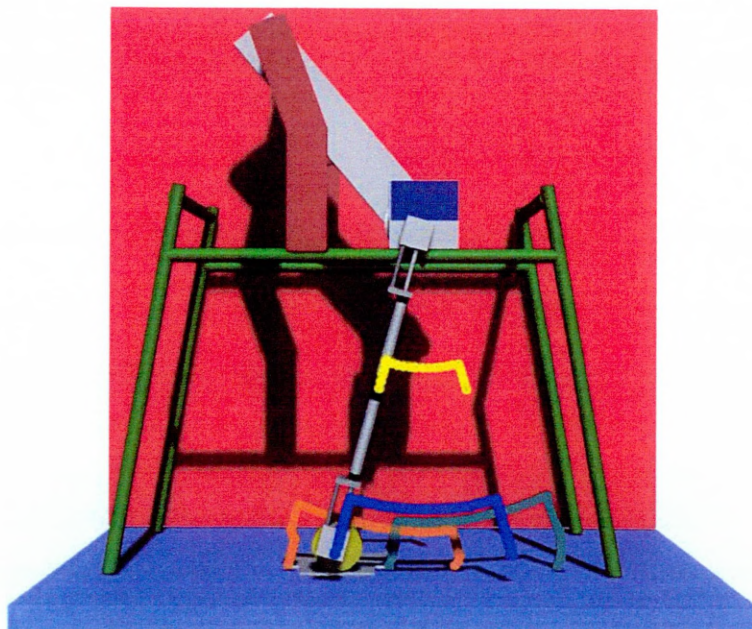


Figure 5.13 : *Simulation of trajectories for Test Rig configured for one leg. Digital markers provide information on the feet, ankle and leg midpoint position*

5.9 Summary

The implementation of a test bed, based on the geometry validated by the analysis of the human and machine gait, as a mechanism with the same number of DoF as specified in Chapter 4, requires the simulation of different models with the same geometrical configuration in order to test the implementation of suitable human-like, gait patterns, movement profiles and trajectories generated by the analysis of human and machine gait described in Chapters 3 and 4.

This extensive simulation has been the topic of discussion and analysis of this chapter, since the telescopic knee implementation was chosen for its advantages^[16] in energy consumption at the ankle joint, and control strategy. The test bed implementation will be the topic of the next chapter, the specifications of this test bed include a leg length of about 15cm with linear actuators connected back to back and rotary actuators at both ends. This test bed will help to validate the capability of achieving suitable trajectories from the simulated movement profiles developed from the gait analysis data, and the human model simulations performed with and without the extra support provided by the use of crutches as shown in Figure 5.14.

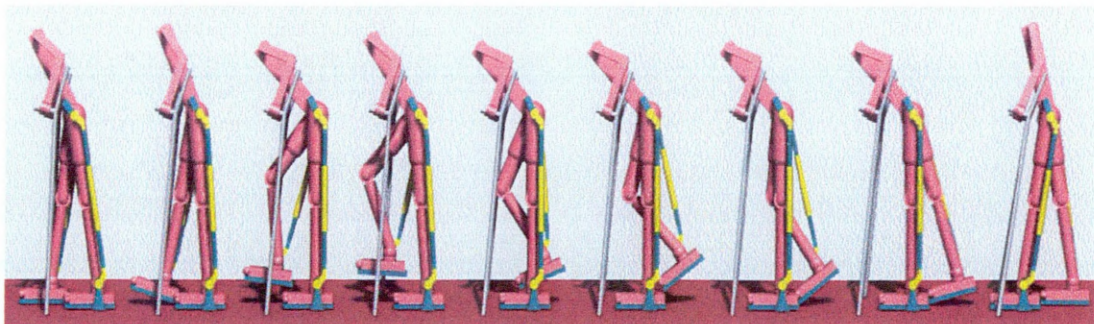


Figure 5.14 : *Simulation of a full step replicating the trajectories obtained with the light studies.*

Chapter 6

TEST BED IMPLEMENTATION

6.1 Introduction

After completing the modelling and simulation (based on the geometry validated by the analysis of the human and machine gait), the information collected made it possible to implement a test bed mechanism with the number of DoF and geometrical configuration specified in chapter 5. This test bed was then used to test the implementation of suitable human-like, gait patterns and movement profiles generated by the synthesis procedure and shown in the different simulations, as shown in chapter 5, and the audiovisual appendix.

6.2 Implementation Considerations

The telescopic knee implementation was chosen (see Figure 6.1) for its advantages^[16] in energy consumption at the ankle joint, and control strategy, as explained in the previous chapter. After careful examination of the simulations and evaluation of feasible mechanisms, pneumatic cylinders were selected to perform this telescopic movement. It became clear very early that for the purposes of the experiments, the actuators could provide the support and become themselves the actual legs of the design.

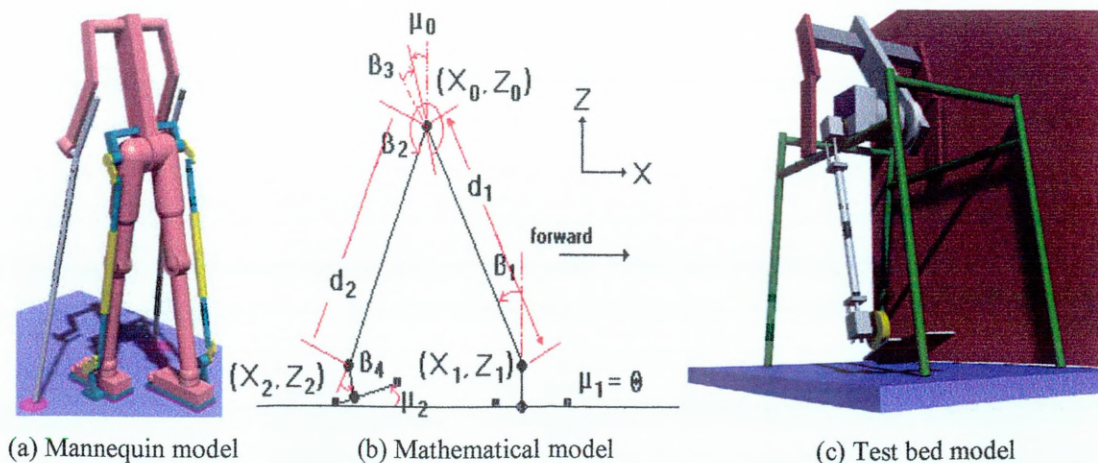


Figure 6.1 : Different models developed for the project

6.3 Actuators and Structural Considerations

Because of the availability of pneumatic cylinders and an ample provision of air supply in a wide range of operating pressures, it was decided to implement the system using pneumatic cylinders as the linear actuators. After examining the different cylinders (available from the university) suited for the specification outlined in chapter 5, it was decided that each leg was to be fitted with two 2.24 cm stroke cylinders (10 cm retracted) connected back to back as this option allowed more flexibility by controlling the two cylinders independently for each leg, removing the need for position control of single cylinders. The two connected cylinders and associated guides became the actual implemented leg.

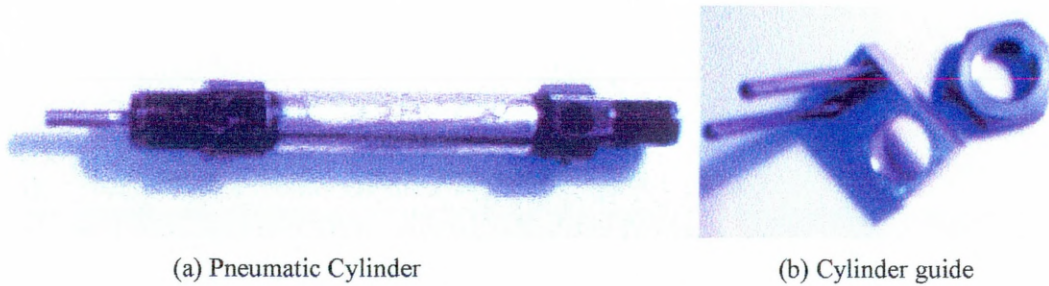


Figure 6.2 : *Pneumatic Cylinder and Guiding Rods*

As the mechanism will replicate the movement of a human leg in the sagittal plane, it is of paramount importance to constrain the movement to this plane by not allowing the cylinder shaft to perform any undesired rotations. In order to avoid this, a guide with two rods (see Figure 6.2) was custom made for these cylinders. To ensure that the legs were straight at all times with minimum energy consumption (another parameter of the design), a spring was located around the shaft in between the body of the cylinder connecting the cylinder with the rotary actuator of the hip or the ankle (see figure 6.3). Appendix B shows a complete geometrical description of the test bench mechanical components.

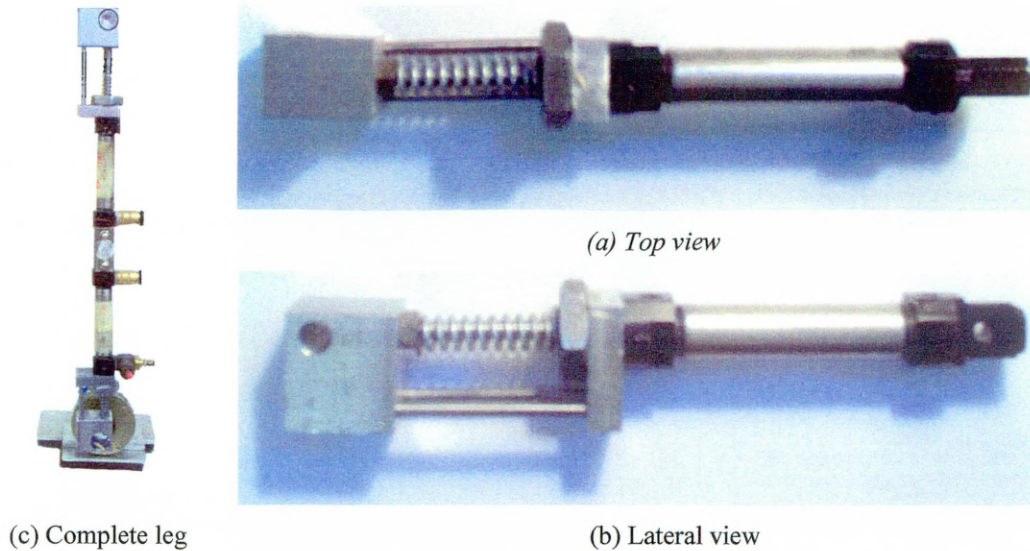


Figure 6.3 : Complete leg with pneumatic cylinders and guiding rods and spring

Four of these supports were custom-built in aluminium for each cylinder so they could be attached to each other back-to-back and connected to the stepper motors at the hip and the ankle to form the complete leg structure (see Figure 6.4). Both legs were assembled in this way and then the stepper motors of each hip were connected.

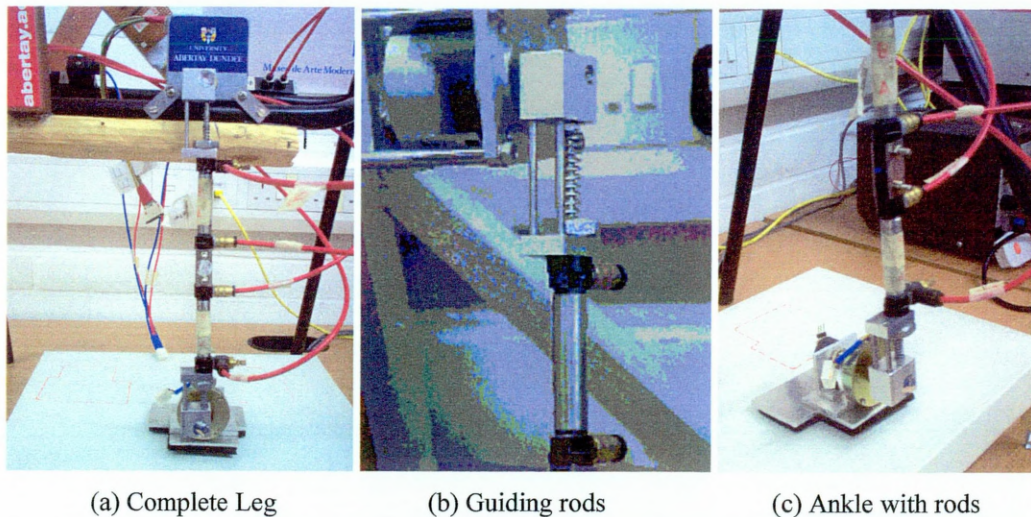


Figure 6.4 : Leg Development : (a) Complete Leg with Hip actuator and Cylinders with attachment mechanisms connected back-to-back; (b) Attachment mechanism for the Hip with Guiding Rods and Spring; (c) Attachment mechanism for the Ankle with Guiding Rods and Spring.

The stepper motors for the hip were also sourced within the university stock and were of a 4-phase 1.8 degree type. This is equivalent to 200 steps per revolution, or 400 if used in half step mode (see figure 6.5).

The ankle joint also was fitted with a smaller stepper motor as the ankle torque requirements are minimised by the use of the telescopic leg.

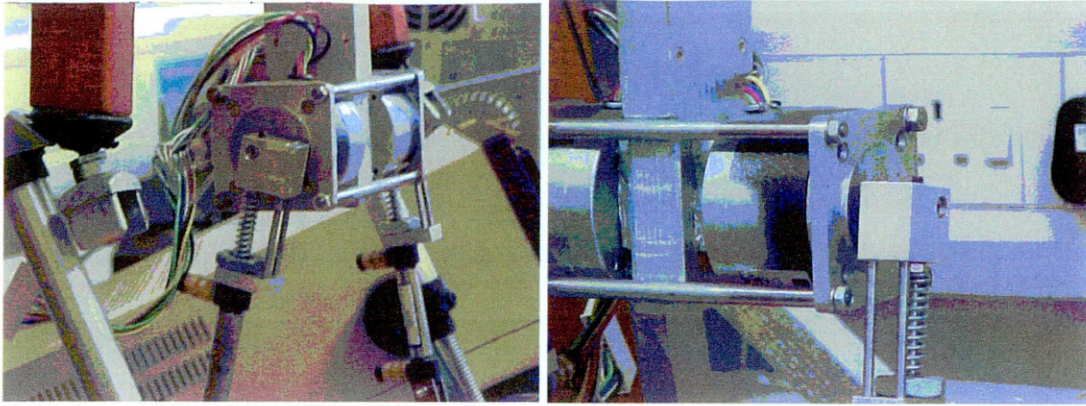
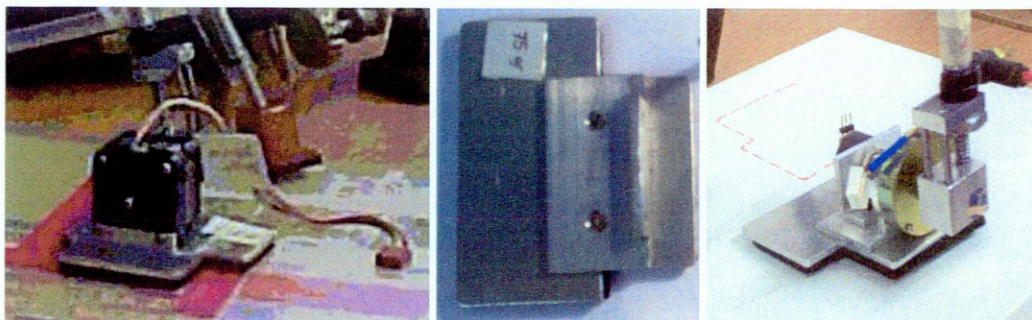


Figure 6.5 : *Hip stepper motors with connecting rods.*

The stepper motor was fitted to a custom-built foot that attached to the leg assembly (see Figure 6.6). The foot was designed to be proportional to the dimensions of the cylinders plus the stepper motors. The complete dimensions can be found in Appendix B.

Once the lower limbs of the robot were built, a torso with upper limbs was added and the hands were fitted with crutches. In order to reinforce the support, a structure holding the Test bed from above was custom-built and a simple hook added to the torso allowing it to be hung from the structure¹ (See Figure 6.7).



(a) Foot with Ankle stepper motor (b) Detail of the foot (c) Ankle Pancake stepper motor

Figure 6.6 : *First Test bed*

¹ The structure came to be referred as the “gallows”.

Once this Test bed was finished, some simple tests were carried out and it became clear that the direct drive of the hip was not adequate for the whole leg with the ankle actuator. More than that, the ankle actuator was too bulky for the application, although the direct drive demonstrated the ability to operate the foot.

The use of the gallows was cumbersome as the torso was not fixed precisely. Controlling its position was a crucial parameter in defining the tests that were to be performed to compare the behaviour of the system when executing the movements to the generated trajectories defined as suitable to implement.

A new simulation procedure was therefore developed. This time the hip was fixed at a point and the leg pivoted along this fixed location. For this concept a support structure was then developed. One leg of the first test bed was then used as the basis of the second one, but different stepper motors were selected (see Figure 6.8).

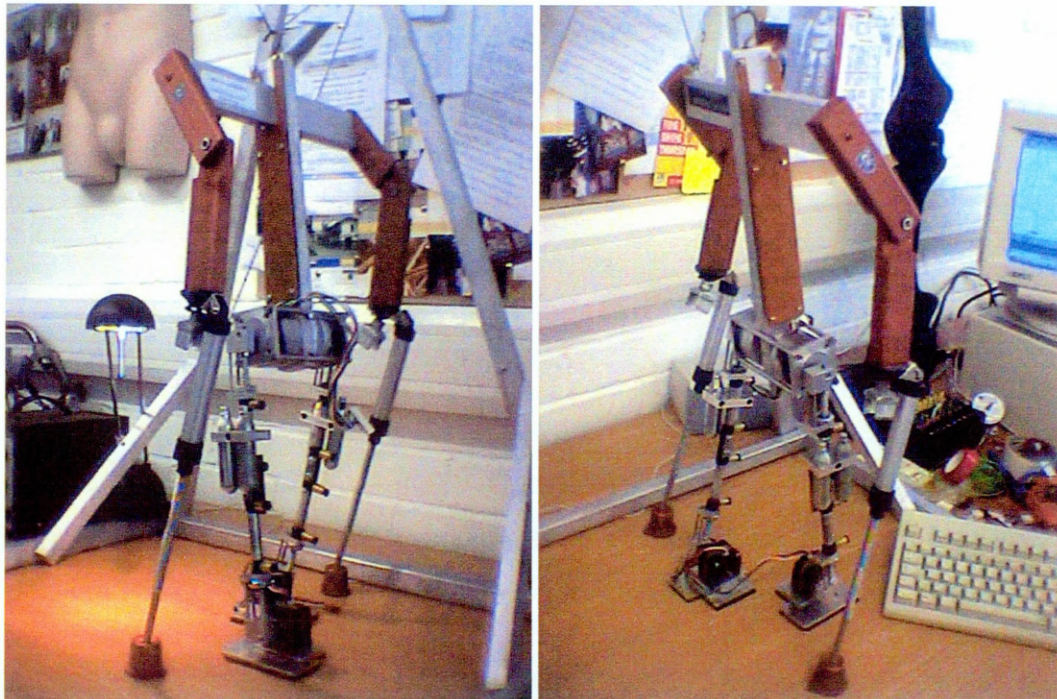


Figure 6.7 : *First Test Bed*

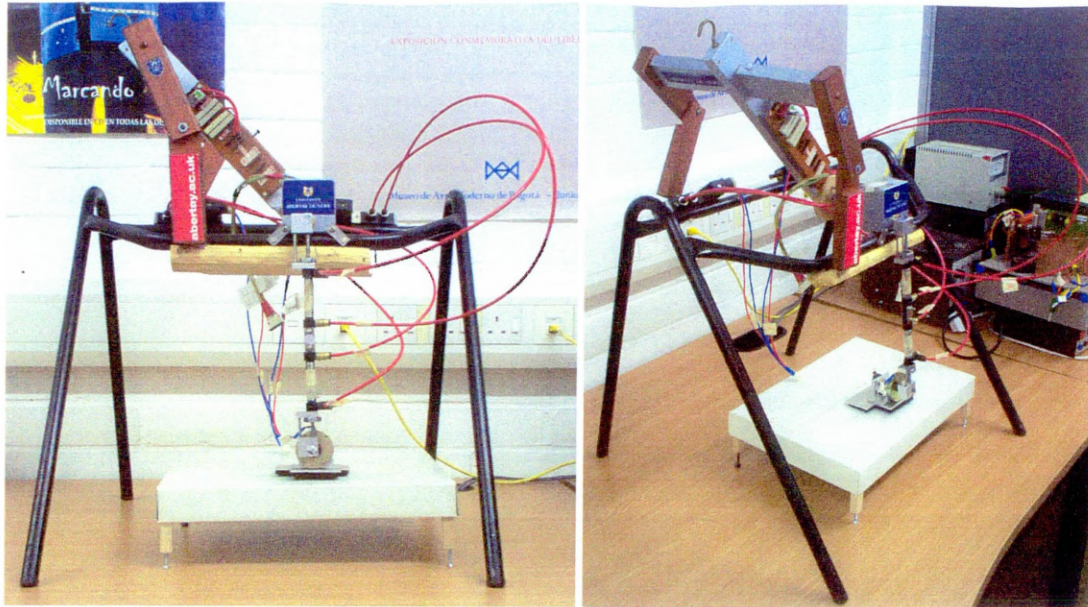


Figure 6.8 : *Second test bed.*

A new hip stepper motor was chosen with a higher detent torque and shafts on both sides so it could accommodate the potentiometer that will sense position, speed and direction. The same 4-phase 1.8 ° type was maintained as the connections were already existent for this configuration.

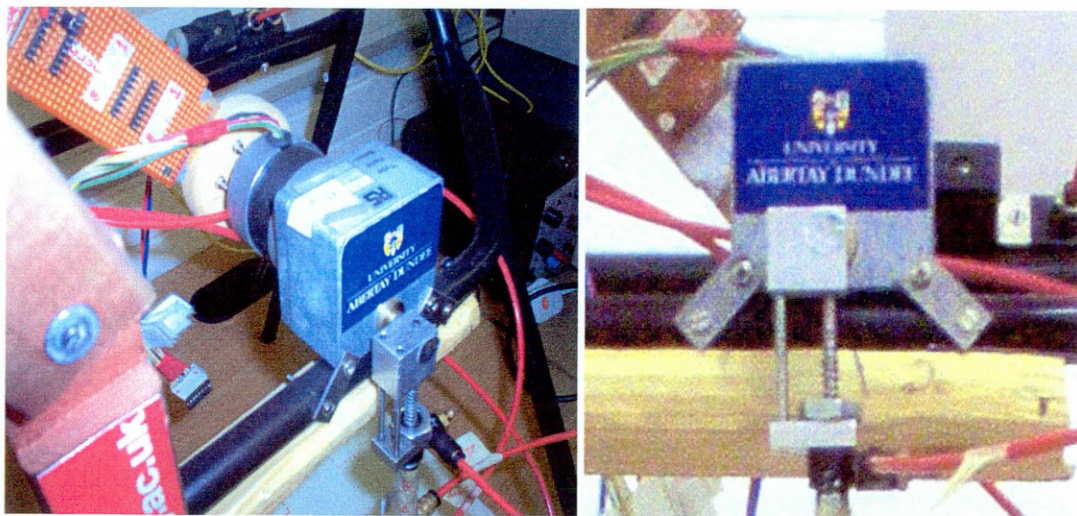
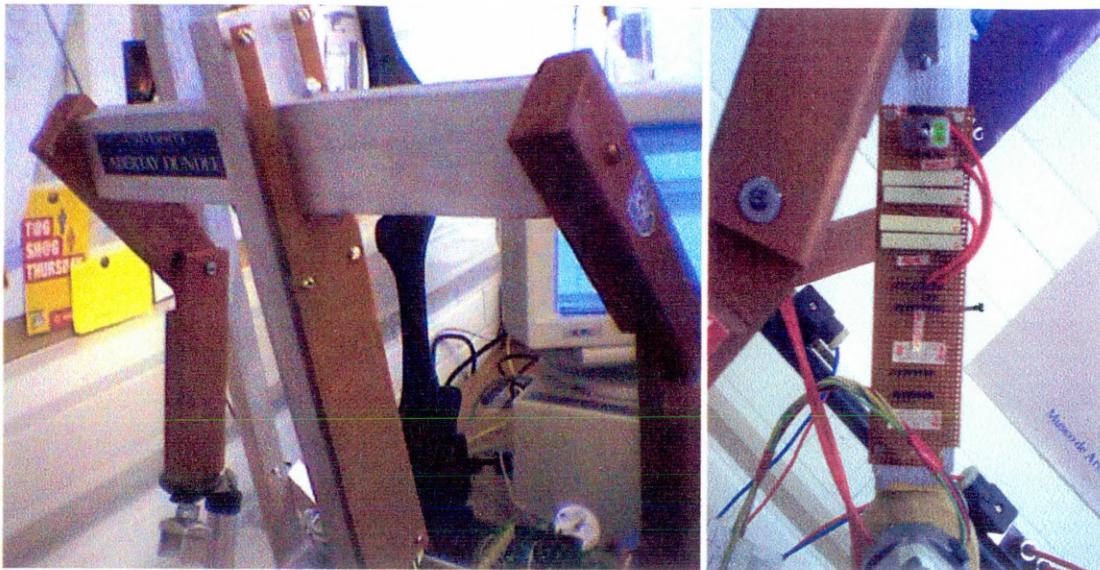


Figure 6.9 : *Hip 50:1 gear box for increased torque actuation*

A 50:1 gearbox was fitted allowing for the required increment on torque to move the complete leg (see Figure 6.9). The gearbox was positioned on the structure of the test bed directly where the hip is located and the shaft of the gearbox acts as the pivot

point for the leg. The shaft of the gearbox was then connected to the head of the cylinder.



(a) Space for additional circuitry

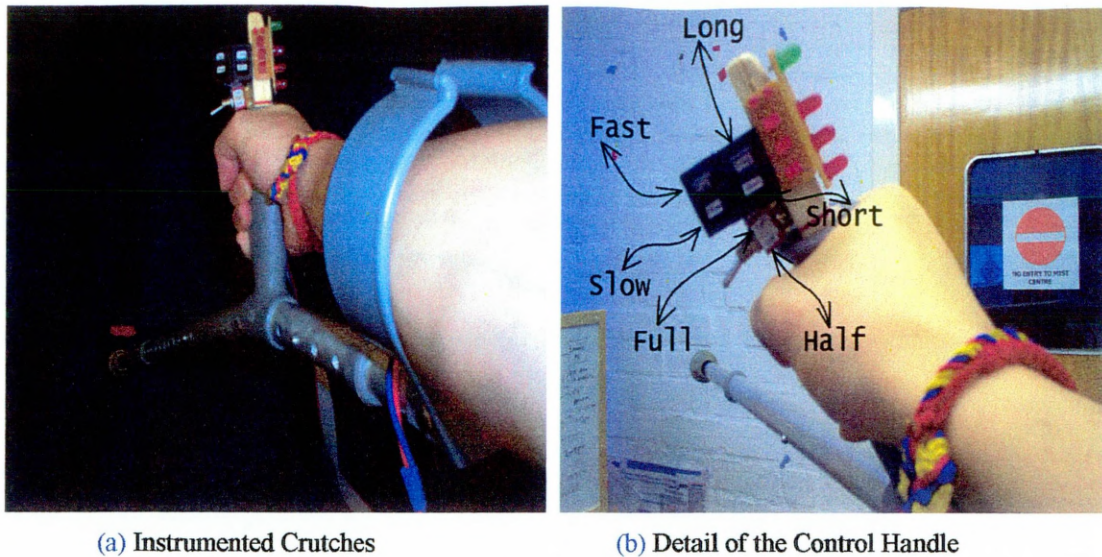
(b) Circuitry and connectors in place

Figure 6.10 : *First Test bed space allocated for additional circuitry and connections; and Second Test bed with circuitry and connectors in place.*

Part of the circuitry was located on the torso from where it was connected to a control box that interfaced with the PC where the trajectories were programmed and monitored (see Figure 6.10). Finally the controls that were located in the crutches (see Figure 6.11) enabled three controllable variables:

- Step Length (Long / Short)
- Step Interval (Slow / Fast)
- Lift Magnitude (Full / Half)

The step length defines the number of rotations the stepper motor goes through to reach the required step, a short step here being half the hip rotation of a long step. Slow or fast determines the time interval in between steps – “slow” having twice the time interval as the “fast” setting. The full and half lift magnitude determines if both cylinders are contracted or only one is used in order to achieve the required toe clearance.



(a) Instrumented Crutches

(b) Detail of the Control Handle

Figure 6.11 : Instrumented Crutches with Control handle.

In order to implement the gait pattern the following information must be made available to the controller:

- The contact at each foot².
- The contact at the crutches².



(a) Crutch with sensor and spring

(b) Crutches with connecting cable.

Figure 6.12 : Instrumented Crutches

This is necessary as the requirement for static balance must be met at any given moment of the movement, the system in fact checks for a 3 or 4 point stance, section 6.5 explains this in more detail.

² The implemented sensor was a microswitch that checked for contact rather than reaction force.

When these signals indicate a stable configuration, it is then possible for the wearer to command the rear leg to lift in order to perform the swing, achieving a safe movement in the sagittal plane.

The crutches are instrumented with microswitches and a tuneable spring (see Figure 6.12). Thus, for any given user's weight it can be adjusted to trigger the microswitch when full or partial weight is located on the crutches. The sequence for a single step is based on the combination of the exoskeleton and the crutches.

Once a stable 4 point state is detected it is possible to operate the exoskeleton to lift the rear leg and swing it forward, thus positioning the leg in accordance with the selected trajectory. The step length could be short or long, the step interval could be fast or slow and the lift magnitude could be either full or half (for a full explanation see section 6.4). Figure 6.13 sketches an implementation with the controls embedded in the crutches and all the basic sensing and actuating equipment required for the test model.

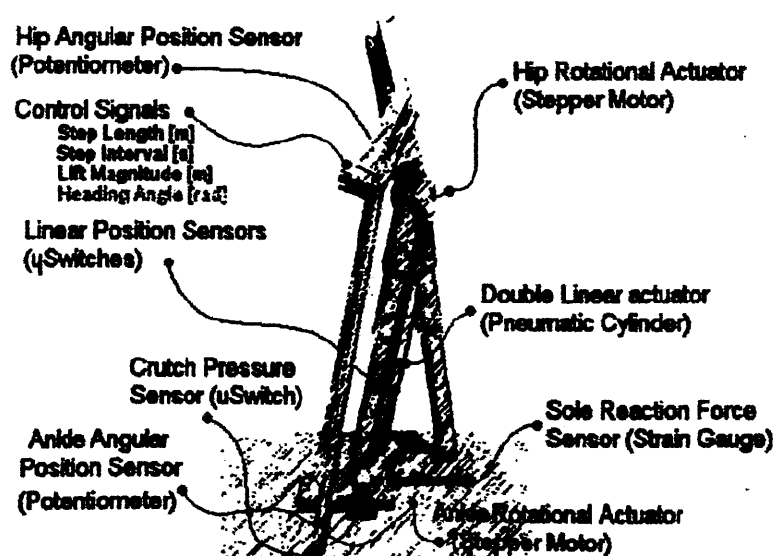


Figure 6.13 : Diagram with the sensors and actuators allotment.

6.4 Control Strategy

The control strategy is based around achieving the path trajectories for the position of the hip, knee and ankle while constantly ensuring that the condition of static balance enforced by at least three points of support is valid for any given moment. The command in the crutches allows for the selection of the specified gait pattern, the linear actuator does not provide a continuously variable lift, the extension and retraction of the cylinders is performed by means of *on/off* valves and no attempt to achieve a proportional control strategy was attempted. Instead, the provision for using proportional valves was left within the program if it was felt that a more precise position-control was required. The hip and ankle stepper motors positioning relies on counting the actual steps, and therefore the initial control approach is an open loop one.

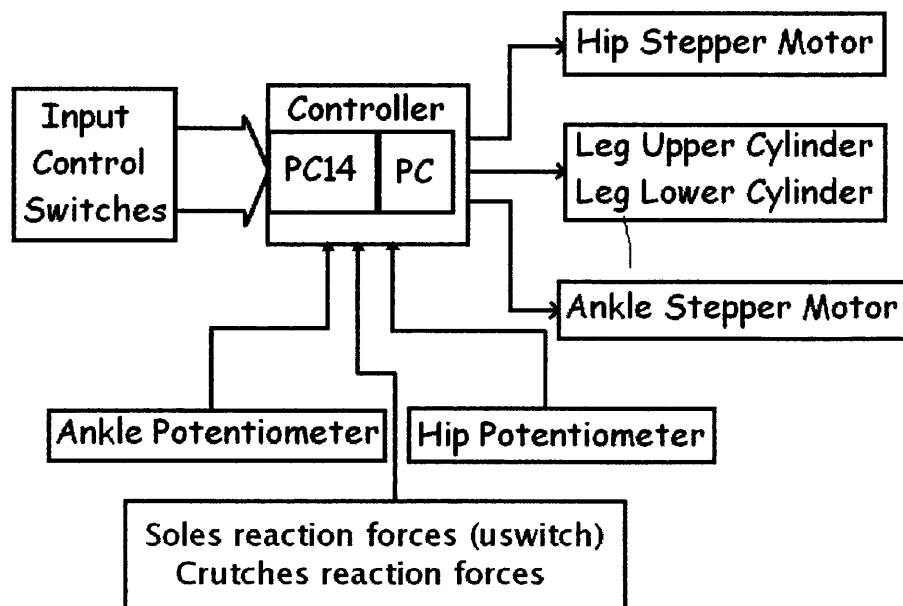


Figure 6.14 : Control diagram with the sensors and actuators interconnection.

Figure 6.14 offers a flow diagram of the different signals derived from figure 6.13, with the controller receiving input and sensing control signals and despatching control signals to the actuators.

A Finite State Control was then selected as it allows the use of simpler controllers and fast numerical processing and the reference trajectory can be stored as a number of finite states. In order to minimise the number of states, stability is maintained by always keeping at least three points of contact with the ground.

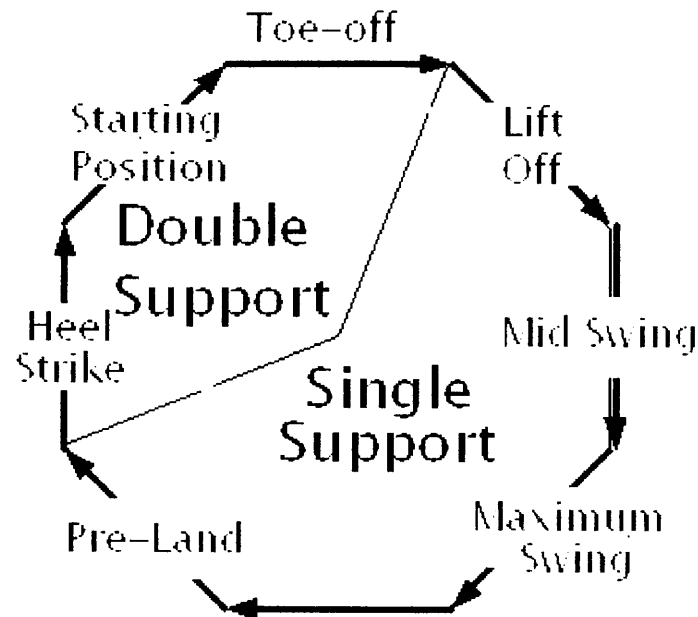


Figure 6.15 : Transition states between single and double support, for a single step.

The Model Reference Controller (MRC) oversees the Finite State Controller (FSC) so there are two levels to consider using this approach. The MRC supplies a stepwise reference trajectory, feedback gains and other parameters to the FSC, which uses the position feedback to track the reference. For a full list of the Reference States see Table 6.1.

The MRC is used because it allows for future expansions of the capabilities of the controller to modify the system. The FSC checks the control inputs and enforces the transitions in between the different states. The MRC checks at first instance the hip potentiometer signal and the contraction and expansion of the telescopic actuators in

order to check the position in space of the leg. It can compare this to the movement that is being carried out and implement corrections. The FSC can be understood as a simple control strategy that implements sequential transitions, e.g. compress the cylinders, rotate hip 5000 steps, extend cylinders, whereas the MRC checks for leg pose and attitude and corrects deviations from the intended trajectory. This approach simplifies the tests and offers a wider degree of flexibility to the user since a key objective is to give the user the feeling of controlling the exoskeleton and not being controlled by it.

Table 6.1: Look Up Table for Control Inputs

If [Step Length And Step Interval And : Lift Magnitude] Then [Perform Profile]	If [(Step Length = Long)And (Step Interval = Fast) And (Lift Magnitude = Full)] Then [Perform Profile 1]	If [(Step Length = Short)And (Step Interval = Fast) And (Lift Magnitude = Full)] Then [Perform Profile 2]
If [(Step Length = Long)And (Step Interval = Slow) And (Lift Magnitude = Full)] Then [Perform Profile 3]	If [(Step Length = Short)And (Step Interval = Slow) And (Lift Magnitude = Full)] Then [Perform Profile 4]	If [(Step Length = Long)And (Step Interval = Fast) And (Lift Magnitude = Half)] Then [Perform Profile 5]
If [(Step Length = Short)And (Step Interval = Fast) And (Lift Magnitude = Half)] Then [Perform Profile 6]	If [(Step Length = Long)And (Step Interval = Slow) And (Lift Magnitude = Half)] Then [Perform Profile 7]	If [(Step Length = Short)And (Step Interval = Slow) And (Lift Magnitude = Half)] Then [Perform Profile 8]

System Definition :

- I. Trajectories are defined by a sequence of transitions from one reference state to the next.
- II. Reference states are statically stable equilibrium points, therefore the biped is able to remain at any reference state indefinitely.
- III. If a direct natural transition from one reference state to the other is disrupted, the system will return to the previous reference state.
- IV. Control flow diagram is as follows:

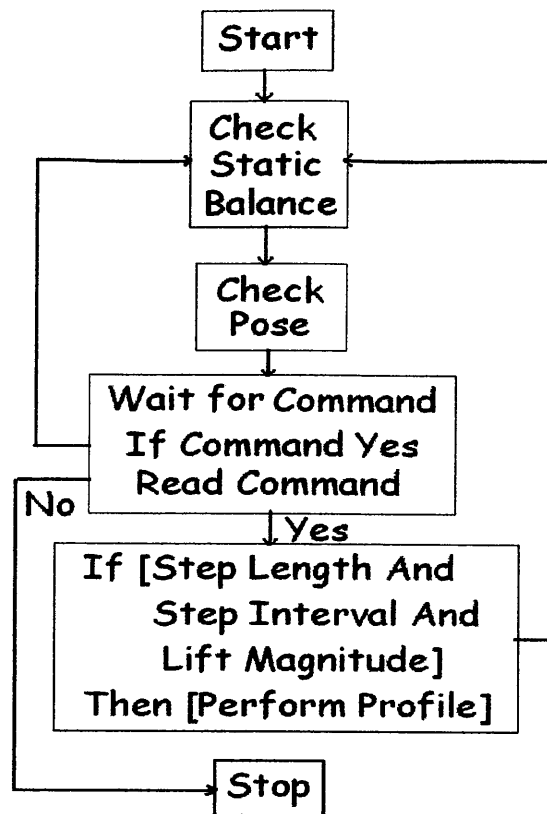


Figure 6.16 : *Control Flow Diagram*

6.5 Electronic System Considerations

A PC based digital acquisition system reads the angular and linear positions and velocities of the joints of the system controlling the stepper motors and pneumatic cylinders. For each leg two more readings must be obtained in order to implement the gait pattern; firstly the ground reaction forces at each sole through specially designed sole contact sheets, and secondly the output of a force sensor at the tip of the crutches.

The DAQ I/O system was developed around PC14 cards (see Figure 6.18) with the respective signal conditioning modules. A more detailed technical description of the electronics is described in Appendix B.

6.6 Step Sequence Considerations

For implementation on the model, the main sensors and actuators are as shown in figure 6.17. In the first instance, each of the individual joints, corresponding to an overall system of 3 DoF only, will be controlled for just one leg to determine the profile of the required control signals and to match the response of the leg to the developed gait pattern.

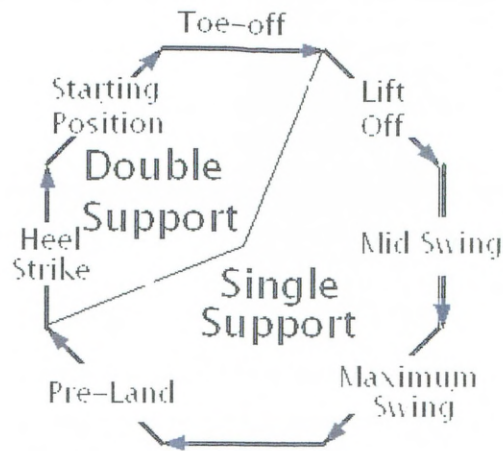
6.6.1 Gait Pattern Considerations

In order to implement the gait pattern the following information must be made available to the controller:

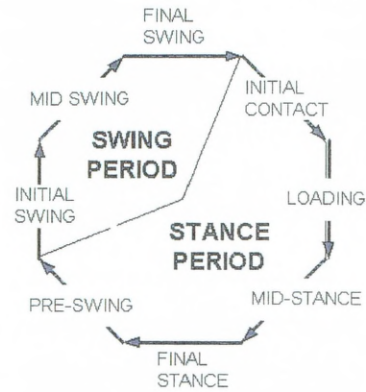
- The contact at each foot.
- The contact at the crutches.

When these signals indicate a stable configuration, it will then be possible for the wearer to command the rear leg to lift in order to perform the swing and achieve a safe movement. In the first instance the exoskeleton will only be able to walk in the sagittal plane.

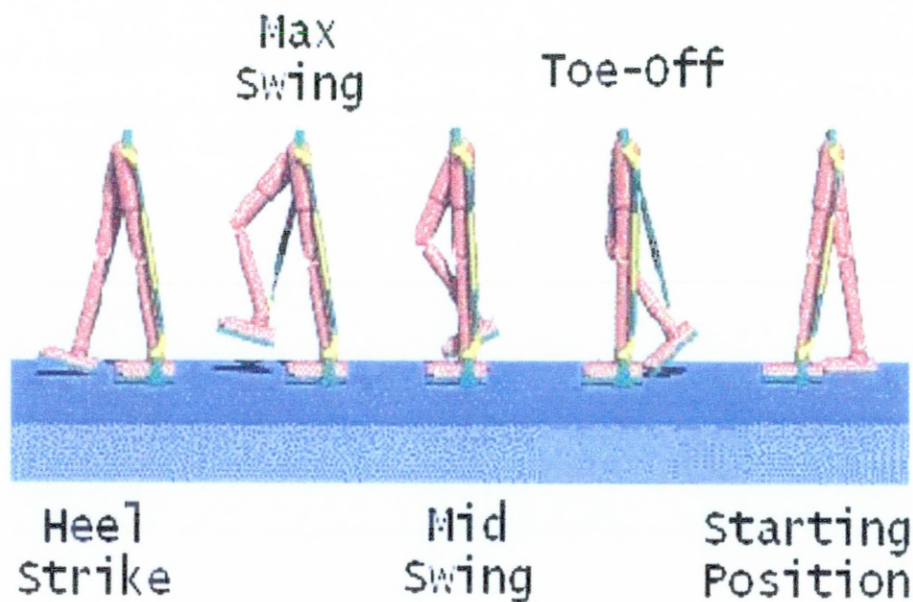
To achieve normal walking, a sequence of steps must be made with each leg moving forward and alternating between states of static and dynamic balance and unbalance. Each step is then a sequence of movements grouped in different states involving the rotation of the lower limb articulations. These states are of two kinds - single support and double support. A step is initiated in a double support phase and terminates in a double support phase.



(a) Project's gait cycle model



(b) Perry's gait cycle model













(c) Gait cycle model

Figure 6.19 : Gait Phases nomenclature (including Perry's nomenclature ^[22] top right)

One way to define these phases is to use the nomenclature developed by Perry ^[20] and shown in Figure 6.19. The idea is to divide a full step in two periods, one is the swing period and the other is the stance period. For the project a different nomenclature and definition was established using the same idea of dividing the gait cycle into periods defined by the transitions between single and double support. Therefore the two periods are defined as single and double support instead of swing and stance periods, as seen Figure 6.19. The requirements for walking based on the exoskeleton are those shown in Table 6.2.

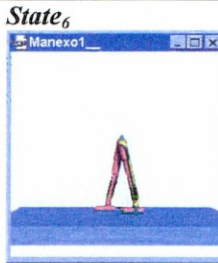
Table 6.2: Gait Cycle for a step

<p>State₀</p> 	<p>In State₀, the exoskeleton is at the beginning of the cycle, one leg in front of the other leg in static balance. This will become the final position with the legs inverted at the end of the step.</p>	
<p>Starting Position</p>		
<p>State₁</p> 	<p>The exoskeleton moves from Phase₀ to State₁, getting ready to lift the rear leg. The front leg becomes the stance leg, the rear linear actuator is contracted, lifting the rear heel but maintaining double support.</p>	
<p>Toe-Off</p>		
<p>State₂</p> 	<p>During State₂ the rear leg is lifted to the determined height or, as in this case, to the maximum contraction of the linear actuators.</p>	
<p>Lift-Off</p>		
<p>State₃</p> 	<p>States 3, 4 and 5 all correspond to the swing period. Once the rear leg is lifted (State₂) it is ready to swing and land in front of the stance leg, thus producing movement. Once the leg has achieved the desired swinging angle it is stretched forward so the foot can be placed on the ground.</p>	
<p>Mid_Swing</p>		
<p>State₄</p> 	<p>Because the crutches are assisting the movement at all times, the process of grounding the foot is not only meant to prevent falling, but also positions the swinging leg appropriately in relation to the position of the crutches.</p>	
<p>Maximum. Swing</p>		

State₃ therefore describes the mid-stance swing position, Phase₄ the uppermost extension that the mechanism can achieve and State₅ the final extension and contact with the ground.

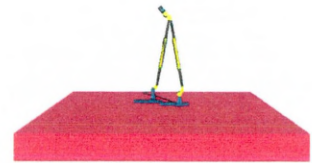


Pre-Land / Heel Strike



Once the swinging leg has grounded and is positioned correctly flat on the ground, the final position is similar to State₀ but with the legs reversed.

Starting Position



6.6.2 Implementation Considerations

The controls for the operation of the exoskeleton are embedded into the crutches and will allow the user to control parameters such as the height of the step, the length of the step and the rate at which the leg swings forward. However, sensors on the crutches work in association with the sensors on the soles of the feet of the exoskeleton to sense the weight of the person as they lean forward before allowing the lifting of the rear (swinging) leg, thus ensuring a stable position at the start of the step.

The sequence for a single step based on the combination of the exoskeleton and the crutches is then as follows:

- I. *Detect stable position – State₀*
- II. *Operator initiates the step sequence using controls on the crutches to set step height, step length and rate – State₂.*

- III. *The exoskeleton initiates the step – State₃, State₄ and State₅.*
- IV. *Step completed – State₆.*
- V. *Reposition crutches for next step.*
- VI. *Return to I.*

6.7 Summary

Based on the design specifications outlined in chapter 5 and the movements profiled with the simulated models, a test bed was developed so that the synthesised gait patterns and movement profiles could be tested and related to the light studies explained in chapter 3. For the hip a 12V, 1.8 ° motors with a holding torque of 500 mNm and a detent torque of 30 mNm were chosen. For the ankle a pancake type stepper motor 9.6V, 1.8 °, holding torque of 20 mNm and a detent torque of 2.5 mNm, expecting to allow angular velocities of a few degrees per second was selected.

The test bed dimensions resulted in a leg length of around 35 cm, a complete description of the test bed can be found on Appendix B.

The test will help to validate the synthesis of the profiles generated with the simulated models, and compare them with the trajectories obtained from the light studies as explained in the next chapter.

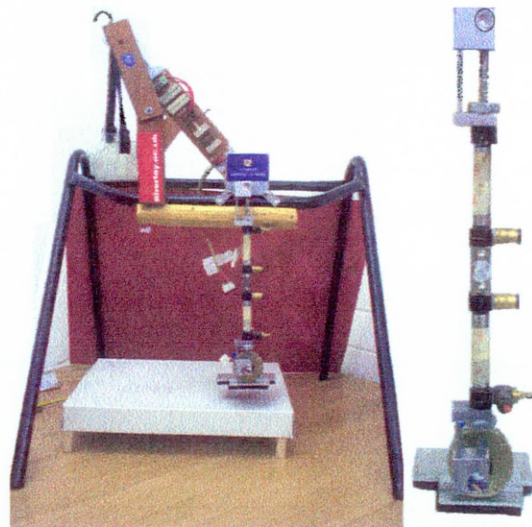


Figure 6.20 : Test bed implementation

Chapter 7

TESTS & RESULTS

7.1 Introduction

The methodology used to assess the system followed a very straightforward format. The trajectories generated by the simulation models were compared against the trajectories recorded from normal individuals with and without crutches for the ankle, toe, heel, knee and hip. The trajectories were then validated as compatible with the criteria established in Chapter 2 as the movement that the exoskeleton is required to achieve must have a human like feel. These type of gait patterns enable the user to feel more comfortable when using the exoskeleton.

The primary criteria for the synthesis of the trajectories was that of robustness and simplicity. With energy efficiency as key to the performance of the exoskeleton. The best fit resulted when the swinging leg was contracted throughout the swing period. This reduced the control complexity by actuating the cylinders for a shorter period of time and obtained a very good toe clearance, crucial when dealing with uneven ground surfaces and very similar to the human gait with crutches.

7.2 Gait Synthesis comparison with Clinical Gait Data

The simulated trajectories (of the foot, ankle, knee and hip) were synthesised for implementation on the test bed and compared against information collected on human gait through clinical gait analysis and light studies¹. Figure 7.1 shows the complete trajectory for a step for the three primary markers (heel, ankle and toe) and for the hip and knee. The results of the comparison were translated into simulated gait trajectories implemented with the virtual mannequin, as seen in Figure 7.2.

¹ For clarification of the human gait analysis process, readers are advised to consult chapter 3.

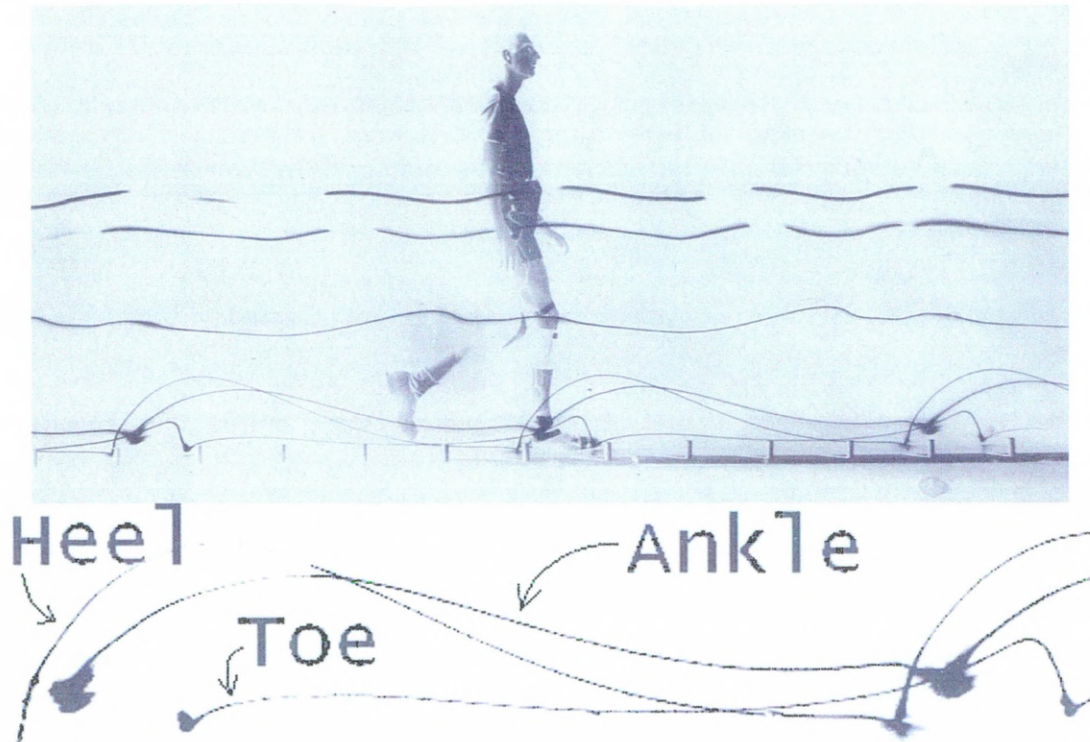


Figure 7.1 : Light studies with healthy individuals (Ankle, Heel and Toe markers) ^[6].

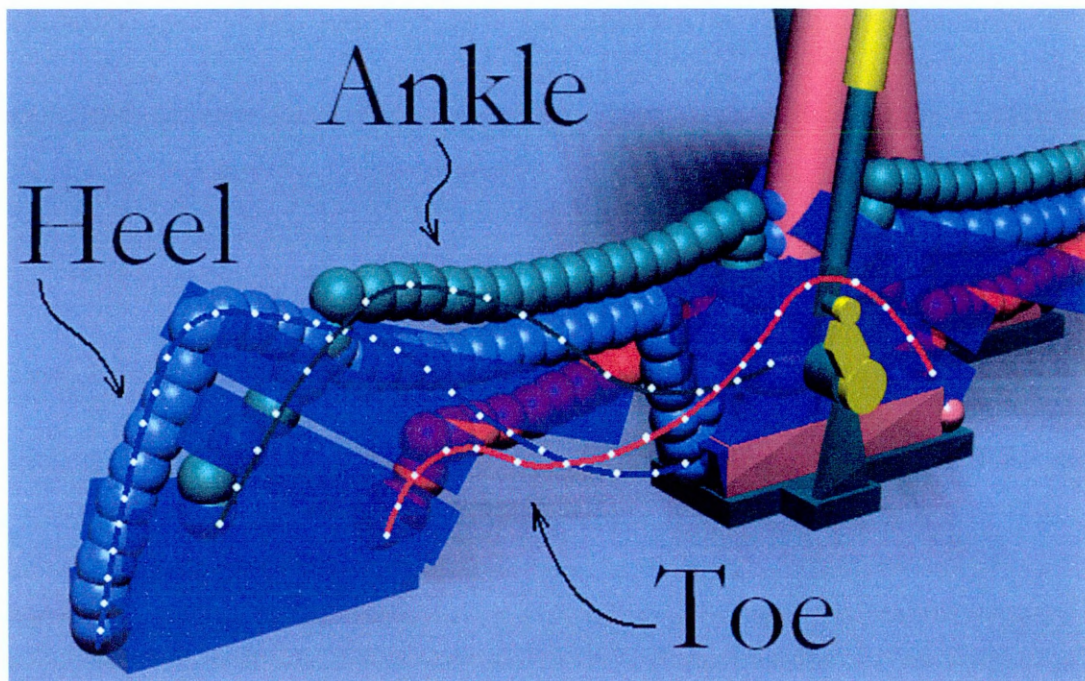


Figure 7.2 : Simulation for mannequin with exoskeleton with superimposed trajectories for light studies of healthy individuals (as shown in figure 7.1).

The movements of the ankle heel and toe were isolated for a complete step and then the simulations compared to the movement of the marker for the ankle, toe and heel

for the virtual mannequin walking with crutches using these movements over a step. The result of different types of implementations can be seen in Figures 7.3, 7.4 and 7.5. The information of most interest was the data collected for the markers at the ankle and the foot, as the idea behind the synthesis of the trajectories is that of matching the foot movement in the single and double support periods.

The first attempt was for an implementation similar to that of a healthy individual walking at 1.2 sec per step, see Figures 7.3 (a), 7.4 (a) and 7.5 (a). From that basic pattern, other different patterns were generated aiming at increasing the toe-clearance, (the basic human gait tends to have a very low clearance). A second sequence (rapid lift-off) was implemented but failed to improve the toe-clearance, see Figures 7.3 (b), 7.4 (b) and 7.5 (b).

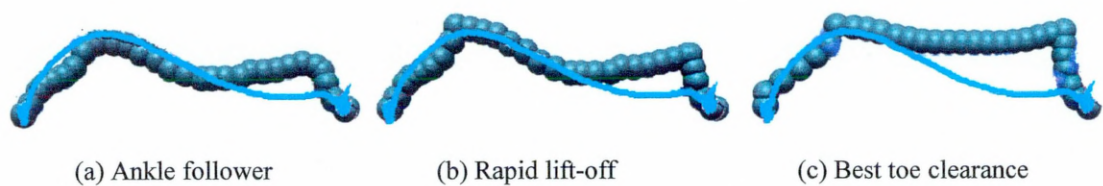


Figure 7.3 : *Ankle Simulated vs Continuous Light Studies.*

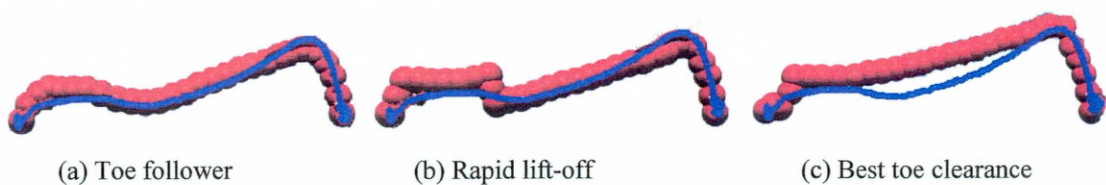


Figure 7.4 : *Toe Simulated vs Continuous Light Studies.*

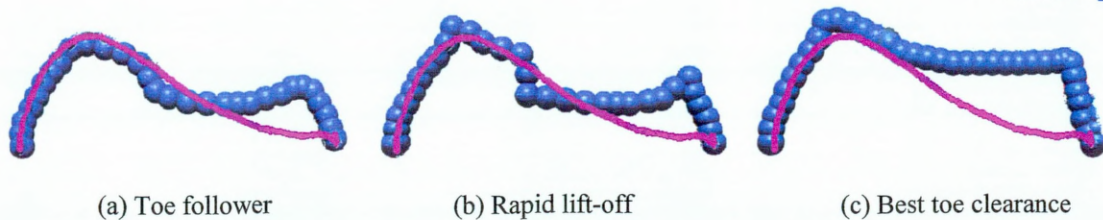


Figure 7.5 : *Heel Simulated vs Continuous Light Studies.*

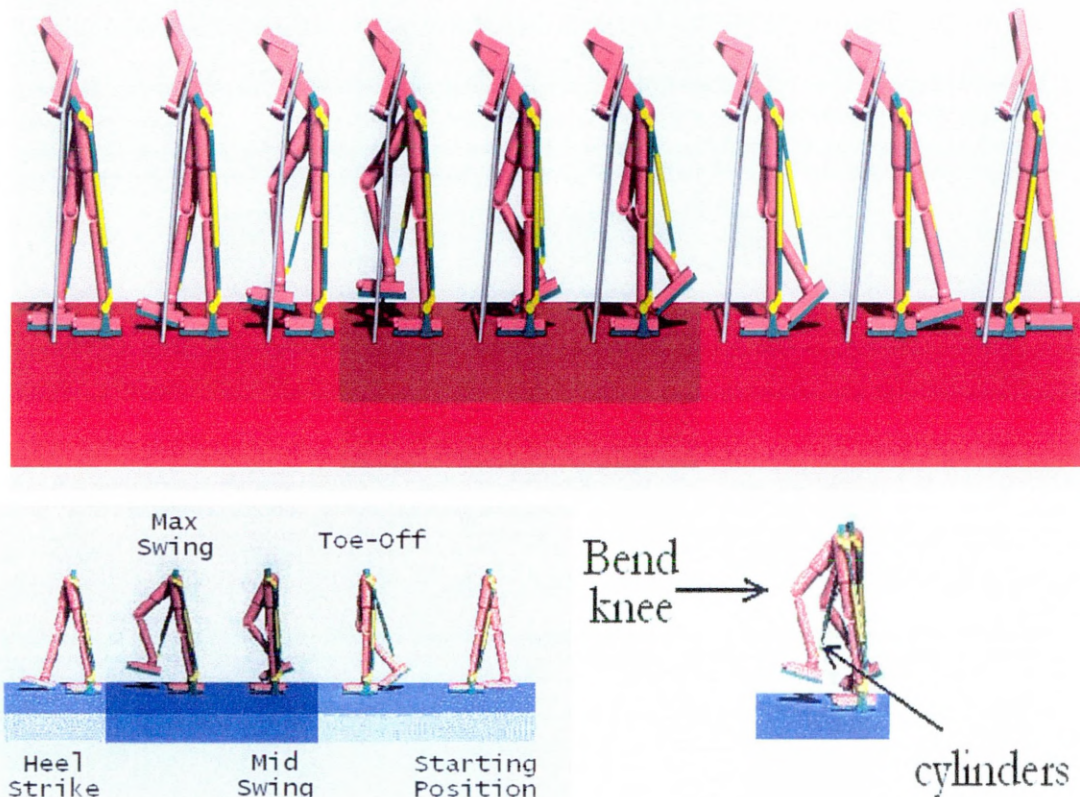


Figure 7.6 : *Maximum toe clearance for a full step*

Eventually, a significantly improved toe-clearance sequence was implemented by maintaining the leg bend during the whole swinging period (Figure 7.3 (c), 7.4 (c) and 7.5 (c)). With both cylinders fully retracted throughout the complete swing period, the toe clearance is maximised for the complete step as seen in Figure 7.6.

With toe-clearance as the main criteria, the best implementation found is a very simple one. The exoskeleton cylinders retract completely, the leg swings in a controlled fashion and when the swing reaches its maximum value, the cylinders are extended and the foot lands in a controlled way, not impacting on the ground as in normal walking, but being gently positioned it on the ground. The videos gait01.avi and gait02.avi in the Audiovisual Appendix (APAV) illustrate this positioning of the foot on the ground.

After the ankle trajectory was modelled, the trajectories of the toe and the heel were also compared to the continuous light studies to establish the best trajectory for the ankle when acting to position the foot on landing at the beginning of the double support period. Figures 7.7 and 7.8 shows the simulation of a complete set of markers for two consecutive steps. The markers are located at the hip, knee, ankle toe, heel and sole level of the mannequin. A complete animation (gait03.avi) of the simulation can be seen in the Audiovisual Appendix (APAV).

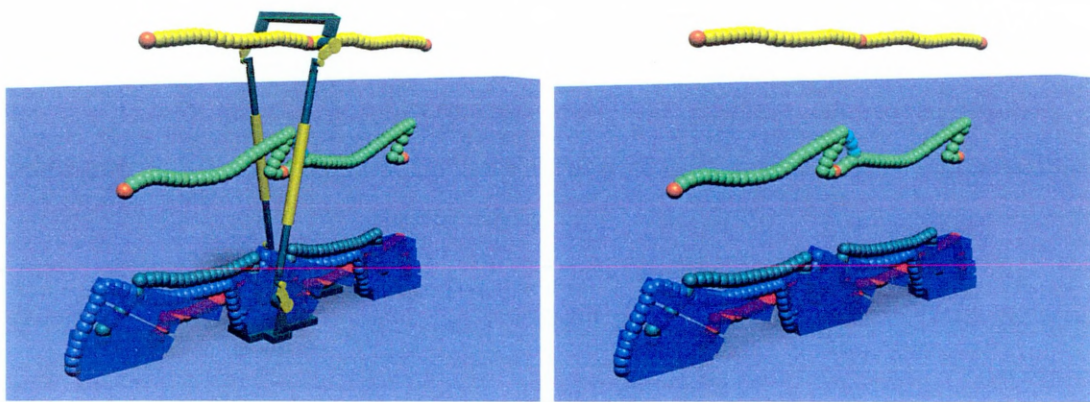


Figure 7.7 : *Simulation with exoskeleton for maximum toe clearance.*

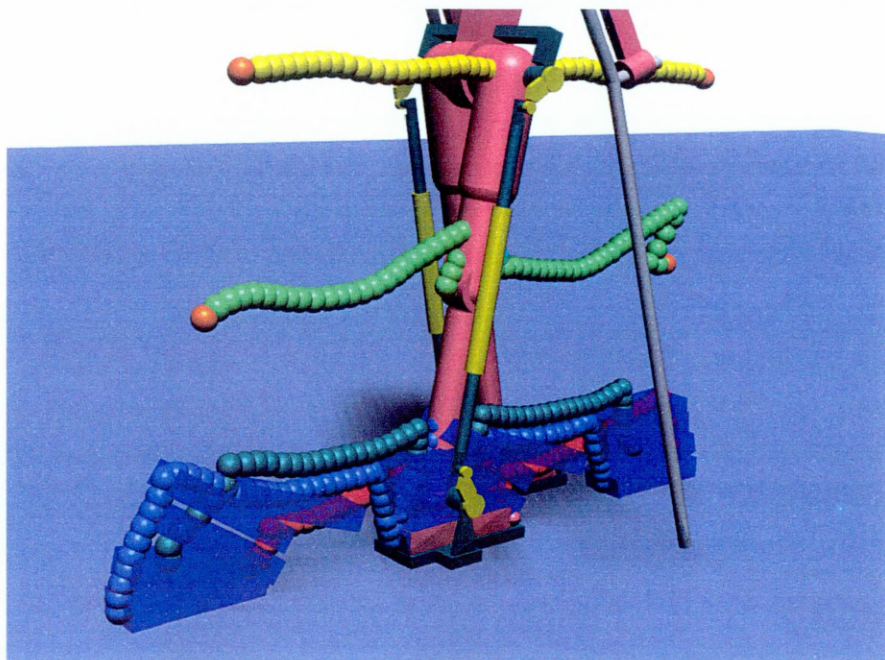


Figure 7.8 : *Simulation with mannequin for maximum toe clearance.*

For the best toe-clearance trajectory as defined, the full movement of the exoskeleton with the user deploying crutches was simulated to synthesise the trajectories seen in Figures 7.5 and 7.6. These movements were then translated to the implementation on the test bench.

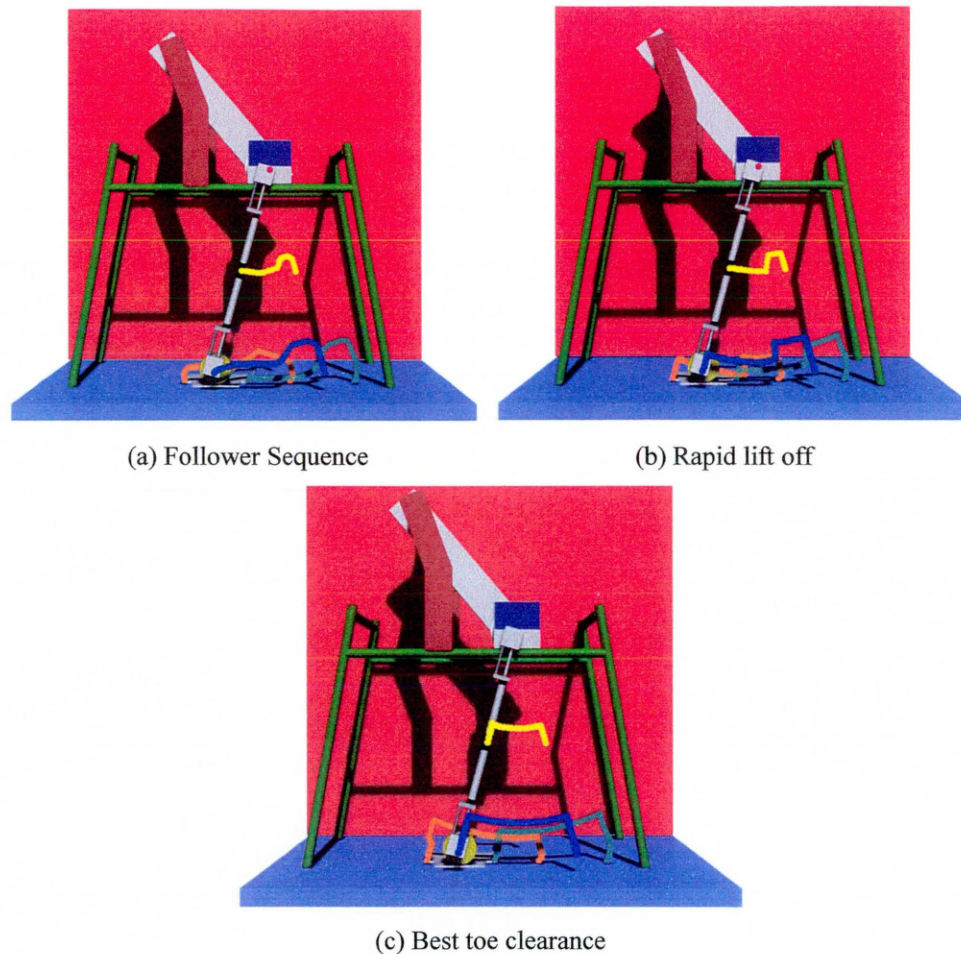


Figure 7.9 : Simulation of test bed with different gait patterns. For a more detailed explanation of the different patterns see section 7.2

The difference being that the test bench leg is fixed at the hip level to the support structure and therefore does not move, so previously defined movements have to be translated to this new fixed frame of reference for comparison purposes. The test bench was then operated with the parameters obtained for the different solutions for the different sequences (follower, rapid lift off and best toe clearance). The comparison started with the simulation of the test bed (See Figure 7.9).

The generated trajectories were then compared against light studies of subjects using crutches, employing the Muybridge technique, carried out in the laboratory. Digital markers were added in software and then traced in the video recordings while trying different gait patterns (see Figures 7.10 and 7.11). The video recordings can be seen in the Audiovisual Appendix (files gait06.avi and gait07.avi). The results can be seen in Table 7.1 where the different parameters recorded for a normal individual aided with crutches are compared with those recorded for normal individuals walking without the aid of crutches.



Figure 7.10 : *Muybridge Technique studies for normal gait subject walking with crutches, subject walking contralaterally at 30 steps /min. Cycle is commencing.*

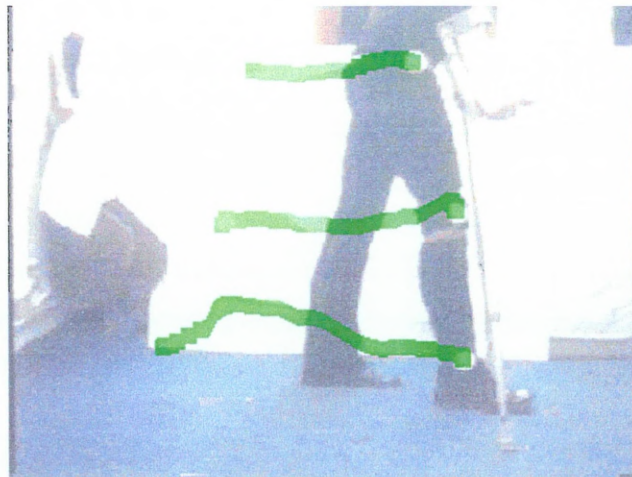
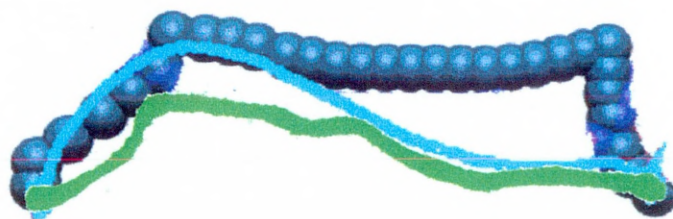


Figure 7.11 : *Muybridge Technique studies for synthesised gait, subject walking with crutches, subject walking contralaterally at 30 steps /min. Cycle has been completed*

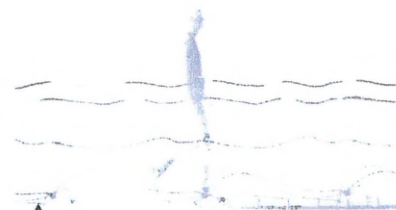
Table 7.1 shows the direct comparison in between simulated and the measured parameters with light studies for a normal individual unaided and aided by crutches. The gait patterns ensure static balance at all times and more than adequate toe clearance.

Table 7.1 *Simulated and measured comparison of light studies for a healthy individual aided and unaided by crutches*

Ankle Simulated vs Recorded :

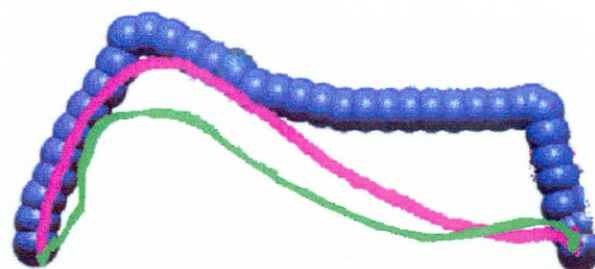


The ankle marker on the individual aided by crutches is well below the mark of the unaided on, toe-clearance for the simulated sequence is the highest of them all.

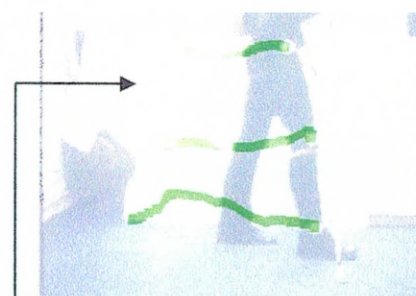


Simulated ■
Recorded Unaided ■
Recorded with Crutches ■

Heel Simulated vs Recorded :

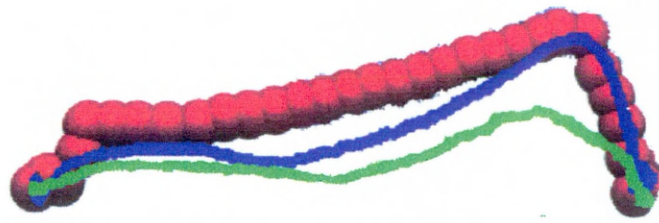


The heel marker also drops a little from the sequence of that aided by crutches to the unaided sequence, emphasising the drop in toe-clearance.

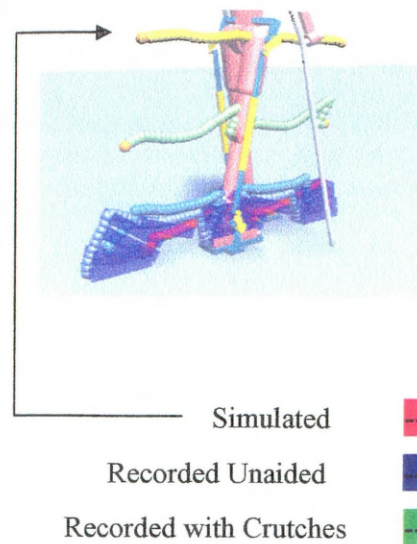


Simulated ■
Recorded Unaided ■
Recorded with Crutches ■

Toe Simulated vs Recorded :



The toe shows a very different second half of the movement where it is lacking in clearance with the floor.



Analysing each marker in Table 7.1, the first noticeable characteristic of the observed gait when assisted by crutches is that the toe clearance reduces significantly. Perhaps the fear of falling diminishes as the user feels subconsciously more secure with the extra support points. The second noticeable characteristic is an ankle sequence very similar in shape to those of the light studies, where the leg is extended as the swing approaches the landing phase. Again the difference with the simulated trajectories lies in the improved toe clearance due to the extended abduction of the leg until it achieves the maximum swing position.

The difference recorded at the knee is due to the improved toe clearance and overall the synthesised gait pattern offers the advantage of being easier to implement without a significant compromise of the movement. This simplification offers a huge advantage in terms of implementation and in terms of safety by reducing the risk of collisions with objects in the ground. Although it exhibits some clear differences from both, normal gait and gait assisted by crutches, the conclusion is that from the point of view of a bipedal machine it is easier to control and less expensive (energy consumption is reduced) to retract the cylinders and bend the leg while performing the

swing. Provided that the differences in the overall gait do not make up for a type of walk that can be rejected by the prospective users, this realisation can be interpreted as a validation of the simulated gait patterns

Although the movements look different, the visual impression from the complete simulated step with the positioning of the crutches and the progression of the gait cycle seems relatively natural (specially when compared with the gait wit crutch aid). The Audiovisual Appendix (APAV) contains the material relevant to this subject and even though cosmetically the movement of the leg seems very familiar, it is not quite exactly the same as the movement of a person walking unaided by crutches. The movement is definitively more similar to that experienced by a person aided by crutches, but with the added feature of the enhanced toe-clearance.

7.3 Gait Implementation and comparison with simulation

The validation of the simulated trajectories, in turn allowed the synthesis of this gait patterns for its implementation in the model of the leg in the test bench. After the previously discussed comparisons and analysis took place, the selected gait patterns were then loaded as trajectories were into the virtual and real test bench and again using the Muybridge technique developed during the project, the recorded movements were directly compared and analysed by superimposing one set onto the other, see Figures 7.12, 7.13, 7.14 and 7.15.

The third implementation besides contracting, swinging and extending, maintains the sole parallel to the ground at all times, by rotating the ankle, in an attempt to maximise the mid-foot clearance.

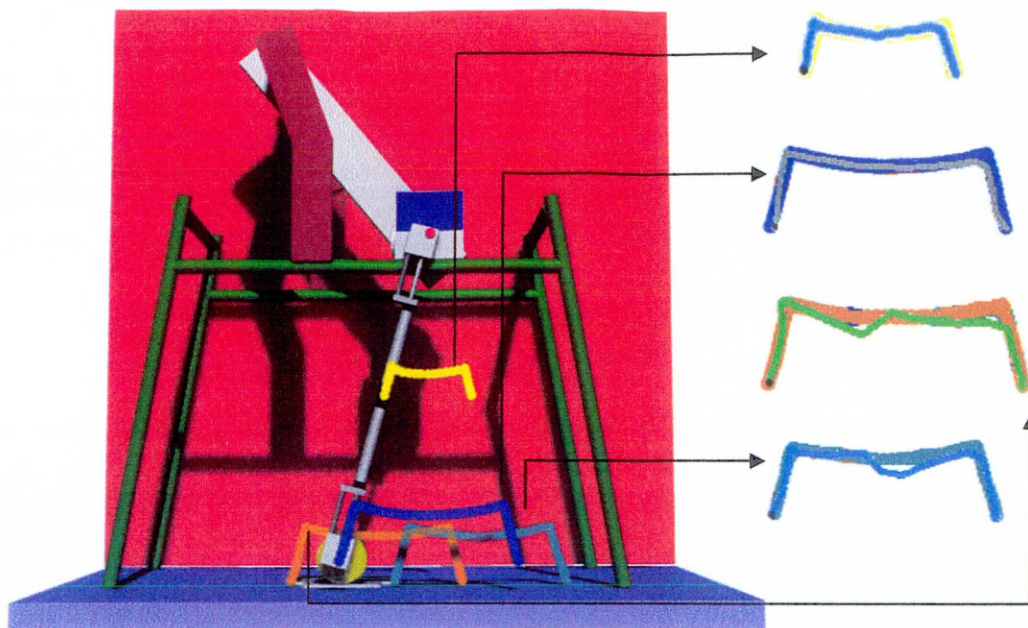


Figure 7.12 : Third implementation (sole is parallel to the ground at all times). The leg is kept contracted during the complete swing during the single support period (Simulation – gait10.avi).

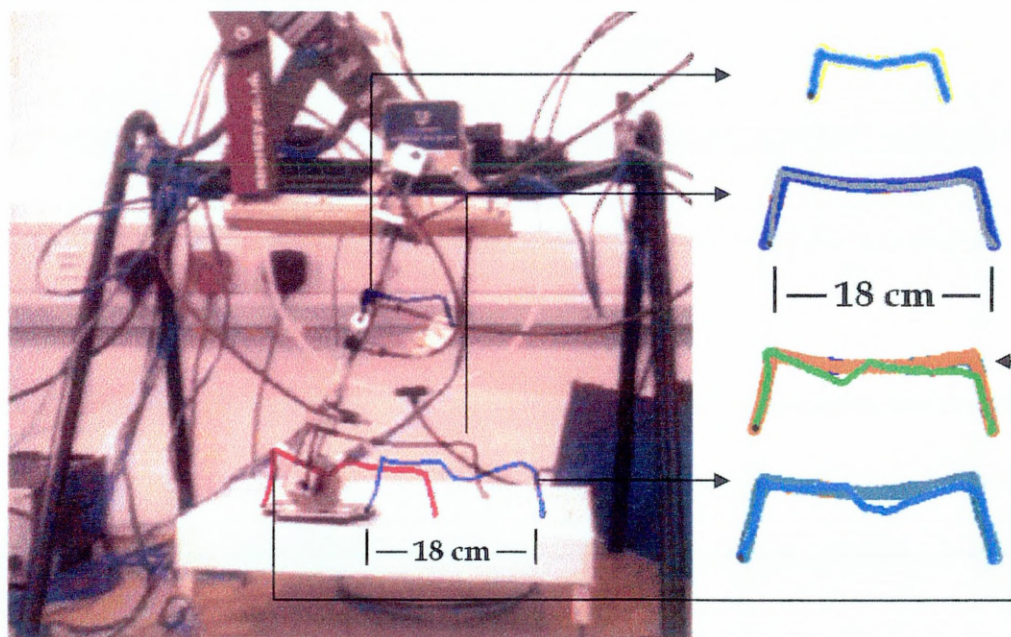


Figure 7.13 : Third implementation (sole is parallel to the ground at all times). The leg is kept contracted during the complete swing during the single support period (Test Bench – gait11.avi).

The leg in the test bench moved as expected when commanded from the control handle in the crutches.

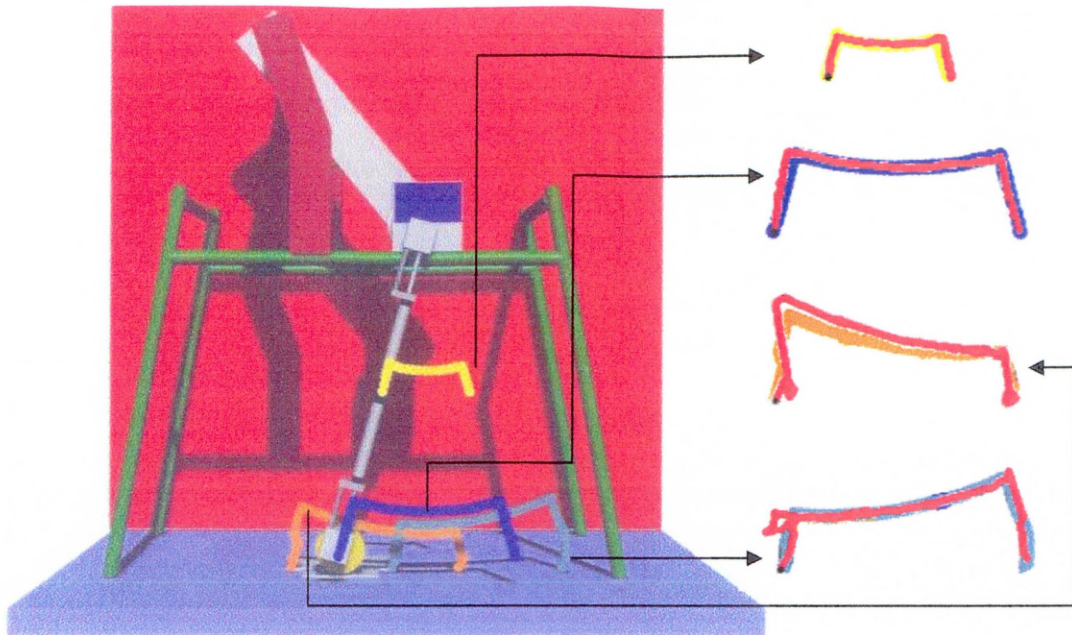


Figure 7.14 : *Fourth implementation (ankle is fixed during the swing). The leg is kept contracted during the complete swing during the single support period (Simulation - gait12.avi).*

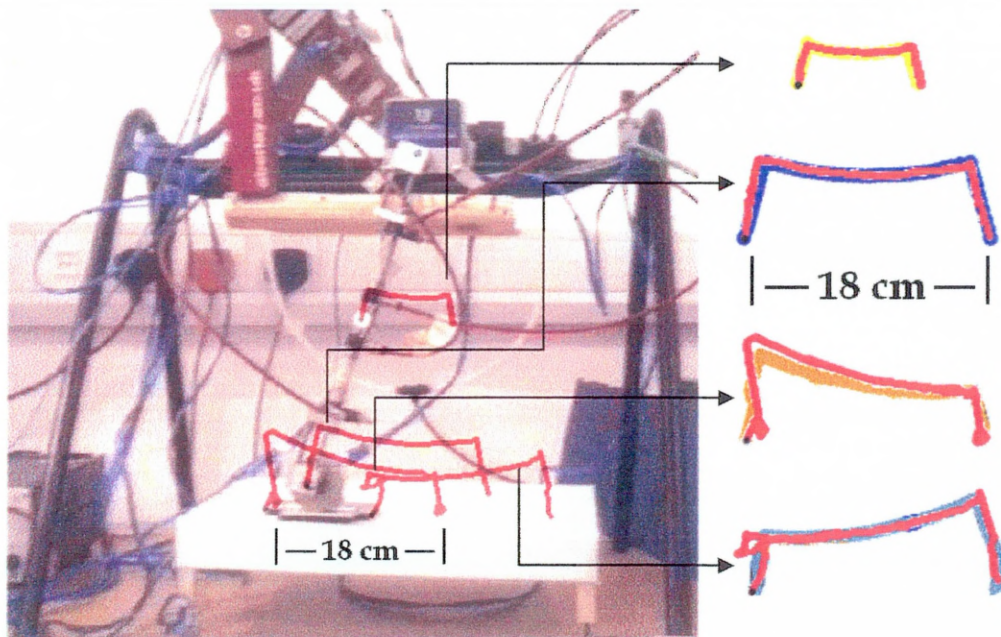


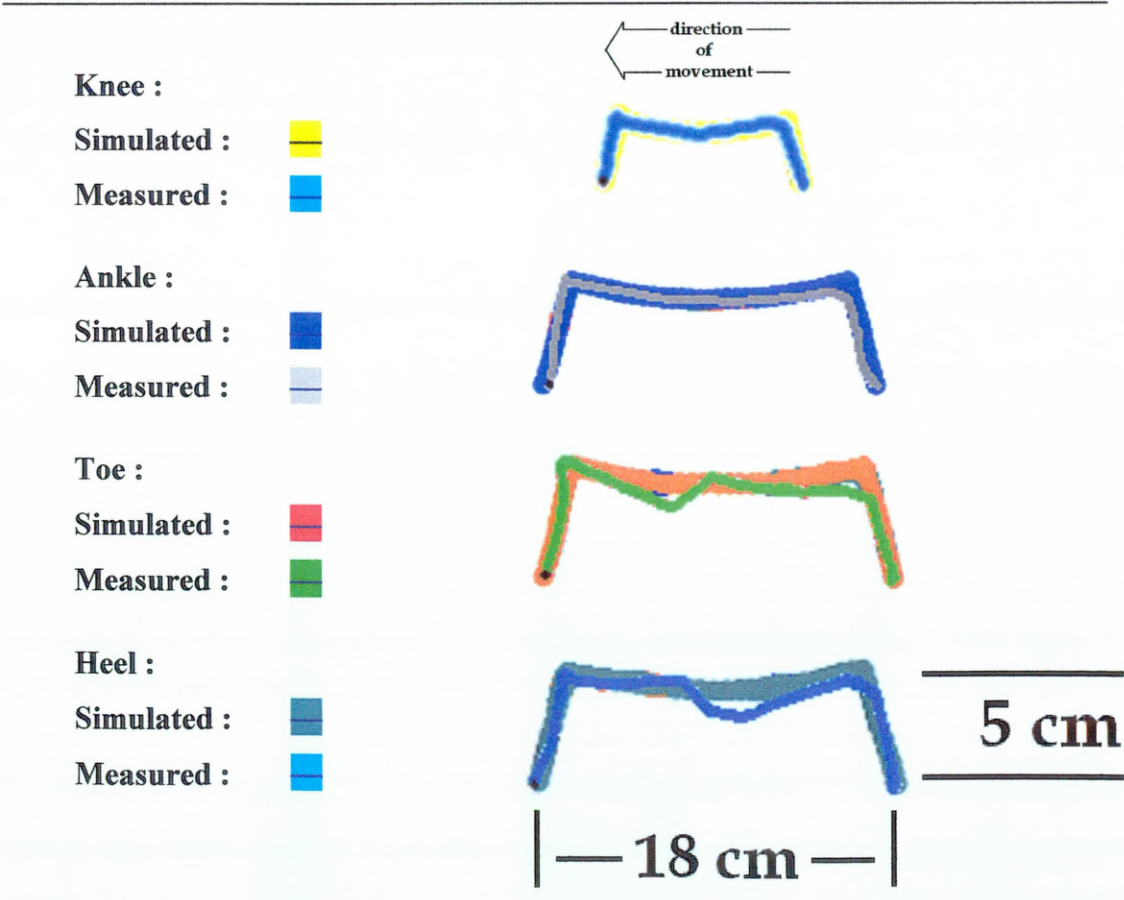
Figure 7.15 : *Fourth implementation (ankle is fixed during the swing). The leg is kept contracted during the complete swing during the single support period (Test Bench - gait13.avi).*

The fourth implementation besides contracting, swinging and extending, fixes the ankle joint in an attempt to maximise the toe-clearance at lift off.

Different setups were tried but the one depicted corresponds to the slow pace, long step.

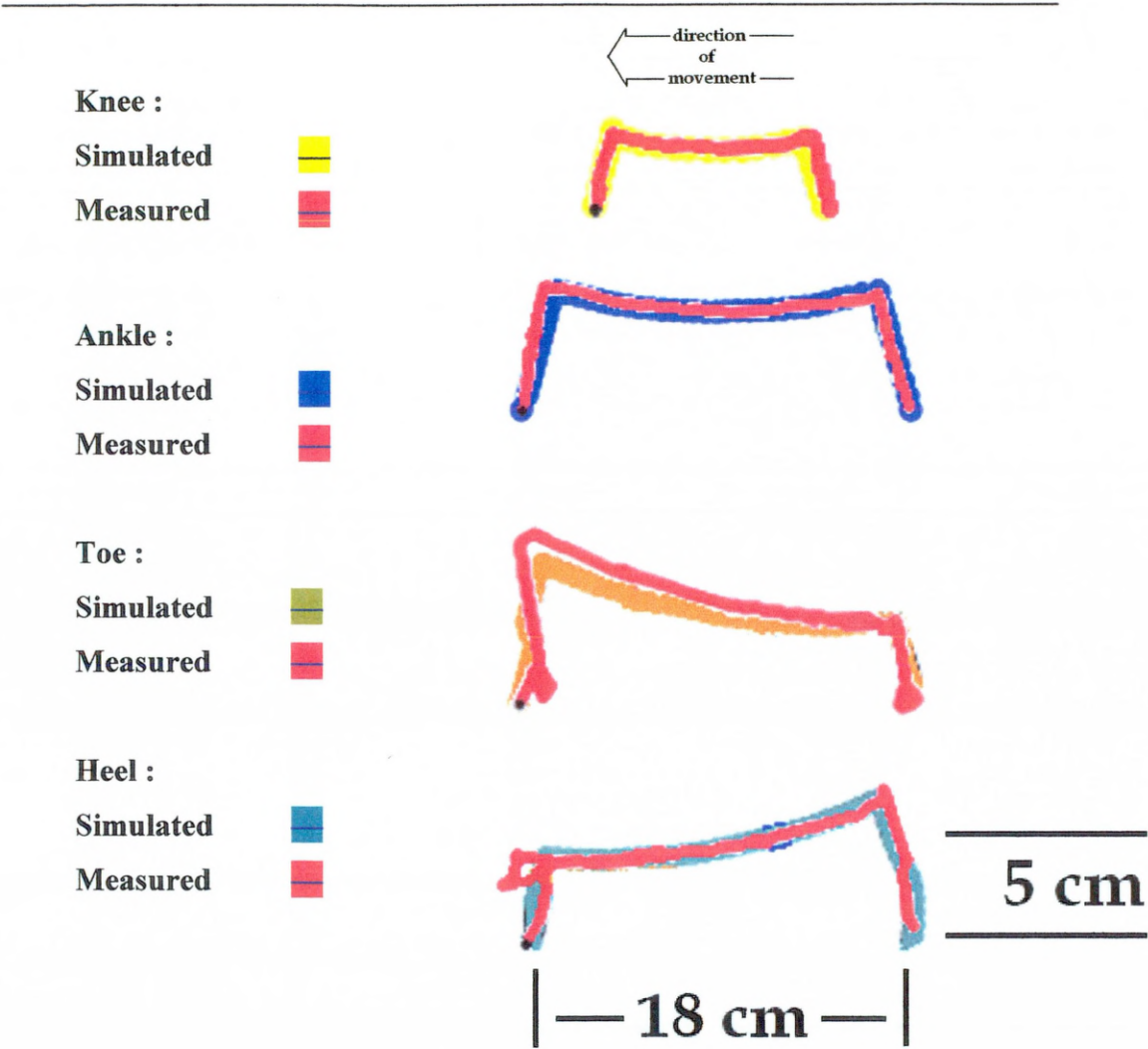
The Audiovisual Appendix (APAV) also contains the files the videos of the movements of the leg being controlled by the crutches with digital markers highlighting the position of the parameters of interest. It also contain videos of the animation of the simulation (files gait10.avi, gait11.avi, gait12.avi and gait13.avi). A full description of the angular values for each marker can be found in Appendix A, however for comparison purposes superimposition is straight forward and allows a quick detection of disparities. The superimposed sequences can be found in Tables 7.3 and 7.4.

Table 7.3 *Comparison of the Test bench Simulation vs. the measured values for the third implementation (see Figure 7.12 and 7.13)*



The superimposition recordings for every marker for the fourth implementation where the ankle is kept fixed while the swing is performed is illustrated on Table 7.4.

Table 7.4 *Comparison of the Test bench Simulation vs. the measured values for a setup with long step and full toe clearance (see Figures 7.14 and 7.15)*



7.4 Summary

Gait patterns were synthesised using the criteria established from the simulation and modelling process. The patterns were then optimised from the point of view of the practicality of the implementation as a legged robot. The trajectories were then compared to those obtained by light studies and refined so they could be safely and

easily implemented and then tried on the purposefully built (quarter size scale) test bench.

The control strategy was examined and the whole process (light studies – simulation – synthesis – test bench implementation) verified in respect of the practical considerations of it being used as a tool for aiding a person to walk in a controlled environment on an even floor and constrained to the sagittal plane. The quarter-scale test bench served its purpose of validating the assumption that the gait patterns synthesised were feasible for implementation and is encouraging for the future development a full size system.

Some aspects of the design of the structure and the control of a full size exoskeleton were addressed in the construction of the quarter-scale size model, and although many improvements can be made (proportional valves, increase in the number of DoF), it provides a good insight into what is required to build a more functional test bed and a fully functional exoskeleton.

The methodology can be applied to all sorts of (robotic aid) equipment for rehabilitation of the lower limbs, the concrete application for this project is clearly that of the walking function, but it could be used in theory for other types of rehabilitation applications based on robots.

The DVD-ROM containing the Audiovisual Appendix also contains the executable programs (PC14-*.c and PC14-*.exe) used to control the test bench with the different implemented gait patterns.

Chapter 8

SYSTEM DEVELOPMENT - NEXOS

8.1 Introduction

When considering the possibilities for the development of the project, two options were clearly identified. The first option centred around the concept of developing the design of the exoskeleton as a full size machine capable of transporting a person around a controlled space with an even surface for a fixed period of time, e.g. around a supermarket.

This option required a significant investment from either private or public funds, given its novelty and the fact that in the market there is nothing that targets the rehabilitation of the walking function. The project can be used as a proof of concept to build a full size machine with the design specifications established during the research.

This chapter provides an introduction to the development of the NeXOS project, and the early stages of the work are presented here to show how the techniques developed for the analysis of the design requirements of the exoskeleton, led to and underpinned the work of NeXOS. A full discussion of NeXOS is the subject of separate peer reviewed publications, and is not considered as part of this thesis.

8.2 The NeXOS Project

Before any proposal around the above option could be developed, a different idea began to materialise as part of a joint research programme between the University of Abertay Dundee and a partnership of Hospitals and Universities in Sheffield and Barnsley (see Appendix C).

The option considered by the team was not to rehabilitate the walking function but to perform rehabilitation tasks dealing with the movement of the leg with the exoskeleton working from a stationary position instead of while walking around.

During informal discussions of this and other ideas with other researchers with expertise based around telehealth and rehabilitation, it was decided that enough interest existed. A formal proposal was therefore submitted to the New and Emerging Applications of Technology Programme of the UK Department of Health, to develop a machine whose design takes into account the exercising of the lower limbs to restore muscle function and control as part of a rehabilitation process for individuals with lower limb injuries such as fracture or stroke.

In addition, moving the limbs passively can act to reduce pain and increase comfort for a range of progressive clinical conditions such as Multiple Sclerosis and Motor Neurone Disease. The provision of all such patient support is, in particular, consistent with the NHS National Service Framework themes for older people ^[62]. Based on the telehealth expertise developed at Abertay, the research proposal advocates the integration of World Wide Web-based strategies for the control, monitoring and provision of user feedback with the rehabilitation technology represented by the research outcomes of the exoskeleton.

The proposal aimed to take advantage of the design of the exoskeleton as established in the previous chapters to support and exercise the lower limbs of individuals with a variety of conditions and needs.

8.3 NeXOS Project Research Objectives

The provision of an intelligent rehabilitation system for the lower limbs capable of providing both motion and resistance to motion can benefit individuals with a wide range of lower limb problems. Typically, rehabilitation involves an exercise regime supported by physiotherapy to enable the patient to regain muscle function and control of the limb(s). This can involve a large number of healthcare professionals such as physiotherapists, occupational therapists and district nurses as part of the patient's support team. Discussions with such healthcare professionals suggests that the development of an effective rehabilitation regime involves regular and close contact between the patient and their support team ^[62].

The exoskeleton will be targeted for use by the patient in their own home. It will be possible to monitor the therapies, and these can be adjusted remotely by the patient's support team in consultation with the patient, without the need to physically visit the home. The system would also provide feedback on progress against defined targets and goals as treatment develops.

While the proposed system will provide support following injury and in relation to a range of neurological conditions, particular benefits are likely in respect of older people by enabling a focused exercise regime to be developed which is properly and closely matched to their individual need in a manner which is consistent with the aims of the National Service Framework for Older People ^[62].

As such the aims of the project as set out in the research proposal were:

- (1) To research the design, control and implementation of an innovative form of exoskeleton that can be used to support rehabilitation of the lower limbs as part of a telehealth environment and which incorporates an appropriate level of intelligence to enable it to autonomously adapt to patient performance;
- (2) To research the use of web-based strategies for performance monitoring by the patient's support team and for the provision of enhanced feedback to the patient on their performance in relation to agreed targets and goals;
- (3) To research the integration of web-based control and feedback strategies with video links as part of a remote rehabilitation network enabling individuals to derive support from other patients as well as from their support team;
- (4) To increase patient involvement and participation in the decision-making processes associated with their treatment.

8.4 Brief Review on Current Robotic Rehabilitation Aids

The literature review and patent search concluded that most of the Robotic Assistive Technologies developed so far deal with the upper limbs. As for instance the work by Reinkensmeyer^[66], Volpe^[67], Eftiring^[68], Rao^[69], Lum^[70], the REHAROB^[71] Project (shown in Figure 8.1 (a)) and the Boston Biomotion Domestic Robot^[72].

For the lower limbs only limited references could be established, these include systems such as the Leg Extension system produced by Monitored Rehab Systems^[74] (see figure 8.1 (b)), the Lokomat produced by Hokoma^[73] (see Figure 8.2), the TEM system by the Yaskawa Electric Co.^[75] (see Figure 8.3) and the Leg Rehabilitation System^[76] (see Figure 8.4) from Marmara University.

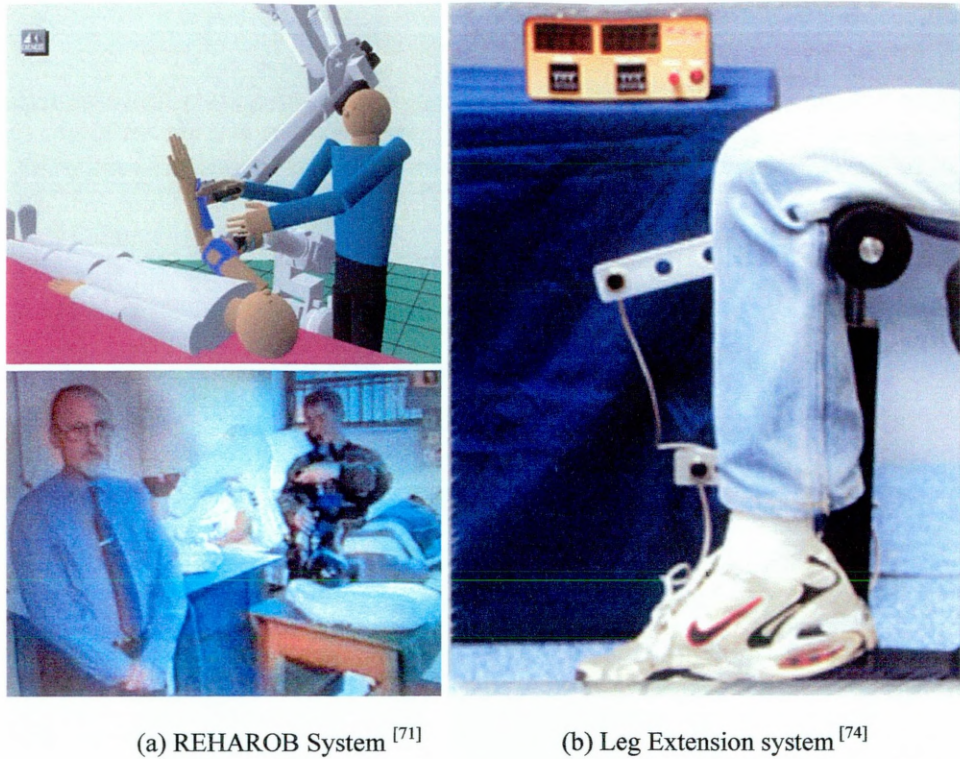


Figure 8.1 : Simulations and visualisations of the REHAROB and the Leg Extension projects.

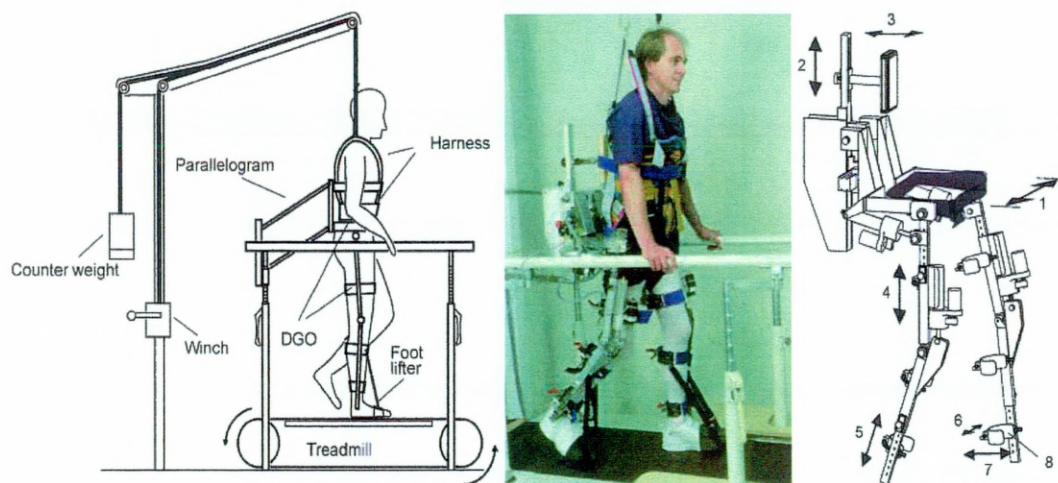


Figure 8.2 : LOKOMAT System by Hokoma, schematic and physical implementation ^[73].

Two machines in particular have similar specifications to the expected exoskeleton. One is the Therapeutic Exercise Machine (TEM) ^[75]. This has been developed as an exercise machine to decrease spasticity by repetitively performing a Range of Motion Exercise (ROM-E) for the hip and knee joints of stroke patients (see Figure 8.3) and has the closest approach to features that the NeXOS project aims to implement.

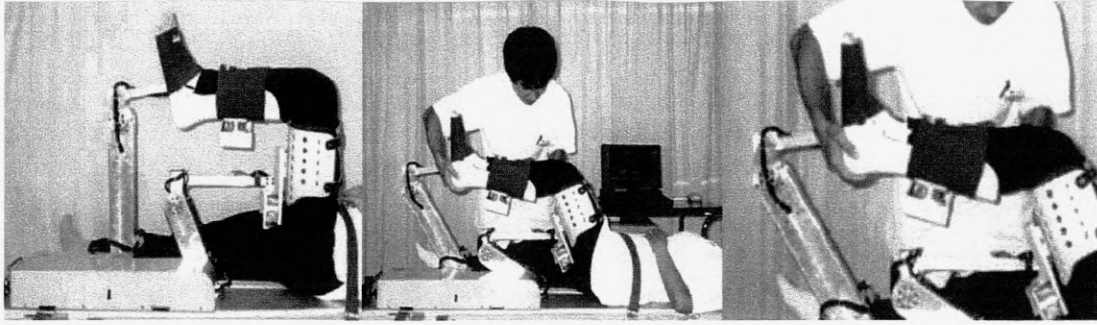


Figure 8.3 : *Therapeutic Exercise Machine*

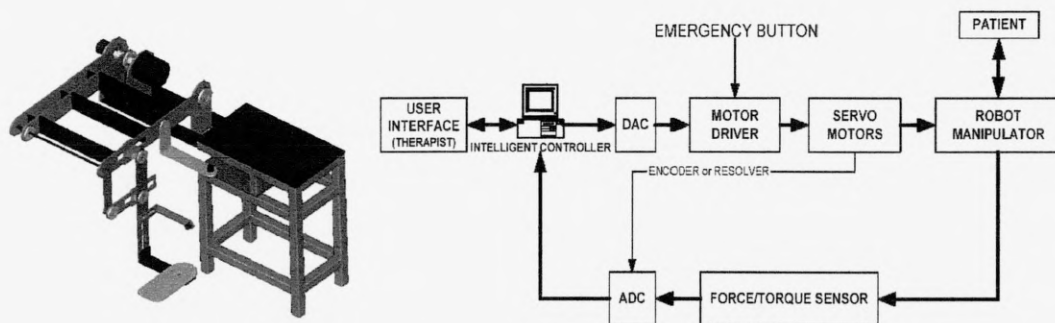


Figure 8.4 : *System visualisation and modular configuration^[76]*.

The other is a conceptual design (currently seeking funding for development) pioneered at Marmara University ^[76]. The idea of this rehabilitation aid is to replace the duties of the physiotherapist in accomplishing routine physical movements. The system works in two stages, learning and therapy (application). The controller is composed of two parts, an impedance controller (see Figure 8.4) and an expert system. The impedance control (force feedback) is very suitable for the implementation of the teach and repeat feature as it allows the extraction of the forces involved in the manipulation process performed by the physiotherapist ^[76].

While in learning mode, the system creates a database by using the data of force and position that the therapist inputs, then in therapy mode (see Figure 8.5) the system processes the information acquired in the learning mode. The robot operates with 3 DoF (see Figure 8.4) and can perform abduction-adduction and flexion-extension

movements for the hip. For the knee it can perform flexion-extension. The safety system is controlled in two stages at hardware and software level.

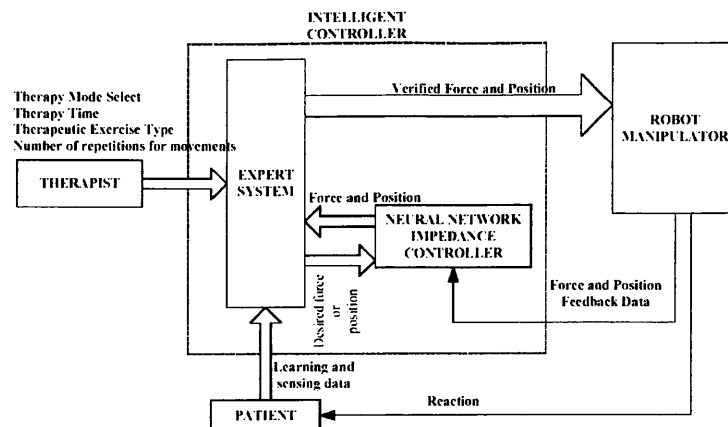


Figure 8.5 : Detailed Block Diagram of Leg Manipulator System (Therapy Mode) ^[76].

In relation to current passive therapy based on the use of automated passive movement machines, the reported research relating to the use of such machines appears to be contradictory as to whether their use is beneficial or not.

Some studies seem to suggest that it reduces hospital stay time by increasing the knee Range of Motion (ROM) more quickly than would otherwise be the case ^[77] while decreasing the need for manipulations post-operation ^[78] and the use of analgesia ^[79]. However, other studies suggest that there is no difference in hospital stay time ^[80] and some other suggests that there is increased wound drainage ^[81], decreased wound healing and increased need for analgesia ^[82]. All of these studies are concerned with purely passive motions as the machines concerned have no ability to provide a resistance to motion. There is also research that suggests that some Knee Arthroplasty (KA) patients would benefit from further rehabilitation in the home following discharge from hospital ^[83].

8.5 System Requirements

Existing systems such as the Continuous Passive Movement (CPM) machine and the TEM system ^[75] of Fig. 8.3 are essentially passive in their operation in that they act only to move the leg through a defined series of movements and require the user to exert no forces during the motion.

A key aim of the NeXOS approach is to achieve a balance of operation in which motion may range from the purely passive with the system entirely responsible for the movement of the leg to active condition where the patient would be working against the system which provides resistance to motion. It was also envisaged that combinations of assisted and resistive motion may be provided within a cycle of operation.

This meant that for the purposes of the initial evaluation it was necessary to specify prospective user groups whose needs would reflect these requirements in terms of providing and resisting motion. Specifically, there was a need to identify:

- A group whose requirement was for passive movements only but where a much greater degree of control over those movements than currently achievable would be beneficial.
- A group whose requirement was for a range of active assisted movements involving the requirement to adjust the response of the system in relation to the movements achieved. These movements would be based around the Oxford scale used by physiotherapists and set out in Table 2.1.

It was therefore decided to establish the initial investigation around patients undergoing Knee Arthroplasty (KA) and Spinal Cord Injury (SCI) patients. The reasons for this decision were:

Knee Arthroplasty - The rehabilitation of patients following Knee Arthroplasty is often based on the use of a CPM machine to provide purely passive articulation of the knee joint following surgery. As already indicated, these need to be reconfigured should the patient move during the therapy or be required to leave the machine for any purpose and, being purely passive, they do not allow for the exercising of the muscles.



Figure 8.6 : Typical exercise performed in the sagittal plane for SCI subjects showing tracks for ankle and knee ^[89].

Spinal Cord Injury - Spinal Cord Injury patients undergo passive motion therapy to help to maintain a basic conditioning of the lower limbs as illustrated in Fig. 8.6 where a physiotherapist is working with a member of the project team. The therapy requires that the physiotherapist move the limb through a sequence of motions while providing support for the limb. A specific problem that is not addressed by this form of therapy is the observed loss of Bone Mineral Density (BMD) leading to an increased risk of fractures ^[3] and is something to which consideration is being given.

Based on the above it becomes possible to establish some requirements for the exoskeletal system in that it should:

1. Enable a variety of motions and motion types as determined by the physiotherapist in conjunction with the user.
2. Enable the rapid reconfiguration of the system to encompass different forms of motion based around a range of basic defined motions.
3. Enable the monitoring and control of the forces applied along the axis of the major bones in the lower leg in order to try to alleviate the loss of BMD.
4. Support a rapid and automatic set-up procedure for any individual patient based on knowledge of that patient's physical dimensions. This would enable patients to set the machine up themselves rather than requiring assistance, enabling its use in the home environment.
5. Autonomously adjust, within definable limits, to any change of patient position during the therapy process.
6. Monitor the forces and motions exerted achieved by the system and the patient throughout a cycle and autonomously adjust these to maintain parameters such as force, velocity and power within agreed and defined limits.
7. Provide motivation to patients through feedback on their performance against agreed norms and by allowing them to assume some degree of control over the rehabilitation process. This could include a dialogue with the machine to establish a baseline of relevant activity prior to using the machine, with the work programme then being adjusted accordingly.
8. Provide for 'jerk free' transitions and operation throughout the cycle.

8.5.1 White Space

There is also a need to establish the barriers therapists and users have in relation to conducting home based practice and to address these in a user centred manner. In the case of the physiotherapists, focus groups identified the requirement for an interactive procedure for setting up and configuring the system with some form of *teach and repeat* procedure in which the system recorded and then played back the actions of the physiotherapist seeming preferable. It was also considered by the physiotherapists that, at least initially, a system such as that which was being proposed would not be used remotely. This was in part due to worries over safety, but was also associated with the concern expressed by the therapists over their perceived need to maintain a direct contact with the patient during the treatment.

There was also identified the need to establish an appropriate, and easily reconfigurable, interface which will enable the user to have a degree of direct control over the operation of the system. For instance in respect of their ability to terminate the operation of the machine in case of discomfort. There is also the requirement to establish mechanisms by which the system can respond to direct measures of patient performance. For example, there is the possibility that passive motion therapy may induce spasticity in some patients ^[84], the onset of which must be detected by the system, which must then respond accordingly.

8.6 Definition of Range of Motion

It is necessary to define the range of motion for the exoskeleton to establish a geometry that can deliver the required movements. To do this, as with the methodology followed with the bipedal exoskeleton, the first step was to gain

knowledge on the different types of movements and exercises needed and establish if movement in the sagittal plane is sufficient or if movement in other planes is required.

Table 8.1 shows the different types of exercises performed in a routine rehabilitation therapy ^[76] :

Table 8.1 *Types of exercises in terms of performed activity*

<i>Passive</i>	Performed for the patient by another person or an exercise device
<i>Resistive</i>	Performed by the patient against a device exercise or manually controlled by the physiotherapist.
<i>Active Assisted</i>	Assistance is provided to the patient as the ability to produce some active movement is developed.
<i>Active Static</i>	No Change in the joint angle occurs when acting against a fixed resistance.
<i>Active Kinetic</i>	The resistance is displaced through a range of motion.
<i>Active Isokinetic</i>	The resistance is applied at a constant speed by constantly checking the muscle effort.

Once the movements were selected and video markers traced the trajectories of interest for the definition of the range of motion, a leg model was developed and the ankle trajectories of the different selected exercises were reproduced and use to refine the final definition of the work space as shown in Figure 8.7 and 8.8.

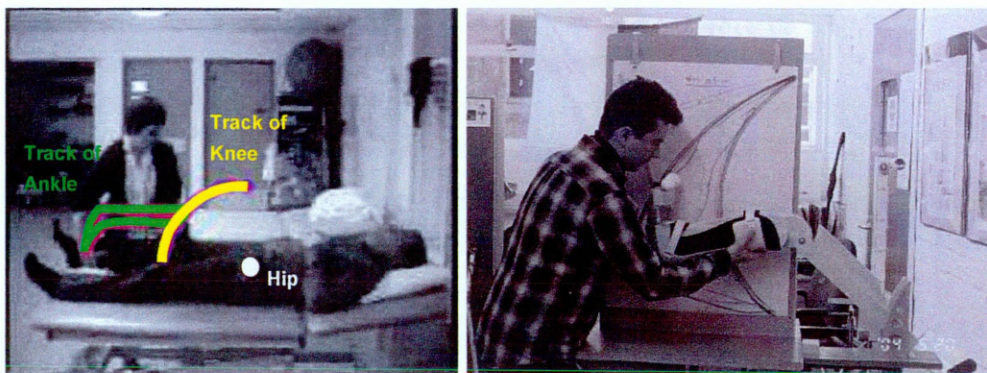


Figure 8.7 : Ankle tracks for typical exercises performed in the sagittal plane for SCI subjects.

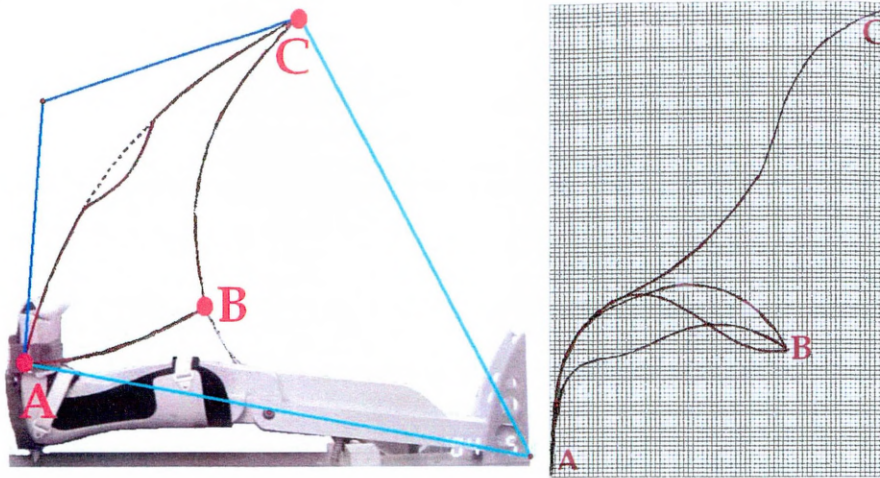


Figure 8.8 : Leg model and ankle track generation in the sagittal plane for different exercises for SCI individuals.

The decision to define the geometry marks the cut off point in between how the PhD led to the NeXOS project and the subsequent independent evolution of the design and implementation process. This process is presented in the next section for clarification purposes as indicated in the introduction of the chapter and, as explained before, beyond the scope of this dissertation .

8.7 System Implementation

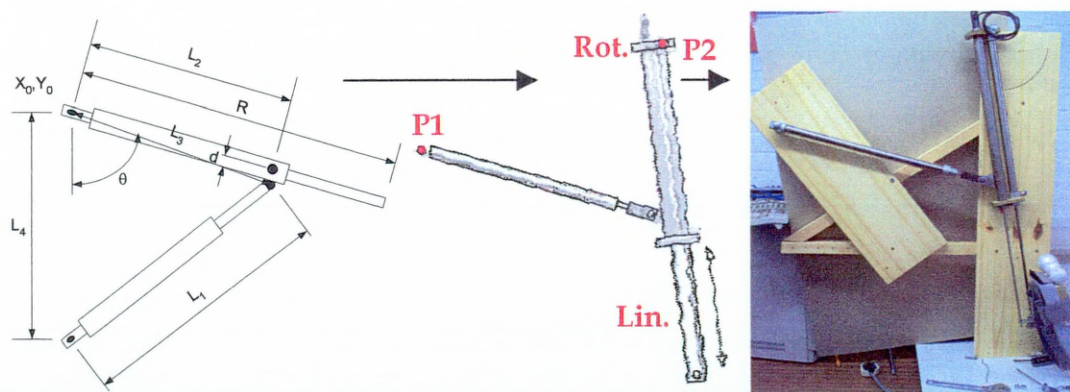


Figure 8.9 : Geometrical model, mathematical implementation and prototype mechanism.

[†] It is not intended to raise a challenge regarding the intellectual property rights of the NeXOS project by disclosing this development process. The commercialisation unit of Abertay University agreed to publicly disclose research material in August 2006.

The chosen geometry for the mechanism covers the pre-defined range of motion for exercises performed in the sagittal plane. Figure 8.9 shows the transition in between the mathematical model, the conceptual model and the implemented prototype.

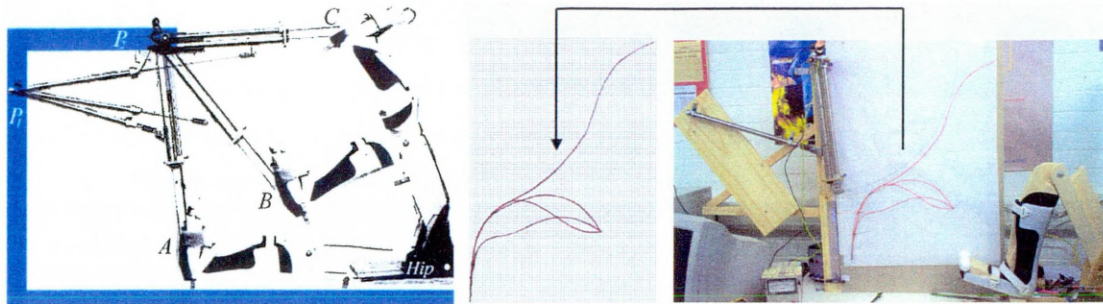


Figure 8.10 : *Prototype mechanism workspace.*

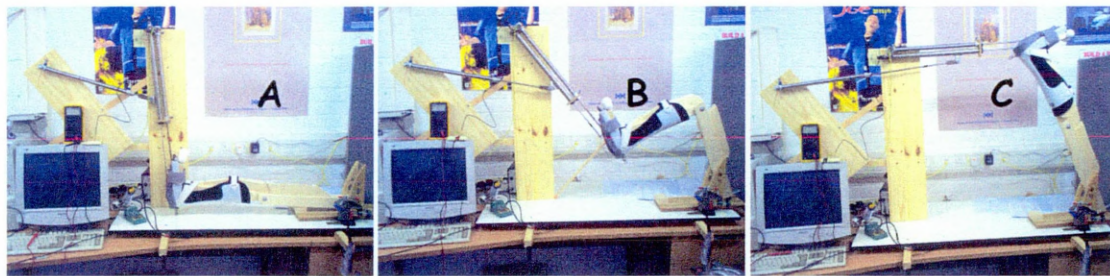


Figure 8.11 : *Range of motions for prototype mechanism.*

Figures 8.10 and 8.11 depict the different range of motions exhibited by the prototype mechanism and compared to the expected profiled movements.



Figure 8.12 : *Prototype mechanism mounted on the test rig.*

Once the prototype mechanism was mounted on the test rig it was shipped to the test facility in Sheffield, where it was used to record different movement profiles resulting from different types of leg manipulation performed by the physiotherapists. Figure 8.13 shows a physiotherapist performing the movements and the DAQ system recording the sensors data.

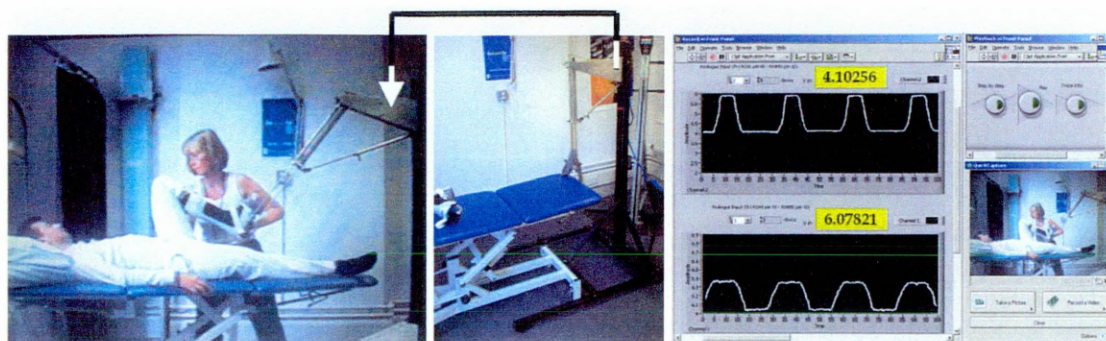


Figure 8.13 : *Physiotherapy performing leg manipulations and DAQ system.*

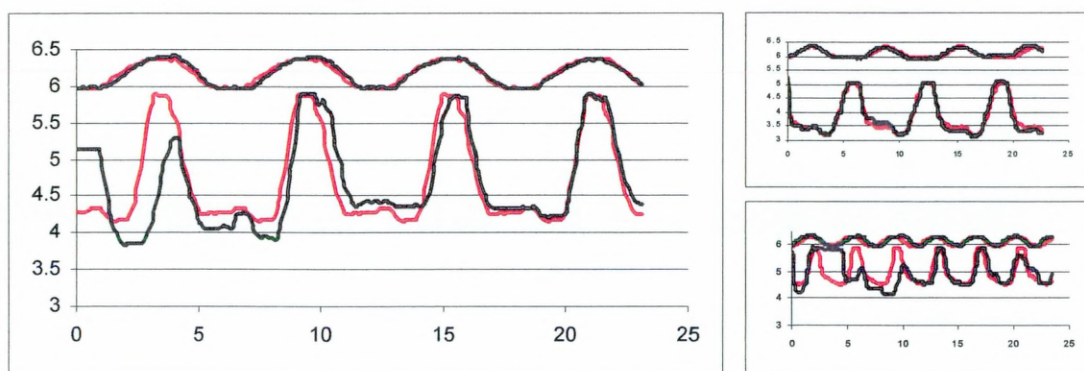


Figure 8.14 : *Recorded and average data set for different physiotherapists.*

The recorded data (shown in Figure 8.14 in black) was then analysed and its cyclic nature used to develop an average measurement intended to replicate the manipulation movement performed by the physiotherapist (shown in Figure 8.14 in red). This average signal was then used to replicate the movement and a Mark II prototype was build and pneumatically powered in order to test the teach and repeat features of the rig as shown in Figure 8.15. Figure 8.16 shows in addition to the recorded data (shown in black) and the average data (shown in red) the implemented data for the replicated movements (shown in blue).

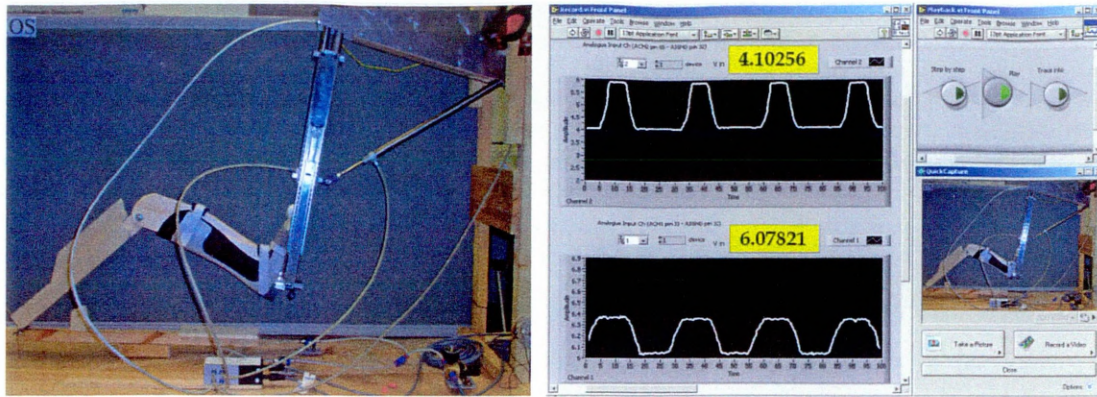


Figure 8.15 : Mark II prototype for teach and repeat tests.

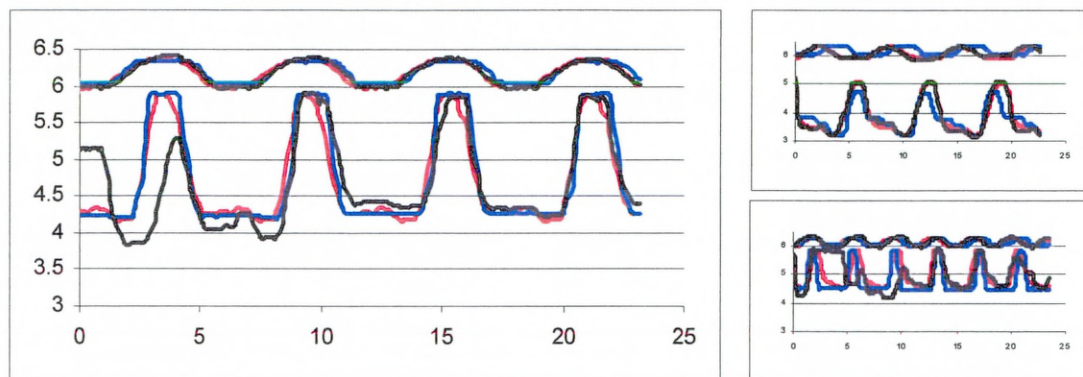


Figure 8.16 : Recorded, average and implemented data set for different physiotherapists.

8.8 Summary

The NeXOS project is an integrated and cross-disciplinary project which involves a range of subject specialists coming together to develop and implement a new form of rehabilitation system. In doing so, conventional engineering design and systems analysis methods are being combined with other forms of analysis to enable the system concepts to be integrated with stakeholder need.

As part of this process, extensive use has been made of system modelling and visualisation tools to support the communication of ideas between the project team and to stakeholders through focus groups.

Following on from the analysis and modelling stages, the developed rig was used to evaluate the different control and operational strategies implemented, as well as to finalise the kinematic geometry of the exoskeleton. Use will be made of web-based strategies as the basis of both local and remote communications with the exoskeleton and with stakeholders.

Finally seven peer reviewed articles ^{[89] [90] [91] [92] [93] [94] [99]} were published describing the design aspects of the NeXOS project, the full text of these papers can be found on the Appendix C.

Chapter 9

CONCLUSIONS

9.1 Synopsis

In this thesis a novel solution for the rehabilitation of the walking function for individuals with lower limb impairments was considered and analysed.

The objective of this thesis was to investigate the design and development of a novel gait assistant exoskeleton for the rehabilitation of the walking function in individuals with lower limb impairments (primarily Spinal Cord Injury) as a replacement for a wheelchair and as an alternative to Functional Electro Stimulation.

One of the main objectives was to construct a quarter-size model of the simulated concept, this model was implemented in the form of a test bed, in which different experiments were carried out in order to obtain the criteria required for the determination of the chosen geometrical configuration and gait synthesis. Although a full-size exoskeleton was not implemented, a therapeutic implementation was favoured instead whereby the exoskeleton design procedure is employed to investigate applications for robots in the field of rehabilitation of the lower limbs. Nonetheless, a very in depth analysis of the issues surrounding the requirements for a full size exoskeleton was undertaken and the criteria required for such design was formalised.

The initial assumption of an exoskeleton able to replace wheelchairs as the main stream aid for mobility in the target population had to be completely revised and changed to a different approach in which a wheelchair contains the exoskeleton that can detach itself and allow a short stroll in the vicinity of the wheelchair. The wheelchair would then contain the bulk of the energy storage and anything else that is not absolutely necessary for the exoskeleton in order to minimise the weight of the

structure. Incorporating an exoskeleton into daily life activities, requires careful consideration of user comfort when long-term use is a priority.

This conclusion is comparable to that offered by the experience of other groups that have developed other exoskeletons, in every single referenced case the implemented configurations have remained constrained to clinical environments. The real problem had always lain in the application for the exoskeleton. In principle, not a single designed examined took into account the possibility of incorporating the exoskeleton with the wheelchair and make full use of the extended potential that such a configuration could offer. That was a costly mistake for every project that was undertaken in the 1970's.

The other clear mistake is the definition of the appropriate task to be performed by the exoskeleton. As established before an exoskeleton can not match (with the current technology) the functionality of a wheelchair, specially a powered wheelchair, this is mainly due to the practicality of the combination of electric power and efficiency of the use of wheels. But these other projects have failed to take into account that there is no need to completely eliminate the use of wheelchairs.

Seeing the definition of requirements from a user point of view and not from a technology lead one, a potential user requires to stand up and walk in very limited scenarios, i.e. basically where a wheelchair is either difficult to take to, for example if the user requires access to the upper level of a building with no lift and whose solely access is through a set of stairs, the user could in theory detach from the wheelchair and climb the required set of stairs performed the activity that led him to that

particular location and return to the wheelchair. Since many buildings do cater for individuals with wheelchairs requirements, this action will only be required to increase the range of autonomy and independence of the user.

Another possible scenario, is that of a social gathering where most people attending are standing up, the user could detach from the wheelchair in order to engage in conversations at eye level, again with the purpose in mind of increasing the independence and self-confidence of the individual using this technology.

As far as determining successful configurations for the previously established requirements, different types of geometrical setups were considered and the telescopic knee was preferred for its practicality, robustness and simplicity. For this geometrical configuration, different models were constructed to establish the true potential of the design. The mathematical models, the 3D models and the quarter-scale model all proved the feasibility of implementing such a design in a form that can render gait trajectories of remarkable similarity to those of normal individuals and of a great level of robustness and efficiency in the light of energy consumption.

At the moment the project has benefited from the injection of resources received as it has transformed from a tool for the rehabilitation of the walking function into a tool for rehabilitation in physiotherapy for illnesses in a different group of the population. Most likely the next transformation will be one that see the exoskeleton becoming a tool for research as well as for teaching physiotherapy to new apprentices of this demanding profession. As the nature of research is to embark upon a journey of discovery where the implementation of a technology developed with a specific

purpose in mind can be moulded to suit the requirements of the markets and the users of the technology being developed, this is seen as a logical and sensible development.

As the exoskeleton continues its journey from one research stage to the next, it may be valid to say that the potential of an idea, of a question and the technological solutions it creates may be the most compelling argument for continuing the pursuit of ever more demanding questions and novel ideas.

Many years ago when the possibility of a project like this was just an idea, the thought of a device that could deliver the same performance as an ideal functional electro stimulation system, but made of a structure composed of actuators that could be controlled in a more efficient way than current FES systems, was very remote. However while writing the conclusions the underlying core concept, the hypothesis remains the same. Such a device could deliver a rehabilitated walking function to wheelchair users and other individuals with lower limb impairments.

Nevertheless, from the moment this idea took form until the moment these conclusions were written, there has been plenty of analysis and useful conclusions for the implementation of a solution to such a problem. Clearly the link between bipedal walking machines and rehabilitation of the walking function has to be the target application of this technology to produce a purpose designed legged machine that can act as a gait assistance device for individuals with certain classes of lower limb impairments.

9.2 Functionality

Taking into account each one of the five basic design requirements (Function, Comfort, Cosmetic Appearance, Trustworthiness, Cost/Benefit Analysis) for the successful implementation of the gait assistant, the first element to be analysed relates to the functional requirements, they can be related to the three different activities developed with the exoskeleton.

9.2.1 Standing

The exoskeleton provides stability to the body preventing it from falling or collapsing. Each leg of the exoskeleton consists of 3 DoF in the form of a rotational hip, a telescopic knee ^[16] and a rotational ankle. As the exoskeleton is maintained in static balance while standing there is virtually no energy consumption in this position.

9.2.2 Walking

When walking the exoskeleton carries out the cyclical joint movements required in the sagittal plane for the forward progression of the body maintaining static balance at all times.

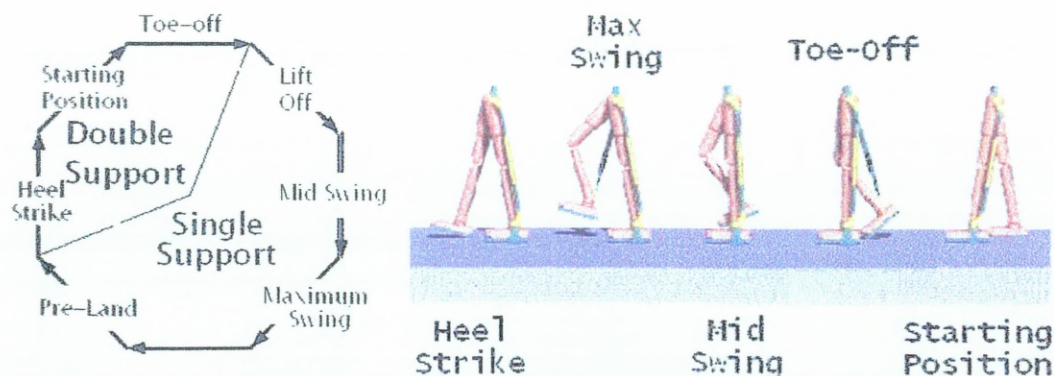


Figure 9.1 : Sequential description of the movements required for the gait

The sequence for a single step based on the combination of the exoskeleton and the crutches is then as follows:

- I. *Detect stable position – State₀*
- II. *Operator initiates the step sequence using controls on the crutches to set step height, step length and rate – State₂.*
- III. *The exoskeleton initiates the step – State₃, State₄ and State₅.*
- IV. *Step completed – State₆.*
- V. *Reposition crutches for next step.*
- VI. *Return to I.*

9.2.3 Transition

The exoskeleton design facilitate the joint movements in the sagittal plane required for vertical displacement of the body with respect to the feet touching the ground. This activity *Standing Up and Sitting Down* activity is performed with the aid of the wheelchair. If the exoskeleton is to stand up from a sit down position on a resting place other than the base wheelchair, external aid is required to perform that transition.

The transitions in between the wheelchair and the gait assistant device can be singled out as the most important factor for the success of the long term acceptance of the gait assistant. To achieve this goal, the wheelchair has to be designed as a cradle that accommodates the exoskeleton and the exoskeleton has to become an integral part of the wheelchair that can be detached for a short walk, the power supply required for the actuators to perform is required to be onboard the exoskeleton.

9.3 User Comfort

Contact between the body and the gait assistant is minimised thus avoiding high local pressure points and its associated problems (irritation of the skin and perspiration problems ^[1]) because the exoskeleton use is limited and the user relies more in the wheelchair. Donning and doffing is not more complicated than that experienced with a normal powered wheelchair, this ensures that the level of independence achieved is comparable to that of wheelchair users.

As the exoskeleton is incorporated into the wheelchair it can be transported wherever the user goes. Since the idea is to allow for a short walk then problems with the contact points in between the exoskeleton and the skin of the user can be minimised. The user can incorporate the exoskeleton into their everyday activities since the use of the wheelchair will remain basically the same.

9.4 Cosmetic Appearance

The aesthetic appearance of the exoskeleton is completely linked to that of the wheelchair. Although the cosmetic appearance of the gait assistant is a very important factor to consider in public social gatherings, the fact that the user has the possibility to switch from a sitting down position to a standing up position is of tremendous value as social interaction is sometimes limited due to the pre-conceptions to which wheelchair users can be subjected.

The gait aid is designed in such a way that it can be made to look inconspicuous as it can be worn under clothes. The geometrical design and size of the actuators is such that this can be achieved.

9.5 Trustworthiness

The gait assistant, as it has been described, poses very minimal risks for the user. The simplicity of its movements make it very reliable and as static balance has been incorporated into the design, safety issues surrounding falling down can be minimised. Since the user is in control of the crutch positioning and step sequence it increases the user's confidence in its own abilities to manipulate the exoskeleton. The gait pattern selected minimises the risk of tripping over obstacles as toe-clearance was top in the design considerations

9.6 Cost/Benefit Analysis

The benefits resulting from the rehabilitation of the walking function are difficult to quantify, and would have to be assessed as part of a treatment or therapy. At the moment there is very little in the way of treatment that offers any hope of a complete rehabilitation of the walking function. There is no equipment that allows a wheelchair user to detach from it and use a powered walking assistive device in order to walk indoors and outdoors.

The synergy in between the exoskeleton extends its autonomy, although the gait assistant does not cater for extended periods of autonomous walking. Considering that there are not commercially available devices that can deliver the improvement that will doubtless be obtained from the ability to stand up and interact with other people at eye level and not from the height predetermined by the wheelchair (which is especially true in many social situations). The possibility that at least a certain amount of independence (albeit relatively little) from the wheelchair is attainable is certainly better than nothing at all, and also very difficult to quantify, but in terms of

components the exoskeleton can be estimated to cost (£2500 - £3000) a small percentage of what an electric wheelchair capable of climbing stairs does cost (£15000 - £20000). The real cost is associated with the process to make the device compliant to the regulations for medical equipment. This is a pre-requisite for any device that qualifies as medical equipment prescribed (and therefore paid for) by the appropriate health insurance bodies.

9.7 Safety Analysis

The main safety aspect involves the user falling down when walking, due to the proposed static balance, the most likely event that could cause it, is the user lifting up one or both crutches while the stepping process is taking place. Because the system evaluates if the static balance criteria of three points of support is effectively achieved just before allowing the step to be carried out, once the step is being performed the system assumes that the condition will be maintained throughout the sequence, if the user removes the crutches they will fall down. The system relies on the user actually not wanting to fall down and exercising a moderate amount of caution.

9.8 Final Discussion

If a robot with the characteristics of a powered exoskeleton able to completely rehabilitate the walking function with total independence from the wheelchair is someday completed, certain other technologies will have to evolve to the standard required for such a design.

Developments in other fields such as the military augmentation devices can deliver technology solutions perfectly suited for a project like this. The relentless

miniaturisation of digital based controllers delivers everyday cheaper, smaller and more practical solutions for the control requirements of the exoskeleton and the associated wheelchair.

More important perhaps, are advances in insurance policies and health services everywhere in the world encouraging an increased market that will help boost the attainment of the resources required to further the project to the state where a user can slip into the exoskeleton and walk about in a safe and practical manner.

The ideas circumscribed within the frame of a technology prescription can also be exploited as aid robots can be tailored to the exact requirements of specific individuals suffering from a well defined pathology instead of designing robotic aids tailored for mass consumption in huge populations suffering from a similar illness.

The issue of clinical tests and approval for a medical device was not considered in the project and if mass solutions are to be achieved this has to be taken onboard, unless specific tailor - made solutions are adopted in the long term.

For the moment the technologies developed as part of the current NeXOS project are very helpful as the way the project is setup, they can be recycled back into the initial concept. The knowledge gained from the use of LabView and the use of proportional valves can all give feedback into the evolution of the implemented test bed. Perhaps more important than all of that is the fact that NeXOS has proven the initial concept to be on the right path, tackling issues in rehabilitation robotics that deserve the consideration of funding bodies willing to support these types of realisation.

The technology and procedures developed for the synthesis of the gait patterns and the analysis of geometries, has been adopted for NeXOS producing excellent results without the need to make extra developments or new models. This is because the same models used for the exoskeleton can be recycled into the modelling of the different aspects analysed within the context of the requirements established by the physiotherapists.

9.9 Associated Publications

Five peer reviewed articles with details of the exoskeleton have been published ^[95] ^[96] ^[97] ^[98] ^[100] with some more planned for submission. The full text papers can be found in Appendix C along with the abstracts of those ready for submission. Seven more peer reviewed published articles of the NeXOS project ^[89] ^[90] ^[91] ^[92] ^[93] ^[94] ^[99] can also be found in this appendix.

EPILOGUE

Personal Reflections

Perhaps every research student, when reflecting back on the development of their work, can confidently say how differently things could have been done and where the focus and main impetus should have lain. It could be assumed that the final result of becoming a professional researcher is to understand the limits of what can be done with the variables of time and resources available for that purpose. Perhaps trying to convince the whole scientific community of how good researchers we were when embarking on the first year of the research is somehow irrelevant compared to how good researchers we have become as a result of what research can teach a postgraduate student.

Very clearly the type of researcher that emerges from such a research programme is more measured and perhaps less over-optimistic and usually more realistic regarding what can be achieved. The importance of publishing and how to deal with the pressures of the shortcomings of the funding system is another very clear lesson that can only be learned through an iterative process of writing and submitting.

The importance of keeping a good log book of the experiments, developments and relevant information to the project is painfully clear when trying to remember events and information that occurred years ago.

The way researchers analyse things is not a mere coincidence, after writing so much, a certain clarity can be found in the way ideas, concepts and explanations are formulated, formatted and delivered to a reader. In the formation of research groups the necessity for mixing research students and postdocs is clear. However, the differences between the two are ever more apparent as although students are cheaper, their results are slower in developing while postdocs can deliver straight away.

More relevant than anything is perhaps the role of the supervisor in deciding what type of researcher a postgraduate student will become. Even the writing style is greatly influenced by the mentors of generations of students who have become professional researchers. The research programme is also a turning point that defines clearly what type of life can be expected when complete dedication to science is required.

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APPENDIXES

APPENDIX A

Complete derivation for set of equations 5.1 - 5.5. The equations for direct $\mathbf{p} = \mathbf{h}(\mathbf{q})$ and inverse $\mathbf{q} = \mathbf{h}^{-1}(\mathbf{p})$, kinematics can be derived as follows

$$\mathbf{p} = \begin{bmatrix} x_0 \\ z_0 \\ \mu_0 \\ x_2 \\ z_2 \\ \mu_2 \end{bmatrix} = \begin{bmatrix} p_0 \\ p_1 \\ p_2 \\ p_3 \\ p_4 \\ p_5 \end{bmatrix} = \begin{bmatrix} x_1 - d_1 \cdot \text{Sen}(\mu_1 + \beta_1) \\ z_1 + d_1 \cdot \text{Cos}(\mu_1 + \beta_1) \\ \mu_1 + \beta_1 + \beta_3 \\ x_0 - d_2 \cdot \text{Sen}(\mu_0 + \beta_2) \\ z_0 - d_2 \cdot \text{Cos}(\mu_0 + \beta_2) \\ \mu_0 + \beta_2 + \beta_4 \end{bmatrix} \quad \text{Eq. 5.1}$$

$$\mathbf{q} = \begin{bmatrix} \beta_1 \\ d_1 \\ \beta_3 \\ \beta_2 \\ d_2 \\ \beta_4 \end{bmatrix} = \begin{bmatrix} q_0 \\ q_1 \\ q_2 \\ q_3 \\ q_4 \\ q_5 \end{bmatrix} \quad \text{Eq. 5.2}$$

$$\dot{\mathbf{p}} = \mathbf{J} \cdot \dot{\mathbf{q}} \quad \text{Eq. 5.3}$$

$$\mathbf{J} = \begin{bmatrix} \frac{\partial p}{\partial q_0} & \frac{\partial p}{\partial q_1} & \dots & \frac{\partial p}{\partial q_5} \end{bmatrix} \quad \text{Eq. 5.4}$$

$$\mathbf{J} = \begin{bmatrix} \frac{\partial p}{\partial \beta_1} & \frac{\partial p}{\partial d_1} & \frac{\partial p}{\partial \beta_3} & \frac{\partial p}{\partial \beta_2} & \frac{\partial p}{\partial d_2} & \frac{\partial p}{\partial \beta_4} \end{bmatrix}$$

Each one of the six columns in the Jacobian matrix is developed as follows :

$$\frac{\partial p}{\partial \beta_1} = \begin{bmatrix} \frac{\partial x_0}{\partial \beta_1} = -d_1 \cdot \text{Cos}(\mu_1 + \beta_1) = z_1 - z_0 \\ \frac{\partial z_0}{\partial \beta_1} = -d_1 \cdot \text{Sen}(\mu_1 + \beta_1) = x_0 - x_1 \\ \frac{\partial \mu_0}{\partial \beta_1} = 1 \\ \frac{\partial x_2}{\partial \beta_1} = \frac{\partial x_0}{\partial \beta_1} - d_2 \cdot \text{Cos}(\mu_0 + \beta_2) = z_1 - z_2 \\ \frac{\partial z_2}{\partial \beta_1} = \frac{\partial z_0}{\partial \beta_1} + d_2 \cdot \text{Sen}(\mu_0 + \beta_2) = 2 \cdot x_0 - (x_1 + x_2) \\ \frac{\partial \mu_2}{\partial \beta_1} = 1 \end{bmatrix} = \begin{bmatrix} z_1 - z_0 \\ x_0 - x_1 \\ 1 \\ z_1 - z_2 \\ 2 \cdot x_0 - (x_1 + x_2) \\ 1 \end{bmatrix}$$

Eq. 5.4.1

$$\frac{\partial p}{\partial d_1} = \begin{bmatrix} \frac{\partial x_0}{\partial d_1} = -\text{Sen}(\mu_1 + \beta_1) \\ \frac{\partial z_0}{\partial d_1} = \text{Cos}(\mu_1 + \beta_1) \\ \frac{\partial \mu_0}{\partial d_1} = 0 \\ \frac{\partial x_2}{\partial d_1} = \frac{\partial x_0}{\partial d_1} = -\text{Sen}(\mu_1 + \beta_1) \\ \frac{\partial z_2}{\partial d_1} = \frac{\partial z_0}{\partial d_1} = \text{Cos}(\mu_1 + \beta_1) \\ \frac{\partial \mu_2}{\partial d_1} = 0 \end{bmatrix} = \begin{bmatrix} -\text{Sen}(\mu_1 + \beta_1) \\ \text{Cos}(\mu_1 + \beta_1) \\ 0 \\ -\text{Sen}(\mu_1 + \beta_1) \\ \text{Cos}(\mu_1 + \beta_1) \\ 0 \end{bmatrix}$$

Eq. 5.4.2

$$\frac{\partial p}{\partial \beta_3} = \begin{bmatrix} \frac{\partial x_0}{\partial \beta_3} = 0 \\ \frac{\partial z_0}{\partial \beta_3} = 0 \\ \frac{\partial \mu_0}{\partial \beta_3} = 1 \\ \frac{\partial x_2}{\partial \beta_3} = -d_2 \cdot \text{Cos}(\mu_0 + \beta_2) = z_2 - z_0 \\ \frac{\partial z_2}{\partial \beta_3} = -d_2 \cdot \text{Sen}(\mu_0 + \beta_2) = x_0 - x_2 \\ \frac{\partial \mu_2}{\partial \beta_3} = 1 \end{bmatrix} = \begin{bmatrix} 0 \\ 0 \\ 1 \\ z_2 - z_0 \\ x_0 - x_2 \\ 1 \end{bmatrix} \quad \text{Eq. 5.4.3}$$

$$\frac{\partial p}{\partial \beta_2} = \begin{bmatrix} \frac{\partial x_0}{\partial \beta_2} = 0 \\ \frac{\partial z_0}{\partial \beta_2} = 0 \\ \frac{\partial \mu_0}{\partial \beta_2} = 0 \\ \frac{\partial x_2}{\partial \beta_2} = -d_2 \cdot \text{Cos}(\mu_0 + \beta_2) = z_2 - z_0 \\ \frac{\partial z_2}{\partial \beta_2} = d_2 \cdot \text{Sen}(\mu_0 + \beta_2) = x_0 - x_2 \\ \frac{\partial \mu_2}{\partial \beta_2} = 1 \end{bmatrix} = \begin{bmatrix} 0 \\ 0 \\ 0 \\ z_2 - z_0 \\ x_0 - x_2 \\ 1 \end{bmatrix} \quad \text{Eq. 5.4.4}$$

$$\frac{\partial p}{\partial d_2} = \begin{bmatrix} \frac{\partial x_0}{\partial d_2} = 0 \\ \frac{\partial z_0}{\partial d_2} = 0 \\ \frac{\partial \mu_0}{\partial d_2} = 0 \\ \frac{\partial x_2}{\partial d_2} = -\text{Sen}(\mu_0 + \beta_2) \\ \frac{\partial z_2}{\partial d_2} = -\text{Cos}(\mu_0 + \beta_2) \\ \frac{\partial \mu_2}{\partial d_2} = 0 \end{bmatrix} = \begin{bmatrix} 0 \\ 0 \\ 0 \\ -\text{Sen}(\mu_0 + \beta_2) \\ -\text{Cos}(\mu_0 + \beta_2) \\ 0 \end{bmatrix} \quad \text{Eq. 5.4.5}$$

$$\frac{\partial p}{\partial \beta_4} = \begin{bmatrix} \frac{\partial x_0}{\partial \beta_4} = 0 \\ \frac{\partial z_0}{\partial \beta_4} = 0 \\ \frac{\partial \mu_0}{\partial \beta_4} = 0 \\ \frac{\partial x_2}{\partial \beta_4} = 0 \\ \frac{\partial z_2}{\partial \beta_4} = 0 \\ \frac{\partial \mu_2}{\partial \beta_4} = 1 \end{bmatrix} = \begin{bmatrix} 0 \\ 0 \\ 0 \\ 0 \\ 0 \\ 1 \end{bmatrix} \quad \text{Eq. 5.4.6}$$

This set of columns corresponds to each one in the Jacobian shown in equation 5.5

$$J = \begin{bmatrix} \begin{pmatrix} z_1 - z_0 \\ x_0 - x_1 \\ 1 \\ z_1 - z_2 \\ 2 \cdot x_0 - (x_1 + x_2) \\ 1 \end{pmatrix} & \begin{pmatrix} -\text{Sen}(\mu_1 + \beta_1) \\ \text{Cos}(\mu_1 + \beta_1) \\ 0 \\ -\text{Sen}(\mu_1 + \beta_1) \\ \text{Cos}(\mu_1 + \beta_1) \\ 0 \end{pmatrix} & \begin{pmatrix} 0 \\ 0 \\ 1 \\ z_2 - z_0 \\ x_0 - x_2 \\ 1 \end{pmatrix} & \begin{pmatrix} 0 \\ 0 \\ 0 \\ z_2 - z_0 \\ x_0 - x_2 \\ 1 \end{pmatrix} & \begin{pmatrix} 0 \\ 0 \\ 0 \\ -\text{Sen}(\mu_0 + \beta_2) \\ -\text{Cos}(\mu_0 + \beta_2) \\ 0 \end{pmatrix} & \begin{pmatrix} 0 \\ 0 \\ 0 \\ 0 \\ 0 \\ 1 \end{pmatrix} \end{bmatrix}$$

Eq. 5.5

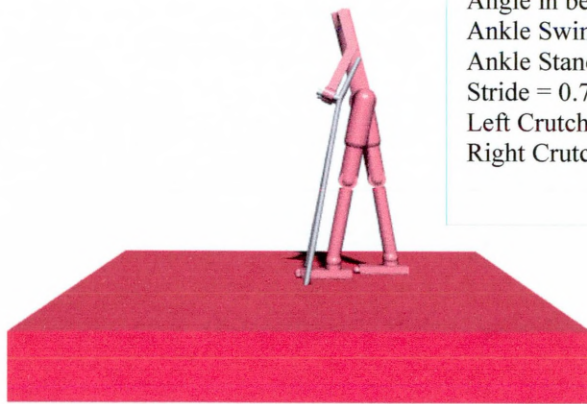
Table APA.1 contains the values for the parameters of the simulation of the model of the mannequin with the crutches for the sequence of a full step, this table corresponds to the values of Table 5.2.

Table APA.1

Starting position :

Stance leg is in front of swing leg.

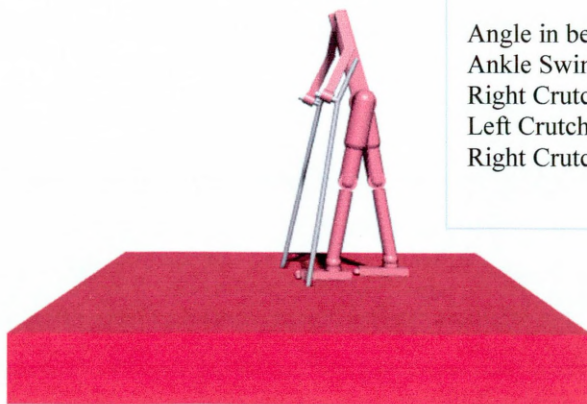
Angle in between legs $\alpha = 20.323^\circ$
 Ankle Swing Leg = (0.466, 0.416, 0.0975)
 Ankle Stance Leg = (0.724, 0.570, 0.0985)
 Stride = $0.724 - 0.466 = 25.8\text{cm}$
 Left Crutch Tip = (0.846, 0.805)
 Right Crutch Tip = (0.842, 0.180)



Swing Leg crutch's positioning :

Both Feet are flat on the ground. Right crutch is positioned ready for next step.

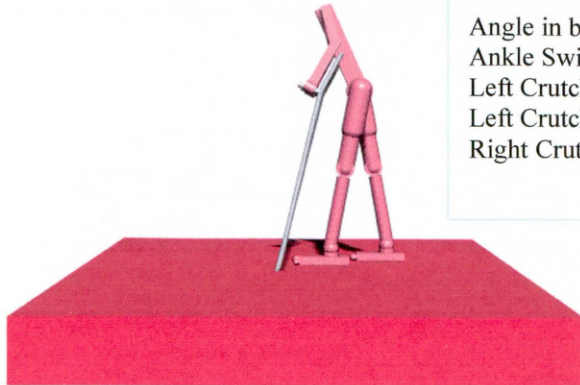
Angle in between legs $\alpha = 20.323^\circ$
 Ankle Swing Leg = (0.466, 0.416, 0.0975)
 Right Crutch Jump = $0.96 - 0.842 = 11.8\text{cm}$
 Left Crutch Tip = (0.846, 0.805)
 Right Crutch Tip = (0.960, 0.195)



Stance Leg crutch's positioning :

Both Feet are flat on the ground. Left crutch is positioned ready for next step.

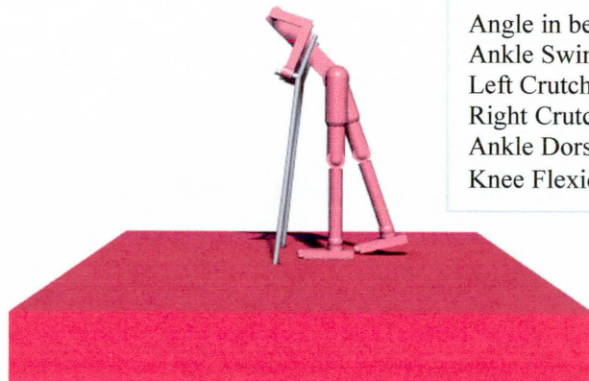
Angle in between legs $\alpha = 20.323^\circ$
Ankle Swing Leg = (0.466, 0.416, 0.0975)
Left Crutch Jump = $0.983 - 0.846 = 13.7\text{cm}$
Left Crutch Tip = (0.983, 0.770)
Right Crutch Tip = (0.960, 0.195)



Preparation for Toe – Off :

Swing Foot is not flat on the ground, toe still is in contact with the ground.

Angle in between legs $\alpha = 20.323^\circ$
Ankle Swing Leg = (0.489, 0.416, 0.137)
Left Crutch Tip = (0.983, 0.770)
Right Crutch Tip = (0.960, 0.195)
Ankle Dorsiflexion = 3.6°
Knee Flexion Angle $\phi = 180.0^\circ$



Toe – Off :

Swing Foot is not about to lift off, but still is in contact with the ground.

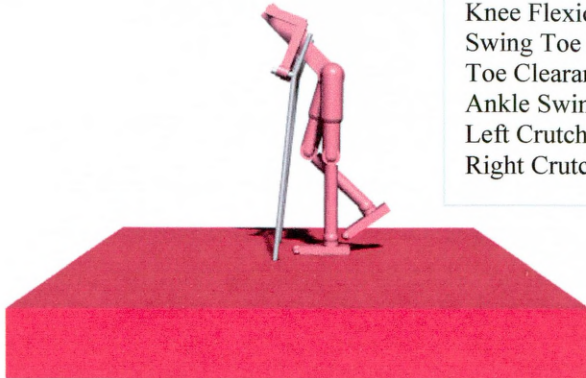
Knee Flexion Angle $\phi = 135.79^\circ$
Ankle Swing Leg = (0.542, 0.416, 0.175)
Left Crutch Tip = (0.983, 0.770)
Right Crutch Tip = (0.960, 0.195)



Lift – Off

Swing Foot Lifts Off the maximal knee flexion value.

Knee Flexion Angle $\phi = 117.36^\circ$
Swing Toe Z Position = 0.0932
Toe Clearance = 5.32cm
Ankle Swing Leg = (0.562, 0.416, 0.223)
Left Crutch Tip = (0.983, 0.770)
Right Crutch Tip = (0.960, 0.195)



Mid-Swing

Swing Foot is airborne going through the swinging sequence

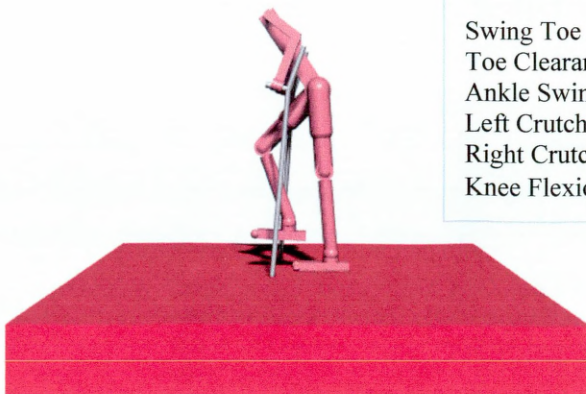
Swing Toe Z Position = 0.103
Toe Clearance = 6.3cm
Ankle Swing Leg = (0.669, 0.416, 0.204)
Left Crutch Tip = (0.983, 0.770)
Right Crutch Tip = (0.960, 0.195)
Knee Flexion Angle $\phi = 117.36^\circ$



Maximal Swing Position

Swing Leg achieves its maximal rotational value and prepares for landing.

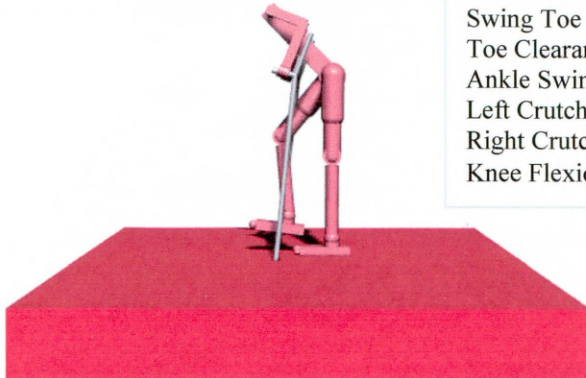
Swing Toe Z Position = 0.165
Toe Clearance = 12.5cm
Ankle Swing Leg = (0.910, 0.416, 0.220)
Left Crutch Tip = (0.983, 0.770)
Right Crutch Tip = (0.960, 0.195)
Knee Flexion Angle $\phi = 117.36^\circ$



Pre-Landing

Swing Leg achieves its maximal rotational value and prepares for landing.

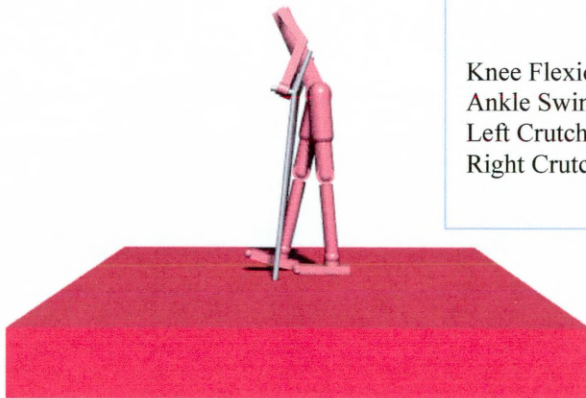
Swing Toe Z Position = 0.118
Toe Clearance = 7.8cm
Ankle Swing Leg = (0.914, 0.416, 0.166)
Left Crutch Tip = (0.983, 0.770)
Right Crutch Tip = (0.960, 0.195)
Knee Flexion Angle $\phi = 134.46^\circ$



Heel Strike

Swing Leg achieves its maximal rotational value and prepares for landing.

Knee Flexion Angle $\phi = 180.0^\circ$
Ankle Swing Leg = (0.937, 0.416, 0.114)
Left Crutch Tip = (0.983, 0.770)
Right Crutch Tip = (0.960, 0.195)



Starting position

Swing Leg achieves its maximal rotational value and prepares for landing.

Ankle Stance Leg = (0.725, 0.57, 0.0985)
Ankle Swing Leg = (0.956, 0.416, 0.0987)
Step Length = 23cm
Left Crutch Tip = (0.983, 0.770)
Right Crutch Tip = (0.960, 0.195)



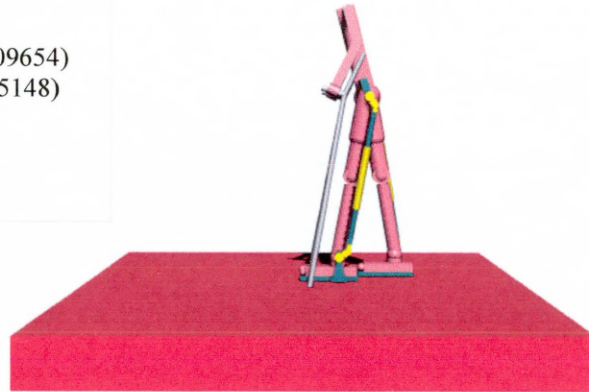
Table APA.2 contains the values for the parameters of the simulation of exoskeleton moving the model of the mannequin with the crutches for the sequences of a full step, this table corresponds to the values of Table 5.3.

Table APA.2

Starting position :

Stance leg is in front of swing leg.
Cylinders are fully extend.

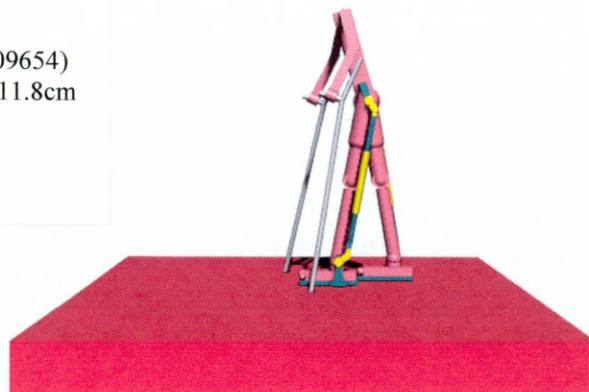
Angle in between legs $\alpha = 20.323^\circ$
 Ankle Swing Leg = (0.46677, y, 0.09654)
 Knee Swing Leg = (0.52987, y, 0.45148)
 Stride = 0.724-0.466 = 25.731cm
 Left Crutch Tip = (0.846, 0.805)
 Right Crutch Tip = (0.842, 0.180)



Swing Leg crutch's positioning :

Both Feet are flat on the ground. Right crutch is positioned ready for next step.

Angle in between legs $\alpha = 20.323^\circ$
 Ankle Swing Leg = (0.46677, y, 0.09654)
 Right Crutch Jump = 0.96-0.842 = 11.8cm
 Left Crutch Tip = (0.846, 0.805)
 Right Crutch Tip = (0.960, 0.195)



Stance Leg crutch's positioning :

Both Feet are flat on the ground. Left crutch is positioned ready for next step.

Angle in between legs $\alpha = 20.323^\circ$
Ankle Swing Leg = (0.46677, y, 0.09654)
Left Crutch Jump = 0.983-0.846 = 13.7cm
Left Crutch Tip = (0.983, 0.770)
Right Crutch Tip = (0.960, 0.195)



Preparation for Toe – Off :

Swing Foot is not flat on the ground, toe still is in contact with the ground.

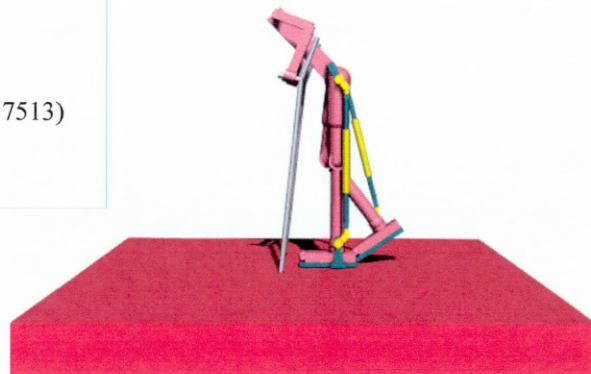
Angle in between legs $\alpha = 20.323^\circ$
Ankle Swing Leg = (0.497, y, 0.13352)
Left Crutch Tip = (0.983, 0.770)
Right Crutch Tip = (0.960, 0.195)
Ankle Dorsiflexion = 3.6°
Knee Flexion Angle $\phi = 180.0^\circ$



Toe – Off :

Swing Foot is not about to lift off, but still is in contact with the ground. One Cylinder is fully extend and the other is fully retracted.

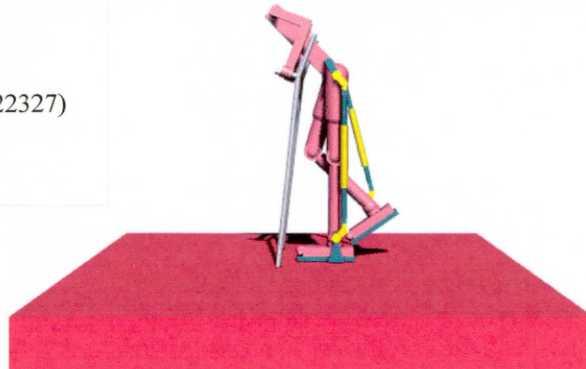
Knee Flexion Angle $\phi = 135.79^\circ$
Ankle Swing Leg = (0.54208, y, 0.17513)
Left Crutch Tip = (0.983, 0.770)
Right Crutch Tip = (0.960, 0.195)



Lift – Off

Swing Foot Lifts Off the maximal knee flexion value. Both Cylinders are fully retracted.

Knee Flexion Angle $\phi = 117.36^\circ$
Toe Clearance = 3.89cm
Ankle Swing Leg = (0.56199, y, 0.22327)
Left Crutch Tip = (0.983, 0.770)
Right Crutch Tip = (0.960, 0.195)



Mid-Swing

Swing Foot is airborne going through the swinging sequence. Both Cylinders are fully retracted.

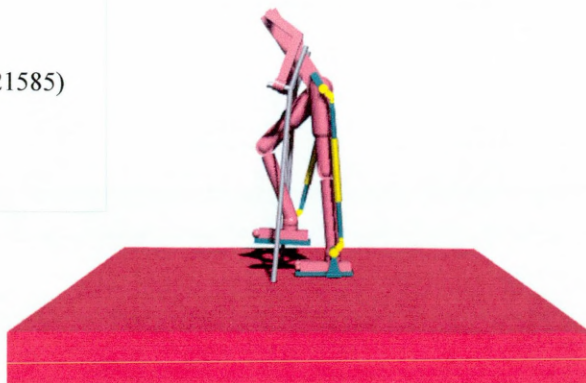
Toe Clearance = 3.185cm
Ankle Swing Leg = (0.66743, y, 0.20494)
Left Crutch Tip = (0.983, 0.770)
Right Crutch Tip = (0.960, 0.195)
Knee Flexion Angle $\phi = 117.36^\circ$



Maximal Swing Position

Swing Leg achieves its maximal rotational value and prepares for landing. Both Cylinders are fully retracted.

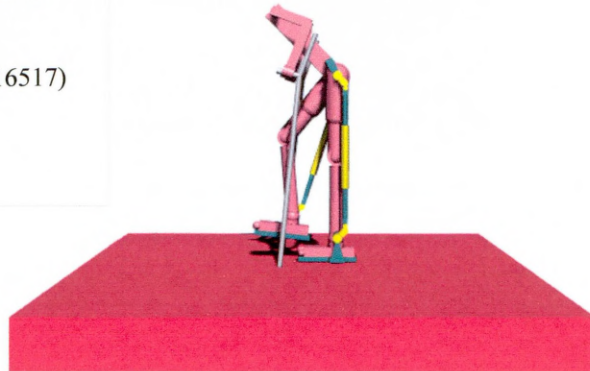
Toe Clearance = 12.072cm
Ankle Swing Leg = (0.91935, y, 0.21585)
Left Crutch Tip = (0.983, 0.770)
Right Crutch Tip = (0.960, 0.195)
Knee Flexion Angle $\phi = 117.36^\circ$



Pre-Landing

Swing Leg achieves its maximal rotational value and prepares for landing. One Cylinder is fully extend and the other is fully retracted.

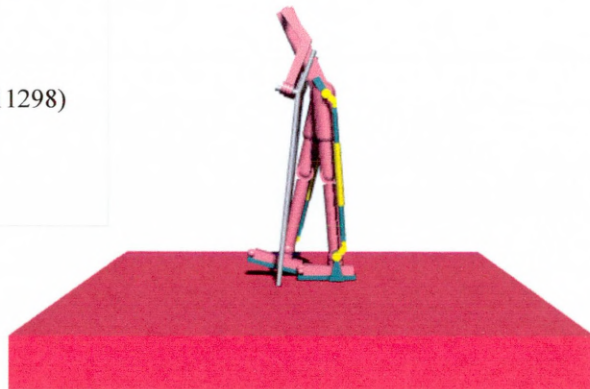
Ankle Swing Leg = (0.91475, y, 0.16517)
Left Crutch Tip = (0.983, 0.770)
Right Crutch Tip = (0.960, 0.195)
Knee Flexion Angle $\phi = 134.46^\circ$



Heel Strike

Swing Leg achieves its maximal rotational value and prepares for landing. Cylinders are fully extend.

Knee Flexion Angle $\phi = 180.0^\circ$
Ankle Swing Leg = (0.93718, y, 0.11298)
Left Crutch Tip = (0.983, 0.770)
Right Crutch Tip = (0.960, 0.195)



Starting position

Swing Leg achieves its maximal rotational value and prepares for landing.

Ankle Stance Leg = (0.7192, y, 0.09877)
Ankle Swing Leg = (0.95637, y, 0.09803)
Step Length = 23.717cm
Left Crutch Tip = (0.983, 0.770)
Right Crutch Tip = (0.960, 0.195)

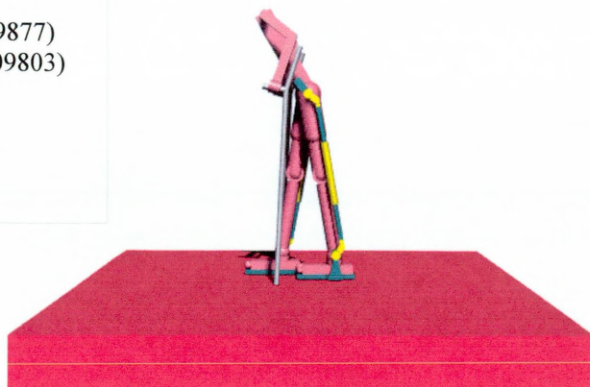
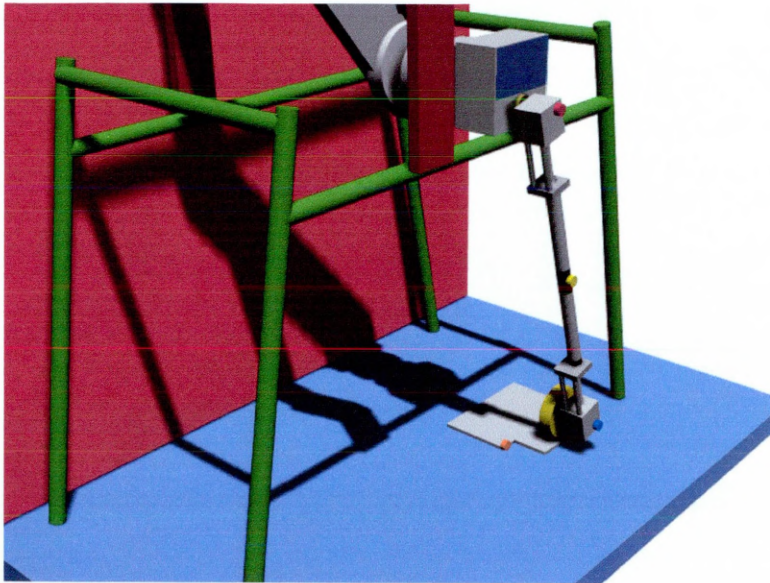


Table APA.3 contains the values for the parameters of the simulation of the test bed with the swinging leg for the sequence of a full step, this table corresponds to the values of Table 5.4.

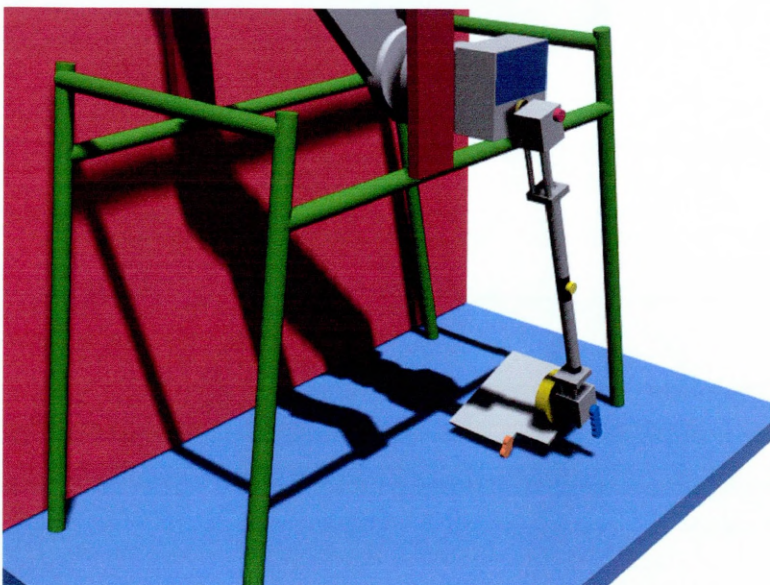
Table APA.3



Starting position :

Stance leg is in front of swing leg.
Cylinders are fully extend.

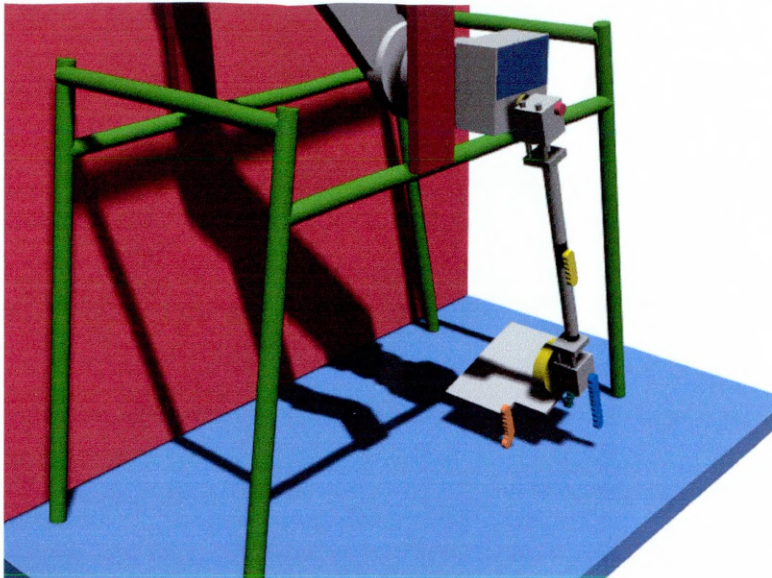
Angle in between legs $\alpha = 20.323^\circ$
Ankle Swing Leg = (0.46677, y, 0.09654)
Knee Swing Leg = (0.52987, y, 0.45148)
Stride = 0.724-0.466 = 25.731cm
Left Crutch Tip = (0.846, 0.805)
Right Crutch Tip = (0.842, 0.180)



Toe – Off :

Swing Foot is not about to lift off, but still is in contact with the ground. One Cylinder is fully extend and the other is fully retracted.

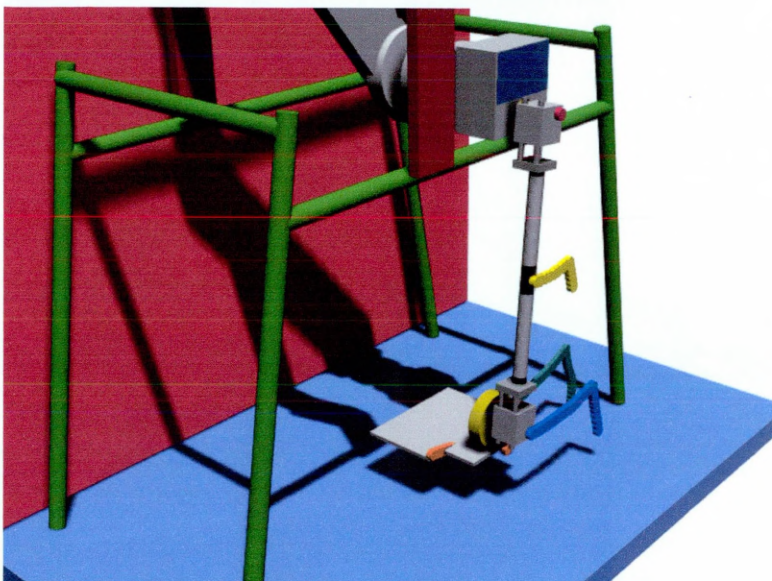
Knee Flexion Angle $\phi = 135.79^\circ$
Ankle Swing Leg = (0.54208, y, 0.17513)
Left Crutch Tip = (0.983, 0.770)
Right Crutch Tip = (0.960, 0.195)



Lift – Off

Swing Foot Lifts Off the maximal knee flexion value. Both Cylinders are fully retracted.

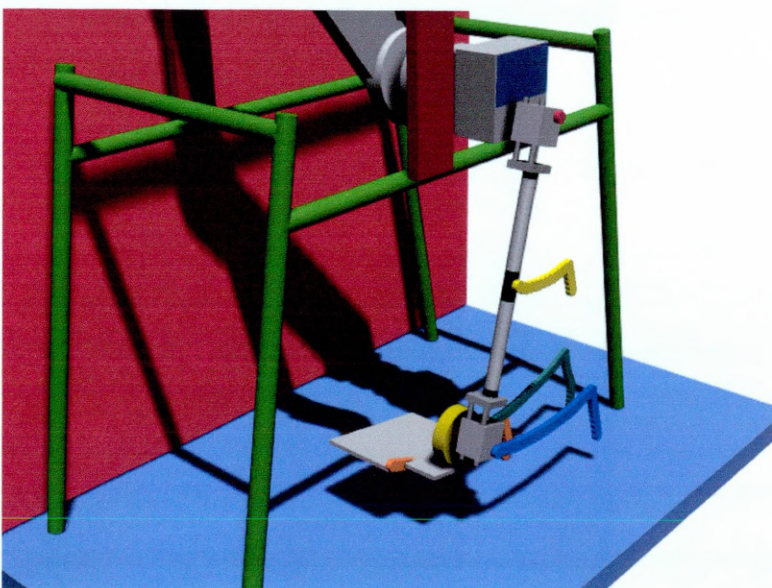
Knee Flexion Angle $\phi = 117.36^\circ$
 Toe Clearance = 3.89cm
 Ankle Swing Leg = (0.56199, y, 0.22327)
 Left Crutch Tip = (0.983, 0.770)
 Right Crutch Tip = (0.960, 0.195)



Mid-Swing

Swing Foot is airborne going through the swinging sequence. Both Cylinders are fully retracted.

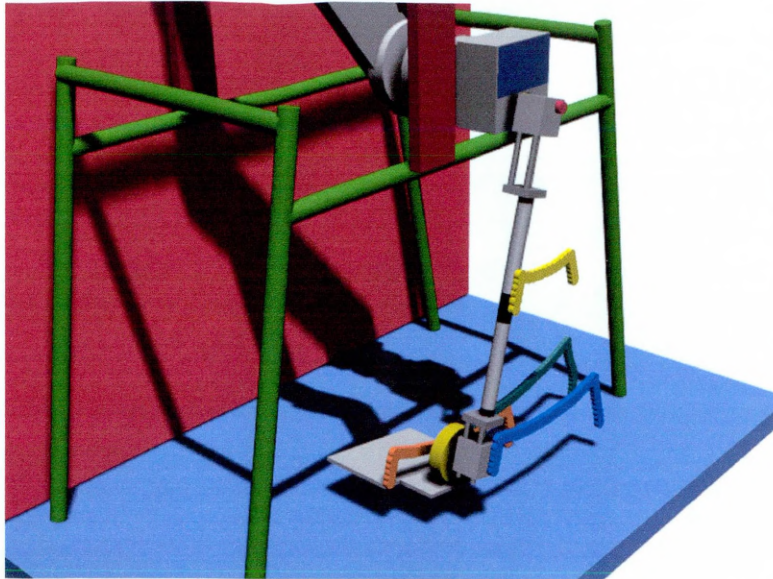
Toe Clearance = 3.185cm
 Ankle Swing Leg = (0.66743, y, 0.20494)
 Left Crutch Tip = (0.983, 0.770)
 Right Crutch Tip = (0.960, 0.195)
 Knee Flexion Angle $\phi = 117.36^\circ$



Maximal Swing Position

Swing Leg achieves its maximal rotational value and prepares for landing. Both Cylinders are fully retracted.

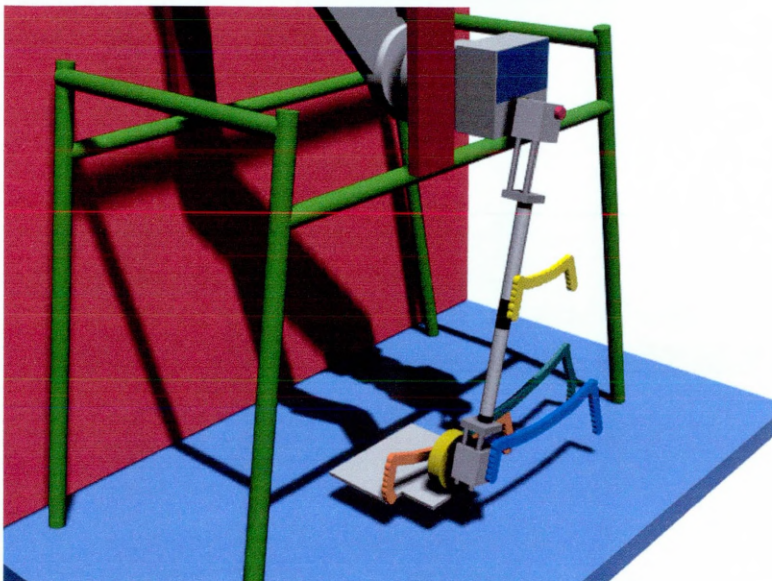
Toe Clearance = 12.072cm
 Ankle Swing Leg = (0.91935, y, 0.21585)
 Left Crutch Tip = (0.983, 0.770)
 Right Crutch Tip = (0.960, 0.195)
 Knee Flexion Angle $\phi = 117.36^\circ$



Pre-Landing

Swing Leg achieves its maximal rotational value and prepares for landing. One Cylinder is fully extend and the other is fully retracted.

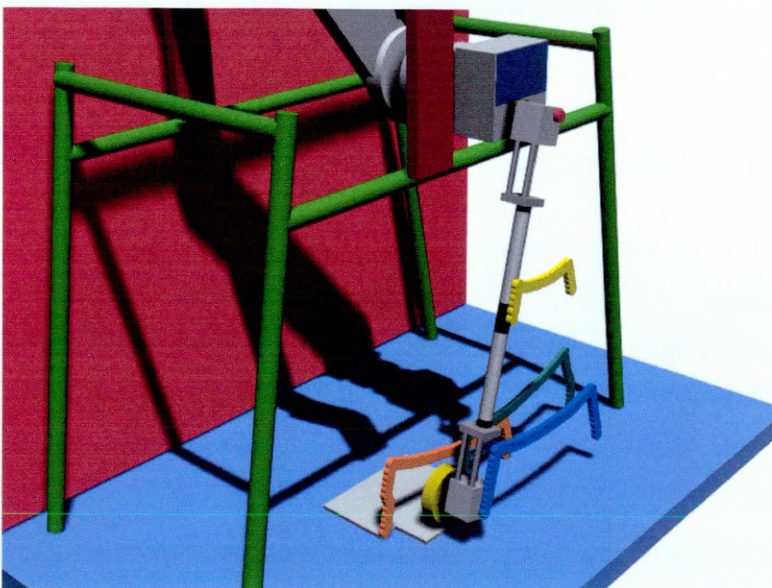
Ankle Swing Leg = (0.91475, y, 0.16517)
 Left Crutch Tip = (0.983, 0.770)
 Right Crutch Tip = (0.960, 0.195)
 Knee Flexion Angle $\phi = 134.46^\circ$



Heel Strike

Swing Leg achieves its maximal rotational value and prepares for landing. Cylinders are fully extend.

Knee Flexion Angle $\phi = 180.0^\circ$
 Ankle Swing Leg = (0.93718, y, 0.11298)
 Left Crutch Tip = (0.983, 0.770)
 Right Crutch Tip = (0.960, 0.195)



Starting position

Swing Leg achieves its maximal rotational value and prepares for landing.

Ankle Stance Leg = (0.7192, y, 0.09877)
 Ankle Swing Leg = (0.95637, y, 0.09803)
 Step Length = 23.717cm
 Left Crutch Tip = (0.983, 0.770)
 Right Crutch Tip = (0.960, 0.195)

Appendix B

Interfacing Considerations : the 8255 PPI

The PC 14 is based on the 8255 PPI interface chip, this chip features three ports that can be setup to be input or output port. Each port is 8 bit long and therefore up to 8 input or output signals can be addressed. The signal conditioning stage must be understand as a means of decoupling the digital low voltage / low current electronics from the high voltage high current electronics. This is important because the digital signals of the computer cannot cope with the current requirements of the relays and coils found on the valves and the stepper motors to be controlled. Therefore the signal conditioning must condition the low power control signal from the computer into a high power signal capable of driving the stepper motor coils and the relays driving the solenoid valves.

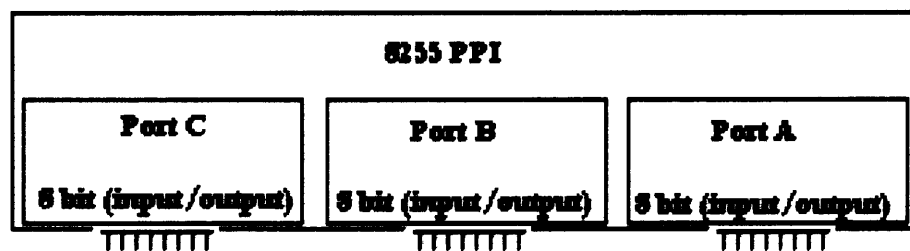


Figure APB.1 : Schematic diagram for a 8255 chip

The factory setting for the PC 14 cards currently in use, is BA = Hex 300, with the control register words for both banks being BA + 03 for PPI 1, and BA + 07 for PPI 2.

A complete description of the software code used can be found in the appendix ???. Each PC 14 AT have two 8255 with an overall 48 lines for digital I/O, plus three 8253 digital timers, that can be used for real time control strategies.

Stepper Motors Drivers

A Host Board was designed then where two 4 Phase Unipolar Driver Cards could be slotted with all the necessary provisions for current management and fail safe.

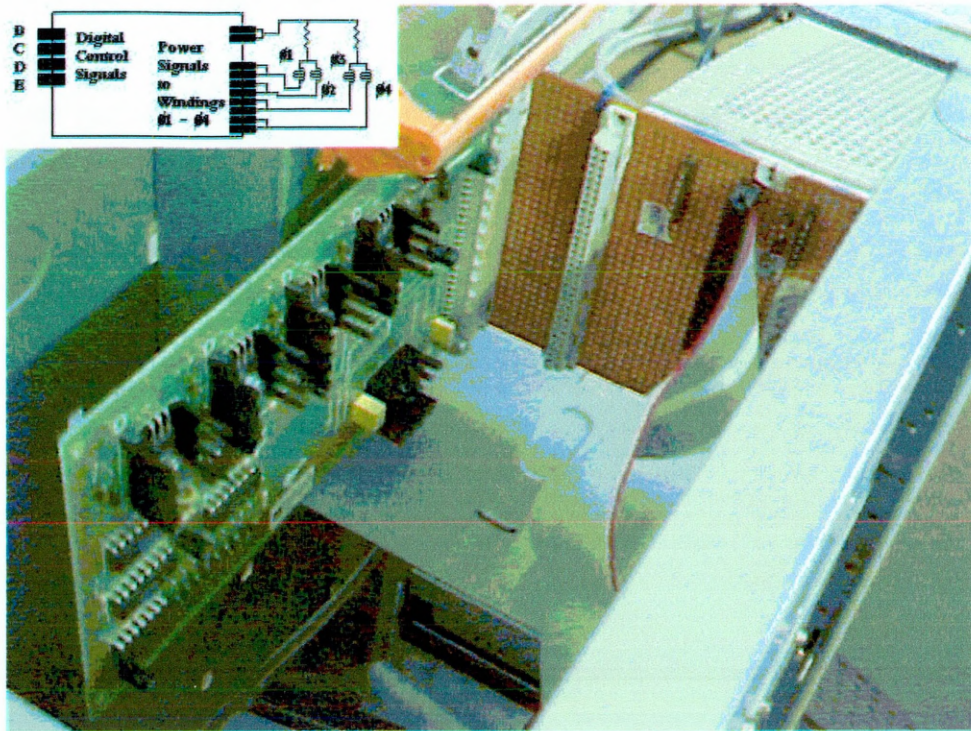


Figure APB.2 : Stepper Motor Driver Card connected to Host Card & schematic diagram.

The resistors used employed where heating considerably so to avoid any probable cause for concern a small fan was fitted directly on top of the resistors. The Host Board was then tested with one driver card controlling the Hip Stepper Motor. Then with the Stepper Motor connected with the leg through the gear box, the tests were successful and no problems were detected apart from that of the heating resistors.

The driver system for the motors is a 4 Phase Unipolar stepper motor Drive board, as the stepper motors are 4-phase, 4 control signals are required and for the complete

bipedal system, 16 control signals are required, so two ports of the 8255 are necessary just for the rotary actuators.

Cylinders Solenoid Valves Drivers

For every cylinder one signal is required, therefore four bits are required which is equivalent to half a port. From these considerations it is clear that one complete 8255 PPI is required for the actuators control signals, and therefore another 8255 PPI can be used for the sensors signals.

Decoupling card ready to interface with the relays for driving the solenoids of the cylinder valves were used. The same design can be used for connecting the control ports of the 8255 PPI with the Stepper Driver Card.

The opto-isolator used in the card is an ISQ 74 Figure APB.3 shows the main digital control signals required to interface the board.

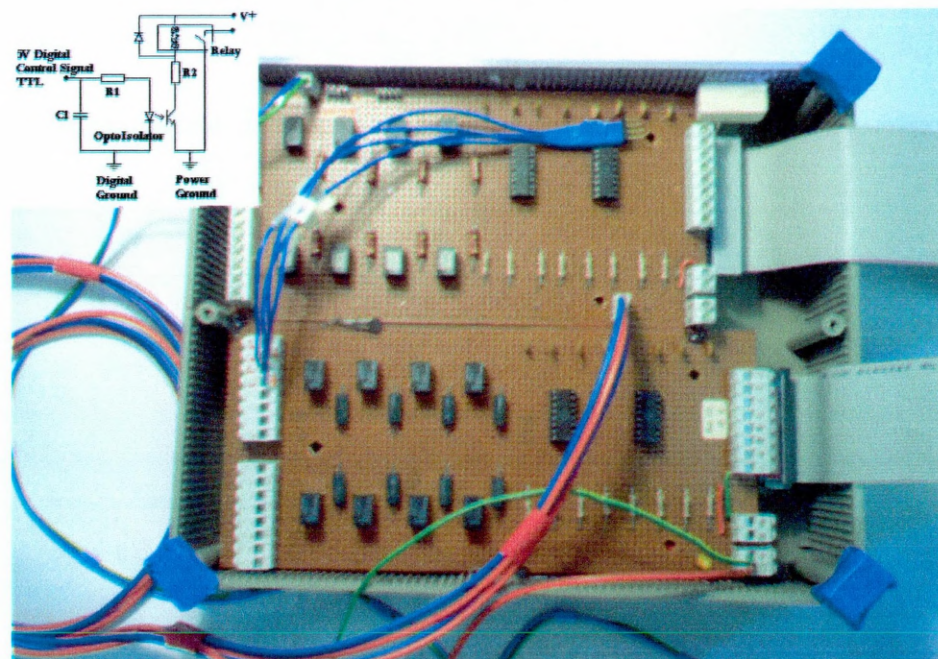


Figure APB.3 : Cylinder Valves Drivers and Schematic diagram.

Considerations for Mapping the Actuators with the I/O Card

In order to map correctly the digital signals controlling the motors and the valves some guidelines must be taken into account. The PC 14 has two banks of ports with on board leds corresponding to PPI 1 (PA0 -PA7) and PP2 (PA0 – PA7) so these two banks are very suitable for checking the controlling signals.

Therefore *full/half step*, *clock*, *direction*, *preset* and *cylinder on/off*, for the right leg hip motor are going to be located in Port A, as the control signals are the first ones to be tested it makes sense to write the code in this way so debugging and testing will be less cumbersome.

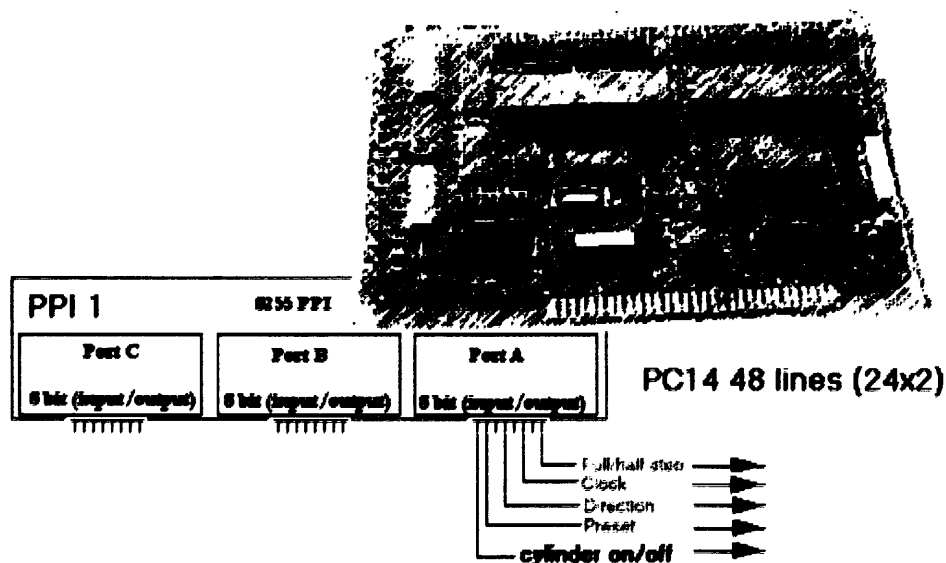


Figure APB.4 : Mapping of Digital Signals required to drive one stepper motor.

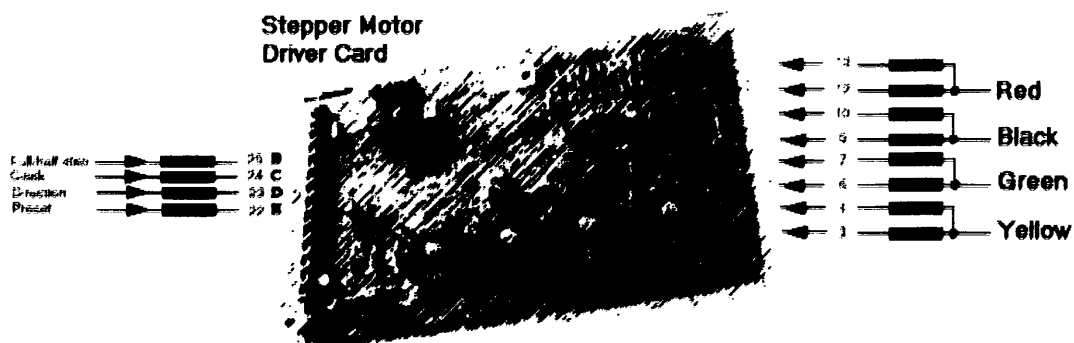


Figure APB.5 : Mapping of I/O signals for Stepper Driver Card.

For the Stepper Motor driver Card, a host card is required so it can fit inside the controlling PC, this card will host two Unipolar driver cards and the connections required to interface the motors and the PC 14, it makes sense to locate it inside the PC because in this way the distance that the ribbon connectors travel from the PC 14 to the driver cards is minimized

For the Hip Stepper motor is important to take into account that for a 1.8 degree type, this is equivalent to 200 steps per revolution or 400 in the half mode.

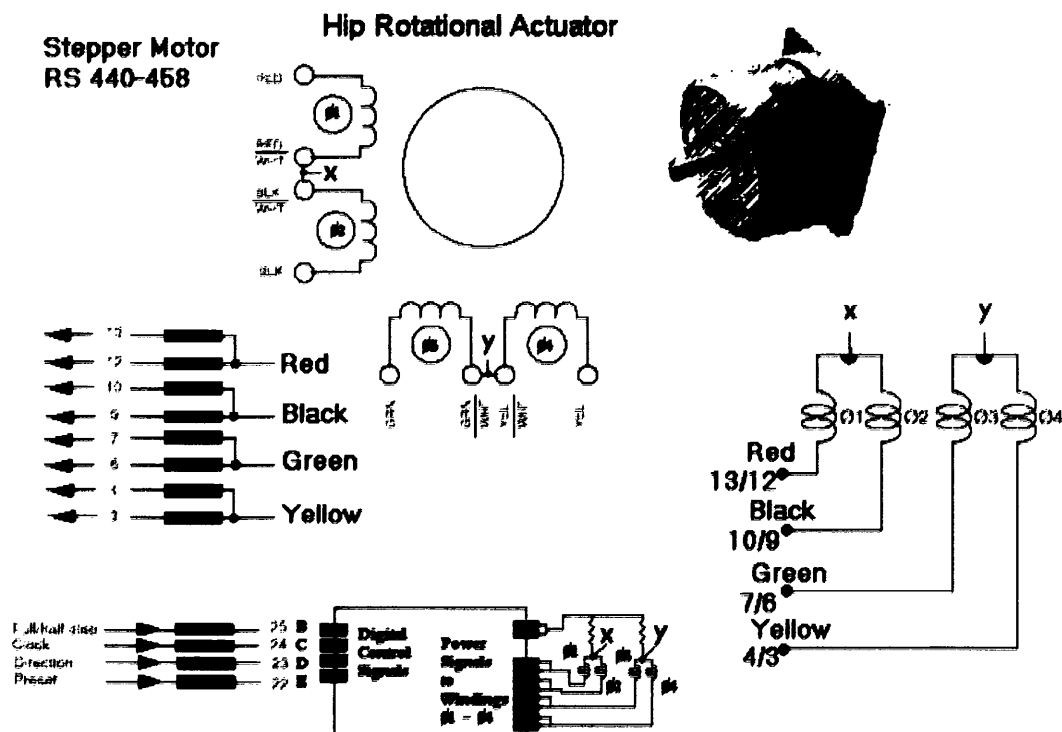


Figure APB.6 : Mapping of signals for the hip Stepper Motor and Driver Card.

The maximum frequency is 25 KHz, but this value is more than enough considering the low speed requirement for the application.

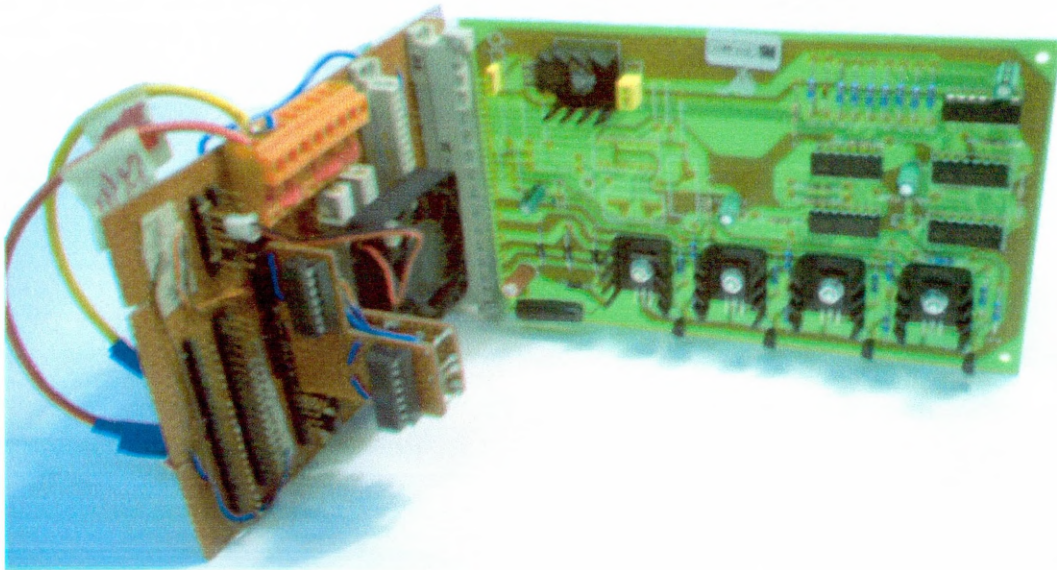


Figure APB.7 : *Stepper Motor Driver Card connected to Host Board.*

Data Acquisition System for the sensors

For the rotary joints potentiometers an ADC is necessary along with a sample and hold circuit. Two strategies can be envisaged, multiple ADC's for every potentiometer or multiplexation of the analogue signals and just one ADC.



Figure APB.8 : *Typical implementation for DAQ System.*

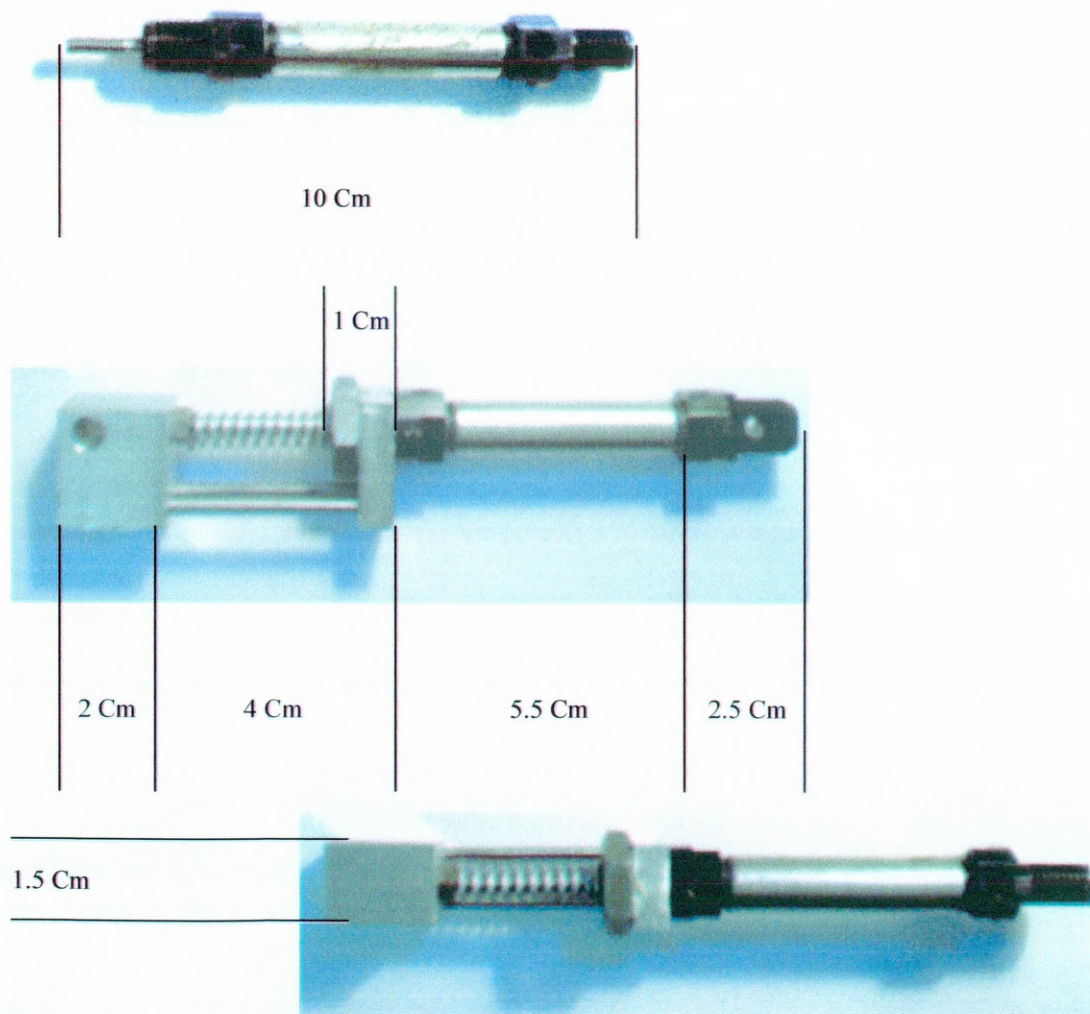
Interfacing of the DAQ with the Microprocessor System

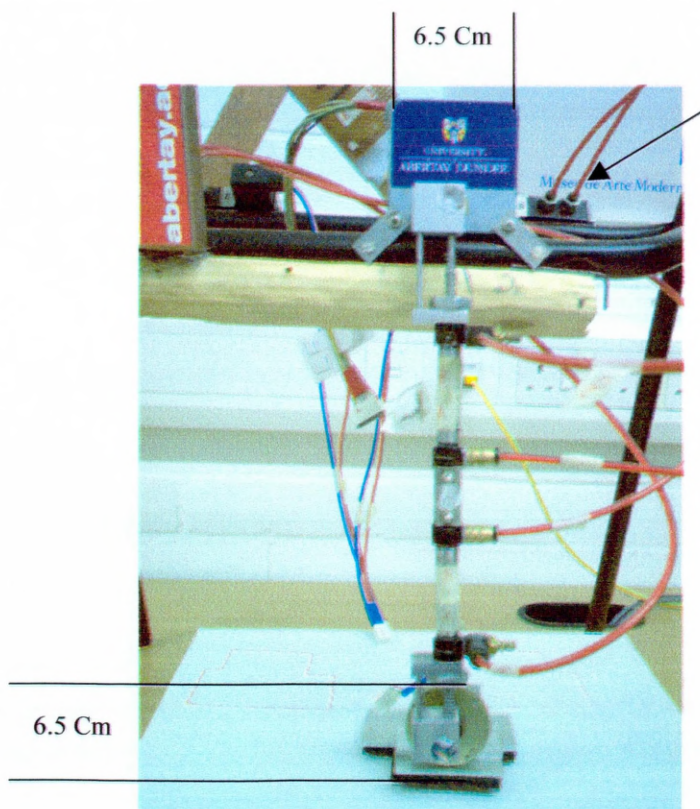
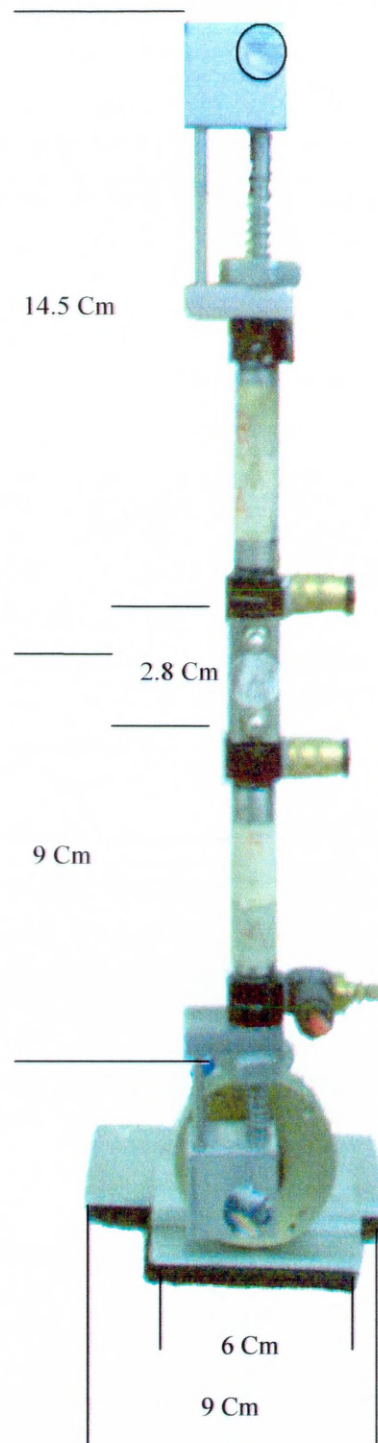
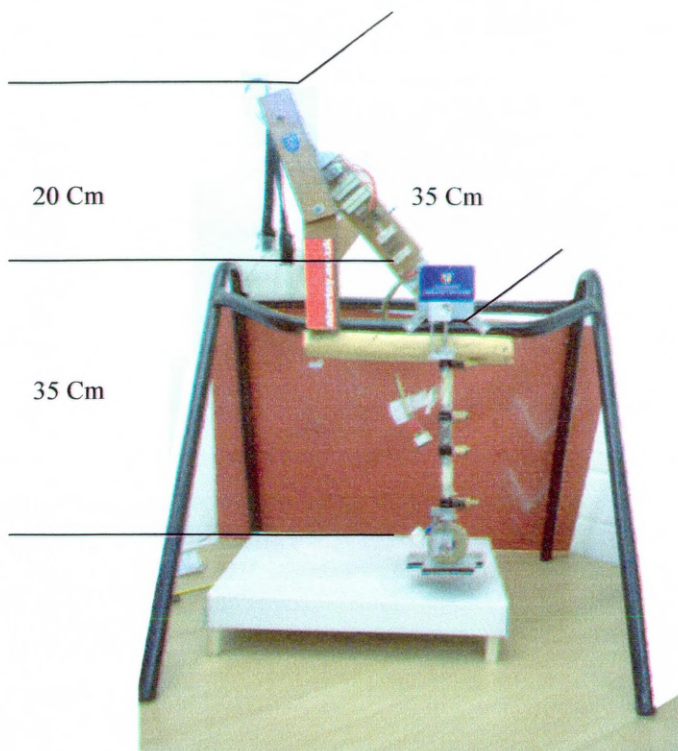
Regardless of the type of ADC what is important are the start /stop control signals required by the ADC and the address control signals required by the multiplexer. Also important are the signals for digitalisation, these ones depend on the resolution of the

ADC, so typically 8 to 12 bit conversion values can be found and therefore at least two ports of the 8255 PPI should be used.

An alternative is to use an AD7572 and multiplex the output signals, it is recommended also that the ADC be a successive-approximation one, like the AD7572JN12, since speed is the key factor when changing channels from the different potentiometer signals.

Test Bench Geometrical Description





Appendix C

Associated Publications

NEAT PROPOSAL (pp 223).

PROVISION OF GAIT ASSISTANCE FOR INDIVIDUALS WITH LOWER LIMB IMPAIRMENTS^[95], *7th Mechatronics Forum International Conference*, Atlanta, September 2000 (pp 227).

AN INTELLIGENT EXOSKELETAL AID FOR GAIT ASSISTANCE^[96], *4th International Conference on Climbing and Walking Robots*, Karlsruhe, September 2001 (pp 230).

THE OPERATION OF A POWERED EXOSKELETON TO SUPPORT REHABILITATION^[97], *8th Mechatronics Forum International Conference*, Twente, June 2002 (pp 234).

THE USE OF SIMULATION AND MODELLING IN THE DESIGN AND DEVELOPMENT OF AN EXOSKELETON TO SUPPORT MOBILITY^[98], *9th Mechatronics Forum International Conference*, Ankara, August 2004 (pp 236).

REMOTE REHABILITATION USING AN INTELLIGENT EXOSKELETON^[89], *2nd Cambridge Workshop on Universal Access and Assistive Technology*, Cambridge, March 2004 (pp 239).

NeXOS - THE DESIGN OF AN ACTIVE EXOSKELETON TO SUPPORT REHABILITATION^[90], *3rd IFAC Symposium on Mechatronic Systems*, Sydney, September 2004 (pp 244).

Additional Publications

NeXOS – Remote rehabilitation using an intelligent exoskeleton^[90], *Proceedings of the 5th International Conference on Gerontechnology*, Nagoya, May 2005 (pp 247).

THE ANALYSIS, DESIGN AND IMPLEMENTATION OF A MODEL OF AN EXOSKELETON TO SUPPORT MOBILITY^[100], *9th International Conference on Rehabilitation Robotics*, Chicago, June 2005 (pp 249).

USER INVOLVEMENT IN THE DEVELOPMENT OF REMOTE ASSISTIVE TECHNOLOGY FOR THE REHABILITATION OF SPINAL CORD INJURY^[91], *XVIIIth World Congress of Neurology*, Sydney, November 2005 (pp 251).

The full text of the following published papers have been removed from the e-thesis due to copyright restrictions:

- **Neat proposal (pp.223-226)**
- **Acosta Marquez, C.A. and Bradley, D. (2000) Provision of gait assistance for individuals with lower limb impairments, 7th Mechatronics Forum International Conference, Atlanta, September 2000 (pp.227-229)**
- **Acosta Marquez, C.A. and Bradley, D. (2001) An intelligent exoskeletal aid for gait assistance, 4th International Conference on Climbing and Walking Robots, Karlsruhe, September 2001 (pp.230-233)**
- **Acosta Marquez, C.A. and Bradley, D. (2005) The analysis, design and implementation of a model of an exoskeleton to support mobility, Sfh International Conference on Rehabilitation Robotics, Chicago, June 2005 (pp.249-251)**
- **Mawson, S., Enderby, P., Bradley, D., Acosta-Marquez, C.A., Hawley, M., Brownsell, S.J., Buri, H., Scott, E. (2005) User involvement in the development of remote assistive technology for the rehabilitation of spinal cord injury, XVIIIth World Congress of Neurology Sydney, November 2005 (p.251).**

THE OPERATION OF A POWERED EXOSKELETON TO SUPPORT REHABILITATION

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Abstract

Following injury, individuals may be faced with an extensive rehabilitation process involving repeated use of the affected limb against a controlled load. In other cases, the ability to move the limb may contribute the individual regaining control of that limb, for instance following a stroke. The paper describes an approach to the rehabilitation of the lower limbs using a web-based control strategy together with an intelligent exoskeleton to provide the necessary motions and resistance for a series of exercises leading to direct walking support. Integration within a telehealth environment is proposed to enhance user feedback and to support the management of the rehabilitation process.

1 Introduction

Following an immobilising injury to their lower limbs an individual may well be faced with an extensive rehabilitation process to regain musculature on that limb. In addition, passive movements of a limb often acts to reduce pain and to increase the comfort level for patients suffering from a range of clinical conditions, and in particular progressive neurological conditions such as Multiple Sclerosis and Motor Neurone Disease while passive muscle contractures provide support for the rehabilitation of post stroke patients. The paper examines a proposal for a lower limb rehabilitation strategy for such individuals based on the use of an intelligent exoskeletal structure, the impedance of which can be dynamically adjusted to either provide support for or offer resistance to motion as required within a simulated step cycle and can then be used with crutches to support walking and mobility as the rehabilitation process proceeds.

It is further envisaged that the exoskeletal structure would be integrated as part of a telehealth environment capable of being monitored and adjusted remotely by the relevant healthcare professionals by means of a web-based communication strategy. This same web-based strategy would also be used to support feedback to the user as to their progression against a defined and agreed set of objectives. This will remove the need for a direct intervention at the user's home every time there is a change to be made to the system set-up.

Previous papers (Acosta, 2000 & 2001) have described the basic design of an exoskeleton structure intended to assist individuals with lower limb disabilities in

regaining mobility by providing support for them to walk while the role of active orthosis in supporting rehabilitation has also been identified (Belforte, 2000). The dynamics of the exoskeleton were developed based on gait analysis to support a variety of different types and forms of gait to allow it to conform to the requirements of different individuals. Figure 1 shows the evolution of the basic gait under the control of a linear actuator to activate the knee while Figure 2 shows



Figure 1: Gait development



Figure 2: Gait sequence

2 Telehealth and Rehabilitation

Telehealth is concerned with the provision of support for healthcare, of which rehabilitation may well be one aspect, using a wide range of communications based technologies and strategies. Typically therefore, a telehealth system may integrate a remote monitoring capability with diagnostic or other equipment in the user's home environment in a way designed to support and enhance user need.

Typically, the rehabilitation process involves an exercise regime supported by physiotherapy to enable the patient to regain musculature and to enhance their control of the limb(s). One of the problems associated with such regimes is that of matching the exercise pattern to the patient as rehabilitation progresses, requiring regular and close contact with the care team. Such a strategy for rehabilitation is often labour intensive and relies heavily on the skills of healthcare personnel such as occupational therapists, who in some areas are hard to recruit. In a telehealth environment the remote rehabilitation system, located in the patients own home, would be capable of being monitored and adjusted remotely without need for a visit to the home. By including a web-based strategy for such control, the patients themselves could then readily be involved in the decision making processes and receive enhanced information regarding their progress while allowing them to keep in touch with their care and support team.

3 Exoskeleton Design and Operation

The basic design of the exoskeleton as used to support walking has been shown in Figure 1 in which the action of the telescopic actuator in moving the leg through the step cycle of Figure 2 can be seen. This proposed structure of the exoskeleton involving a linear actuator is well suited to support the rehabilitation process for a range of lower limb injuries as it can be programmed to move from a situation where it drives the limb through a motion cycle to providing resistance against which the limb can work to support muscle development. In addition to this impedance change, the configuration also lends itself to a range of geometries in which the actuator operates in parallel with or in series with the limb as required with appropriate support structures. By controlling the operation of the actuator appropriately, various motions of the limb can then be achieved to allow for the different stages of the rehabilitation process.

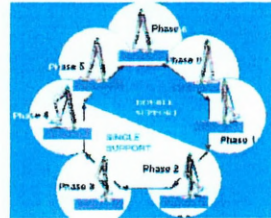


Figure 3: Motion sequence

3.1 Mode 1 operation – Motion sequencing

In this mode the actuator would be programmed to move the leg through the sequence of motions shown in Figure 3 representing a single pace. Thus the action of the leg would be similar to that experienced while walking but with the user in a seated or lying position and would be intended to provide a basic exercising of the leg. In this case the user would not be working against the exoskeleton but would be using the action of the

exoskeleton to enable the leg to be put through a defined series of movements. It should also be noted that though the configuration of Figure 3 places the actuator in parallel with the leg, it is also possible to place it in series with the leg to achieve the same set of motions. The choice of which configuration is to be used would be arrived at in discussion between the user and the supporting healthcare professionals.

4.2 Mode 2 operation – Resistance

In the second mode, the exoskeleton would be programmed to provide resistance to motion, allowing the user to use the exoskeleton for the purpose of exercising particular groups of muscles. It is possible that when it is being operated in this configuration the exoskeleton would be responsible for providing part of the motion, for instance the flexing of the leg, phases 0 to 3 in Figure 3, while providing a controlled resistance for other parts of the overall motion cycle such as the extending of the leg, phases 4 to 6 in Figure 3. The nature of the exoskeleton means that it is possible to monitor the range of motions achieved and to readily control the forces involved allowing its use for rehabilitation purposes to be matched to user needs and to be developed as rehabilitation proceeds.

4.3 Control

The exoskeleton will be operating under the direct control of its local processor which will assume responsibility for the actuator operation throughout a single step cycle in response to instructions received from the web-based controller. Because of the nature of the system involved there is a need to adapt the operation in response to user performance, for instance to take account of variations or to respond to monitored changes in performance in relation to previous usage. For this reason, the possibility of using a fuzzy controller to integrate velocity, force and power within the control environment is being investigated.

4 Web-Based Controller

The ability of the exoskeleton to be programmed and adaptive to the user makes it suitable for remote control using the web. This would then allow the person responsible for overseeing the rehabilitation process such as an occupational therapist or physiotherapist to receive an update on the use of the exoskeleton and of the user performance relative to set goals or targets. Similarly, the user could be provided with information on their performance against these same goals and targets, allowing them to see for themselves how they are progressing. Once targets are achieved, the exoskeleton could then be programmed to automatically move to the next target or to generate a report for the occupational therapist or physiotherapist to allow a new programme to be entered remotely.

The concept of using the web to control and monitor remote functions is by now reasonably well established and a number of projects have looked at applications, for instance in manufacturing, where the web is used to transfer non-real-time information, including command settings (Bright, 2000). For the rehabilitation process, the provision of a web-based controller for the exoskeleton which is used to define the rehabilitation cycle and to provide information to both the user and the support team is seen as offering a number of advantages including:

- Flexibility – Installation of the system can be achieved anywhere there is a telephone line.
- User feedback – Each user would have an individual page on which they would see their own current performance indicated along with the targets agreed with their support team.
- Control – The support team will be in a position to receive regular updates on achievement and performance and to communicate with the user to adjust settings without the need for a visit. This does not mean that visits will be eliminated, rather that there will be no need to wait for a visit before making adjustments.
- Access – The increased availability of information will support access by all participants in the rehabilitation process.

Overall therefore it is suggested that the web is a suitable medium for achieving both the reporting and the control of the exoskeleton as part of the overall rehabilitation process.

6 Conclusions

An exoskeleton originally designed to provide walking assistance to individuals with lower limb disabilities also has the potential to support and facilitate applications associated with the rehabilitation process where this involves the motion of the lower limbs. The configuration of the exoskeleton is such that it is capable of being used either to move the leg through a sequence of motions or to provide controlled resistance to motion or some combination of the two.

By integrating the exoskeleton with a web-based controller as part of a and it is suggested that this could be achieved using a web based strategy for the remote programming of the device by an occupation therapist or physiotherapist. The provision of such a web-based system therefore serves to increase the flexibility and ease of use of the system and will support rapid updating with user progression.

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THE USE OF SIMULATION AND MODELLING IN THE DESIGN AND DEVELOPMENT OF AN EXOSKELETON TO SUPPORT MOBILITY

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Abstract

Gait analysis and other techniques allow the definition of an individual's gait which can then be used to establish the operation of an exoskeleton to support mobility. In order to enable a realistic gait to be achieved the information obtained from the gait analysis must be integrated with knowledge of system operation to generate motion profiles. The paper considers the development of such profiles together with the use of video capture and analysis methods to link theoretical profiles with those achieved using a quarter-scale model of the exoskeleton.

1. Introduction

The development of an exoskeleton to support mobility in individuals with lower limb impairments requires the detailed consideration of the needs of those individuals and the integration of those needs within a functioning system. Previous papers [Acosta-Marquez 2000] [Acosta-Marquez 2001] have reported on the background to, and arguments for, the provision of an exoskeleton aimed at enhancing mobility and of the extension of that concept into a system to support rehabilitation of the lower limbs [Acosta-Marquez 2002]. Here, the intention is to consider the way in which simulation and modelling, including video analysis, was used to develop the functional structure of the exoskeleton and to relate the behaviour of the simulated system to that of a quarter-scale model of the exoskeleton operating in the laboratory.

This progression from concept to the scale model requires the bringing together of a number of forms of modelling and analysis beginning with the establishment of the gait profile through to the generation of the control strategies necessary to achieving a functioning exoskeleton. Various modelling techniques and tools were used in the course of this development including; gait analysis, mathematical modelling, kinematic modelling and behavioural modelling.

For the purpose of validation there was also a need to link predicted performance with that achieved by the quarter-scale model. This was achieved using video analysis to generate the tracks for markers located on the model for comparison with the tracks generated by the kinematic model using Visual Nastran. Using this technique, a generally good comparison between the theoretical tracks and the achieved tracks for the model can be demonstrated.

2. Gait and Gait Analysis

There are available a number of gait analysis systems [Vaughan] [De Lisa] which permit individual gaits to be recorded together with information on factors such as muscle activity, joint forces and ground reactions as shown in Fig. 1. Working from information generated by gait analysis, the requirements for walking in the sagittal plane can be established and used to construct a 'gait mannequin' using Visual Nastran software (R11) as shown in Fig. 2. Once the gait mannequin was available, it can then be used to carry out a behavioural analysis of possible actuator systems in terms of their ability to reproduce the required gait patterns.



Figure 1 Output from gait analysis system showing joint forces (left) muscle activity (centre) and ground reaction (right)

The gait patterns having been established, it was necessary to give consideration to the requirements for walking and the ability of a user to achieve the required levels of control. In particular, consideration was given to the question of stability during walking and it was decided that crutches would be used to provide additional support.



Figure 2 Gait mannequin created using Visual Nastran

This enabled each stride to be initiated from a position of 4-point stability as in Fig. 3 with linear actuators being used to provide the movement of the leg. A stride could only be initiated when both crutches and both feet were in contact with the ground and carrying load, as indicated by sensors in the crutches and on the soles of the feet. A 3-point stability (1 foot and two crutches) was then maintained throughout each stride. The introduction of crutches also enabled the controls for the exoskeleton to be embedded in the handgrips where they are directly available to the user.



Figure 3 Gait mannequin with upper trunk and extremities, balance is simplified by the use of crutches

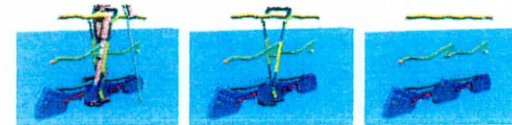


Figure 4 Examples of generated tracks using Visual Nastran

Working from the gait analysis models a series of dynamic models were created to allow all aspects of the stride to be analysed and to generate a series of tracks for the ankle, knee and hip of a person using the exoskeleton to support walking. Examples of these tracks are shown in Fig. 4 for the hip, knee, ankle, toe and heel, while Fig. 5 shows a complete step cycle.

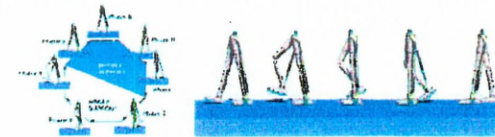


Figure 5 Step cycle with transitions in between single and double support

3. Quarter-Scale Model

In order to verify the simulation a quarter-scale model of the system (Fig. 6) was constructed in the laboratory to test the operational concepts and to link the operation of the model to the use of the crutches for control.



Figure 6 Views of the quarter-scale model

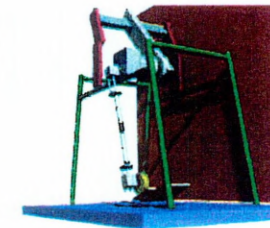


Figure 7 Quarter-scale model in Visual Nastran

This model reproduced the operation of the linear knee and also provided a stepper motor for control of the position of the ankle during the forward motion of the leg, with operation of the hip being achieved by a further stepper motor. The crutches were deployed separately with initiation of the step profile being under the control of the system user once the appropriate stance conditions had been achieved.

At the same time, the quarter-scale model was reproduced using Visual Nastran as shown in Fig 7. This meant that there were now three models for which comparisons required to be made as follows:

- ? The gait model in Visual Nastran, with and without crutches.
- ? The simulation of the quarter-scale model in Visual Nastran.
- ? The quarter-scale model itself.



Figure 8 Tracks generated from quarter-scale model using video markers



Figure 9 Tracks for quarter-scale model using Visual Nastran

Once the quarter-scale model was completed, its operation was recorded on video with markers placed at the ankle, toe, heel and 'knee'. These markers enabled the track of the model to be compared with the equivalent tracks generated for the gait mannequin and the quarter-scale model using Visual Nastran and shown in Fig. 9.

4. Results

Figure 10 illustrates the use of light studies and video capture to obtain the tracks for the hip, knee and ankle when walking with crutches. The figure also shows the equivalent tracks generated by simulation using the mannequin with crutches.

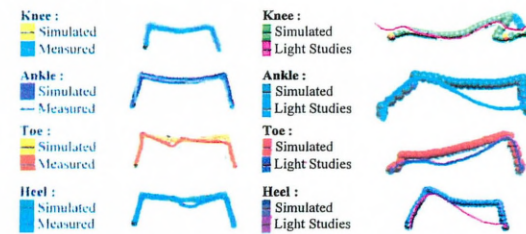


Figure 10 Step cycle with transitions in between single and double support

Using the technique of Fig. 10, Fig. 11 now shows on the left the results of a direct comparison between the measured values from the test rig and the simulation and to the right a direct comparison between the simulated gait of the mannequin and light studies

for a subject using crutches. From these it can be seen that the values for the test rig are very nearly equivalent to those obtained from the simulation, with some minor differences in the movement of the sole of the foot due to the way in which the ankle movement on the test rig had to be adjusted for a flat pickup and put-down.

However, the values from the gait mannequin model are seen to somewhat different for those obtained from a real person for the following reasons. In the case of the knee, this is because the simulation was constrained to achieve static balance throughout the forward swing of the moving leg but without the ability exhibited by a real person to control the trunk and upper body. In practice, a user would be able to adjust using their trunk and upper body so that achieving a high degree of fit at the knee is not considered to be a particular requirement. In the case of the ankle, heel and toe, the variations are due to the fact that the cylinders used to control the lift were only set to full extension or full retraction. If the position of the cylinder was controlled for intermediate positions, a higher degree of fit would be possible.



(a) Simulated and measured results for test rig (b) Simulated and measured result for mannequin and person with crutches

Figure 11 Comparison between simulated and recorded behaviour

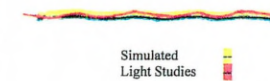


Figure 12 Results for motion of the hip

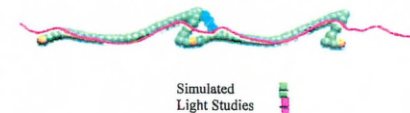


Figure 13 Results for motion of knee

Figures 12 and 13 complete the picture by showing the equivalent comparisons for the motion of the hip and knee for both simulation and the light studies. The hip movement shows a good comparison while the knee again shows the differences as identified and explained above.

5. Conclusions

The development of the exoskeleton has shown the effectiveness of linking different types of model throughout the development process. The way in which these models have been linked has supported both the conceptual development of the system and details such as actuator configurations and associated control strategies. The ability to generate motion tracks using the simulation and to compare these with those obtained using virtual markers on the quarter-scale laboratory model proved effective in determining the effectiveness of the control strategy.

In the medium term, development of the exoskeleton is proceeding not to support the walking function directly, but to facilitate the rehabilitation of the lower limbs. This project, supported by the UK Department of Health through its New and Emerging Applications of Technology (NEAT) programme aims to provide a system which can be used both local and remotely as part of an integrated programme of therapy for a wide range of individuals (Bradley 2004)

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Remote Rehabilitation Using an Intelligent Exoskeleton

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S.J. Brownsell, S.J. Mawson and C. Acosta-Marquez

1 Introduction

The exercising of the lower limbs to restore muscle function and control is an important part of the rehabilitation process for many individuals, for instance following an injury or after a stroke. Passive movement of the limbs can also act to reduce pain, prevent contractures, and increase comfort for a range of clinical conditions, as for instance patients with progressive neurological conditions such as Multiple Sclerosis and Motor Neurone Disease. However, with the increasing demands on rehabilitation teams, and resource constraints, many patients may well receive less than optimal attention. The paper reports on the initial stages of a 3-year research programme, funded by the Department of Health New and Emerging Applications of Technology programme and commencing in March 2003, which aims to address these and other related needs. Specifically, the research aims to

- Investigate the design, control and implementation of an innovative form of exoskeleton incorporating an appropriate level of intelligence to enable it to autonomously adapt, within defined limits, to patient performance in support of the rehabilitation of the lower limbs as part of a telehealth environment;
- Research the use of web-based strategies for performance monitoring by the patient's support team and for the provision of enhanced feedback to the patient on their performance to agreed targets and functional goals;
- Research the integration of web-based control and feedback strategies as part of a remote rehabilitation network, enabling individuals to derive support from other patients as well as from their support team;
- Increase patient involvement and participation in the decision making processes associated with their treatment

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the system, the exoskeleton must be able to detect and respond to external forces and torques. This should include the system's ability to control the forces exerted by the exoskeleton to support and enhance the force exerted by the patient's muscles. The system should be able to detect the activity of the muscles and provide appropriate support. The system should be able to provide resistive motion to the patient's muscles. The exact configuration of the system is at the time of writing still undetermined as this will be dependent on the input from physiotherapists and others as to the range of motion and forces expected. The exact configuration is still under discussion and is subject to frequent review. Examples of the types of motion that could be expected are:

- Allow the patient to pass through a sequence with them in a seated or lying position passively controlled by the system with manual handling of the support and appropriate support. Here, the patient would not be working against the force of the system, but the system would be supporting them in a defined range of movement.
- The exoskeleton is programmed to provide resistance to motion to exercise particular groups of muscles. In this configuration it could be expected to be responsible for part of the motion such as the flexing of the leg, while the patient's muscles would be responsible for the other part of the motion such as the extending of the leg.

The system should be able to respond to external self-adaptation, such as changes in patient performance, based on pre-programmed settings and feedback parameters set by the support team in consultation with the patient. This will remove the need for a direct intervention at the patient's knee, except in the case of a change to be made to the system set-up as their rehabilitation progresses. The use of the World-Wide Web to support communication between users of the support team will serve to increase flexibility in operation and allows for the deployment of support groups in different locations in different locations.

2 Identification of Patient Groups

Based on an analysis of the prospective patient groups it was identified that the following groups to be considered initially should include:

- A group of patients who are unable to passively control their own motion and require active forces of control over these movements that is currently supplied by a physiotherapist.
- A group whose requirement was for a combination of passive and active control, in order to be able to obtain the ability to adjust the recovery of their motion as the rehabilitation process progresses.

It is also noted that it is important to note that to consider a group that will be able to undertake of a resistive passive strategy, and another group requiring active support strategies. These groups could then act to define the boundaries of

the performance requirements for the system, and other groups could then be evaluated in relation to these. It has also been determined that the groups should include a requirement for intelligent feature, such as the autonomous adaptation to user performance and automated set-up and configuration, that enables the basic system to be used for a wide variety of purposes and with a wide range of patients as possible. Following the decision as to the general nature of the target groups, meetings with physiotherapists and a literature and patent review indicated that in the first instance attention should be given to Knee Arthroplasty (K.A.) patients as one target group with complete and incomplete Spinal Cord Injury (S.C.I.) patients forming the second target group. The reasons for the decision are as set out below.

2.1 Knee Arthroplasty

The rehabilitation of patients following knee Arthroplasty often includes the use of a Continuous Passive Motion (CPM) machine that provides a purely passive articulation of the knee joint and does not allow for the exercising of the muscles (in a resistive capacity) during the process. The proposed exo-skeleton system should:

- Enable the rapid and automatic set-up of the machine for any individual patient based on a knowledge of that patient's physical dimensions and using identifiable reference points and measures.
- Autonomously adjust, within definable limits, to any change of patient position during the therapy process.
- Monitor the forces exerted by the machine and the patient throughout the cycle and autonomously adjust these to maintain parameters such as force, velocity and power within agreed and defined limits.
- Enable patients to obtain feedback on their performance against agreed norms and allow them to assume some degree of control over the rehabilitation process. This could, for instance, include a dialogue with the machine to establish a baseline of relevant activity prior to using the machine, with the work programme then being adjusted accordingly.

2.2 Spinal Cord Injury

Complete Spinal Cord Injury (S.C.I.) patients undergo passive motion therapy to help maintain a basic conditioning of the lower limbs. This requires the physiotherapist to move and provide support to the limb through a sequence of motions (Figure 1). In the case of incomplete S.C.I. patients there is the further need to provide active assisted motions as part of the therapeutic process.

A problem not addressed by this form of therapy is the observed loss of Bone Mineral Density (BMD) leading to an increased risk of fractures (Duffy *et al.* 2000) and the research is seeking to address this. Specifically therefore the exoskeleton should:

2.1.2.2. User Acceptance

- Initially, the most basic and common of the forces applied from the axis of the motor, located in the leg, has in order to try to alleviate the loss of MBDs.
- To make the range of motion of the leg, respect the constraints of the exoskeleton control mode, has been established with them physiotherapist.
- Finally, the capabilities of the system to encompass different forms of motion based around a defined basic motion.



Figure 1. Exoskeleton in the sagittal plane, tracking tracks of all the robot and the exoskeleton.

In developing the system there is a need to establish the problems therapists and user have in conducting home based practice and to address these in a user centred manner. There is also a requirement to develop appropriate interfaces which will enable the user to have a direct control over the exoskeletons operation, to enhance the ability of a user to safely terminate operation of the machine in case of discomfort. There is also the need to establish mechanisms by which the system can respond to direct measures of patient performance. For instance, the literature search revealed that there is the possibility that passive motion therapy may induce spasticity in some patients (Moffatt *et al.* 1999), the onset of which must be detected by the system.

Interviews have been conducted with the superintendent physiotherapist and other senior physiotherapists at the Spinal Cord Injury Unit at Northern General Hospital in Sheffield and as a result it was decided to initially perform movements only in the sagittal plane. The considerations led to this decision:

- There is a safety issue in that the team was concerned with problems arising in the hip due to movements in other plane, as it is relatively easy to cause injury or dislocation because of the difficulty in tracking the proximal point for the hip joint which can only be calculated and not measured directly.
- By exercising the leg in the sagittal plane, most of the necessary therapeutic movements can be achieved and in the case of SCI the strategy for improving MBD is completely dependent on the loads imposed in this plane (Daly *et al.* 2006).

Patient and therapist focus groups have been established to provide feedback into the design process. This is an important aspect of the research as the physical appearance, size, and user interaction are all key to eventual user acceptance. Issues surrounding the wider use of rehabilitation robotics are also being pursued

in the focus groups along with efforts to understand why certain people accept or reject different forms of rehabilitation. The work of the focus group is currently in progress and the results and outcomes from these groups will be reported separately.

2.3 Other Systems

A literature and patent search concluded that most Rehabilitation Robotic Assistive Technologies deal with the issues of the upper limbs. For the lower limbs, while there are a number of references to systems to support standing and gait rehabilitation, including the use of Functional Electrical Stimulation (Holomat Matsui *et al.* 2001), only two major references to lower limb therapy and rehabilitation using robotic technology could be established (Sakaki *et al.* 1999; Matsuoka *et al.* 1999). Though the focus of the current research is associated with a wider group of individuals than those requiring gait rehabilitation, it is envisaged that it would support and indeed feed into the use of gait rehabilitation systems. Thus, an individual having undergone knee arthroplasty may use the exoskeleton to restore motion and function before progressing to other forms of rehabilitation.



Figure 2. Therapeutic Exercise Machine (Sakaki *et al.* 1999).

Of the machines identified as overlapping with the intended functions of the exoskeleton, the Therapeutic Exercise Machine (TEM) of Figure 2 (Sakaki *et al.* 1999) is the closest in terms of features. The TEM was developed as an exercise machine to decrease spasticity by repetitively performing a Range of Motion Exercise (ROME) on the hip and knee joints of patients and is based around the continuous modulation of the actual load in the patient's leg. The TEM is intended purely for passive motions and if the patient exerts an external force, as in the case of an spastic event, the mechanical arms simply move compliantly. Further, it does not incorporate the inbuilt intelligence, and hence the flexibility, of the system being proposed, nor does it have any provision for a Web-Based strategy for controlling, monitoring and implementing the user feedback.

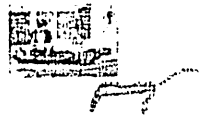


Figure 3. Path of motion for SCT therapy exercise (sagittal plane)

A machine intended for domestic rehabilitation and designed to operate without on-line supervision was developed by Boston Biomed Inc. (Matsuoka *et al.* 1990). This shares some features with the proposed web-based strategy as it can act as a virtual therapist linked to the Internet. This enables a physician to have online access to the user's performance, with the possibility of changing some machine parameters for different exercises.

In relation to CPM machines, reported work relating to their use appears to be contradictory as to their benefits, or otherwise. Some studies appear to suggest that they reduce hospital stay time by increasing the knee Range of Motion (ROM) more quickly than would otherwise be the case (Maloney *et al.* 1990) while decreasing the need for manipulation post-operation (Lachiewicz 2000) and the use of analgesia (Colwell *et al.* 1992). However, other studies suggest that there is no difference in hospital stay time (MacDonald *et al.* 2000) and that there is increased wound drainage (Pope *et al.* 1997), decreased wound healing and increased need for analgesia (Johnson 1990). All of these studies are concerned with purely passive motions as CPM machines have no ability to provide resistance to motion. There is also research that suggests that some KA patients would benefit from further rehabilitation in the home following discharge from hospital (Worland *et al.* 1998).

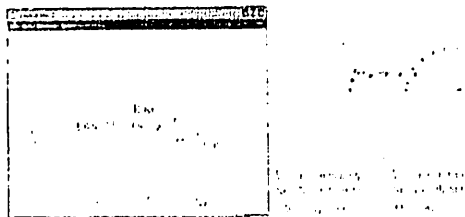


Figure 4. 2D Model for lower limb work space mapping exercise (sagittal plane)

3 System Definition, Configuration, Interfacing and Control

The video recordings of a physiotherapist working with one of the project take as subject were digitised and virtual markers used to trace the trajectories followed by the knee and the ankle as illustrated by Figure 3 which shows a frame from the video and the extracted track. A 2D model of a leg was then constructed incorporating the key measurements as in Figure 4. This then enabled the trajectories to be defined analytically and the effective region of interest defined taking into account different leg lengths. Once the leg model, tracks and region of interest had been established different kinematic configurations were then evaluated in relation to the workspace in terms of their ability to reproduce the movement achieved at the ankle by the physiotherapist. As indicated earlier, the final kinematic form has not been confirmed as the model and the associated motions, are still being developed in association with physiotherapists.

If real-time access for the control of the exoskeleton is required, then the choice of hardware and software for control and interfacing should include a development environment that allows for the provision of external monitoring. If real-time monitoring is not essential, then a simple database manager with a windows interface could be enough, removing the necessity to access the embedded controller directly.

3.1 Intelligence and Control

In order to achieve the selected movements the exoskeleton has to track the path of the ankle. To ensure this condition, accurate positioning of the body of the patient is important, though the system should be able to autonomously adjust for slight variations in that position during use. In order to achieve this profile of movements, a minimum of two active degrees of freedom are required.

The control strategy currently suggested and envisaged is based on that used in the Lancaster University Computerised Intelligent Excavator (CUE) project (Bradley & Seward 1998) which is a proven and robust method for achieving track following under conditions of variable load and which has proven capable of integrating with an intelligent control strategy which adjusts and adapts performance in relation to changing conditions. The method is based on the use of a velocity vector, with the tracking errors then being used to determine system response. The major difference is that the exoskeleton will probably require the use of a fuzzy rule set instead of the crisp rule set used by the excavator to achieve the intelligent adaptation to patient use that is envisaged, but not yet defined.

In particular, the internal intelligence should support, without direct intervention by either the user or the therapist, features such as the ability to change the current therapy routine in response to user performance. For instance, by increasing or decreasing the speed of motion or adjusting the applied forces. The boundaries within which these changes could take place would be determined in discussion between the patient and their support team. It should also support the

3.2 Web Options

From the perspective of the user, the adoption of a web-based system could enable them to have access to the therapist from their home whenever they are using the system and to obtain direct feedback on their progression. It could also facilitate the development of a support network, made up of individual users able to communicate with each other from their individual homes.

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Figure 8. Functional Viewpoints

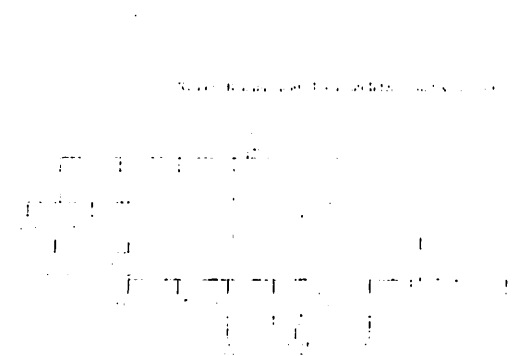


Figure 6. Step 1: Extracted Components

3.2 System Design

- Be sufficiently abstract to capture physical and information characteristics.
- Be able to include human and other non-technical forms of interaction.
- Support a multi-disciplinary approach involving individuals with a wide range of backgrounds and expertise.
- Present the information in a clear and simple fashion.

Viewpoint analysis has been used to identify the key system elements and to support communication between team members. The results of this exercise are shown in Figures 5 and 6 in which the functional viewpoints are those related to the way the system works and the non-functional viewpoints are the remaining

4 Conclusions

There are few examples of the use of motion systems to support the rehabilitation of the lower limbs. This paper reports on the preliminary concepts involved in the design and development of such an aid intended to provide both posture movement of the leg and resistance to motion. User views, including those of therapists, are being used to gain important insights into how the exo-device should work, and interact with the user, including the use of the web for both communication and control. The initial range of motions required has been established and a kinematic

analysis carried out to identify possible actuator configurations to achieve the desired motion profiles.

5 Acknowledgements

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NeXOS - The DESIGN OF AN ACTIVE EXOSKELETON TO SUPPORT REHABILITATION

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Abstract: Increasing pressures on physiotherapy services, the need to increase the involvement of patients in their treatment and the development of telecare technologies all mean it is necessary to consider new ways of providing the levels of support and treatment required. The concept of an integrated and intelligent system based around the use of an exoskeleton to support patients and physiotherapists during rehabilitation is at the heart of the NeXOS project. The paper discusses the development of the NeXOS concept, and of the interdisciplinary nature of the programme, aimed at providing a prototype system capable of being integrated with patient support services using web-based strategies. *Copyright © 2004 IFAC*

Keywords: Mechatronics, Engineering Design, Automation, Rehabilitation

1. INTRODUCTION

Physiotherapy is an important component in the rehabilitation process for individuals with a wide range of lower limb disabilities. In order to maintain soft tissue flexibility and promote the relearning of motor patterns, repetition of the movement, over an extended period of time, of the lower limbs through a series of controlled motions is essential.

Typically, this is achieved by the physiotherapist directly moving the lower limb, requiring a significant degree of physical effort on behalf of the physiotherapist and meaning that they can only deal with a single individual at a time. Further, the requirement that the manipulation is carried out manually means that it is not possible to provide additional specialist treatment at other times, for instance in an individuals home environment as part of an extended telecare package.

Where mechanical systems are used, as for instance in the case of a Continuous Passive Movement (CPM) machine that might be used following knee surgery, these are currently relatively limited in their operation and capability, are difficult to set up and need frequent monitoring and adjustment, as for instance when a patient moves position slightly. This is not however to say that there have not been a number of attempts to produce automated and semi-automated systems to support the rehabilitation of the upper and lower limbs. These include systems such as the Lokomat produced by Hokoma [Hokoma, 2003], the TEM system developed by Sakaki *et al* [Sakaki *et al*, 1999], the REHAROB project [Tóth *et al*, 2001] and the Leg Extension system produced by Monitored Rehab Systems [CMDSport, 2003], all of which aim to provide a range of motion types for different purposes. In addition, use is being made of Functional Electric Stimulation (FES) based on external electrodes to provide for and enhance lower



Fig. 1. Simulated stride pattern

limb activities for purpose of therapy, as for instance the cycling and standing systems developed at Glasgow University in the UK [Hunt, 2003][Jaime *et al.*, 2002].

The paper discusses the initial stage of the design and development of a novel form of rehabilitation system for the lower limbs which makes use of an exoskeleton to provide the required motion and which is capable of being used either locally or remotely, using the World Wide Web as a communication and control medium to link the patient with their support team during the rehabilitation process. The project, known as NeXOS, is supported by the UK Department of Health through its New and Emerging Applications of Technology programmes and brings together engineering design and mechatronics with clinical and operational requirements in meeting patient need.

The key aspect of the development of the NeXOS project, and of the exoskeleton concept around which it is based, is that of meeting patient needs in relation to the rehabilitation process. The paper therefore concentrates on the design processes necessary to meeting that need rather than the technologies to be used. Consideration is given to the solving of problems such as the communication between individuals with different domain expertise and ensuring that there is a common understanding of the requirements and of the means of solution required. Tools such as viewpoint analysis and focus groups have been used to assist in identifying requirements, and the information gained has been used to support the mathematical modelling of a range of solution options for the kinematic structure.

2. BACKGROUND

The NeXOS project was developed out of a research programme to develop an intelligent exoskeleton capable of providing assisted walking to individuals with a range of lower limb disabilities [Acosta-Marquez, 2001]. This exoskeleton system uses a linear actuator to activate the knee and to provide support for the walking function as suggested by Figure 1. Additional support during walking was provided by the use of crutches, as shown in Figure 2, which had the controls for the walking function embedded in their handgrips. Using this strategy, the user would be able to pre-define up to three gaits and to select the appropriate one as required. Additional control enabled the user to adjust the height of the lift and, where appropriate, the length and speed of the step. Balance was facilitated by the crutches and



Fig. 2. Computer generated mannequin with crutches



Fig. 3. Quarter-scale laboratory model

sensors in the crutches and shoes ensured a 4-point stance before a step could be initiated.

The exoskeleton was developed into a quarter-scale laboratory model (Figure 3), the performance of which was compared with the predicted performance using video capture and analysis to contrast the theoretical and achieved motions. Once the quarter-scale model was functioning, consideration was given to the possibility of using the exoskeleton as the basis of a rehabilitation system for use with the lower limbs. Evaluation of the required modes of operation for both walking and rehabilitation resulted in the decision to proceed in the first instance with the implementation of the rehabilitation function as this seemed to offer more immediate benefits to a wider range of prospective user groups. The project team that was assembled to take the programme forward provided a balance of technical and clinical skills typical of many mechatronic projects, these skills are set out in Table 1.

While this grouping provided the necessary skills, it created problems of communication that had to be



Fig. 4. Typical exercise performed in the sagittal plane for SCI subjects showing tracks for ankle and knee.

Table 1: Project team structure

Team Member	Role
University of Abertay Dundee	System design and programme management
Barnsley District General Hospital	Clinical expertise and patient management
Sheffield University	Remedial systems and user needs
Sheffield Hallam University	Physiotherapy and user needs

Table 2: Oxford scale

Grade	Definition
0 or 1	No activity
2	Active assisted
3 and 4	Resisted
5	Functionally active muscle

solved if a viable system was to result. Specifically, each of the team members viewed the system from a different perspective and required a different form of interaction, expressed in relation to their own specific skills. Thus, one set of team members were concerned with an ability to express the operation of the exoskeleton in terms of its motions and forces in order to define and implement control strategies, while others were concerned with establishing patient relationships and others with the defining the interaction with the physiotherapists who would deploy the system.

This difference of perspectives and the requirement and ability to ensure a robust communication between team members is a characteristic of most mechatronic projects, and is thus an area to which particular attention has to be given. In the case of the NeXOS project, this has meant using a variety of tools ranging from observation to analysis to establish a basis from which development can take place. It is the use of these tools and the integration of the information that they provide that will be considered in the following sections.

3. SYSTEM REQUIREMENTS

3.1 Patient Groups

Systems such as the CPM machine and the TEM system referred to earlier are essentially passive in their operations in that they act only to move the leg through a defined series of movements and require the user to exert no forces during the motion. A key feature of the NeXOS exoskeleton is the ability to

provide a balance of operation in which motion may range from the purely passive in which the exoskeleton is entirely responsible for the movement of the leg to active where the patient would be working against the exoskeleton, which would then be providing a resistance to motion, to various combinations of active-assisted and resistive motions within a cycle of operation.

This meant that for the purposes of the initial evaluation it was necessary to specify prospective user groups whose needs would reflect these requirements in terms of the ability of the exoskeleton to both provide and resist motion. Specifically, there was a need to identify:

- A group whose requirement was for passive movements only, but where a much greater degree of control over those movements than currently achievable would be beneficial.
- A group whose requirement was for a range of active-assisted movements involving the requirement to adjust the response of the system in relation to the movements achieved. These movements would be based around the Oxford scale used by physiotherapists and set out in Table 2.

Based on meetings with physiotherapists, and supported by a literature and patent review, it was decided to establish the initial investigation around patients undergoing Knee Arthroplasty (KA) and Spinal Cord Injury (SCI) patients. The reasons for this decision were:

Knee Arthroplasty - The rehabilitation of patients following Knee Arthroplasty is often based on the use of a Continuous Passive Motion (CPM) machine to provide purely passive articulation of the knee joint following surgery. As already indicated, these need to be reconfigured should the patient move during the therapy or be required to leave the machine for any purpose and, being purely passive, they do not allow for the exercising of the muscles.

Spinal Cord Injury - Spinal Cord Injury (SCI) patients undergo passive motion therapy, as illustrated by Figure 4 where a physiotherapist is working with a member of the project team, to help to maintain a basic conditioning of the lower limbs. This requires that the physiotherapist move the limb through a sequence of motions while providing support for the limb. A specific problem that is not addressed by this form of therapy is the observed loss of Bone Mineral Density (BMD) leading to an



Fig. 5. Identification of viewpoints using brainstorming

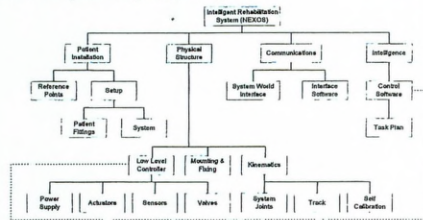


Fig. 6. Hierarchy of functional viewpoints

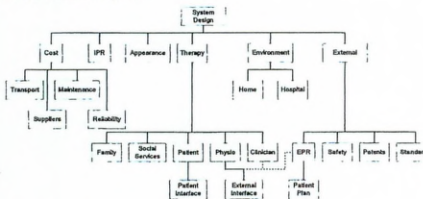


Fig. 7. Hierarchy of non-functional viewpoints

increased risk of fracture [Kiratli *et al.* 2000][Dauty, 2000] and addressing this problem is something to which consideration is being given.

Based on the above it becomes possible to establish some requirements for the exoskeletal system in that it should:

1. Enable a variety of motions and motion types as determined by the physiotherapist in conjunction with the user.
2. Enable the rapid reconfiguration to encompass different forms of motion based around a basic defined motion.
3. Enable the monitoring and control of the forces applied along the axis of the major bones in the lower leg in order to try to alleviate the loss of BMD.
4. Support a rapid and automatic set-up procedure for any individual patient based on knowledge of that patient's physical dimensions. This would enable patients to set the machine up themselves rather than requiring assistance, enabling its use in the home environment.
5. Autonomously adjust, within definable limits, to any change of patient position during the therapy process.

6. Monitor the forces exerted by the machine and the patient throughout a cycle and autonomously adjust these to maintain parameters such as force, velocity and power within agreed and defined limits.
7. Provide motivation to patients through feedback on their performance against agreed norms and by allowing them to assume some degree of control over the rehabilitation process. This could, for instance, include a dialogue with the machine to establish a baseline of relevant activity prior to using the machine, with the work programme then being adjusted accordingly.
8. Provide for 'jerk free' transitions and operation throughout the cycle.

3.2 Physiotherapists

In the case of the physiotherapists, the initial focus groups identified the requirement for an interactive procedure for setting up and configuring the system. This could perhaps take the form of a 'teach and repeat' procedure in which the system recorded and then played back the actions of the physiotherapist. It was also considered that, in the first instance, a system such as was being proposed would not be used remotely, and safety issues were identified

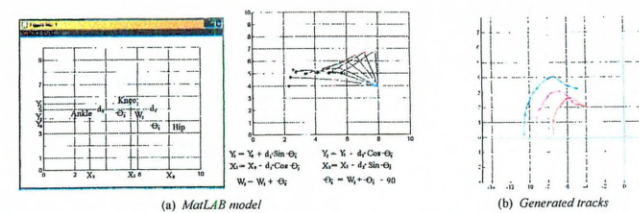


Fig. 8. Leg motion during rehabilitation using a track representation derived from video analysis

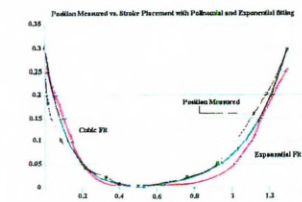


Fig. 9. Results of curve fitting routines for path generation



Fig. 10. Velocity vectors for path generation and representation

which would become the subject of detailed study and evaluation as the kinematic geometry of the exoskeleton became established.

They also identified the need to establish an appropriate, and easily reconfigurable, interface which will enable the user to have a degree of direct control over the operation of the machine, for instance in respect of their ability to terminate the operation of the machine in case of discomfort. There is also the need to establish mechanisms by which the system can respond to direct measures of patient performance. For instance, there is the possibility that passive motion therapy may induce spasticity in some patients [Skold, 1999], the onset of which must be detected by the system, which should then respond accordingly.

3.3 Functional Analysis

Having established the basic requirements for the system, it was necessary to perform a functional

analysis in order to be able to define the key objectives in relation to meeting the performance requirements for the exoskeleton. The method chosen for carrying out this analysis was viewpoint analysis as this enabled the division of the identified viewpoints into those directly concerned with the physical operation of the exoskeleton (the *functional viewpoints*) and those concerned with its integration into clinical and other systems (the *non-functional viewpoints*) [Bradley *et al.* 2000]. Using this approach, a brainstorming session is first used to identify the viewpoints, which are then clustered to create the groups of functional and non-functional viewpoints. These can then be arranged to form hierarchy trees representing the interrelationships between the individual viewpoints. Figure 5 shows the results of the brainstorming session using a whiteboard while Figures 6 and 7 show the resulting hierarchies.

4. SYSTEM DEFINITION

Using the information obtained from the patient groups and physiotherapists and combining this with the structure provided by the viewpoint analysis, the research moved to consider both the development of the physical structure of the exoskeleton and its control in parallel with the development of user views through a combination of analysis and focus group studies.

4.1 Analysis

By introducing virtual markers into the video images of a physiotherapist performing a series of manipulative exercises (Figure 4) it is possible to extract information as to the range and types of movements involved, including information about the velocity profile established over a complete manipulative cycle. This information can then be used to construct a mathematical model of the motion of the ankle which can then be used to represent patients of different leg lengths as shown in Figure 8. Having established a mathematical model of the track of the ankle, curve-fitting routines can then be used to establish a mathematical representation of the tracks generated by the physiotherapist during manipulation as in Figure 9. This representation then forms the basis of the control strategy that is proposed for the control of the exoskeleton in which

actuators are controlled to move the ankle along a track defined as a series of linked velocity vectors as in Figure 10, each of which represents the desired movement in a specified time interval. This control strategy is a modification of that which was used for the Lancaster University Controlled Intelligent Excavator (LUCIE) which demonstrated that the velocity error and the rate of change of that error in a track following situation could be used as the inputs to an expert controller based around a production rule structure [Bradley, 1998].

4.2 Focus Groups

Work with focus groups has to date been with orthopaedic physiotherapists as these have been identified as a key clinical groups and has concentrated on establishing their views on the relevant target groups of users and of the requirements for the introduction of a system such as NeXOS within a rehabilitation environment. The feedback from these groups has to date advised the team on factors such as the structure of the interface and the information it should contain and the target environments for system introduction. As the project develops, additional focus groups involving users will be established to obtain their input into the design process.

5. SYSTEM DEVELOPMENT

On the basis of the requirements analysis, viewpoint analysis, system analysis and focus groups it becomes possible to create a system schematic incorporating the principal features of the exoskeleton and its related system components. The system is thus based around establishing appropriate communication between the physiotherapist and the patient, supported by reference to relevant clinical databases and references, for instance to establish patient identification and to store previously defined motion profiles.

6. CONCLUSIONS

The NeXOS project is an integrated and cross-disciplinary project which involves a range of subject specialists coming together to develop and implement a new form of rehabilitation system. In doing so, conventional engineering design and systems analysis methods are being combined with other forms of analysis to enable the system concepts to be integrated with stakeholder need. As part of this process, extensive use has been made of system modelling and visualisation tools to support the communication of ideas between the project team and to stakeholders through focus groups. Following on from the analysis and modelling stages, the development rig will be used to evaluate the different control and operational strategies as well as to finalise the kinematic geometry of the exoskeleton. Use will be made of web-based strategies as the basis of both local and remote communications with the exoskeleton and with stakeholders.

7. ACKNOWLEDGEMENTS

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NEXOS - REMOTE REHABILITATION USING AN INTELLIGENT EXOSKELETON

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Rehabilitation of the lower limbs is important to maintain or restore muscle function and control while moving the limbs passively can maintain soft tissue length and act to reduce pain for a range of clinical conditions. There are few examples of the use of robotic technologies to assist in rehabilitation of the lower limbs. With continued time pressure on rehabilitation services, there is a desire to develop such robotic aids to assist in the repetitive nature of some exercises or Range Of Movement tasks. This paper reports on the development of one such robotic aid that targets the lower limbs and can be operated in the user's own home, at a super clinic or controlled via the Internet. The range of motion required has been established and a kinematic analysis carried out to identify possible geometrical configurations to achieve the desired motion profiles.

BACKGROUND

Exercising of the lower limbs to restore muscle function and control is an important part of the rehabilitation process for many individuals after injury such as fracture or stroke. In addition, moving the limbs passively can act to reduce pain and increase comfort for a range of clinical conditions, for instance for patients with progressive neurological conditions such as Multiple Sclerosis and Motor Neurone Disease. There are many indications in the research literature to show that changing motor control requires repeated and intensive practice¹. However, with the increasing demands on rehabilitation teams, and resource constraints, many patients may well receive less than optimal attention.

This project seeks to develop an intelligent exoskeleton (a jointed structure which supports and moves the leg) to support and exercise the lower limbs of a wide range of individuals with a variety of conditions and needs. The exoskeleton will also be capable of limited self adaptation (within defined parameters) to patient performance. This removes the need for direct intervention by physiotherapists every time there is a change to be made to the system set-up. It also removes the requirement for the physiotherapist to be physically with the patient, enabling the possibility of intensive physiotherapy in the patient's own home.

METHODS

The development of the exoskeleton is reported in two sections, namely:

- (1) user requirements
- (2) technical development.

USER REQUIREMENTS

User involvement in the development of healthcare technologies is an increasingly accepted principle² that has a positive effect on user satisfaction and may be an effective means of capturing product requirements³. Consequently, focus groups with providers and patients were conducted. Data was collected using a semi-structured interview schedule and all focus groups were recorded and transcribed for subsequent content analysis.

Provider requirements

Four focus groups with physiotherapists in South Yorkshire (UK) were conducted (n = 32). All participants were purposively selected from the clinical fields of either orthopaedics or spinal cord injuries, or were community based physiotherapists. The following themes emerged:

- *Limitations of current equipment:* Therapists indicated that there was no evidence to suggest that the use of devices currently in use (continuous passive motion) impacted upon recovery. There appeared to be few formal procedures for its use and it was limited to a few complex cases. The current machines are only capable of passive movements and they have a tendency to pull the patient down the bed and therefore repeated readjustments are required.
- *Usefulness of robotic aid:* Considered particularly relevant for repetitive movements to increase the range of movement (ROM), for prolonged courses of therapy, bilateral and specialised conditions, such as sports injury and leg lengthening. It was suggested that at times patients

are not discharged as the ROM was not appropriate and a device such as that being proposed could be useful in such circumstances.

▪ **Compliance with therapy:** Some suggested that the robotic aid could make some patients reliant on it, while others suggested that only highly motivated patients would benefit.

▪ **Control of therapy:** Therapists indicated they would envisage a requirement to monitor the robotic aid and check all is working well. They also indicated a need to feel and touch joints. One physiotherapist commented "Nothing can substitute that [exoskeleton] for a pair of hands and skill in actually being able to analyse that joint."

▪ **Location of therapy:** It was indicated that some patients would benefit from home based therapy with the robotic aid, but some concerns over legal and safety issues were raised. For example "I suppose you've got to be careful, because you don't want to get on it with stiff legs, do you, and have it sort of wrenching around." However, all of the groups discussed the concept of 'super clinics' where several robotic aids were physically supervised by one physiotherapist. This, they suggested, was an all win, as throughput was improved while maintaining the physical contact and interaction with patients.

Patient requirements

Four structured focus groups were held with patients (n = 12). Three key themes emerged:

▪ **Monitoring function of robotic aid:** Participants valued the potential ability to monitor and measure more accurately their progress and improvement. They considered that a machine could perform this more accurately than a physiotherapist.

▪ **Motivational factors:** Feedback was important to participants. They wanted to know if they were improving and identified a robotic aid as having the potential to accurately provide them with this information. They suggested that people stop complying to their therapy when they no longer see improvement and the robotic aid could resist this.

▪ **Contact with physiotherapist:** The robotic aid was viewed as part of a package of therapy, and that they would need to have access to a therapist by telephone, over the internet or receive a visit from a community physiotherapist. While it was suggested the therapist can help to motivate them, the robotic aid was seen as a way of monitoring and measuring progress in the absence of sufficient numbers of physiotherapists and for reducing the number of visits to rehabilitation sessions.

Overall the focus groups suggested that the present technology is not performing well and a robotic aid that functions appropriately would be useful in a clinical setting, although there is some initial resistance to using such a device in the

community. Physiotherapists indicated their computer skills were weak and this is likely to cloud the issue. Issues surrounding a requirement to feel joints and control the therapy were also reported. Patients viewed the system positively and believed it would speed their recovery. Based on the focus groups, a statement of requirements was written which is being used to develop the exoskeleton. This focuses on nine key areas:

1. Security – ensuring different levels of access to information, exercise routines, physiotherapy notes and so forth.
2. Patient Identification – ensuring that the correct exercise routine is performed on a specific patient.
3. Teach & Repeat Profiler – to enable a physiotherapist to manoeuvre a patient's leg through a cycle of exercises and for the system to store this information. This should then be repeated by the exoskeleton for the required number of cycles.
4. Exercise Monitoring – recording relevant data for reports to the patient and therapist.
5. Assessment Questionnaire – ensuring the 'wellness' of the user prior to and immediately after each exercise routine.
6. Treatment Report – providing relevant data to the patient and physiotherapist.
7. Patient Administered Treatment – under the overall control of the physiotherapist, enabling the patient to choose exercise profiles, and adapt them within defined limits to their requirements.
8. Patient Interface: "Deadman's handle" hardware – providing realtime information and encouragement to the patient, with appropriate safety controls such as pause and emergency stop.
9. Remote Controller Unit – enabling the system to work remotely irrespective of geographical location of the patient or physiotherapist.

TECHNICAL DEVELOPMENT

Hardware development

The focus groups indicated that in order to gain acceptance in clinical usage, at present, the system must function in a hospital setting and perhaps in the future be applied to the community. Consequently the device must be capable of providing therapy with a patient in a hospital bed, increasing the range of possible applications for which the device could be utilised. A number of geometries were investigated and that of Figs 1 and 2 chosen. While this increases the range of possible uses of the exoskeleton in a clinical setting, it increases the physical size of the device and therefore is likely to require future adaptation for home usage.

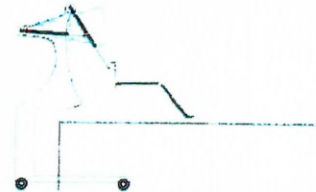


Figure 1: Geometry for hospital and home usage



Figure 2: Leg model for motion validation

The development of this configuration was based around the extensive use of kinematic modelling and analysis to ensure conformation with the desired range of motions. The control strategy was based around the concept of velocity tracking in which the error build up can be used as a performance indicator. The choice of pneumatic cylinders as the primary actuators also supported the ability to provide resistance to motion, an important component of some therapy strategies.

Software development

Clearly, while the technical performance and robustness of the hardware is important, if such a system is to be used effectively the user interface and operational aspects should be such that physiotherapists, who may not have highly developed computer skills, can easily operate the device. One of the major outputs from the focus groups was that physiotherapists felt that they would be unable to program an exercise using a software package. Therefore, a 'teach and repeat' function is being developed, enabling the physiotherapist to place the patient in the device, move the leg as required and record this movement. The exoskeleton can then replay this motion as

necessary for as long as appropriate. The exercise profile for that patient can also be saved and therefore enables the patient to undertake 'expert' physiotherapy without the direct supervision of the physiotherapist.



Figure 3: Example of the physiotherapist's interface

Prototype interfaces such as those shown in Figs 3 and 4 have been developed thus far, including short 'questionnaires' to be completed by the patient prior to and on completion of the exercise cycle. The design strategy used is one which is based around a modular software structure, enabling new modules to be added or changes made with minimum disruption. The system design is also such as to support the remote use of the system. In the short-term this is likely to be on a local area network within a clinical setting, perhaps in a 'super clinic' as suggested in the focus groups, but in the long term the ability to work over the internet, with features such as automated message passing and control of the system from a remote site could be easily included.

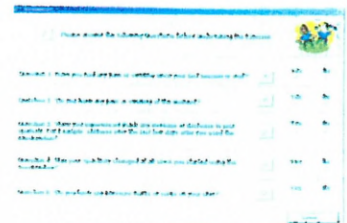


Figure 4: Example of user's questions prior to commencing therapy

DISCUSSION AND CONCLUSIONS

It was originally envisaged that the exoskeleton structure would be used in a home setting. This could enable extensive rehabilitation through an expert device and therefore suggests the possibility of earlier hospital discharge and improved clinical outcome. What we have discovered is that while this may ultimately be achievable, in the short term, the development of such a device would be counter productive – although technically possible. The user needs analysis indicated that service providers were reluctant to use such a device and the computing expertise required by professionals may not currently be appropriate.

This again highlights the importance of a detailed and effective user needs analysis, and the acceptance by developers to listen to this important aspect of development. The focus of the exoskeleton has therefore changed to enable home useage, but firstly to be operational in a hospital setting. It is hoped that by developing the exoskeleton in this manner that professionals will become familiar with its operation and begin, themselves, to push for its use in a community setting in future product developments.

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