The feasibility of enhancing postural stability using externally applied forces

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The Feasibility of Enhancing Postural Stability using Externally Applied Forces

by

C. Rodgers

A Master's Thesis

Submitted in partial fulfilment of the requirements for the award of Master of Philosophy of the Loughborough University of Technology

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Abstract

This report describes work carried out on a pilot study to investigate the feasibility of using externally applied forces to enhance the postural stability of standing patients.

The long term aim of the research is the development of a new treatment system for the rehabilitation of patients with a variety of postural or ambulatory disabilities. The concept is based upon applying directional forces to a patient by means of electronically controlled actuators, in order to provide the stability which they lack. Eventually the treatment system would incorporate a structure to support the actuators and patient allowing the training of stability in walking as well as standing.

The report covers all stages in the development of an initial prototype actuator system, covering fundamental design specifications through to commissioning and testing. Although rudimentary, some clinical tests are described and conclusions are drawn regarding the feasibility of the concept of externally enhancing stability. More exhaustive trials are left to a further study.
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Cadbury the Bear for comfort when times were bad
The EHB Bar for when things were really bad
## Contents

1. **INTRODUCTION**  
   1.1 Existing Therapy Regimes  
   1.2 Conceptual Treatment System  
   1.3 Prototype Evaluation System  
   1.4 Literature Search  

2. **CAUSES OF DISABILITY**  
   2.1 Loss of Limb  
   2.2 Progressive Illness  
   2.3 Stroke  

3. **SYSTEM SPECIFICATION**  
   3.1 Supportive Forces  
   3.2 System Measurements  

4. **THE PHYSICAL ACTUATOR SYSTEM**  
   4.1 Forceplate  
   4.2 Harness  
   4.3 Actuators  
   4.4 Transducers  
   4.5 Power Amplifier  

5. **ELECTRONIC CONTROL SYSTEM**  
   5.1 Force Control System  
   5.2 Controller Front Panel  
   5.3 Analogue Controller  
   5.4 Digital Controller  
   5.5 Safety Features  
   5.6 Computer Interface and Data Capture  
   5.7 Force Command Files  

Page  
1  
2  
5  
7  
8  
9  
10  
11  
15  
15  
21  
25  
26  
28  
30  
32  
34  
36  
37  
40  
41  
42  
45  
48  
51
# CONTROL STUDY AND SYSTEM COMMISSIONING

6.1 System Modelling
6.2 Evaluation Of System Parameters
6.3 System Analysis
6.4 Effects of Parameter Variations
6.5 Bandwidth Testing
6.6 System Accuracy and Commissioning

# DATA COLLECTION AND PRESENTATION

7.1 Data Capture
7.2 Presentation Techniques

# PATIENT TRIALS AND CONCLUSIONS

8.1 Methodology
8.2 Patients
8.3 Patient Trials
8.4 Conclusions

# APPENDICES

9.1 F241 Load Cell Specifications
9.2 Printed Armature DC Motor
9.3 A/D - D/A Interface Package
9.4 System Diagrams
9.5 Software Structure

# REFERENCES

10. REFERENCES

# BIBLIOGRAPHY

11. BIBLIOGRAPHY

# PROJECT PUBLICATIONS

12. PROJECT PUBLICATIONS
CHAPTER 1

INTRODUCTION

Significant medical resources and funds are taken up in the rehabilitation of patients having walking problems. These problems may exist either because of a specific neurological disorder or as a consequence of trauma. Hemiplegia is by far the most common disorder which leads to the need for intensive rehabilitation techniques, and recent statistics [1] highlighted the tremendous costs involved in the treatment of hemiplegics. In the United Kingdom 5% of National Health Service resources are used up in their rehabilitation, with associated bed occupancies of 13% in general medical wards and 25% in geriatric wards.

Physiotherapy departments are heavily involved in the process of rehabilitation, and again stroke patients are a major proportion of their workload. Although there is a continuing development of existing appliances such as crutches and walking frames, these seem to have a fundamental disadvantage in that they create an unnatural posture and gait, largely because the support forces are applied via the arms, and it is often necessary to strengthen the upper limbs prior to the actual process of walking rehabilitation.

The work described in this report is the first stage in a research programme aimed at a radically new form of therapy for walking rehabilitation, the essence of which is to provide stability by the use of carefully controlled forces applied externally at or around the patient’s centre of gravity. The mechanism by which these stabilising forces are applied will be under electronic control, and the patient will be able to exercise in a normal erect posture and with bipedal gait. The electronic control will enable the level of external stabilisation to be progressively withdrawn as therapy proceeds, thereby encouraging the patient to redevelop stability on his own.
Although the proposed system may be of benefit to any patient suffering from a temporary loss of reduction of balance and walking ability, its predominant application will be the walking rehabilitation of stroke victims.

1.1 Existing Therapy Regimes

Regimes of therapy exist designed to accommodate patients with widely varying levels of stability, ranging from people requiring minimal support to those who are unable even to stand. Most stroke sufferers will eventually regain walking ability, and much physical therapy involves exercising in some manner the muscles involved in ambulation. For a patient who in the initial stages is confined to a hospital bed, nursing staff will exercise the lower and upper body, stretching, extending, twisting, and flexing the limbs. This form of exercise eliminates a number of potential problems associated with medium to long term bed occupancy, however the advantage for future mobility is that upper-body strength is increased, muscle wastage is reduced, and the patient becomes re-accustomed to using and controlling his limbs.

Therapy generally then proceeds with more extensive exercises involving the patient being out of bed. At this stage the patient may still not possess any locomotive ability but is able to take a more active part in the therapy, performing exercises under their own strength with the guidance of a physiotherapist. These exercises will again involve stretching and flexing the limbs and may involve rudimentary balance and co-ordination exercises, but not necessarily in the standing position.

Once a level of competence is achieved in non-posture related exercises, it is natural to progress to exercises involving standing posture and walking. Here the patient is not only gaining the benefit of the physical exercise itself (which is usually very hard work for them), but is also having to learn about posture and gait, and must start to employ the various sensory feedback systems of the body (eg visual) to aid their walking ability.

For stroke victims having one side of the body affected by the trauma, this presents immense problems. Since sensation and response can be greatly reduced on the affected side, a patient automatically compensates by shifting weight to
the affected side, leaving the unaffected side free to generate compensatory movements to retain standing balance. This uneven weight distribution may well work in the standing position, but proves difficult when walking is attempted as walking inherently involves alternately shifting weight from one side of the body to the other.

To help patients in the initial stages of walking, one or two physiotherapists will give physical assistance, guiding the patient through the walking action and ensuring that the patient is protected from falling. Clearly this is extremely labour intensive. If the patient possesses sufficient upper-body strength they may be able to make use of apparatus such as parallel bars or walking frames, but again fairly intensive supervision is essential.

As therapy proceeds a patient using walking aids will gain confidence. The physiotherapist must then progressively steer them away from such equipment and encourage them to walk unaided, although this will not be possible in all cases. In the final stages of therapy with the patient having regained much of their original walking ability, exercises involve tasks that would be encountered in everyday life. Thus the occupational therapist would encourage the patient to climb steps, get in and out of a motor vehicle, carry weights etc., ensuring that the patient is able to cope with situations outside the hospital when expert assistance is not available.

The supportive equipment used in re-learning the walking process is invariably passive, usually consisting of some form of mobile supportive framework, or rigid fixed support, see Fig. 1.1. Although the use of such equipment is beneficial, most possess inherent drawbacks. For elderly people in particular it is not uncommon for a significant amount of time to be expended in simply strengthening the upper limbs prior to therapy, since a significant degree of upper-body strength is required to cope with the task of handling the various frames, crutches etc. Ironically it is often necessary during therapy to possess more upper-body strength than before the trauma occurred. Strengthening muscles for this process is a slow task, and clearly adds to the overall timescale required for recovery.
Fig. 1.1. Walking rehabilitation using passive walking aids to assist stability. Note unnatural body stance in each case.
Chapter 1 Introduction

Of more significance is the fact that most supportive devices force a patient to adopt an unnatural posture and gait, that is the body position and stature are different to that encountered during normal unaided walking. This is clearly visible in Fig. 1.1, and is due to the fact that supportive forces are applied via the arms (a wholly unnatural way to walk), hence the need for upper-body strength. This is often not fully appreciated, and means that a patient must further expend time and energy becoming acquainted with the characteristics of a particular piece of equipment. Indeed if an able-bodied subject were forced to use crutches, early efforts would be clumsy and poorly-balanced.

One further consequence of therapy involving static supports is that it is difficult to remove the assistance given by these devices in a progressive manner. Once a dependence has been formed it can be very difficult to remove this support. What is required of a patient is almost a quantum-leap in ability to achieve the step from supported to unaided walking. Bridging the gap between the two is difficult and invariably involves much time and effort and physical assistance from therapy staff.

1.2 Conceptual Treatment System

A conceptual system that proposes to overcome many of the problems associated with existing regimes and equipment is shown in Fig. 1.2. Ideally the apparatus would be attached by simple means to the trunk of a patient's body using a comfortable harness. The patient would then be given whatever level of support is needed to be able to walk and exercise in as near a normal manner as possible.

Horizontal and vertical support would be provided by actuators mounted at each corner driven by an overall control system. The same control system would also monitor the linear tracking movement of the apparatus and would have to respond to the patient's movements. If the patient walked forward, the system would detect the movement and react accordingly by powering the system in the correct direction, whilst still supplying supportive forces. For the system to be considered safe, the control system would have to be able to differentiate between a patient's normal motor actions in walking, and a potentially dangerous situation such as a fall.
Highly controllable dynamic supportive forces that could support the patient and react to the patient’s movements would be essential for the system to be of any real benefit. There are many advantages to be gained from such a system. The system may cope with people with a very wide range of stabilities. Being controllable the system could be made to give very positive support to an unstable patient, or could give minimal support to a patient with a good level of stability. Most patients would be expected to fall between these two extremes, but the controllability of the system would still allow them to be fully accommodated in terms of therapy.

A significant consequence of this level of control would be that, as therapy proceeds for a particular patient, the level of supportive forces generated by the system could be progressively reduced, perhaps over a number of weeks, encouraging the patient to increasingly depend upon and hence redevelop their own stability.

A further benefit is that all external forces may be applied at the patient’s centre of gravity, ie at approximately waist height. This allows the patient to adopt a normal unhindered standing or walking posture which is an ideal basis from which therapy can proceed. The stooping posture, characteristic of people using static supportive devices, can be eliminated. In addition there would no longer be a need to strengthen arms before therapy starts, allowing therapy to commence at an earlier stage in the rehabilitation regime.
Chapter 1 Introduction

An electronically controllable system could allow sophisticated safety measures to be implemented. Ideally these would not affect the patient in any way during normal operation, but would operate when a potentially harmful situation might occur. If a patient swayed too far to one side the system could detect the danger and could immediately respond with appropriate restraining forces to ensure that the patient does not fall, and is then allowed to recover. This would ensure the patient’s safety at all times, allowing more of the physiotherapist’s time to be spent on teaching and directing rather than simply providing physical support.

1.3 Prototype Evaluation System

The prototype system discussed in this report allows the feasibility of external postural enhancement to be examined in a physical system involving patients with balance deficiencies. It is not intended as a finished rehabilitation system, but may be utilised as a testing ground so that ideas and concepts relating to a longer-term solution may be examined and proven.

The hardware and associated electronic control system comprising the prototype system was developed in the Department of Electronic Engineering at Loughborough University in collaboration with the Orthotics and Disability Research Centre at Derbyshire Royal Infirmary. Initial testing of the system was performed at Loughborough. The system was then transferred to Derby for software development and further testing before trials commenced involving able-bodied subjects.

Fig. 1.3 shows the prototype system in operation, in this case with an able-bodied subject. To keep the system relatively simple, no vertical support is given to the patient. Horizontal forces are generated by two orthogonally-positioned actuators, mounted horizontally and attached to the patient by a harness at waist level. Transducers in the system monitor the position and speed both of the patient’s centre of gravity (at waist height), and centre of pressure at floor level. Forces acting upon the patient can also be monitored.
An analogue control system allows both static and forces to be generated. A digital computer and controller is responsible for system management, and data collection and analysis. The computer may also be used to generate a standard series of changing force patterns for each patient, to allow valid comparison of data from different patient trials.

The levels of static and supportive forces are finely controllable so tests may be performed using varying levels of support over a period of time, allowing any improvement in the patient’s stability to be detected.
1.4 Literature Search

A literature search was performed at the start of the 20-month project and was repeated mid-way through as development of the prototype system progressed. Two things were initially considered important - what methods had been developed to quantify a patient's level of stability, and what models if any were available that could be adopted or modified to represent the response of a standing patient subjected to external forces.

Numerous publications existed on both topics. A large number of related posture models were revealed [5,7,8,10] but it became clear that each particular model examined had been developed to suit a specific area of interest - each model was quite correct in itself but it represented only those features directly relevant to the research. Thus for example if twisting occurred in an arm between two joints but this had no significant effect upon the system's performance, it was ignored in the model by representing the arm as a rigid connection. This has the effect of simplifying the overall model and ultimately making the task of predicting the behavior of the system easier.

No models were found that proved suitable to the work undertaken in this pilot study, but the ideas of simplicity and relevancy are adopted in the models featured in Chapter 3 and Chapter 6 to predict static and dynamic responses of a standing patient. Each model is different and quite specific and each is kept simple by only considering features that are relevant to the analysis. The models are used both to allow initial system specification to take place and to predict subsequent system performance and response.

A number of different techniques were found to exist designed to quantify a patient's level of postural stability [6,9,11,12,14] and these were found to have varying levels of accuracy and practicality. The better analysis methods involved patients standing on a force-measuring platform [13] that could yield centre of pressure information. When used in conjunction with a computer a number of powerful analysis techniques were achievable. The computer was employed to sample and store real-time data and subsequently analyse and display it in various forms.
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Some of the documented analysis and display techniques are adopted in this report and results obtained from this research may be directly compared with results obtained by other groups since a number of the analysis and display techniques have been widely adopted. The sophistication of the prototype system has also allowed other analysis techniques to be developed. Whilst much information has been recorded about the behaviour of a person’s centre of pressure point (derived from a force-sensing platform), no accurate information was found concerning measurements taken around the patient’s centre of gravity. Using the analysis and display techniques described in this report it is possible to use the same analysis procedures to examine the behaviour of centre of pressure and also centre of gravity points in free-standing trials, and also during tests involving external forces or disturbances. Results from both of these sources give a much clearer understanding of how a patient controls his posture. The ability of the system to generate accurate data from both sources under repeatable test conditions has been found from the literature search to be unique.

The following chapter examines causes and extents of disabilities in patients who are likely to benefit from using the conceptual treatment system as part of a rehabilitation therapy routine.
CHAPTER 2

CAUSES OF DISABILITY

It is envisaged that the prototype system discussed in this report will be able to generate therapy routines beneficial to patients having short-term balance and walking deficiencies. The eventual proposed form of therapy is not rigorously discussed - it is the feasibility and technical engineering aspects involved in designing a working prototype system that are examined in detail.

Despite the clear emphasis on the engineering solution, clinical aspects are considered at every stage in the design process. Before initial system design specifications may be attempted it is necessary to examine the background and causes of patient disabilities, and to consider ways in which existing therapy routines and programmes may be enhanced by the proposed system. Consequently various forms of trauma causing disability are discussed, and in each case the suitability of the proposed eventual system to provide useful assistance in therapy is reviewed.

2.1 Loss of limb

The problems associated with the loss of a limb are both physical and psychological, and both aspects will affect the extent and speed of recovery. A patient must recover physically from the accident or trauma making amputation necessary, and must then adapt to the inherent problem of restricted mobility. Psychological problems can be enormous. Amputation can cause depression, loss of interest, a reduction in motivation and will-power, and many amputees attempt to isolate themselves. Coupled with an envisaged reduction in mobility and potential loss of prowess, a patient's psychological problems can be the most difficult factor in recovery, requiring expert and possibly long-term counselling.
However despite the enormous problems encountered by amputees, it is significant that the motor faculties possessed by the patient before the trauma are unimpaired - since the brain and physical nerve pathways to limbs are undamaged the only obstacle in the way of the patient recovering a normal walking function is the obvious physical handicap and potentially psychological problems, eg loss of will-power.

These problems can be overcome given time, and it is considered that a therapy routine utilising the conceptual actuator system could prove beneficial to lower-limb amputees. In particular it is expected that the system's ability to provide restrained support and respond to intentional movement would allow an amputee to progress more rapidly during walking therapy, and hence regain a more advanced level of walking competence at an earlier stage.

2.2 Progressive Illness

There are a number of conditions that are progressively debilitating, making it increasingly difficult for a person to retain a level of physical mobility. Diseases such as muscular dystrophy and multiple sclerosis are typical, either causing muscles to waste directly, or affecting the central nervous system and leading indirectly to muscle wastage.

Diseases of this type are suspected of being hereditary, although little is known about how and why they occur. The best that may be achieved with drugs is a slowing down of the effects of the disease. The diseases are progressive in that in the later stages a sufferer may possess muscles so weak that they are unable to move around without physical help and are likely to require use of a wheelchair. One thing that is significant in helping to check diseases of this type is regular (but not vigorous) exercise. Massage and swimming are frequently employed as methods of exercise to keep muscles active and prevent excessive muscle degeneration.

The conceptual system discussed could be used as a powerful exercise tool. The system could be set to provide sufficient support for the patient to be able to exercise without fear of instability, and the adaptability of the system would allow it to be 'tailored' to suit a wide range of disabilities. The system would
would also be capable of assessing and monitoring a patient's level of stability in the initial stages, and could be used to detect trends in the longer term.

It is also possible that the system could take a more active role in a patient's exercise by generating a series of forces and movements designed to cause the patient to respond in a prescribed manner. Thus it might be possible to help the patient exercise in a particular way, using carefully designed exercise schedules.

Due to the progressive worsening of diseases such as multiple sclerosis and muscular dystrophy, it is likely that the conceptual system could play an important part in exercise therapy for sufferers, but it is unlikely to be of use in their long-term rehabilitation.

2.3 Stroke

Despite the widespread occurrence of stroke, few people understand the cause of the trauma, though many are familiar with the subsequent physical manifestations. Stroke sufferers can be affected in many different ways and to differing levels of severity. Mentally a patient can become confused and disoriented. Physical effects can range from minor coordination problems to full body hemiplegia. Speech and sight can also be affected. The extent and rate of recovery is also highly variable and will depend upon many factors.

The three case studies below illustrate the variability of stroke severity and recovery.

Case Study 1. A 67-year old woman suddenly felt dizzy whilst standing at the kitchen sink; the room seemed to go round and round and this sensation (vertigo) persisted for an hour. She felt nauseated. Three hours later, the vertigo returned and she had great difficulty in walking; it felt as if her left side was not properly co-ordinated. Her husband took her to a hospital casualty department, where she was fully conscious but still complaining of vertigo. Her speech was slightly slurred although its content was quite normal. Although her vision was unimpaired, eye movements were noted to be abnormal, the eyes slowly moving back to a central position before quickly flicking back again. She had weakness of the lower half of the right side of the face but there was no weakness in the left arm or leg, although tone was slightly reduced. She found it was impossible to perform rapid movements of the hand and, when trying to touch her nose with her left forefinger, the incoordination of the left hand was obvious. When she tried to stand, she had difficulty balancing and could only walk with assistance.
Chapter 2 Causes of Disability

The patient in this case had a stroke affecting the brain stem. This type of stroke is rare and is frequently fatal. Those that do show any sign of recovery usually do very well.

**Case study 2.** A 24-year old nurse was admitted to hospital whilst on holiday in Greece with headache and a general feeling of malaise. Within 48 hours, she developed obvious signs of a left hemiplegia. She had been on the contraceptive pill for 3 years. There was no family history of vascular disease. She was transferred back to this country and a CT scan confirmed the diagnosis of a right cerebral infarction. The patient made steady progress and the eventual outcome was excellent, although she still had residual weakness of the left foot. She returned to her former nursing duties after 9 months.

The contraceptive pill, particularly the high-dose oestrogen pill, has been linked with an increase in blood viscosity which in turn can increase the possibility of thrombosis (clotting).

**Case study 3.** A 60-year old taxi-driver was admitted to hospital after a stroke had caused weakness, but not total paralysis, of his left arm and leg. After four weeks he was transferred to a rehabilitation centre, and at this time he could walk without a stick, but still had difficulty with hand movements and weakness of the shoulder. After three weeks at the centre he was discharged home and asked to attend three times a week for out-patient physiotherapy and occupational therapy. His only remaining problem was the shoulder weakness and fine movements of his left hand. He was completely independent in self-care, and could travel on public-transport.

Three months after his stroke, he tried driving his taxi and had no problems, but he could not manage one activity essential to his trade - he could stretch out his left arm but could not flick down the taxi's clock meter which he would have to do when he was operational again. It was almost six months after the stroke before he was able to do this and only then could he return to work.

The stroke in this case was relatively minor, having more effect upon the upper left side of the body than the left leg. Although walking was affected in the early stage, recovery of mobility for this patient was very rapid.

The three case studies involve strokes of quite different types. In each case one of the immediate effects of the trauma is a fundamental reduction in the balance and walking ability of the patient. Not all strokes generate this kind of disability, but the great majority of stroke sufferers are affected in this way.

All strokes are caused by a cessation of the blood flow to an area of the brain. Each blood vessel in the brain supplies blood only to a particular part of...
the brain, so if a blood vessel becomes blocked the brain tissue it serves is immediately deprived of blood and hence its oxygen supply. This invariably causes a cerebral infarction, meaning that affected brain tissue dies due to lack of blood.

If a blood vessel haemorrhages, the situation is similar but usually more severe, since blood escapes through a weak spot in the blood vessel wall and damages widespread brain tissue by compression. In addition the area of brain supplied by the ruptured artery may receive insufficient blood, causing further damage to the brain.

This is a very simplistic view of a highly complex subject, however two points are significant. Firstly since almost any blood vessel in the brain may be damaged, a victim may be affected in many different ways and to varying extents depending upon the location and severity of the damage. Hence stroke victims will possess widely varying disabilities. Secondly for the great majority of stroke sufferers some degree of recovery is possible, as is seen in each of the case studies. Although an area of the brain will have died, the functions that it controlled can invariably be re-learned. This is fundamental to all rehabilitation regimes.

The structure of the brain is such that if a cerebral infarction occurs on one side of the brain it is the other side of the body that is affected. Thus if the left side of the brain is damaged by stroke, movement of the right arm, leg, and face can be disturbed, along with the right field of vision. A patient that suffered such would be termed a right hemiplegic. In the majority of people the speech controlling centers are on the left side of the brain so a right hemiplegic may also experience speech problems. Hearing is interpreted on both sides of the brain from both ears and hence is not damaged by a stroke.

Some strokes are so slight that full recovery is possible within just a few weeks - they may even occur with the patient noticing no more than a slight dizziness. Others are so severe that little long-term recovery can be expected. Between these two extremes lie most stroke sufferers.
Chapter 2 Causes of Disability

Most people who suffer any degree of disabling stroke experience a weakness of the arm, hand, or leg on one side of the body, combined with a numbness and clumsiness in the affected limbs. Balance may also be reduced. When these symptoms occur the walking action that once had been totally natural is severely affected. In all cases of hemiplegia resulting from stroke the level of eventual recovery will depend upon many factors, the most significant being the severity of the stroke itself. Most stroke sufferers may be expected to recover much of their original mobility and physical capabilities, but may experience some degree of difficulty performing certain tasks.

Given the nature and characteristics of the disability suffered by stroke patients, and the existing rehabilitation regimes, it is believed that the conceptual system discussed in this report would prove beneficial as part of a coordinated therapy routine. Specifically the ability of the system to provide widely differing levels of support, and to gradually reduce this support over a period of time links the system closely to the real-life recovery pattern of stroke patients. It is considered that the assistance given in this way would be of significant benefit in the rehabilitation of a wide range of stroke sufferers.
CHAPTER 3

SYSTEM SPECIFICATION

The most basic requirement of the prototype system is that it must be able to generate and precisely control forces to enhance the stability of a patient. Ideally the resultant compensating force generated by the system should be in the form of a three dimensional vector, having a highly controllable magnitude and direction, and being able to supplement a patient's own stability efforts. Thus it should be possible to compensate in both the anterior/posterior and medial/lateral directions, and also generate support vertically. In reality control of the vertical component would add greatly to the complexity of any controlling system since it would be necessary to apply vertical forces in a minimum of three places. This would necessitate three force actuators and a sophisticated controller linking the three actuators to generate controllable coherent forces.

To simplify the overall system design the feasibility of using external forces is only examined in standing patients - ie postural, not supportive stability, is generated. This necessitates the restriction that only patients having the ability to stand unaided may take part in stability trials.

3.1 Supportive Forces

Assuming that a patient has the ability to support himself or herself in the standing position (but not necessarily balance unaided), supportive forces may be applied in two dimensions, there being no requirement for vertical support. A two-dimensional force vector may be resolved into two orthogonal force components, so any two-dimensional force may be generated by two force actuators mounted in the same plane but at right angles. This is illustrated in Fig. 3.1.

For supportive forces to be of most benefit to a patient they should be applied at or around the patient's centre of gravity. This point typically occurs in the
centre of the body at the height of the iliac crests (approximately at the point where the hip bones slightly protrude).

Two force actuators are employed to generate the stabilising force, one generating side-to-side forces (medial/lateral direction), and the other generating front and rear forces (anterior/posterior direction). This is illustrated in Fig. 3.2. This affords a conceptually simple system, but more importantly it allows a patient's stability performance to be easily assessed independently in each of these directions by using one or other of the actuators. This configuration also allows the patient to assess and respond to directional forces more easily, ie those coming from the side, or those from front and rear.

Since forces are required to be applied from both medial/lateral, and anterior/posterior directions, each actuator must be capable of applying positive and negative forces, ie they must be able to generate forces both pushing and pulling. The speed of response of each actuator must also be sufficiently fast to accommodate the fastest expected movements performed by a patient. The response and bandwidth of the actuators and control system is discussed in Chapter 6.

A system that allows forces to be applied in two dimensions is simplistic but allows rudimentary enhancement of a patient's stability. The advantage for the overall system is that it is only necessary to generate two force vectors, so only two actuators and associated controllers are required.
Chapter 3 System Specification

With two force actuators mounted horizontally at right angles, a preliminary specification can be assumed for the medial/lateral, and anterior/posterior actuators:

- Maximum displacement: +/-0.15m
- Maximum speed: 0.3m/s
- Maximum static force: 100N
- (Average force: 30N)

The displacement and speed parameters are derived from simple tests. The figure for the maximum force relates to the opposing force required to stabilise a $10^\circ$ lean of a 60kg patient - i.e. a lean that could result in a fall if left uncorrected.

To enhance the postural stability of a patient the controller associated with each actuator must be capable of generating static forces and also forces of a supportive nature. For a force to be supportive it must supplement the patient’s own stability, ideally only when required. For example when a patient leans too
far in one direction the controller should respond by making an actuator generate a progressive compensatory force, increasing with displacement. If a patient moves too quickly it should also be possible to restrain by generating a force related to the patient’s speed. These compensating forces can be likened to spring and damping forces, ie they should increase with increases in displacement or speed.

These spring and damping response characteristics are shown in Fig. 3.3. For the displacement profile (a), there should ideally be a dead-band region where the patient may sway and perform movements necessary for balance with no

![Diagram](image)

**a)** Support related to sway magnitude.

![Diagram](image)

**b)** Support related to sway speed.

Fig. 3.3 Ideal supportive force characteristics.
assistance from the system. If the patient were to move out of this ‘safe’ region, perhaps by swaying too far forward or to one side, the system must produce a progressive stabilising force opposing the patient’s direction of movement. A patient with poor stability would require this to happen at smaller excursions from the centre balance position than with a well-balanced patient.

It must be possible to vary the characteristic to accommodate widely varying levels of patient stability. This variation in the characteristic is indicated by the shaded region in Fig. 3.3(a). Control of the characteristic also allows the supportive assistance generated by the system to be progressively reduced as a patient becomes more competent, encouraging patients to develop their own stability without receiving more assistance from the system than is necessary.

The situation is similar for the speed (damping) profile shown in Fig. 3.3(b). Again there should ideally be a dead-band region within which the patient is free to perform balance movements without any help from the system. This dead-band region does not relate to a physical area as in the case of the spring profile, but relates to the speed of the patient - if the patient performs slow movements as might be the case for a well-balanced person, he remains in the dead-band region of the graph. However, if movements become rapid as might occur in a fall, the patient enters the shaded region on the characteristic and the system generates an opposing force increasing in relation to the patient’s speed of sway - ie the system generates damping.

The threshold at which supportive forces are generated must be variable, and this corresponds to the shaded region in both the characteristics. For patients with poor stability the characteristic is likely to lie on the inner-edge, implying a small dead-band region. A patient with good stability might use the characteristic on the outer edge, or may even require no assistance. The shaded regions between these two extremes contain the range of characteristics that may be set to accommodate and assist patients with a wide range of balance deficiencies.

The displacement (spring) and speed (damping) characteristics shown in Fig. 3.3 may be approximated to the characteristic shown in Fig. 3.4. This is a linear characteristic where the level of opposing force generated by the system is
directly related to the displacement or speed of the patient. This approximation makes it very straightforward to generate these supportive forces within the actuator control systems. The penalty is the loss of the dead-band region, however since the approximating characteristic is linear, supportive forces generated by the system will be very small in this region.

The magnitude of the restraining spring force may be derived from simple geometry. The maximum horizontal static force needed to restrain a leaning patient will relate to the patient's weight and height. For a 60kg patient with centre of gravity 1m above floor level, this force will be zero with no lean and will increase with the lean angle (Fig. 3.5). The sideways force characteristic will relate to the tangent of the lean angle, but for small angles the force/displacement characteristic will be approximately linear. This gives a
stiffness figure of 600N/m to counteract the patient’s basic instability. The actuator system must be capable of generating greater stiffnesses, and a figure three times this level was chosen as an initial stiffness specification.

A figure for the maximum supportive damping force must be derived by speculation. If a patient were to move at the maximum speed the actuator system allows, i.e., 0.3m/s, it is feasible that the system may need to arrest the patient’s speed very rapidly - certainly within one second and more likely a factor of ten times faster than this. Working with this latter figure yields a speculative damping figure of 200N/ms$^{-1}$ for a 60kg patient. Again a figure in excess of this was adopted in the final specification.

Given these criteria and assumptions, each actuator must be able to generate the following types and levels of stabilising force:

a) A steady force independent of the patient’s movements in the range -100N to +100N

b) A force to give a spring characteristic in response to the patient’s displacement, variable from 0 - 2000 N/m.

c) A force to give a damping characteristic in response to the patient’s speed, variable from 0 - 500 N/ms$^{-1}$.

3.2 System Measurements

The system must have the potential to assess and quantify a patient’s level of stability. This is required initially to ascertain what level of support is necessary, and in the longer term to monitor progress as therapy proceeds.

Various well-documented methods of measuring postural stability exist [2][3][4] and some are discussed in a later chapter. Many assessment methods involve monitoring over a period of time either the patient’s centre of pressure or centre of gravity. The concept of centre of gravity of the body has already been discussed. The centre of pressure developed by a person in the standing position relates to the way in which the person’s weight is distributed at the person’s feet. When standing unaided and motionless, the centre of gravity point may be projected directly downwards and will correspond to the centre of
pressure point at the person’s feet. This is the case even if the person is leaning significantly - provided that the person is not moving these two points will correspond (Fig. 3.6 (a)).

When the person moves, the centre of pressure and centre of gravity points will move apart. If one foot is quickly lifted slightly from the floor the centre of gravity point will not move significantly, but the centre of pressure point must move beneath the foot still remaining on the floor, since this is the only point at which pressure may be applied. The patient must then rapidly adjust his body position, either placing his foot down or leaning to one side to move the centre of gravity back above the centre of pressure, otherwise imbalance occurs. This is an important condition of balance - that the two points must be closely related in the vertical plane otherwise instability and toppling will occur.

If a side force is applied to the person the two points will again move apart. If the subject resists the force and remains standing in the same position with the same posture, the centre of gravity point will not move. However to oppose the
force the person must cause a force reaction with the floor, and this will cause
the centre of pressure to move (Fig. 3.6(b)). Effectively the person is relying
upon friction in the floor to give sufficient grip to maintain his position and
oppose the force. If the sideways force is increased, causing the centre of
pressure point to move outside the person's foot profile, the person will topple
over.

Centre of gravity information may be derived easily from the position of each
actuator shaft. Centre of pressure data cannot be derived directly from the
actuator system, so the patient must stand on a force sensing platform commonly
known as a force plate. Detailed information can be derived from a force plate
about a patient's position and stance, not only in a static standing position, but
also dynamically as a patient moves. The information generated by the force
plate can be processed to give centre of pressure data for a standing patient
whilst connected to the actuator system.

Fig. 3.7 Signals generated and sampled by the actuator system.
Fig. 3.7 indicates what signals are generated and used in the system. These signals are:

a) Force Fx and Fy on the patient at waist level (hence acting approximately at the centre of gravity).

b) Displacement Dxg and Dyg of the patient’s centre of gravity.

c) Displacement DxP and DyP of the patient’s centre of pressure at floor level.

All of these signals appear in orthogonal component form - ie as an x and y component. From the displacements b) and c) the velocity and acceleration of the patient’s centre of gravity and centre of pressure may also be derived.

The method by which forces are generated and signals are monitored is described in the following chapter.
A schematic diagram of the overall actuator system is shown in Fig. 4.1. There are two sub-systems - an actuator sub-system, and a computer sub-system. The actuator sub-system consists of the actuators themselves, power amplifiers to drive the actuators, transducers, and an electronic controller. The computer sub-system consists of a processor and associated devices, and an interface to the physical system. The force plate necessary to detect ground reaction data is interfaced to the processor.

This chapter details the physical hardware comprising the actuator sub-system. The computer sub-system is described in Chapter 5, along with the electronic control system that links all of the various elements.
Chapter 4 The Physical Actuator System

4.1 Forceplate

The force platform used with the system is embedded at floor level. Having the force plate flush with the surrounding floor makes it convenient for patients, since there is no step up, allowing a normal unconcerned posture to be adopted. The platform consists of a solid metal plate approximately 60cm x 40cm x 5cm (24" x 16" x 2") mounted on four load cells, one near each corner, see Fig 4.2. None of the load cells is visible from above.

![Image of a force plate with load cell positions and dimensions.](image)

Fig. 4.2 Kistler Force Plate showing physical dimensions and load cell positions.

Referring to Fig. 4.3, the total downward force exerted by the patient can be derived from the sum of all four load cell signals:

\[
\text{Total Force } = \Sigma F = F_A + F_B + F_C + F_D
\]

For a force plate with load cells located at the exact corners, Fig. 4.3 (a), the X and Y components of the patient's centre of pressure are:

\[
X = \frac{(F_A + F_B) - (F_C + F_D)}{\Sigma F}
\]

\[
Y = \frac{(F_B + F_C) - (F_A + F_D)}{\Sigma F}
\]

The variables X and Y are ratios in the range -1 to +1 and may be scaled by the physical force plate dimensions. With each load cell located within the
external boundaries of the force plate, as in Fig. 4.3 (b), the dimensions $x_1$, $x_2$, and $y_1$, $y_2$ must be taken into account. Thus:

\[
X = \frac{(F_A + F_B) - (F_C + F_D)}{\Sigma F (1 + x_1/x_2)}
\]

\[
Y = \frac{(F_B + F_C) - (F_A + F_D)}{\Sigma F (1 + y_1/y_2)}
\]

For the force plate dimensions given, the $X$ and $Y$ centre of pressure components resolve to:

\[
X = \frac{(F_A + F_B) - (F_C + F_D)}{1.67 \Sigma F}
\]

\[
Y = \frac{(F_B + F_C) - (F_A + F_D)}{1.5 \Sigma F}
\]

These equations are used to convert the four force plate load cell signals into two signals, representing in Cartesian form the centre of pressure of the patient.
4.2 Harness

There are a number of important constraints governing the design of the harness. It must be comfortable to wear and yet provide a reasonably rigid connection to the body around the centre of gravity. It must also be easy to get on and off, and must connect simply to each actuator when commencing and finishing a patient trial.

It is crucial that the harness applies forces firmly to a patient’s centre of gravity. Rigidity of the harness must be compromised with patient comfort. By careful choice of materials it is possible to ensure that any force on the harness is transferred firmly to the patient, whilst leaving the patient comfortable.

Provided that any twisting or displacement movement made by the patient is small, it is valid to apply horizontal supportive forces at only two points, using two actuators mounted at 90° (Fig. 4.4 (a)). If displacement or twisting becomes excessive the geometry of the system would mean that either the forces are no longer orthogonal (Fig. 4.4(b)), or that undesirable torques are generated (Fig. 4.4(c)). The assumption that the patient only causes small movements of the harness is reasonable. If one considers a person wearing a belt around the waist, as the person walks normally, twisting and side to side movements of the belt are small - the belt stays approximately horizontal, and only minor twisting occurs. In the standing position these movements are considerably smaller, so that during a patient trial forces applied at two points at right angles approximately retain their orthogonal nature.

Fig. 4.5 shows the harness design adopted. It is conceptually simple and overcomes many of the problems associated with connecting patients to the actuator system. The padded interior sections are of rigid Plastazote (a very firm sponge material) and are contoured to fit firmly around the iliac crests. The plastazote ensures that a patient is comfortable and maintains a reasonably non-elastic connection between the patient and each actuator.

To fit the harness to a subject it is simply split into two halves and offered around the body at waist level, and then brought together until firm and comfortable. A ratchet arrangement in each harness arm allows the four sections
to be tightened with minimal effort. A spring lever releases the ratchets to allow for removal. This process is very rapid and is intended to minimise patient stress.

Two spherical bearings on the harness allow it to be connected to the actuators by means of spring clips. The bearings take up any minor twisting that a patient may perform, as well as accommodating minor changes in geometry when the actuators are moving.
4.3 Actuators

Two electric actuators are used to exert the carefully controlled forces necessary for patient support. The basic design of each actuator is shown in Fig 4.6. At one end of the body is a small electric motor. This directly drives a lead screw running the full length of the actuator. Riding on the lead screw is a high-efficiency recirculating ball nut which is connected to the moving shaft, moving in and out of the actuator body. Applying a current to the motor generates rotation in the lead screw, causing the shaft to move inwards. Any rotational movement of the motor is transformed into a linear movement of the actuator shaft. This in turn can be used to generate a linear force. By varying the current through the actuator motor it is possible to alter the torque applied to the lead screw and hence the resultant linear force applied via the moving shaft.

The motor current is not a good indicator of the resultant output force due to the dynamic nature of the system. Consequently to allow the system forces to be finely controllable, a load cell within each actuator is incorporated into a closed loop control system, with force as the command. In this way the precise forces necessary for stability trials can be generated and monitored.

Fig. 4.7 shows an actuator mounted on a framework as it would be used with a patient trial.
Chapter 4 The Physical Actuator System

Fig. 4.6 Illustration of internal design of actuator, incorporating displacement transducer in the main body.

Fig. 4.7 Photograph of actuator mounted on gimbal stand, illustrating the degree of vertical and horizontal movement available.
Each actuator pivots freely in a gimbal arrangement allowing a limited degree of horizontal and vertical movement. This allows the patient sufficient freedom to move and twist a small amount as might be expected in the standing position. The small degree of freedom also ensures that the system can cope with patients who do not distribute their weight evenly - i.e., they may lean heavily to one side causing one actuator to move horizontally. The entire actuator and gimbal arrangement may also be wholly shifted vertically, ensuring that for a wide range of patient heights, forces may still be applied horizontally to the standing patient.

4.4 Transducers

Each actuator incorporates two transducers, one measuring displacement, and the other measuring force. The displacement transducer allows the position of the moving shaft to be accurately monitored, and a force transducer (load cell) generates a signal proportional to the force exerted by an actuator which corresponds to the sideways force acting upon a patient.

The force transducers (load cells) are sandwiched between the driving actuator and any mass or body attached to the system. Fig 4.8 indicates how each load cell is connected rigidly to the end of the actuator shaft. The sensing element of the load cell connects to a spherical bearing and this clips simply onto the patient harness. The bearing accommodates any minor twisting of the

![Diagram of actuator-harness connection, showing load cell, flexible rose joint, and detachable coupling.](image)

Fig. 4.8 Actuator-harness connection, showing load cell, flexible rose joint, and detachable coupling.
patient harness whilst keeping firm the connection between actuator and harness. Due to the physical position of each load cell, a signal will be produced proportional to the actual force applied to the body. Technical details relating to the load cells are given in Appendix 1.

The position of each actuator shaft is determined by using two infra-red detectors and a circular slotted disc. These components are mounted within each actuator body, the disc being mounted on the actuator motor shaft, and the detectors positioned close to the disc (Fig. 4.9). Any rotary movement causes the disc to cut the infra-red beams intermittently. The detectors are spaced apart by a constant multiple of the disc slot pitch, plus or minus a quarter of the pitch. This spacing enables direction to be determined. The pulse trains generated by the detectors contain sufficient information to determine the direction and degree of rotation of the actuator shaft. Logic circuitry is used to decode this information to produce a digital position count, and this subsequently generates an analogue position and velocity signal. The decoding methodology and circuitry is described in Chapter 5.

Incorporating the transducers described above into the control system allows constant monitoring of the static and supportive forces applied by each actuator, as well as the position and velocity of each actuator shaft.

Fig. 4.9 Optical transducer used to derive position of actuator shaft.
4.5 Power Amplifier

Each actuator incorporates a motor and lead screw arrangement, and by powering the motor, linear forces can be generated at the actuator shaft. To drive the motor a commercial power amplifier is used. Fig. 4.10 shows a schematic diagram of one amplifier unit. Two independent amplifiers are required, one being used for each actuator, each having an analogue command signal input and a number of control inputs.

The command input takes the form of a bipolar voltage signal (generated by the electronic controller) and this dictates what current is fed to the motor. This in turn controls the motor torque and consequently the linear force generated by the actuator. Current feedback is used internally in the power amplifier to provide tighter control of the motor current. Digital inputs provide limited control of amplifier function and also allow the implementation of safety features.

Fig. 4.10 Amplifier unit showing analogue command input and digital interlock signals.
Each power amplifier can only operate in one of three states:

**ENABLED:** The amplifier is fully operational, accepting current commands from the electronic control system and generating appropriate currents in the actuator motor. In this state the actuator will be capable of generating supportive forces to enhance a patient's stability.

**DISABLED:** The amplifier no longer drives the actuator motor, having the effect of 'freezing' the operation of the actuator. Each actuator may still be moved but a large external force is required to overcome inertial forces and friction. When the system is in this state it effectively provides very strong support for a patient, and is an inherent safety feature. The amplifier may be switched between the ENABLED and DISABLED states by using the two enable control inputs (these are controlled by the electronic controller).

**FAULT:** If an internal fault is detected by the amplifier unit, all functions are disabled and a fault control signal is generated. The system adopts the disabled (ie stable) state and becomes locked in this state. The cause of the fault must be corrected and the system must be powered-down and re-started. A fault condition will occur if current, voltage, or speed limits are exceeded.

The digital enable inputs allow control of amplifier operation. Only when both signals are present can the amplifier be enabled. Directional control is also provided - when either direction inhibit control is disabled the motor may only rotate in one direction. Since the position of each actuator shaft is known, this feature is employed to provide electronic end-stops by disabling the amplifier when actuator travel approaches its limit. Both actuator amplifiers are housed in the same cabinet, and indicators show in which state each actuator is operating. Control and drive signals are fed to the unit via a 25-way D-type connector from the electronic controller.

This chapter has discussed in detail the physical components that comprise the actuator sub-system. The electronic controller that links each component is discussed in the following chapter.
CHAPTER 5

ELECTRONIC CONTROL SYSTEM

The electronic controller forms the main element of the actuator sub-system (Fig. 5.1). It is responsible for implementing the control system function and monitors the status of the system continuously. The controller also communicates with the computer sub-system, allowing the computer to sample and store system measurements and generate force command profiles during patient trials.

Fig. 5.1 Configuration of electronic controller within actuator sub-system. Shaded areas represent those parts of the electronic controller that are implemented digitally.
Each actuator requires its own independent electronic controller and each of the two controllers contains analogue and digital circuitry. The analogue section implements the force control system, and the digital section allows supportive forces to be generated. System management and safety features are implemented with digital circuitry.

5.1 Force Control System

A diagram of the electronic force control system associated with each actuator is shown in Fig. 5.2. The controller consists of two distinct sections - a force loop, and a command loop. The force loop controls in real time the operation of an actuator, monitoring the resultant output force and correcting any imbalance between this and the desired force. The command loop generates a

![Fig. 5.2 Schematic diagram of force control system.](image-url)
force command signal which is applied to the force loop. The force command may be a non-varying (static) force, a force related to an actuator’s position (i.e., a spring characteristic), or a force related to the actuator’s speed (i.e., a damping characteristic). Each type of command may be varied between specific limits and may be applied in isolation or in combination.

The force control loop has two inputs, a force command signal (generated from the command loop) and a signal directly from the actuator load cell. The force command signal represents the desired force, i.e., the force that is to be generated, and the load cell signal represents the actual measured force generated by the system.

With no force command (i.e., force command signal at zero volts) the system will attempt to maintain approximately zero force on an actuator load cell. If a force is applied externally, i.e., if someone connected to the system were to lean towards an actuator, the force signal generated by the load cell feeds through the system and energizes the actuator motor, causing the actuator shaft and load cell to move away from the force, and hence attempt to reduce the resultant applied force to zero. Under these conditions very small forces are required to move the shaft of the actuator - the control system is made to do the work. The implication for anyone connected to the system is that if they lean or move around, each actuator will move in sympathy with them. Since the gain of the system can be set to be high, the force required by the subject to move each actuator is very small - too small to be noticed physically. Consequently even though being connected to the system, with no electronic force command present, a subject can be considered as standing unaided - i.e., the system is neither aiding nor hindering the balance and standing ability of the patient. This is an important aspect of the system, as it allows a person’s standing ability to be monitored without the system causing significant effect.

The command loop shown in Fig. 5.2 can be used to introduce static forces into the system. Static forces merely involve the introduction of a fixed force command signal into the force loop. The system will endeavour to ensure that the difference between this fixed demand and the actual measured force is close to zero by dynamically shifting the position of the appropriate actuator.
static force will be maintained even if a subject moves around, due to the
dynamic nature and response of the force control loop.

The displacement transducer in each actuator is used to generate a signal
proportional to the position of each actuator shaft, and hence the physical
position of the patient. Digital pulses are generated when any actuator motion
occurs, and these are processed and scaled to produce an analogue signal
proportional to the linear position of an actuator shaft. By using this signal as a
force command a spring characteristic may be generated. As a force is applied
the actuator moves, causing the position signal to rise accordingly. Since the
system always attempts to keep the measured force and the force command
equal, an increase in applied force will be required to generate an increasing
displacement. Since the position signal is directly proportional to the actuator
displacement, the force required to cause a displacement of 10cm will be twice
that required for 5cm - ie the spring characteristic will be linear.

The magnitude of the force command signal may be varied. In the physical
system this has the effect of altering the stiffness of the spring characteristic. By
centering the characteristic about the mid-point of an actuator's travel,
supportive forces may be applied to a patient. Any swaying will cause actuator
movement and will be opposed by forces increasing with the degree of
movement. Varying the spring stiffness has the effect of varying the level of
support - high spring stiffness giving strong support, and low stiffness giving
reduced support. By carefully selecting a spring stiffness appropriate to the level
of a patient's stability, a patient is allowed the freedom to stand almost unaided
if sway movements are small, but is given much stronger support if larger
potentially dangerous sway movements occur, ie those that might result in a fall.

Differentiation of the position signal gives a signal proportional to the speed
of an actuator. This may also be used as a force command, and has the effect of
introducing damping into an actuator's movements. With damping selected the
system will oppose rapid changes in a patient's position. Since the positional
signal generates a linear spring characteristic, the velocity signal that generates
damping will also have a linear characteristic.
Static force, spring, and damping force commands may be selected individually or in combination to produce various supportive effects. Having the capability of varying the level of each type of force command allows the system to offer widely varying types and levels of support, accommodating patients with a wide range of postural instabilities.

5.2 Controller Front panel

The electronic controller incorporates the force control system, and also allows the user to set the various levels of spring support, damping support, and static force required for a patient trial. Digital circuitry handles all safety interlocks, ensuring that the system is operated safely.

The user front panel gives access to the control system and also indicates the system status. Fig. 5.3 shows the design of the controller front panel. The analogue section of each controller (A1/A2) implements the force loop control system and the generation of static and supportive force commands. Spring support, damping support, and static forces are selected by switches, and levels may be set by adjusting calibrated controls. The static force command also has an additional direction selection switch (push or pull) and has a multi-turn pot for more accurate force level setting.

The digital section of each controller (D1/D2) indicates the status of the system and incorporates a number of safeguards to help prevent potentially dangerous situations. It is designed to be simple to set up, but inherently safe.

At switch-on, the digital controller for each actuator adopts an initial RESET state, shown by a red indicator on each of the panels D1 and D2. At switch-on the absolute position of each actuator shaft is unknown, so an initialisation procedure is required to give the position decoding circuitry a datum position. This is accomplished after switch-on by manually moving each actuator shaft to its mid-travel position where a contrasting dark band on the actuator shaft is detected by a reflective optical detector. When this occurs, the appropriate digital controller resets its displacement counter and adopts a READY state, shown on the front panel by an amber indicator. The READY state shows that the system is correctly initialised, but the force control system is not yet enabled.
When in the READY state, the system may be toggled between READY and ENABLED by using two push-button switches on the digital controller front panel. In the ENABLED state (green indicator), the system is fully operational, and the appropriate power amplifier and actuator becomes active. This is the state that would be adopted during a patient trial, with supportive forces having been previously set on the analogue control panel.

5.3 Analogue Controller

The analogue controller implements the force control system. Fig. 5.4 shows a simplified design of the controller - functional diagrams are included in Appendix 9.4.

There are three inputs to the controller - a signal from the actuator load cell representing the measured force (force loop), an analogue position signal generated by the digital board (command loop), and an external force command from the host computer, used to generate controlled force commands during patient trials.

Internally the load cell is configured as a strain gauge bridge and consequently generates a differential force signal which is buffered and amplified by op-amp (1) in Fig. 5.4. The output from this op-amp is fed to the main amplifier (4) and represents the actual measured force in the force loop.
The command loop can generate static forces, forces from an external source, and supportive forces with a spring or damper characteristic. Static forces merely involve the introduction of a fixed voltage signal into the loop, and this is generated by amplifier (5).

The digital controller decodes and counts displacement pulses from the actuator displacement transducer, and these are used to generate an analogue actuator position signal. This signal is amplified by op-amp (2) and is fed to op-amp (4) as a displacement (spring) force command. By differentiating this signal, (op-amp 3), an analogue velocity signal is derived and this is also fed to op-amp (4) and forms a velocity (damping) force command.

The force error derived from all of these signals is used to drive the power amplifier which controls each actuator's motor current, which in turn is related to the linear force generated by each actuator, thus closing the force loop. Compensation is utilised to modify the response of the system to ensure that the system bandwidth is sufficient.

5.4 Digital Controller

The digital controller performs two main system functions. It decodes pulses from the displacement transducer and converts these into an analogue position
signal, and it also controls overall system management and implementation of safety features.

Each actuator contains a displacement transducer that consists of two infra-red sources and detectors, and a circular slotted disc. These components are arranged so that as the actuator rotates the disc turns and interrupts each detector. The two detectors are mounted so that only one experiences a transition at any time. Fig. 5.5 shows typical pulse waveforms generated as rotary motion occurs. From the phase relationship of the two waveforms, directional information may be derived.

Clockwise rotation occurs if:

\[
\begin{align*}
(A=1 \text{ AND } B=1) \text{ AND } B \text{ changes} \\
\text{OR } (A=1 \text{ AND } B=0) \text{ AND } A \text{ changes} \\
\text{OR } (A=0 \text{ AND } B=0) \text{ AND } B \text{ changes} \\
\text{OR } (A=0 \text{ AND } B=1) \text{ AND } A \text{ changes}
\end{align*}
\]

This may be simplified to:

\[
((A=B) \text{ AND } B \text{ changes}) \text{ OR } ((A\neq B) \text{ AND } A \text{ changes})
\]

Similarly for counter-clockwise rotation:

\[
((A=B) \text{ AND } A \text{ changes}) \text{ OR } ((A\neq B) \text{ AND } B \text{ changes})
\]

Fig. 5.5 Quadrature waveforms from displacement transducers emphasising signal phase relationship.
If \( A \) and \( B \) represent existing values, and \( A_- \) and \( B_- \) represent previous values (i.e., those before a transition), each expression becomes:

- Condition for CW rotation:
  \[
  (A \oplus B)(B \oplus B_-) + (A \oplus B)(A \oplus A_-)
  \]

- Condition for CCW rotation:
  \[
  (A \oplus B)(A \oplus A_-) + (A \oplus B)(B \oplus B_-)
  \]

Where \( \oplus \) represents an exclusive OR function

and \( /\oplus \) represents an exclusive NOR function.

These two expressions can be implemented in digital logic and are used to generate UP and DOWN counting pulses as shown in Fig. 5.6. The directional pulses are used to drive a binary counter which retains a count of the linear position of each actuator shaft.

The digital count is converted into an analogue signal by a D to A converter, and this signal is used as a force command to generate supportive forces with a spring characteristic. The characteristic is configured so that at the centre of an actuator’s travel, no supportive forces are generated. However, increasing displacements in either direction away from the central position are opposed by an increasing force related to the position signal. Circuit diagrams of the digital circuitry used to generate supportive force commands are given in Appendix 9.4.

The other main function of the digital controller is to handle safety interlocks and to control the overall enabling/disabling of the actuator system. The digital

![Diagram](image-url)
controller constantly monitors the state of the system and only allows the system to be used if all operating conditions and criteria are satisfied.

Fig. 5.7 gives an indication of the major checks performed by the controller. At switch-on the system must be initialised, and before the actuators may be enabled, correct handshaking must be achieved with the Apple Computer, and ‘all clear’ signals have to be received from the operator (via the front panel), the actuator power amplifiers, and all integrated safety enables. If any of these conditions fail, the system cannot be enabled. Before successful system operation is possible, the cause of the failure or inhibit must be located and rectified. Only digital controller functionality is shown in Fig. 5.7. The controller consists of basic logic circuits which are not detailed in this report.

![Digital controller functionality diagram](image)

**Fig. 5.7** Digital controller functionality.

### 5.5 Safety Features

Clearly a system that involves physically exerting forces upon people with balance deficiencies must be entirely safe to operate and must be secure in the event of component failure or operator error. The prototype actuator system incorporates a significant number of specific and general safety features that ensure the patient retains a high level of security during patient trials. These safety features are detailed below.
PANIC Button

It is extremely important that any patient attached to the actuator system must be comfortable and unafraid. The reasons are twofold. Firstly if the system causes the patient distress or concern, then on ethical grounds trials should not be allowed to proceed and the hardware and methodology of testing must be re-examined. Secondly if a patient performs under stressful conditions, any results derived from the trial are likely to be affected, with the patient performing badly or at least inconsistently.

When a new patient is introduced to the system, it is emphasised to them that they have the power to terminate a trial at any time, and this is achieved by means of a PANIC button. This is a push-button switch with a large actuator button mounted approximately 30cm in front of the patient at waist height. Patients are encouraged, given the slightest concern, to hit the button and terminate the trial. Indeed, before a series of trials commence, the system is powered up and the patient is asked to press the button as a matter of routine to be reassured that the system halts and to feel the affects.

The PANIC button works on the loop-disconnect principle - when the button is depressed it breaks a circuit and in this way signals a panic situation. If the button were to become disconnected in some way, the system would still register this as a fault, and enabling the actuator system would not be possible.

When the PANIC button is activated, the electronic controller registers the signal, illuminating an indicator on the front panel and switching the system to the RESET state. The actuators are disabled and hence become very stiff and inherently supportive, holding a patient who may have been in difficulty. Before a trial may recommence (if appropriate), the system must be powered down and re-initialised.

Connector Coding

The main elements of the system, ie the actuators, controller, power amplifier, and computer are connected together by shielded cables. Two safeguards exist to ensure that the system can only operate in the correct configuration. Firstly each connector is uniquely coded so that it is not possible to make an incorrect
connection. Secondly the digital controller has sufficient intelligence to detect whether or not a connection is present. If the absence of a connector is detected, the system locks in a disabled state and may only be enabled when all connection requirements are satisfied.

Front Panel Design

The electronic controller front panel is designed to give the operator full control of the features available within the actuator system, and selection of each feature is arranged to be straightforward. Thus supportive or static forces may be generated to suit a particular patient trial, and each may be easily adjusted by means of the analogue controller settings.

Similarly the status of the system is shown clearly on the digital controller by means of three indicators, arranged in a traffic-light configuration, with red implying a disabled condition, amber a ready condition, and green a fully enabled condition.

There is still the potential for the operator to surprise or worry a patient without using the system incorrectly, so any operator must have a full working knowledge of the actuator system. Potentially one of the more dangerous possibilities is to exert a very sudden force upon the patient without warning. The system is capable of exerting a sideways force of up to 100 Newtons, and this can easily have the effect of toppling even a well-balanced person. To lessen the possibility of this happening, a safeguard has been incorporated. If a static force is selected and is switched into an actuator, the controller automatically causes the force to slowly ramp up to the desired level. Although ultimately the same force will be exerted, the sudden step situation that could prove dangerous to a patient is avoided.

It is possible to override the ramping feature if this is required. This may only be done by depressing a button on the digital controller at the same moment that the intended static force is selected. Thus a two-hand operation is required to bypass the safeguard, and consequently it is much less likely to happen inadvertently. This also applies to step force commands generated by the host computer.
Power Amplifier Shutdown

The dual power amplifier incorporates monitoring circuitry to ensure that operating conditions outside of its specifications do not occur. If actuator currents or voltages become excessive, the unit will shut down and indicate a fault condition. Similarly, if either unit becomes excessively hot, the unit again shuts down. A fault condition on either power amplifier will cause the digital controller to disable the system and lock it in the RESET state until the cause of the fault is rectified.

Actuator End-Stop Detection

Each actuator is prevented from running against its mechanical end stop by an electronic end-stop detector, thereby reducing the risk of damage to the system and possibly shock to the patient. Since the position of each actuator shaft is constantly monitored by the digital controller, the position count is used to detect when the actuator shaft is at either extreme of travel. When this occurs the relevant direction enable on the power amplifier is disabled, but the system as a whole continues to function. Thus once an end stop is reached it is still possible to move the actuator away from the end stop. Once the actuator shaft leaves the end-stop region, full operating conditions are regained with the actuator able to be powered in both directions.

The system features outlined above are primarily intended to protect the patient’s safety during patient trials. Other precautions include the constant and close presence of a qualified physiotherapist during testing, and before each series of trials, tests involving an able-bodied subject to check system functionality.

5.6 Computer Interface and Data Capture

The mechanical and electrical hardware so far described is in itself a complete functional system. It is quite possible to connect both actuators to a patient, set various combinations of spring and damping forces for support, and using the analogue control panel apply static forces whilst monitoring the patient’s response. However, it would be very difficult using this manual technique to generate consistent test conditions. It would also be extremely
difficult manually to quantify the patient's response in terms of stability and subsequently demonstrate any improvement.

To make the system more usable an Apple IIe Microcomputer is employed to monitor and control the system during a patient trial. With suitable interfacing the computer can deal with the large number of signals involved, sampling and recording each channel. The processing power of the computer makes it possible to analyse data and display it in various forms. It can also be made to control each actuator by generating accurately defined force command signals, avoiding using manual switching on the analogue controller. Employing the computer in this manner creates more consistent test conditions than would be possible manually, and can be expected to generate more consistent and reliable results.

Signals from and to the microcomputer are sent and received using a commercial interface package having the capability of generating up to sixteen independent analogue output signals, and receiving and monitoring a further sixteen analogue input channels. The configuration and basic design of the interface package is shown in Fig. 5.8.

Fig. 5.8 Schematic design of commercial interface package showing internal structure.
The architecture of the computer allows the interface to be memory-mapped, hence the monitoring of an input channel merely involves examining a particular memory location, ensuring that processing and storing input samples is straightforward and rapid. Similarly an analogue output signal may be generated by writing to another memory-mapped location. There are frequency and resolution limitations associated with the interface package, but these are not of significance in comparison to the frequency response and accuracy of the system. Each interface channel operates well beyond the bandwidth of the actuators and control system.

For consistency and ease of comparison each patient trial is always of sixty seconds duration. During a patient trial the following eight analogue signals are sampled:

From Actuator 1:
- Signal 1: Force in medial/lateral direction (side to side)
- Signal 2: Displacement in medial/lateral direction

From Actuator 2:
- Signal 3: Force in anterior/posterior direction (front/back)
- Signal 4: Displacement in anterior/posterior direction

From Force plate:
- Signal 5
- Signal 6: Four corner load cell signals used to derive
- Signal 7: position of patient’s centre of pressure
- Signal 8

The system samples each input channel at a frequency of 32Hz to ensure that all significant frequency components of sway will be detected. During the sixty seconds of a patient trial the microcomputer does no processing of any of the eight input channels, but performs a rapid sample-and-store function to record all signals. As part of the same operation it uses two output channels to generate signals to control each force actuator. The software is structured so that all of these operations are performed automatically, requiring no operator intervention. All that is necessary is to specify a force command file and initiate the trial. The system then takes control either until the successful conclusion of the trial, or the point at which it is aborted by the operator or patient.
Chapter 5 Electronic Control System

A self-check is performed by looping-back one input and one output channel and generating a test signal for the duration of a trial. Although this does not verify the correct operation of individual channels, it does confirm the functionality of the interface. The operator is only alerted if a problem is detected.

5.7 Force Command Files

During a patient trial, the role of the Apple computer can be made purely passive, i.e. simply input and store signals from the actuator system. However, the computer can also be made to control the force exerted by each actuator whilst still sampling and storing each of the eight system signals discussed previously.

The software developed for the system allows the operator to design and save a force command file. The file consists of a mathematical force profile for each actuator in the form of a series of numeric samples. During a patient trial these samples are sent as force commands to each actuator controller, and appear as forces exerted by each actuator. This happens at the same frequency as incoming data is sampled.

Each actuator may be controlled independently by a force command file to apply between zero and one hundred Newtons of force, either pushing or pulling. Force command files are created by using the graphics capability of the Apple computer. For each actuator in turn the computer generates a pair of blank axes, representing force against time. The operator then designs the desired force profile by using the computer keyboard to move a cursor around the screen, hence mapping out a profile in Cartesian form. Once this has been done for both actuators the resulting profiles are processed to generate a sequence of numeric samples, subsequently being stored as a single command file on floppy a disc.

Typical force command files are shown in Fig. 5.9. Combinations of rising and falling ramps and steps may be used to generate a comprehensive and varied array of force commands for one or both actuators (Fig. 5.9(a)). Sinusoidal force commands may also be used by specifying a magnitude and frequency for each actuator (Fig. 5.9(b)).
For patient trials involving free-standing it is useful to have a command file with no force command on either actuator, Fig. 5.9(c). Using the NULL file causes the system to monitor and record the performance of a patient effectively under free-standing conditions, even though the patient is still attached to the system.

When a force command file is created it is the shape or profile that is important rather than the absolute force magnitudes. Consequently the ramp and step, and sinusoidal commands shown in Fig. 5.9 contain no scale on the force axis. The maximum force that the system will generate is specified by the operator at the time of a trial, and this corresponds to the peak value on the profile. This gives the system considerable flexibility since the command file profile may be used many times at different force levels. For example the ramp
and step profile could generate a very weak disturbance by specifying a low maximum force of 20 Newtons, or a significantly stronger disturbance by specifying a high force of say perhaps Newtons. The same profile may be used in each case - it is simply the maximum force specified by the operator that changes. Having the capability of generating force command files in this way allows a particular force profile to be retrieved from disc and used in many different patient trials. This ensures that for each trial precisely the same forces and timings are maintained. Such a technique generates very consistent test conditions, and guarantees that results from using the same force profile on different patients may be validly compared.

With the computer sampling data from the system and controlling each actuator, there is generally no requirement for operator intervention during a patient trial. Force, spring, and damping controls on the analogue controller remain active and may be employed to supplement a dynamic force command with spring or damping effects, or static forces. In this way standard force command files may be used during a patient trial, with varying levels of supporting force. Varying combinations of spring and damping forces may be employed, and the benefit of supportive forces can be assessed.

This chapter has discussed the operation of both the electronic control system and the computer interface. The electronic control system consists of analogue and digital circuitry that implements the control system responsible for the operation of each actuator. By using the controller front panel an operator may enable or disable the actuator system, and can introduce static forces, and supportive spring and damping forces. The microcomputer provides a more sophisticated method of controlling the system. It is responsible for capturing and storing data, and may be used to generate very consistent sequences of forces in order to observe and monitor the response of the patient.

The physical system discussed in Chapter 4 and the electronic control system described in this chapter comprise the full actuator control system. Further chapters discuss the testing of the system, and the processing of data to generate potentially useful methods of quantifying patient stability.
CHAPTER 6

CONTROL STUDY AND SYSTEM COMMISSIONING

Each actuator provides a means of applying forces by varying an electric current supplied to the actuator motor. Force transducers allow the resultant forces to be accurately measured. Displacement transducers allow the position and speed of each actuator shaft to be monitored, and hence the position and speed of any patient connected to the system.

To derive adequate control of the forces applied, the force transducers are incorporated into a closed loop control system (Fig. 6.1), and this is used to drive each actuator motor.

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Fig. 6.1 Closed loop force control system with simple force command input.
The input to the system is in the form of a force command, and this dictates what force is desired as an output from an actuator. The comparator compares the actual measured force with the desired force command signal. Any difference between the two signals is used to adjust the output so that the actual force is maintained approximately equal to the force command. Compensation may be required to ensure that the performance and stability of the system are acceptable over the required frequency range. To determine the response of the system, and if necessary the type and level of compensation required, the system must be modeled to allow for subsequent frequency response prediction and analysis.

6.1 System Modelling

Each actuator may be modelled by conceptually breaking it down into identifiable physical components and describing the overall operation by means of a set of simultaneous differential equations. Fig. 6.2 shows the mechanical

Fig. 6.2  (a) External view of actuator.  
(b) Internal workings of actuator.  
(c) Translational model of actuator elements.
components comprising each actuator and indicates how each element may be represented as an ideal translational model, excluding frictional factors and mechanical non-linearities such as backlash.

Torque is applied to the system from the actuator motor and acts upon the inertial mass $J$ in the system. This mass comprises all rotating elements (motor, armature, lead screw), and also linear mass referred through the gearbox (linear shaft, load cell case, and ball nut case), since all of these elements may be considered to be rigidly connected. The high-efficiency recirculating ball nut converts the rotary motion of the lead screw into linear motion of the actuator shaft, and may be represented as a simple gearbox, translating force and displacement inversely. The spring element $k_a$ is present in the load cell, since it comprises a dual strain gauge element, and also in the padding of the patient harness. The mass $m_h$ is rigidly connected to the sensing element of the load cell and represents the mass of the harness.

The patient is held within the harness, but cannot be considered as a simple floating mass. The dynamics of the body and a person's response to sideways force stimuli are both highly complex, so for simplicity the patient is represented as a mass that is grounded by a parallel spring/damper arrangement. Thus the expected response of the patient is represented by the dynamic response of the spring/damper arrangement. Although not ideal, careful selection of the mass, spring, and damping constants can permit rudimentary testing to proceed.

Each of the three element types (mass, spring, damper) may be characterised by a component equation defining a particular force-velocity relationship, and these component equations fully define the behaviour of individual elements. The interconnection of the elements to form a system structure must also be described in terms of structural equations. The translational model of Fig. 6.2(c) is shown more concisely in Fig. 6.3, with each element numbered and given its characteristic equation. Since the harness and patient may be considered to be connected, the two masses have been combined as a single unit.
Component equations for the system are:

\[ T = J \dot{\theta}_1 \]
\[ f_4 = k_4 v_4 \]
\[ f_2 = k_2 v_2 \]
\[ f_3 = m v_3 \]

Structural equations linking the above equations may be derived by inspection of the topology of the system:

\[ v_2 = n \dot{\theta} - v_3 \]
\[ f_3 = f_2 - f_4 - f_5 \]
\[ T_1 = T_m - n f_2 \]

where

- \( T_m \) - is the torque generated by the actuator motor
- \( n \) - is the gearbox ratio in metres per radian

From the two sets of equations a block diagram may be derived to model the system (Fig. 6.4). The system shown has a torque command as an input, but to derive adequate force control, the input should be in the form of a current command. To do this the remaining elements of the system, ie the motor and power amplifier, must be incorporated into the model.
Chapter 6 Control Study and System Commissioning

For the motor:

\[ V_A = I ( R_A + sL_A ) + k_v \dot{\theta} \]
\[ T_m = k_T I_A \]

where
- \( V_A \) = armature voltage
- \( I_A \) = armature current
- \( R_A + sL_A \) = complex armature impedance
- \( k_v \) = back emf constant
- \( k_T \) = torque constant
- \( \theta \) = angular displacement of armature

These two equations may be modelled by the block diagram shown in Fig. 6.5, which may subsequently be incorporated into the main system model.
The power amplifier is current driven and incorporates current feedback to gain tighter control of the actuator torque. Its inclusion in the system block diagram is relatively trivial. Adding the motor and current feedback gives a system where the motor armature current may be carefully controlled. The full actuator force control system is represented in block diagram form in Fig. 6.6. Each section relates to a particular element of the actuator and controller, and these sections are indicated in the diagram. Feeding back the force signal $f_2$ (i.e., the force exerted upon the patient by the system) allows the system to be configured for force control, accepting a force command rather than a current command.

An advantage of having the system model represented in block diagram form is that the function of the system may be relatively easily visualised. Additionally, when in this form the frequency response of sub-sections of the system, or the system as a whole, may be analysed using a commercial software package.

### 6.2 Evaluation of System Parameters

The control system block diagram shown in Fig. 6.6 contains a number of fixed parameters. Many of these relate to physical parameters within the system, e.g., the torque constant for the motor, and their values are known. Spring and damping constants relating to the patient's body are unknown and are likely to vary from patient to patient. These must be estimated initially to allow frequency response analysis to proceed. The compensation stage $H_r$ is also unknown but initially is given a gain of unity with no phase lag or lead.

The frequency response of the force control system must be sufficiently high to allow it to respond to any movements that a patient may perform, and hence generate supportive forces to compensate for the patient's imbalance. Research has indicated [3] that sway movements encountered during free-standing trials contain a maximum frequency component of around 14Hz, suggesting that an analogue force-loop bandwidth target of 20Hz is reasonable.
Fig. 6.6 Block diagram of full actuator force control system.
With the system bandwidth speculated, it is possible to provisionally evaluate all of the control system parameters. The following physical parameters relate to identifiable components within the system, and are all known:

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Description</th>
<th>Value</th>
<th>Source</th>
</tr>
</thead>
<tbody>
<tr>
<td>R</td>
<td>Armature resistance</td>
<td>1.0 Ω</td>
<td>Derived from motor specification</td>
</tr>
<tr>
<td>L_A</td>
<td>Armature Inductance</td>
<td>10 mH</td>
<td>Derived from motor specification</td>
</tr>
<tr>
<td>K_T</td>
<td>Torque constant</td>
<td>0.051 Nm/A</td>
<td>Derived from motor specification</td>
</tr>
<tr>
<td>K_v</td>
<td>Back emf constant</td>
<td>0.05 V/rad/s</td>
<td>Derived from motor specification</td>
</tr>
<tr>
<td>J</td>
<td>Armature inertia</td>
<td>4.4x10^-5 Kg m^2</td>
<td>Calculated from rotary mass of motor, armature, lead screw, and referred mass of shaft, load cell, and gearbox</td>
</tr>
<tr>
<td>n</td>
<td>Gearbox ratio</td>
<td>7.95x10^-4 m/rad</td>
<td>Derived from pitch of lead screw</td>
</tr>
<tr>
<td>k2</td>
<td>Load cell and harness padding spring constant</td>
<td>1.2x10^3 Nm^-1</td>
<td>Derived from load cell specification and simple tests</td>
</tr>
</tbody>
</table>

The parameter Gv controls the bandwidth of the power amplifier current loop. Typically a current loop bandwidth of ten times the force loop bandwidth may reasonably be expected, since the stage is purely electronic. A higher bandwidth is quite feasible but would generate no overall increase in system performance. To calculate the value of Gv, the current loop may be considered as a simple first order system (Fig. 6.7) where the gain may be expressed as:

\[
a / (1+bs)
\]

with a frequency breakpoint occurring when \( b = 1 \)

For the current loop, the breakpoint occurs when \( R + G_v = 2\pi f L_A - R \). Substituting known physical parameters to give a current loop bandwidth of 200Hz requires a value of \( G_v = 25 \).
The only remaining unknown parameters relate to the response of the patient connected to the actuator system - i.e. the mass, spring, and damping values described in the translational model (Fig. 6.2(c)).

The mass \(m\) upon which the system acts will comprise the mass of the harness and patient combined. Although patient masses will vary, the actual mass used during a patient trial or during system testing may be accurately determined. A working figure of 60kg will initially be adopted.

It is not possible to determine the spring and damping constants accurately. These values are difficult to estimate and are likely to vary between patients and may change dynamically in the same patient during a trial. The spring and damping characteristics may also be non-linear, so for example the body’s resistance to stabilising forces may increase non-linearly with displacement, or may change with speed, and will inevitably exhibit a different characteristic in the medial-lateral and anterior-posterior directions due to the nature of a stroke. The best that may be accomplished is to derive a rough estimate for spring and damping levels, and then to use these values for the purpose of analysing the stability and performance of the control system. The response of the system may then be determined as these values are varied to ensure that the system can accommodate subjects with a wide range of stability levels.

Simple experimentation indicates that the maximum static force applied at waist-level that can be resisted by an able-bodied subject is approximately 100 Newtons. This maximum is a function of height and weight and foot separation, rather than strength, and is reasonably consistent across a varied sample of able-bodied subjects. A force exceeding the maximum causes
toppling. The change in displacement of the centre of the body between free-standing and the commencement of toppling is approximately 15cm, so if the displacement characteristic was linear the stiffness value would be unlikely to exceed an approximate value of 650 N/m (ie 100N/15cm) for large displacements. It is possible for the stiffness to be larger for smaller displacements but an estimate of these levels is not realistically possible and would be of little worth due to the likely variation in values encountered across the spectrum of subjects. The stiffness of 650 N/m, whilst depending upon a number of assumptions, does at least give an initial figure with which to work.

Determining a realistic estimate of the damping factor associated with a patient’s response is much more difficult. It is feasible that rudimentary tests could take place by moving the patient small distances at known speeds and measuring the opposing force, but inertial effects, spring effects, and the patient’s dynamic response would be likely to invalidate any force readings making the results unusable. One way to derive a maximum damping figure is to consider the maximum linear speed of an actuator and the maximum resistance that a subject may generate. As for the spring calculations, a sideways force of 100 Newtons is the most that may be resisted and is hence the maximum force a patient may exert upon an actuator. The maximum speed attainable by an actuator is 0.3 ms\(^{-1}\) so a maximum damping rate over large displacements might be 330 N/ms\(^{-1}\) (ie 100N / 0.3ms\(^{-1}\)). Once again this figure depends upon a number of assumptions, but gives a starting point from which analysis may proceed.

Although the estimates of spring and damping levels may not be accurate it is quite possible that these levels may be encountered at some stage during patient trials. They may be used initially to examine the predicted performance of the control system, and can then be varied to examine the effects upon this performance.

6.3 System Analysis

With values having been assigned to all parameters within the actuator force control system, initial analysis may proceed with a view to determining the performance of the overall system. A very effective way of examining the
theoretical system response is to use a commercial software analysis package. With the basic system defined on the package the processing capability of the host computer allows quick analysis and generation of graphical and numerical data. The speed at which this information may be obtained allows the effects of modifications to the control system to be rapidly assessed. The package employed was Simbol (Cambridge Control Ltd, Cambridge) - an interactive simulation and analysis program running on an IBM PC computer. The package can analyse the stability characteristics of a wide range of linear and non-linear control systems. Systems are described interactively on the screen using standard block diagram notation, and their dynamic characteristics may be investigated by using Nyquist, Inverse Nyquist, Nichols, Describing Function, and Bode plots.

The response of the system was investigated using Nichols charts, allowing an estimation of the closed-loop system response to be derived from the open-loop response. A Nichols chart of the basic force response is shown in Fig. 6.8.

The response indicated by the Nichols chart is characteristic of a third-order system, having a low frequency phase response of \(-90^\circ\), and tending towards

![Nichols chart indicating response of force control system with forward gain constant $H_f = 1$.](image)
-270° at high frequencies. With the system parameters as initially determined the closed-loop system response passes very close to the chart origin, indicating that although the closed-loop response would be stable, it is likely to be highly underdamped with unacceptable settling times. The response also indicates that the maximum closed-loop gain occurs at a frequency of approximately 50 rads/sec, suggesting a likely closed-loop bandwidth of only 8 Hz.

The response of the system is lacking in both bandwidth and stability. To achieve the desired closed-loop response a simple increase in the forward-loop gain is not feasible due to the poor response of the initial system. Increasing the forward gain would further decrease the gain and phase margins. Hence compensation is required to generate the required bandwidth increase whilst maintaining both margins at suitable levels.

The system response may be improved by the inclusion of a phase advance stage in the forward loop. The desired bandwidth of the closed-loop system is 20 Hz, a point that lies well below the 0 dB line of the Nichol's chart in Fig. 6.8, indicating that a gain increase is likely to be necessary in addition to a phase advance at this point. Since the final bandwidth is not too critical, one method of applying the compensation stage is to design the phase advance to have maximum effect at 20 Hz, and then adjust the forward gain to give a response that lies tangential to one of the lower-value constant-gain circles on the Nichols chart. A lower tangential gain circle implies a flatter closed-loop response of the control system and is clearly more desirable. The resulting closed-loop bandwidth is a little uncertain using this method, but a bandwidth anywhere near to 20 Hz is acceptable.

The response of a typical phase advance network is illustrated in Fig. 6.9. Assuming for convenience that the two break frequencies may be related, the network will have a transfer function of the form:

\[ H_f(s) = \frac{a(1 + sT)}{(1 + sTa)} \]

with a maximum phase advance \( \phi_c \), where

\[ \phi_c = \sin^{-1} \left[ \frac{(1 - a)}{(1 + a)} \right] \]
Given that a phase margin of $45^\circ$ can be considered desirable for reasonable system performance, the compensation network must advance the phase by at least $40^\circ$ since the uncompensated phase margin is only $5^\circ$. As the phase margin will reduce as the forward gain is increased, a figure of $45^\circ$ can be taken, yielding a value for the constant $a$ of 0.17, giving the corresponding gains and breakpoints shown in Fig. 6.10.

The compensated response curve for the force control system is shown in Fig. 6.11. Here the attenuating property of the compensating network has been corrected by including a gain factor of $(1/a)$, and the forward gain has been increased to give $H_f = 3$. The gain and phase margins are now acceptable and a closed-loop bandwidth of roughly 18Hz has been achieved. The system possesses a slight gain increase of 2.9dB at the break frequency, but this is characteristic of systems of this type and is not considered to be detrimental to
the overall system performance. The compensation network that achieves this phase advance is given in Appendix 9.4.

6.4 Effects of Parameter Variations

With the system shown to be stable and possessing adequate performance, all that remains is to examine the effect upon the system of variations in the uncertain parameters discussed in Section 6.2, i.e., the mass, spring, and damping constants associated with the patient's response to force stimuli. These figures were derived by making a number of approximations and assumptions and it is important that the system performance is shown to be maintained irrespective of parameter variations.

The mass of the patient is the most predictable of the three parameters. Although it will vary between patients it remains constant during a patient trial and can be accurately determined. A good estimation can be made about the range of masses that will occur. It is considered reasonable that masses outside of the range 40kg - 90kg are unlikely to be encountered, hence the response of the system need only be examined within this range. The spring and damping constants present more difficulty since estimation of levels was accomplished by
making some broad assumptions. Values of 650 N/m for the spring constant and 330 N/ms\(^{-1}\) for the spring constant were selected to represent the possible response of the patient, although clearly these values are very approximate. Any examination of the effects of variations in these values must take this into account by investigating the system response as they are varied over a significantly wide range.

As the response of the system was initially investigated by computer modeling, using the same technique to examine the broad effects of parameter changes within the system becomes trivial, merely involving the alteration of defined parameters within the software package and the subsequent display of the predicted response.

The results of the investigation of parameter changes show the control system to be highly immune to quite significant changes in the mass, spring, and damping levels. Each parameter was varied in isolation and then in conjunction with the remaining two. The mass of the patient ranged from 40kg - 90kg, the spring level was varied between 60 N/m and 6000 N/m, and the damping level was varied between 30 N/ms\(^{-1}\) and 3000 N/ms\(^{-1}\). Investigation of the system response using Nichols charts revealed only very minor variations in the gain/phase relationship as the mass, spring, and damping levels were varied. These changes occurred only at very low frequencies - the closed-loop gain remained broadly flat at these frequencies but the phase response changed. However these variations are not considered to be of significance. The closed-loop bandwidth was retained under all conditions.

The conclusions from these tests is that the control system can be expected to perform adequately regardless of the ability or response of the patient. This is an important criterion as data extracted from trials can only be considered valid if the control system performs effectively for all patients. The immunity of the system to variations in levels of mass, spring and damping indicates that the system is able to cope with patients with widely differing responses, and that comparisons may validly be made between different patients with differing stabilities, or the same patient whose stability changes over a period of time.
One further aspect of importance is the behaviour of the system in the absence of a patient - i.e., when an actuator is driven by the control system but the patient and harness are not attached. This situation is likely to occur at the commencement of a trial as a patient and harness are about to be attached to an actuator, and similarly at the termination of a patient trial. Having the actuators driven with no force command as the patient is being connected is advantageous in that each actuator may be moved freely by hand with very little effort. Furthermore, since the control system is enabled there is no switch-on jolt as would be the case were the system enabled after patient connection has taken place.

With no patient or harness attached to an actuator the control system model changes slightly, with the spring and damping constants no longer relevant, and the spring $k_2$, which comprised the load cell and harness springs in series, becoming solely the load cell spring. Under these conditions the system still remains completely stable, possessing a flat closed-loop response at low frequencies and decreasing gain as the frequency increases. This implies that an actuator will remain static when not coupled to a patient, but will move if an external force is applied to create a force error. If the sensing element of the load cell is pushed, the actuator will attempt to move away from the force to retain a zero force error. This is an ideal situation, making the task of connecting harness and patient to the actuator system not only simple, but entirely safe.

This study has illustrated that the predicted bandwidth and response of the control system is sufficient to cope with patients exhibiting a wide range of stabilities, and that under all envisaged conditions the control system is inherently safe. The following section details how the actuator and computer sub-systems were tested to ensure that the actuator control system responds as expected, and that the two sub-systems work in unison.

6.5 Bandwidth Testing

To ensure the validity of data derived from patient trials it is important to confirm the frequency response of the system. This not only applies to the actuator force control system, but also to the associated components, i.e., the computer interface, transducers, and forceplate electronics.
A commercial frequency response analyser was initially employed to verify the response and bandwidth of the force control system. This device generates a precise dynamic test signal which may be used as a force command and by monitoring the resultant output force from each actuator it can accurately determine the system frequency response. Generally these devices work extremely well, however it proved impossible to derive any valid results as the signal from the analyser repeatedly caused the power amplifier to shut down. The power amplifier contains monitoring circuitry that detects any over-voltage signal at its input. If this situation occurs the drive shuts down, preventing possible damage to the unit. It was not possible to reduce the test signal generated by the analyser to a sufficient degree to obtain frequency response results. Nor was it possible to easily remove the power amplifier protection circuitry - indeed it is possible that the response of the system would have been altered and hence any frequency response obtained after removing this feature would not necessarily be valid.

Another simpler though less accurate technique was employed to investigate system behaviour. By using a frequency generator it was possible to inject a sinusoidal test signal as a force command of known amplitude and frequency. Rudimentary testing of the system progressed by varying the test signal frequency and monitoring the resultant force output signal on an oscilloscope. Although the phase response was not noted, it was possible to note the relationship between input and output magnitudes over a range of frequencies. The response of the force control system was found to be broadly as expected, possessing a flat response at low frequencies with increasing attenuation noted around 15Hz. However this test was not ideal - the force signal contained a large amount of high frequency noise which made estimation of levels difficult, and the phase relationship between input and output signals was not examined. Despite this, the results were considered to be valid in broadly demonstrating the response of the system.

The force plate and associated interface electronics possess a bandwidth well in excess of that required. Each position transducer can detect positional changes significantly above the actuator limit speed of 300cm/s, ensuring that the generation of spring and damping supportive forces is sufficiently responsive.
The computer interface and force command generation routing operate at an update rate of 32Hz, allowing frequency components of movement approaching 16Hz to be detected.

6.6 System Accuracy and Commissioning

For patient trials, absolute accuracy is not as important as repeatability. It is important when results from different patient trials are being directly compared that the patients were subjected to the same force stimuli, but provided the forces remain the same on each occasion, the absolute accuracy of the forces is not crucial.

Simple tests showed that force generation was accurate to within 10% of the level set either on the analogue controller or generated by the computer. This tolerance was maintained over the full force range -100N to +100N. The position transducers and associated circuitry that monitored the displacement of each actuator possessed an accuracy to within 3mm over the full 300mm range. Further tests showed that spring and damping supportive forces were within 10% of the value set, and that the characteristic remained linear in each case.

Test software was developed that allowed centre of pressure data from the force plate to be displayed on a computer screen in real time. As a person standing on the plate swayed, the position of the centre of pressure was shown on the screen as a cross, moving with the patient. By applying localised point forces to the plate and monitoring the computer generated centre of pressure, the algorithm for deriving centre of pressure information was determined to be accurate to within a few millimetres over the entire area of the plate. The accuracy was also confirmed to be irrespective of the mass involved. This software was later developed to form part of the overall stability analysis package.

The accuracy of the distance traversed by the centre of pressure point, ie the locus length, was also determined. This can be used as an indication of stability and is discussed in the following chapter. The real-time display software monitored the locus length over a set period, giving the result as a distance traveled. This was checked by moving a heavy stylus around the force plate.
along a precise path and subsequently comparing the computer monitored
distance against the distance moved by the stylus. The predicted and measured
distances correlated to within a few per cent, this accuracy being maintained
irrespective of both the distance traversed (ie for both short and long locus
lengths), and the speed at which the centre of pressure was moved.

The system was found to generate highly repeatable static and supportive
forces. Thermal effects were not a problem, with the system performing equally
as well at switch-on as after several hours of operation. The position monitoring
circuitry performed well over extended periods, and did not exhibit any tendency
to drift or progressively lose the count. The charge amplifier that comprised the
force plate interface was subject to significant drift after only a few minutes, but
operated well over the initial sixty second period. This presented no problem as
it was standard practice when using the unit to reset the charge amplifier before
each monitoring session.

With the system examined for both response and accuracy, the only way to
extensively test all aspects of the system was to perform real trials, and this was
done during the testing phase using able-bodied subjects. A large number of test
trials were performed that allowed comprehensive testing of static and
supportive force generation, and of signal monitoring routines. This inherently
allowed the data analysis, display, and retrieval routines to be used and tested
with real data. These were all found to perform well. Using able-bodied subjects
also allowed other aspects of the system to be examined - the task involved in
getting patients in and out of the harness and actuator system, the response of
the system to a panic situation, and the general management and housekeeping
involved in recording and storing the large number of data files derived from
each subject.

Few problems were encountered during the commissioning phase. Difficulties
that did arise were overcome either by minor modifications to the system or by
changes in the methodology involved in managing the trials, thus ensuring that
the overall system would efficiently record the required data, and also at all
times provide a safe environment for the patient.
CHAPTER 7
DATA COLLECTION AND PRESENTATION

The Apple Microcomputer plays an essential role in the operation of the actuator control system. When not supervising a patient trial it allows the operator to design detailed force command profiles that will be employed during patient tests. Before a trial commences the computer will perform safety checks and will communicate with the operator upon the status of the system. The computer performs continuous checks as a trial progresses as well as capturing and storing key signals from the system and generating force commands for each actuator as dictated by the selected force command file. At the conclusion of a trial the data is stored as an unprocessed data file, allowing further trials to be performed immediately. These raw data files may then be processed at a convenient time once a series of trials has finished and the results may then be analysed and displayed. The structure of the software system that performs these operations is shown in Appendix 9.5.

7.1 Data Capture

Data from a raw data file is processed in two stages before generating a final data file to be used by the analysis and display software. These processing stages involve conversion of the four force plate signals into separate x and y displacement signals (see Section 4.1), and noise suppression.

Since many of the sampled signals are low-level voltages and involve small changes in the signal levels, noise is inherent in the system. Although small, digitisation noise is also introduced by the interface conversion process. Noise occurs on each of the eight sampled input signals, i.e., the two force signals, two actuator displacement signals, and four force plate signals. When sampled by the computer the noise appears as a continuous variation in the least significant bit or bits of a sampled data byte. A number of the analysis techniques that have been developed monitor the difference between successive bits, so to ensure that
the results from these forms of analysis are valid a software routine is used to remove the low-level digital noise. The technique adopted involves introducing a small level of hysteresis into the digital sampling procedure. This is performed by rejecting any input level change of only one bit unless the change occurs in the same direction as in the previous sample. Thus a series of samples changing by only one bit due to noise (eg 16, 16, 17, 16, 16, 15, 16) would be regarded as a constant signal. This technique is more than simply reducing the sample resolution, as the sampling system still retains 1-bit accuracy. The technique is illustrated in Fig. 7.1.

Tests indicated that the two actuator position signals and four force plate signals required only 1-bit suppression to generate clean signals. The force signals from each actuator load cell were subject to more noise and required 2-bit suppression.

![Diagram](image)

**Fig. 7.1 Illustration of software noise suppression.**
Chapter 7 Data Collection and Presentation

After processing, a final data file contains the force signals from both actuators, the displacement signals from both actuators, and centre of pressure data in xy coordinates. The file is also time and date stamped and contains details of the force command file employed and the system operator responsible for the particular trial. In this form the data may then be used to generate the displays discussed below.

7.2 Presentation Techniques

Processed data resulting from a patient trial are stored as a data file on a floppy disc, and may be processed and displayed in a variety of ways to highlight particular features of stability during successive patient trials. The display techniques detailed here are by no means comprehensive. They represent an attempt to extract and display some of the better documented and well known stability parameters, along with some novel display methods not previously adopted. All of the data files shown result from valid trials with hemiplegic or able-bodied subjects. They are shown here to illustrate the display methods available, and more specifically what information may be extracted from each style of display.

The system supports three styles of display: Cartesian, area map, and distribution profile. Cartesian can be used to display any of the system signals, ie force, actuator displacement (centre of gravity), and force plate displacement (centre of pressure). Area maps and distribution profiles are only used to display centre of gravity and centre of pressure information.

Cartesian Displays

During a patient trial the actuators are configured to apply forces at right angles to each other, so the x and y components of any force vector are immediately available. Similarly, once the signals have been processed, centre of pressure and centre of gravity data also resolve into orthogonal x and y components. Consequently a form of Cartesian display is ideally suited to the system since all signals appear in component form.
Using a Cartesian form of display, with time as the major axis, the following signals from the system may be displayed:

- **Centre of pressure** of patient (from force plate data)
- **Centre of gravity** of patient (from actuator movement)
- **Force command** generated by the system
- **Measured force**

Examples of the first three displays are shown in Fig. 7.2. The first display indicates the force command generated by the system during a patient trial. The two time axes display each force applied over a sixty second period, the two vertical axes giving the changing magnitude and direction of each force. The particular force profile used in this example has no force command in the anterior/posterior (front and back) directions, but has a rising and falling ramp command in the medial-lateral (side to side) direction. The medial/lateral force commences at zero, ramps slowly to 30 Newtons applied to the patient’s left side (i.e., pushing to the right), ramps back through zero to apply 30 Newtons of force to the right side (i.e., pulling from the left) and ramps back again to zero. This procedure then repeats twice more during the sixty second trial.

The patient’s response to the force stimuli displayed in Cartesian form can be seen in Fig. 7.2(b). The centre of pressure trace which represents the patient’s reaction with the floor indicates a fairly good response. In the anterior/posterior directions (i.e., the top trace) the trace is reasonably constant, and lies well centered about the zero displacement line. As is expected, the medial-lateral (lower) trace mimics the force profile, since to oppose the applied force the patient must generate a reaction with the floor to remain upright and stable. However it is noticeable that this trace is also well centered about the zero displacement point. Having both traces well centered implies that the patient is distributing weight evenly on both feet - a very desirable goal for a hemiplegic.

The actuator position trace, Fig. 7.2(c) which approximates to the patient’s centre of gravity, reveals more information about the patient’s stance. Although the anterior/posterior position trace is fairly constant and hence stable, it reveals a significant offset in the patient’s forward body position. It is not quite a lean,
Fig. 7.2 Cartesian displays resulting from patient trials.
as this would also show on the centre of pressure trace. It is more like a reverse stoop with waist slightly forward and shoulders slightly back.

In the medial/lateral direction, ie the direction in which the changing ramp force has been applied, the patient has some difficulty in coping. This may be deduced from the amount of waist movement shown by the lower trace - an able-bodied person generally has more confidence and is able to resist the force with less displacement of the body. Also the trace is significantly offset from a central position, indicating a bend of the body to the left side.

In conjunction the two Cartesian displays showing centre of pressure and actuator displacement indicate that the patient is able to distribute body weight fairly evenly between the two feet, but still possesses an unnatural body posture.

The data used to generate the traces that have been discussed were derived from a hemiplegic stroke victim in the latter stages of recovery, and much of what has been speculated from the test displays is very typical of someone who has undergone extensive therapy and is at an advanced point in the recovery path.

For convenience, Cartesian displays derived from the same data file may be overlayed as shown in Fig. 7.3. All of these displays are useful in that they give a good visual indication of how a patient may have performed during a trial. Although they are non-mathematical they may be used to compare visually the performance of different patients, or the same patient over a period of time. The
timescale for stroke recovery could mean that traces taken months or even more than a year apart may need to be compared. The ability to detect trends occurring over the sixty second trial period can also be beneficial - a patient may become visibly more unstable towards the end of a trial owing to fatigue. Cartesian displays allow problems in stance to be detected and also permit short-term and long-term performance improvements to be appraised.

Area Maps

Data relating to a patient's centre of pressure or centre of gravity (actuator position) may also be displayed in the form of an area map. This is simply a graphical representation of a plan view of the floor-mounted force platform on which the patient stands, with relevant positional information superimposed. It may be used as a real-time active display - i.e., the display alters as the patient sways, or a cumulative display where all the measured positions over a sixty second period are displayed. The real-time display of centre of pressure is more of general interest rather than direct use and yields no quantitative data about patient performance or stability. It is the cumulative area maps that give the most information. Typical area maps are shown in Fig. 7.4. All data relate to patient trials involving no force command. With no external forces applied, the patient is effectively performing an exercise in free standing, with the system passively monitoring sway and movement generated by the patient.

By default, each area map shows a physical area 8cm x 12cm inside which excursions of a patient's centre of pressure or gravity occur. If patient sway is excessive and exceeds the defined boundary, the system scales the display accordingly to ensure that no information is lost.

The magnitude of sway that takes place during a free-standing trial is a well recognised and documented indication of a patient's stability. Accordingly the system indicates and draws the minimum size of a rectangle which encloses all sampled positions. The area of this rectangle is also indicated at the top of each display. The average position is the effective centre of the patient's sway and is indicated by a cross. The enclosing rectangle can yield quantitative information relating to a patient's level of stability. In simplistic terms the larger the area of the rectangle, the more unstable is the patient. Directional instability may also be
recognised from this method of assessment when excursions are far greater in one direction than in the other (Fig. 7.4(b)). If the superimposed cross does not lie centrally on the display, it may be concluded that the patient is not distributing his weight evenly. The patient may be leaning forward or backward, or more commonly for victims of stroke, shifting body weight heavily onto the unaffected side.

Fig. 7.4 Typical force plate maps showing variation in patient's centre of pressure. Cross-wires indicate mean position. Enclosing rectangle indicates maximum excursion in each direction.
A number of successive area maps can give a reasonable indication of a patient's stability, and can also indicate any imbalance between anterior/posterior and medial/lateral stability. When used with an active force command file the maximum displacement area can also yield valid information relating to how well the patient manages to retain stability against external forces of varying magnitude and direction.

Distribution Profiles

Positional data from the force plate and actuators may also be displayed in the form of a graphical distribution profile. Two displays of this type are shown in Fig. 7.5 using centre of pressure data taken from the force plate. During a sixty second trial a patient will sway and will spend time at varying distances from the centre of the force plate. The times spent at each position are totalled and are used to generate a displacement profile for each direction. The percentage of time spent at each displacement position is represented as a vertical bar, the magnitude of which is related to the vertical axis scaling. In each example the upper trace represents front and rear movements, and the lower trace side to side movements.

The profile shown in Fig. 7.5(a) indicates a very compact trace with a high time-percentage value. From this it may be concluded that the patient's centre of pressure on the force plate deviated very little from the ideal centre position - ie much time was spent at or very close to this centre position. Hence the time percentage reaches a high peak around the centre (ie zero displacement) point on both traces and rapidly drops to zero for larger excursions. This trace indicates very good stability, and uses data derived from a patient trial involving an able-bodied subject.

In contrast the second display, Fig. 7.5(b) shows two broadly distributed traces, both having low peak time-percentage values. This demonstrates that the subject is much less stable than the previous one - the broadness of the traces indicates that the patient's centre of pressure moved around significantly, spending much time at relatively large displacements away from the central position. This is also reflected in the low time-percentage value - the patient's...
centre of pressure did not remain in one stable position for very long due to sway and instability.

Two other facts may be derived from this particular display. Firstly the medial/lateral and anterior/posterior traces in the second example are greatly offset from the centre position, implying that the patient is distributing his weight unevenly (more so in the anterior direction). When this occurs the shape of the profile is unaffected, but the position of the profile becomes offset. Hence valid comparisons may be made between able and less able stroke victims since

![Fig. 7.5 Centre of pressure displacement profile recorded during freestanding trial.](image)
their differing weight distributions do not affect the important parameters of the displacement profile display. The second thing of significance is that the anterior/posterior trace is much broader than the medial/lateral trace, indicating a poorer stability in one direction than in the other.

So, from a displacement profile, two significant parameters may be extracted and used to grade the stability of a subject - the maximum time percentage spent in any one position, and the broadness of each profile trace. For good stability a large proportion of a trial would be spent in one place with very little sway movement occurring, hence a high time-percentage and compact trace would be expected. For poor stability the reverse is true - very broad traces and low time-percentage values.

The distribution profiles so far discussed take centre of pressure data and generate a displacement distribution profile that can yield quantitative statistics relating to a patient's stability. The same method of display may be used to investigate the velocity of a patient's movements. Two velocity profiles are shown in Fig. 7.6. These velocity profiles are analogous to the distribution profiles already described. Each horizontal axis represents directional velocity in line with each actuator, and the vertical axes give the percentage of time spent traveling at each velocity. As with displacement profiles, the peak value and broadness of the traces give an indication of stability - a compact trace and high time-percentage value again implying good stability. The two profiles shown are taken from two subjects, one having good stability, the other being less stable.

Velocity profiles can give an indication of a patient's speed of sway. With careful design of a suitable force command file, perhaps using step functions, it is conceivable that information relating to patient reaction times could be derived.

The anterior/posterior and medial/lateral velocity traces may be combined to generate a non-directional speed profile, the horizontal axis representing the scalar speed (rather than velocity) of a patient's centre of pressure (Fig. 7.7). Once again a patient's stability has a bearing on the peak value and spread of the profile.
Two additional quantitative parameters may be derived from this form of display. During a patient trial the distance traversed by the position of the centre of pressure may be totalled, giving an overall locus length. This is shown as PATH LENGTH on the speed profile. The average speed of the patient is also available and is displayed as MEAN SPEED. Both of these measurements have been used in stability research and have been found of relevance in determining the stability of patients.

All of the profiles discussed so far have used data taken from the movements of a patient’s centre of pressure derived from the force plate. Positional
information from both actuators may also be used to produce four similar profiles relating to harness movement. Since the harness attaches around the iliac crests, the four profiles approximately relate to the movements of a patient's centre of gravity. Centre of gravity variations therefore may also be examined by using the profiles, and locus length and mean speed data may be derived.

Use of the profile displays, force plate maps, and Cartesian displays gives a reasonably broad method of investigating and analysing results from patient trials. The method of storing incoming data allows other display techniques to be developed in the future. By using the display techniques already developed it is intended that specific characteristics of stability may be identified in individual patients so that suitable therapy specifically geared to the individual patient may be ascertained. This may only realistically be done by examining and analysing large quantities of data files from many patient trials with a view to developing procedures and techniques to make it possible to quantify with more confidence a patient's relative stability.
The objective of this pilot study was to investigate the viability of using highly controlled external forces to enhance a patient's stability. The main objective of this report is to describe the development and operation of the prototype system that generates the precise forces suitable for stability enhancement and monitors and records a patient's responses to these force stimuli. The project was initially of 18 months duration and it was envisaged at the start that most of this time would involve development work and commissioning, leaving only a small amount of time for patient trials. Consequently the results discussed here are used only to prove that in principle the concept is valid - ie a system based upon applying highly controlled dynamic external forces could prove highly beneficial in the short and long term rehabilitation of people suffering impaired balance. A further study would be required to build upon the results obtained here by determining the most suitable methods and therapy regimes, and the effects upon patients in the longer term.

The timescale and large amount of data files inevitable with this type of long-term patient analysis puts this task well beyond the scope of this study. Nevertheless the responses of a number of patients were obtained and results inferred from these tests are highly encouraging.

8.1 Methodology

The procedure adopted for each patient was to fit the harness to the individual and obtain a null file, giving a measure of the inherent balance of the patient without any supportive forces being applied. The actuators were configured to provide no support but simply follow the movements of the patient. This null procedure was repeated at the end of a session and served as a comparison for all of the data.
Following the null procedure the patient was given support by means of a simulated spring, damper, and/or both together with in each case a high value followed by a medium and then a low value. Further to this basic procedure the patients were also subjected to a series of force profiles from each actuator using a specific force command profile. These would enable the balance and stability of the patient to be quantified in engineering terms and provide some basic information regarding the behaviour of patients with impaired balance. All of these procedures were repeated with subjects with normal balance for comparison purposes.

This scheme of force profiles was repeated at least twice for each patient and the data printed out for analysis.

8.2 Patients

In the original proposal it was envisaged that about thirty patients would be involved in the study. However there was an unaccountable shortage of patients when the system was ready for operation. There were many stroke patients being treated locally either as in-patients or out-patients at the Derbyshire Royal Infirmary where the project was based, but many of these were unsuitable. This was because they either had additional problems which would have made interpretation of the results too difficult or they were too early in their rehabilitation programme. It was clear from an early stage that the only patients that were suitable for this initial study were those who had had a stroke but who were in the latter stages of recovery and were unaffected by any other conditions likely to affect their balance and so invalidate the results.

Because of these problems it was only possible to examine four patients. It was intended that these patients would be seen on at least four occasions but again this proved impossible. Due to problems with transport, timings of visits and the inevitable down-time associated with this type of prototype engineering system it was only possible to see each patient on two occasions.
The patients were as detailed below:

<table>
<thead>
<tr>
<th>Patient</th>
<th>Age</th>
<th>Sex</th>
<th>R/L</th>
<th>Time past stroke</th>
<th>Assessment</th>
</tr>
</thead>
<tbody>
<tr>
<td>CMcL</td>
<td>62</td>
<td>M</td>
<td>L</td>
<td>6 weeks</td>
<td>3</td>
</tr>
<tr>
<td>NB</td>
<td>64</td>
<td>F</td>
<td>L</td>
<td>8 months</td>
<td>2/3</td>
</tr>
<tr>
<td>JJ</td>
<td>83</td>
<td>M</td>
<td>L</td>
<td>6 months</td>
<td>1/2</td>
</tr>
<tr>
<td>VL</td>
<td>50</td>
<td>M</td>
<td>R</td>
<td>2 months</td>
<td>3</td>
</tr>
</tbody>
</table>

All of the patients possessed inherent stability without external support whilst standing although they all needed some assistance with their walking. The assessment level for each patient is an estimate of ability assessed on an arbitrary scale by the physiotherapist involved with these patients. A score of zero implied very poor stability, with the subject having difficulty in standing unaided. Control volunteers were included to obtain an estimate of the kind of results achievable by stable people. These people were all in the age range 20 - 30 which was not ideal as it would have been better to have people of the same age as the patients. This was not feasible due to problems of logistics. Details of the volunteer subjects are as follows:

<table>
<thead>
<tr>
<th>Volunteer</th>
<th>Age</th>
<th>Sex</th>
</tr>
</thead>
<tbody>
<tr>
<td>AB</td>
<td>21</td>
<td>F</td>
</tr>
<tr>
<td>DT</td>
<td>28</td>
<td>M</td>
</tr>
<tr>
<td>DP</td>
<td>31</td>
<td>M</td>
</tr>
<tr>
<td>PR</td>
<td>27</td>
<td>M</td>
</tr>
<tr>
<td>KS</td>
<td>20</td>
<td>M</td>
</tr>
</tbody>
</table>

It is inevitable that with so few patients the results will be far from conclusive. However, there are patterns emerging that would indicate that the principle outlined in the original proposal is valid.

8.3 Patient Trials

Despite the small number of patients there are a large number of computer files to examine for each patient in each combination of support type. It is clear that for all patients some form of support proved advantageous. In all cases the support produced a more stable stance which was more evenly balanced between the feet.
To illustrate the effect of the stabilisation of the patients by the use of spring support, damping, or both together, the data from patient NB will be examined. Only two displays of the data are printed out of all the combinations possible. The displays chosen were the displacement map of the centre of pressure on the force plate and the displacement profile of the actuators which relates to the patient’s centre of gravity.

On the centre of pressure maps (Fig. 8.1) the points representing the data form an irregular pattern and this has an enclosing rectangle drawn around it, the area of which is indicated at the top of the display. Fig. 8.1(a) shows patient NB

![Displacement maps for patient NB.](image)
with no supportive forces applied and indicates an area of 3.71 cm$^2$ for centre of pressure under free-standing conditions. With a supportive spring force of 2000Nm$^{-1}$ this area reduces significantly to 2.3cm$^2$ (Fig. 8.1(b)), indicating that the patient’s swaying movements are reduced by the supportive force.

A similar improvement can be observed with profile displays rather than area maps (Fig. 8.2). Here the data are taken from the actuator positions and represents the position of the patient’s centre of gravity. The first display shows the patient’s performance with no supportive forces and gives a relatively broad displacement profile which is grossly offset to the hemiplegic side indicating
that the patient is performing large sway movements whilst adopting an
unnatural leaning posture.

The situation changes when the same spring force as in the previous example
is introduced (Fig. 8.2(b)). Here the profile is considerably narrower and is
well-centred. This shows a smaller level of sway and indicates that the patient is
being held upright in a more normal erect posture rather than with the traditional
lean to the affected side.

Fig. 8.3 shows the situation with damping of 500 N/ms\(^{-1}\). Here the area of the

![Diagram showing displacement map and profiles for patient NB.](image_url)
centre of pressure is slightly less at 1.9cm² than with spring restraint; the displacement profile is narrower but offset to the affected side as indicated in Fig. 8.1. This suggests that a strong spring force will provide an erect posture but the patient is not sure of this posture which feels unnatural to them. Thus a relatively large area of centre of pressure is produced by the patient almost fighting to return to his or her natural position. With damping the patient is allowed to return to their comfortable natural offset position and allows them to remain there in a stable state.

A compromise situation is found in Fig. 8.4 using spring restraint of

![Diagram showing displacement map and profiles for patient NB.](image-url)
500 Nm\(^{-1}\) and damping of 500 N/\text{ms}\(^{-1}\). This produces the smallest area of centre of pressure at 1.1 cm\(^2\) and a very narrow displacement profile which also indicates a nearly erect posture. The situation presented using NB as an example is repeated for the other patients although there are detail differences. However, in the most unstable patient (JJ) the differences were large but more erratic.

As mentioned earlier in the report the patients were also involved in the testing of their responses to known disturbances. These included sine waves, ramps, and other regular forms aimed at producing uniform displacement and velocity profiles. Fig. 8.5(a) shows the displacement profile of the actuators

![Displacement profile](image1)

![Cartesian display](image2)

**Fig. 8.5** Displacement profile and Cartesian display for patient VL with sideways ramping force command and no restraining forces.
derived from a trial involving no spring or damping support and a ramping force generated by the actuator system. With this ramping force command file the profile should ideally consist of a narrow vertical band on the upper trace and a broad band of uniform height on the lower one. Fig. 8.5(b) shows the same force command file plotted against the centre of pressure and actuator motions. The instability of the patient is clear when the axis at the top of this display is examined, which should show no movement because the stimulus is in the direction at right angles. To make any conclusions from this would involve studying more patients on many more occasions, but the results demonstrate the engineering principles of the tests.

It would have been more satisfactory to have measured each patient more times and examined the effects of the stabilising forces on the rate of improvement of their walking. It would also have been better to have had some earlier stage hemiplegic patients to try and assess their progress with this form of therapy. However this was not possible until the basic reliability and safety of the equipment could be demonstrated.

8.4 Conclusions

A smaller number of patients were tested than had been expected, but the results of the project are clear - that it is technically feasible to generate carefully controlled stabilising forces, and that these do indeed enhance the postural stability of patients. Clearly though, the tests performed and results described here are rudimentary. All of the patient trials involved patients in the standing position responding to force stimuli or supportive forces, or effectively free-standing. Some of the methods of quantifying postural stability that have been described in this report are expected to make a valuable contribution in the field of measurements in medical engineering. However, rehabilitation comes mainly from objective exercise which can only be provided when the concepts are extended to include walking. The next stage therefore would be to develop the conceptual system described in Section 1.2, ie to develop a system that gives a patient the necessary dynamic support whilst allowing them to exercise and walk in as normal a manner as possible. Only then can the likely advantages of this form of therapy be fully investigated and appreciated.
9. Appendices

9.1 F241 Load Cell Specifications
9.2 Printed Armature DC Motor
9.3 A/D - D/A Interface Package
9.4 System Diagrams
9.5 Software Structure
APPENDIX 1: F241 LOAD CELL SPECIFICATION

Description

The four-arm active bridge transducing element is fully temperature compensated and housed in a flat cylindrical case of high intrinsic strength, containing positive compression overload stops.

Attachment is by an axial threaded hole at one end face and at the opposite end an internally threaded sensing shaft is fitted for tensile attachment.

The load cell is designed to weigh-system standards and has high cross-load strength and off axis error compensation.
Specifications

Rated load: 300 Newtons
Load point movement: 0.25mm at rated load
Over-range: 50% without change of parameters
Sideload: 100% rated load can be applied at 90°, 10mm from end face without damage to the unit

Case material: Aluminium
Temperature range: -10°C to +50°C
R_in unnormalised: $415\Omega +/-10\Omega$
R_out: $350\Omega +/-3\Omega$
R_insulation: $2000M\Omega$ at 100V
Supply: 10V maximum a.c. or d.c.
Signal output: 2.3mV/V +/-10%
Zero balance: +/-3%
Electrical connections: Glanded flying lead, 4 core

Connections

<table>
<thead>
<tr>
<th>Color</th>
<th>Connection</th>
</tr>
</thead>
<tbody>
<tr>
<td>Red</td>
<td>Supply +ve</td>
</tr>
<tr>
<td>Blue</td>
<td>Supply -ve</td>
</tr>
<tr>
<td>White</td>
<td>Sensing +ve</td>
</tr>
<tr>
<td>Black</td>
<td>Sensing -ve</td>
</tr>
</tbody>
</table>

Supplier

NOVATECH MEASUREMENTS LTD
83 Castleham Road
St. Leonards-on-Sea
East Sussex
TN38 9NT
APPENDIX 2: Printed Armature DC Motor

Description

In the printed armature motor the wound coils of a conventional cylindrical armature are replaced by flat conductors formed on a non-magnetic disc. A separate commutator is unnecessary as the brushes bear directly upon the armature conductors. The axial field is produced by a high-coercitivity ferrite magnet system which, because of the short length, allows a shallow form of construction to be adopted. The magnet system is mounted on the rear end plate carrying the brush holders, and the flux, after passing through the armature, returns by way of the front end plate. The attractive force between the two end plates is used to join the two halves of the motor so that securing screws are unnecessary.

Torque Performance

In conventional motors, torque modulation is produced by reluctance changes as the slots pass under each pole tip, and this variation can be troublesome at very low speeds. The undesirable cogging characteristic is absent in the printed motor as the armature does not contain any magnetic material. This feature, together with a high number of commutation periods, gives a smooth torque down to zero speed.
The absence of magnetic material in the armature also means that the torque output is directly proportional to the armature current and is not limited by saturation. This allows a pulse torque of up to five times the normal full load figure to be developed, although the motor must be derated due to thermal limitations.

**Electrical Characteristics**

**Rated Full Load**
- **Nominal Voltage**: 24V
- **Input Current**: 5A
- **Speed**: 377rads/sec
- **Power Output**: 75 Joules/sec
- **Torque**: 0.02kgm
- **EMF Constant**: 0.05V/rads\(^{-1}\)
- **Torque Constant**: 0.051Nm/A
- **Damping Constant**: 6x10\(^{-4}\)kgm/rads\(^{-1}\)
- **Thermal Resistance**: 2\(\Omega\)
- **Comm Periods**: 141
- **Armature Inertia**: 1.64x10\(^{-5}\)kgms\(^{2}\)

**Supplier**

Printed Motors Ltd
Bordon Trading Estate
Oakhanger Road
Bordon
Hants GU35 9HY
APPENDIX 3: A/D - D/A Interface Package

Description

The A/D and D/A board uses an eight bit digital to analogue converter (DAC) for analogue output and an eight bit successive approximation analogue to digital converter (ADC) for sampling analogue signals.

There are sixteen analogue output channels from the DAC. Each channel may be programmed to output a voltage in the range of -5V to +5V.

There are also sixteen channels of analogue input to the ADC. Each channel will accept voltages in the range -5 to +5 volts and convert that voltage to an eight bit digital value. The ADC is capable of conversions in 9 microseconds.

The interface package is memory mapped so reading any ADC channel and outputting to any DAC channel is accomplished by reading and writing RAM memory as appropriate.

Specifications

Analogue to Digital Conversion

<table>
<thead>
<tr>
<th>Specification</th>
<th>AC</th>
<th>DC</th>
</tr>
</thead>
<tbody>
<tr>
<td>Input Impedance</td>
<td>1KΩ</td>
<td>1MΩ</td>
</tr>
<tr>
<td>Voltage Range</td>
<td>-5V to +5V</td>
<td></td>
</tr>
<tr>
<td>Conversion Time</td>
<td>9 Microseconds</td>
<td></td>
</tr>
<tr>
<td>Accuracy</td>
<td>Absolute +/- 3% FSR</td>
<td></td>
</tr>
<tr>
<td></td>
<td>Relative +/- 1 bit</td>
<td></td>
</tr>
</tbody>
</table>
Digital to Analogue Conversion

Output Voltage: +5V to -5V
Output Current: Source 2mA  
               Sink 2mA  
Accuracy: Absolute +/- 3% FSR  
          Relative +/- 1 bit  
Slew Rate: 10V/ms  
Dynamic Zout: 10Ω  
Outputs: All monotonic

A/D and D/A Conversion Chart

Analogue voltages on input and output range from -5V to +5v. These analogue signals are represented by digital values in the range 0 to 255. The chart below shows the mapping relationship between input and output voltage, and digital value. The gradient is approximately 39mV/bit, which indicates the minimum resolution of which the system is capable.
APPENDIX 4: System Diagrams

The main circuit components comprising the force control system for one actuator are shown below. Two complete systems are required. A detailed circuit diagram for each element is given in the following pages, with the exception of the power amplifier and digital controller. The power amplifier is a commercial unit and is not detailed. The digital controller implements all system interlocks and a number of safety features. It comprises very basic logic electronics and is also not detailed.
Quadrature Position Signal Decode Circuitry

To slider gate switches in actuator body

GND

CLOCK MODULE 20KHz

VCC

120R

P1

S1

S2

Ck

74HCT175

74HCT86

74HCT51

To tachometer counter circuit

UP pulse

DOWN pulse
All supplies decoupled 0.1uF to ground.

All resistors 10% tolerance

Analogue Circuit
APPENDIX 5: Software Structure

The memory capability of the Apple Microcomputer was very limited so for a software system as extensive as the actuator control system it was necessary to reload relevant code as different software functions were required. The total software system (including code, runtime files, data storage etc.) comprised around 120kbytes of code, but only 35kbytes of Random Access Memory (RAM) was available to the software user. A proportion of this space was required for the Basic interpreter, Disc Operating system, and resident code to drive the main Apple graphics screen. The remaining space had to be allocated between incoming data, force command data, and executable software code.

The Basic language was used to write code that implemented operator functions and controlled the actuator system. All Basic code was ultimately compiled in the final version, adding further pressure to the available space as compiled code took roughly 60% more memory than uncompiled code. The compiled code executed significantly faster than the uncompiled code, a feature that was considered more important than the space penalty. Speed of operation was important. The computer switched very slowly between each main software function (since it had to re-load the relevant code each time) but the speed of execution within each section was very rapid. In practice this arrangement worked very well since changes between each function were relatively infrequent, with much time being spent within each function. It was also important that during the data collection stage the operator and patient did not have to wait during computer processing time. Using compiled code allowed very rapid turnaround within this function.

There are four independent software functions and these are accessed through a top-level main menu. The four functions are: Force Command File Generation, Data Collection, Data Analysis, and Data Display. These are shown in the software structure diagram. The function of each is discussed below.
Force Command File Generation (17kbytes)

A command file is used during a patient trial to generate a force profile for both actuators. Each file contains two independent force profiles - one for each actuator. Each profile is designed by the operator using the computer keyboard and the main graphics screen. The operator can use the cursor keys to design the shape of each profile or can specify the magnitude, frequency, and phase of a sinusoidal characteristic. During the design phase the profile is contained in RAM and can be easily modified. When the required profile is achieved it is given a title and is saved to floppy disc. With a large number of command files it proved very easy to forget the profile shape of individual files so a manual index was maintained relating filename to profile shape. Each profile was also momentarily displayed before being employed in a patient trial.

Data Collection (10kbytes)

This software function deals with the initiation and control of a patient trial, data collection, and subsequent data storage. Operator response is important before a trial commences as the software generates an itemised checklist that the operator must acknowledge item by item. Thus the operator will be required to check the selected command file, analogue controller settings, patient status and general system status. The data collection code also performs status checks to ensure system functionality before a patient trial commences.

After initiation of a trial, system operation and data collection are automatic. Sampling may be aborted at any time by the operator (or patient if a panic
situation occurs) but if a trial is successful the software will process the data and will then store it as a floppy disc file. This process is very rapid, ensuring that the processing overhead between trials is minimal.

**Data Analysis (10kbytes)**

This software function generates profile displays from stored data files. It is used 'off-line' - ie when data collection from trials has ceased. The operator drives a simple menu to select the profile type and also the file from which the data is retrieved. Each profile is displayed on the graphics screen. Since it is the data file that has been stored rather than the profile display, the data must be analysed each time a profile display is chosen. This process is quite slow but a hard copy function exists for each type of profile display so it need be processed only once to generate a permanent record of the trial.

**Data Display (15kbytes)**

As with the data analysis function this code is used 'off-line' to display data files generated from patient trials. No data analysis is required. Three types of display are available: Cartesian, Real-time and Area maps. Each of these has been discussed in detail in Chapter 7.

The functionality of the code is simple. The operator is given a number of menu options which must be used to select display type, data file, and display options within each display type (eg Cartesian display can show centre of gravity, centre of pressure, and force command individually or in combination). Generation of each display is very rapid since the data within the data file has already been processed at the data collection stage into a form suited to Cartesian and area map displays. A hard copy facility exists for each style of display.

A third option allows the operator to select a real-time display of force plate information. This represents the patient's centre of pressure on the floor and is particularly useful in educating a patient about posture. As the patient moves, the display responds by moving a cross within a rectangle representing the force plate. Thus the patient can see the immediate effect of posture change. The software to do this formed a substantial part of the display code. Compiled Basic proved incapable of performing the sample and display functions with sufficient
speed so the code was written in assembler and called from the compiled Basic software.

Inevitably the system produced a large number of disc files. Force command files were always well documented and kept separate from patient data files. A new floppy disc was used for each patient in each session and manual records were kept detailing system settings and general comments. This record-keeping function could have been handled by the computer but a manual system was considered a simpler option.
10. References

[1]. Tallis, R. Neurological Rehabilitation - The Next 30 Years Journal of Physiotherapy 1984, 70, 5

[2]. Guimaraes, R.M. Characteristics of the Gait in Old People Who Fall International Rehabilitation Medical 1980, 2, 177-180

[3]. Harris, G.F. A method for the display of Balance Platform Centre of Pressure Data Journal of Biomechanics 1982, 15, 741-745


*Postural Movements During Normal Standing In Man*  
Journal of Anatomy, 1959, 93, 524-539

[10] Duggar, B.C.  
*The Centre Of Gravity Of The Human Body*  
Human Factors, 1962, 131

*Power Spectral Density Analysis Of The Standing Sway Of Males*  

*Correlation Analysis Of The Body Sway In Standing Posture*  
Agressologie, 1976, 17B, 7-14

[13] Shipley, R.E. and Harley, R.J.  
*A Device for Estimating Stability of Stance in Human Subjects*  
Psychophysiology, 1971, 7, 287

*Normal Postural Stability and Steadiness: Quantitative Assessment*  
J. Bone Jt Surg, 1975, 57, 510-516
11. Bibliography

Duncan, P.  
*Stroke Rehabilitation*  
Year Book Medical 1987

Licht, S.  
*Stroke and its Rehabilitation*  
S. Licht 1975

Rose, F.C.  
*Advances in Stroke Therapy*  

Rose, F.C.  
*Stroke, The Facts*  

Sarno, J.E.  
*Stroke, The Condition and The Patient*  
McGraw-Hill 1979
12. Project Publications

12.1 Pages 113 to 121

Enhancing Postural Stability in Hemiplegics Using Externally Applied Forces

International Journal of Rehabilitation
Supplement No. 5 to Vol.10, No. 4 (1987)
Goodall, R.M., Pratt, D.J., Rodgers, C., Murray-Leslie, C.M.

12.2 Pages 122 to 129

Control and Instumentation Systems in the Enhancement of Postural Stability

Progress Reports on Electronics in Medicine and Biology,
The Institution of Electronic and Radio Engineers, 1988
Rodgers, C., Goodall, R.M., Pratt, D.J.
Enhancing postural stability in hemiplegics using externally applied forces

R. M. Godall, D. J. Pratt, C. T. Rogers, and C. M. Murray-Leslie, Loughborough & Derby (United Kingdom)

1. Introduction

Significant medical resources and funds are taken up in the rehabilitation of patients having walking problems. These problems may exist either because of a specific neurological disorder or as a consequence of trauma. Hemiplegia is by far the most significant disorder which leads to the need for intensive rehabilitation techniques, and recent statistics (1) have highlighted the tremendous costs involved in the treatment of hemiplegics: In the United Kingdom 5 % of National Health Service resources are used up in their rehabilitation, with associated bed occupancies of 13 % in general medical wards and 25 % in geriatric wards.

Physiotherapy departments are heavily involved in the process of rehabilitation, and again stroke patients are a major proportion of their work load. Although there is a continuing development of existing appliances, such as crutches and walking frames, these seem to have a fundamental disadvantage in that they create an unnatural posture and gait, largely because the support forces are applied via the arms, and it is often necessary to strengthen the upper limbs prior to the actual process of walking rehabilitation.

The work described here is the first stage in a research programme aimed at a radically new form of therapy for walking rehabilitation, the essence of which is to provide stability by the use of carefully controlled forces applied externally at or around the patient's centre of gravity. The mechanism which applies these stabilising forces will be under electronic control, and the patient will be able to exercise in a normal erect posture and with bidepal gait. The electronic control will enable the level of external stabilisation to be progressively withdrawn as the therapy proceeds, thereby encouraging the patient to re-develop stability on his own.

The paper describes a pilot study carried out to prove the technological and clinical concepts applied to stability in posture only; this has formed the basis for extending the research to the investigation of equipment which will enhance the patient's stability during ambulation. A brief description of the engineering aspects of the experimental apparatus is included, but the emphasis is upon the methods developed for collecting and processing data collected to quantify postural stability, and upon the results of clinical trials carried out with able-bodied and hemiplegic subjects.

2. The experimental apparatus

The ability to provide carefully controlled stabilising forces in response to the patient's movements is central to the whole concept. This has been achieved by means of electrically-driven actuators, a schematic representation of which is given in Figure 1. The electric motor turns a lead screw, and a high efficiency nut using ball bearings runs up and down the screw. A shaft is connected to this screw, and thereby rotary motion is converted into linear movement. The electronic control scheme, also shown in Figure 1, has two distinct actions. The first involves the addition of a force transducer to measure the net force applied by the actuator; this measurement is compared with a force command signal, and the error is amplified and used to drive the electric motor via a power amplifier. If the force command is zero, then it is possible to move the shaft of the actuator in and out with very small
restraining forces - the feedback action causes the motor to drive so as to try and maintain zero force. The second action is to create force commands in a highly controllable manner depending upon the movements of the patient attached to the end of the actuator. These movements are measured by a counting-type transducer built into the actuator itself, which produces a signal proportional to the displacement from the central position, and by applying this signal as a force command creates a spring-like characteristic; varying the gain of the position feedback enables the rate of this spring to be electronically varied. A similar process using the actuator velocity gives a controlled damper-like characteristic. A third option is to apply a constant command signal, and this causes a constant force to be applied independently of the actuator's movement. It is of course possible to apply various combinations of all three types of force, as may be appropriate for stabilisation of the patient.

Figure 1: The actuator and its control scheme

The actuators were designed to have a stroke of 300 mm, a maximum speed of 300 mm/s, can produce a maximum force of 100 N. They are mounted in a gimbal arrangement in frameworks as shown in Figure 2, such that horizontal forces can be applied to a harness fitted to the patient. This harness was specially designed to provide a firm coupling between the actuators and the patient, and also to give an easily adjustable and comfortable fit for a range of patients. Two actuators are used in order to apply forces in the antero-posterior and medio-lateral directions. Figure 3 shows the way in which they are arranged. It also illustrates the other components of the complete system. The patient stands on a force measuring plate; the signals from this and the displacement and force signals from the actuators are monitored by a microcomputer. Signals from the computer to the actuators' electronic controller mean that the experimental investigation is directed from the computer's keyboard.
Figure 2: *The actuator in its framework*

Figure 3: *The experimental arrangement*

3. The computer system

The microcomputer, an APPLE IIe, performs an number of functions: overall supervision of the test procedure (including provision of force commands for the actuator controllers), collection and storage of the test data, and subsequent processing and presentation of the results.
3.1 Supervisory function
A particular test to assess a patient's postural stability consists of a pre-determined period during which the various measurements are recorded. Initiating the recording process, identifying the test run and storing the data are all carried out by the computer. Two different types of test are possible. In the first, the patient simply stands upright; the patient's movements with and without the external stabilising forces activated can be compared. Alternatively it is possible to create a sequence of force commands on the computer; during the trial the actuators apply this sequence of forces to the patient, and his response to the disturbances helps to assess stability. Typical force command sequences may be purely sinusoidal or combinations of falling and rising ramps and steps. Sequences of force command can be saved on floppy disc and retrieved for use in many different patient trials, which ensures that for each trial the same forces and timings are maintained. Such a technique generates very consistent test conditions and ensures that results from using the same force profile with different patients (or with different levels of external stabilisation) may be validly compared.

3.2 Data collection and storage
The computer samples 8 channels of information: two actuator forces, two actuator displacements, and four vertical forces from the plate. The actuator force signals indicate the level of stabilising action being applied, the actuator displacement signals give the movement of the patient's centre of gravity and the force plate signals enable the position of the patient's centre of pressure to be determined. All these signals are sampled at 32Hz (a lower sample rate has been found to be less satisfactory), and in the course of a trial of 60 seconds duration some 15,000 data samples are stored temporarily in the computer's memory. At the end of the trial the four force plate signals are processed to give the co-ordinates of the centre of pressure, and this and the other information are saved in a suitably named file on floppy disc.

3.3 Presentation of results
A number of different display techniques for the results have been developed. These may be sub-divided into two categories - straightforward time histories or overall profiles of the trial.
For the time histories, the results are presented in Cartesian form with time as the major axis. Each display has separate graphs for the M-L and A-P directions, and combinations of the following signals may be plotted: patient centre of pressure at the force plate, actuator movement, force command and actual force. Figure 4 shows a typical set of results displayed in this form. The displays are useful in that they give a good visual indication of how a patient may have performed during a trial. Although they are non-mathematical, they may be used to compare visually the performance of different patients, or the same patient over a period of time. In addition, trends occurring over the 60 second period may be noted. For example, a patient may become noticeably less stable toward the end of a trial, perhaps owing to fatigue.
Overall profiles of the results of the trial can be obtained by processing the data. One of the major techniques used is to present the Cartesian co-ordinates of the patient's centre of pressure in the form of a force plate map. This is a graphical representation of a plan view of the force plate, with the position of the centre of pressure superimposed. The magnitude of the patient's sway over the trial period may be quantified by the area of an enclosing rectangle - a typical force plate map is shown in Figure 5. Another method is to use histograms. Both the centre of pressure co-ordinates and the actuator displacements (i.e. the movements of the patient's hips) can be plotted as statistical distributions, the vertical
Figure 4: Example of time history presentation of results

axis being the percentage of time spent at a particular displacement; means and standard deviations can be derived to quantify these distributions. Figure 6 shows typical profiles for the centre of pressure position for an able-bodied and a hemiplegic subject. These not only demonstrate the reduced sway of the able-bodied subject, but also highlight the typical hemiplegic 'lean' away from the upright position. The coordinates which give the A-P and M-L displacements can alternatively be used to derive a single figure for the distance away from a reference point - a similar statistical presentation can be used. All these profile type of presentations are valuable in that they give a more easily assimilated indication of a patient's performance during a trial. The area of the enclosing rectangle of movements on the force plate map has been adopted as a primary indicator of stability because it provides a single parameter for comparison between patients or between different levels of external stabilisation.

Figure 5: Example of force plate map

4. Clinical trials

Limited trials have been carried out with both able-bodied and hemiplegic subjects using the system which has been described. Most of the trials with able-bodied subjects were used to commission the equipment, validate the instrumentation and establish a sensible methodol-
ogy, but the results also provide a base line for comparison with those from the hemiplegic subjects. This section details the methods used in these trials and discusses the preliminary findings.

4.1 Methodology

The procedure adopted for each patient was to fit the harness to the individual and obtain a 'null' file, i.e. a 60 second trial with no force commands. This gave a measure of the inherent balance of the patient without any supportive forces being applied. (The actuator controllers were set to provide no support but just to follow the movement of the patient). This 'null' procedure was repeated halfway through the session and once again at the end; this served as a basis for comparison for all the data. Following the 'null' procedure, the patient was given support by means of a simulated spring (S), damper (D) and/or both together (SD) with, in each case, a high value followed by a medium and then a low value. Further to this basic procedure, the patients were also subjected to a series of force command profiles from each actuator as described in the computer system section. These enabled the balance and stability of the patient to be quantified in engineering terms and to provide some basic information regarding the behaviour of patients with impaired balance. All of these procedures were repeated with subjects with normal balance for comparison purposes. This scheme of force profiles was repeated at least twice for each patient and the data printed out for analysis.

4.2 Patients

Suitable hemiplegic subjects were selected for the research programme on the basis of being in the latter stages of recovery and also being unaffected by any other conditions likely to affect their balance and so confuse the results. All of the patients had inherent stability without external support whilst standing, although they all needed some assistance with their walking. The trials have so far concentrated on four patients, each of whom have attended for two sessions. Patient details are given in Table 1. The assessment rating was given by a physiotherapist and is on an arbitrary scale from 0 to 5.

<table>
<thead>
<tr>
<th>Patient</th>
<th>Age</th>
<th>Sex</th>
<th>R/L</th>
<th>Time post stroke</th>
<th>Rating</th>
</tr>
</thead>
<tbody>
<tr>
<td>CMcL</td>
<td>62</td>
<td>M</td>
<td>L</td>
<td>6 weeks</td>
<td>3</td>
</tr>
<tr>
<td>NB</td>
<td>64</td>
<td>F</td>
<td>L</td>
<td>8 months</td>
<td>2/3</td>
</tr>
<tr>
<td>JJ</td>
<td>83</td>
<td>M</td>
<td>L</td>
<td>6 months</td>
<td>1/2</td>
</tr>
<tr>
<td>VL</td>
<td>50</td>
<td>M</td>
<td>R</td>
<td>2 months</td>
<td>3</td>
</tr>
</tbody>
</table>

Control volunteers were included to obtain an estimate of the kind of results achieved by stable people. These subjects were all in the age range 20-30, which is not ideal as it would have been better to have people of the same age as the patients. Nevertheless the results give a quantification for what would be considered to be a high level of postural stability.

4.3 Results

Despite the small number of patients a large number of computer files were created, and these were examined for each patient in each combination of support type. It became clear from the results that for all the patients some form of external support was advantageous. In all cases the support produced a more stable stance and a more even balance between the feet.

To illustrate the effect of the stabilisation of the patients by the use of spring support, damping and both together, the data from patient NB is used. Only two displays of the data
are presented out of all the possible combinations of displays available. The displays chosen were the displacement map of the centre of pressure on the force plate and the displacement profile of the actuators. These present a picture of the displacements of the patient both at the level of the hips and at their feet. On the centre of pressure map the points representing the data form an irregular pattern. The enclosing rectangle is not shown, but its area is given at the top.

Figure 7 shows results for patient NB with no supportive forces applied, and indicates an area of 3.71 cm$^2$ for centre of pressure and a narrow displacement profile but grossly offset to the hemiplegic side. The picture is very different in Figure 8, which shows NB with a $S_s$ of 2000N/m. There is a reduced area of centre of pressure (2.3 cm$^2$) and a narrow displacement profile which is centred. This indicates that the patient is being held upright in a more normal erect posture than with the traditional lean to the affected side. Figure 9 shows the situation with a $D_s$ of 500N/ms$^{-1}$. Here the area of the centre of pressure is slightly less than with $S_s$ at 1.9 cm$^2$, but the displacement profile is narrower but offset to the affected side. This suggests that a strong $S_s$ will provide an erect but uncertain posture which the patient feels to be unnatural. A relatively large area of centre of pressure is produced by the patient almost fighting the system to return to their 'natural' position. With a $D_s$ the patient is allowed to return to their happy 'natural' offset position and allows them to remain there in a steady state. A compromise situation is shown by the results of Figure 10 using a $S_s$ of 500N/m and a $D_s$ of 500N/ms$^{-1}$. This produces the smallest area of centre of pressure at 1.1 cm$^2$.
and a narrow displacement profile, which indicates a nearly erect posture. The situation presented using patient NB as an example is repeated for the other patients although there are differences in detail.

Figure 7: Results for patient NB with no stabilisation forces

Figure 8: Results for patient NB with $S_s$ equals 2000 N/m

Figure 9: Results for patient NB with $D_s$ equals 500 N/m$^{-1}$

5. Conclusion

Although tests have only been carried out with a small number of patients, the results have established that it is technically feasible to provide carefully controlled stabilising forces, and that these do indeed enhance the postural stability of hemiplegic patients. It remains to assess patients regularly over the period of their recovery in order to identify any trends
shown by the quantification of postural stability. Although the long term aims of the research are towards a system of therapy, it had not been expected that there would be any therapeutic benefit from enhancing postural stability alone. However, there were some indications that a series of force inputs applied to the patients by means of the actuators could be used for a limited form of therapy similar to the principles of 'rhythmic stabilization' commonly used by physiotherapists at present. In addition to the understanding of how postural stability in hemiplegic patients may be enhanced, the research programme has also established a sound basis for extending the concepts to include walking.

6. Reference


ACKNOWLEDGEMENT
The authors wish to thank the Trent Regional Health Authority, based in Sheffield, U. K., who have provided the funds to support the research described in this paper.
CONTROL AND INSTRUMENTATION SYSTEMS IN THE ENHANCEMENT OF POSTURAL STABILITY

C. T. Rodgers*, R. M. Goodall† and D. J. Pratt‡

SUMMARY

The electronic system associated with a project to assess ways in which external forces may be used to enhance postural stability is described. The system compromises a set of electronically controlled actuators which are used to apply the forces, and a microcomputer which captures, analyses and presents the results of clinical trials.

1 Introduction

A pilot study has been carried out to assess the benefits of using carefully controlled external forces to enhance the postural stability of patients having a variety of pathologies, with particular application to hemiplegics. The objectives of the study have been to establish the engineering and clinical principles with a view to extending the concept to stability in walking. In the course of this research control engineering principles have been employed; also a microcomputer has been used both to control the experimentation and to provide novel means of presenting postural stability factors. The paper describes the electronic and computer systems which have been developed, with emphasis upon the engineering aspects of the project (the results have been described elsewhere).

2 System Specification

The overall system requirements are shown diagrammatically in Fig. 1. It can be seen that there are two major sub-systems: an actuator sub-system and computer sub-system.

The actuator sub-system consists of the actuators themselves, power amplifiers to drive the actuators, transducers and an electronic controller. Transducers are needed to measure the forces being applied to the patient, and also to measure position and velocity in order to characterise the patient's movements. A preliminary specification for the actuators was assumed as follows:

<table>
<thead>
<tr>
<th>Specification</th>
<th>Value</th>
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<tbody>
<tr>
<td>Total travel</td>
<td>0.3m</td>
</tr>
<tr>
<td>Speed</td>
<td>0.3m/s</td>
</tr>
<tr>
<td>Maximum force</td>
<td>100 N</td>
</tr>
<tr>
<td>Average force</td>
<td>30 N</td>
</tr>
</tbody>
</table>

The electronic controller was required to enable a range of stabilising forces to be applied by means of simple controls:

a) A steady force independent of the patient's movement in the range -100N to +100N.

b) A force to give a spring characteristic in response to the patient's movements variable from 0 to 2000 N/m.

c) A force to give a damping characteristic, variable from 0 to 500 N/m/s

The computer sub-system has three main requirements: control of the test, capture and storage of the data during the test, analysis and graphical presentation subsequent to the test. Control of the test involves management of its initiation and termination, and also sending a series of force commands to the actuator control system; the patient's response to this can be used to quantify postural stability with different levels and types of stabilising force. Measurements to be taken in by the computer are those from the actuators (force, displacement and velocity at the patient's centre of gravity) and also signals from a force measuring platform which indicate the size, direction and position of the reaction force at the patient's feet. Methods for analysis and presentation of data which can be used to quantify postural stability are numerous, but it is most important that those chosen are effective because an easily assimilated means of monitoring a patient's progress will form an essential part of the eventual treatment system.

3 The actuator sub-system

3.1 Actuators

In order to exert the carefully controlled forces necessary when applying support to a patient, two electric actuators are used. Fig. 2 is a photograph of the actuator, and Fig. 3 shows details of its design. The configuration of electric motor, lead screw, ball nut, and moving shaft allows rotational movement of the motor to be transformed into a linear movement of the actuator shaft.

By varying the current through the motor it is possible to alter the torque applied to the lead screw, and hence the resultant linear force applied by the moving shaft.

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Chapter 12 Project Publications

Fig 1 The System

Fig 2 Photograph of actuator

Fig 3 Details of actuator and its control
For the system to be finely controllable, a method of accurately measuring the resultant force applied by each actuator shaft is required. These force signals may then be incorporated into a control system, allowing the generation of precise forces necessary for stability trials.

3.2 Transducers
Each electric actuator incorporates two transducers, one to measure displacement, and the other, force. The displacement transducer allows the position of the moving shaft to be accurately monitored, and a load cell generates a signal proportional to the force exerted by an actuator, and consequently the force acting upon a patient. The force transducers (load cells) are secured to the end of each moving shaft such that they are sandwiched between the driving actuator and any mass or body attached to the system. As a result of their physical position they produce a true force signal proportional to the actual force applied to the body.

The position of each actuator shaft is determined by using two infra-red detectors and a circular slotted disc. These components are mounted within each actuator body, the disc being mounted on the motor shaft. Any rotary movement causes the disc to rotate and intermittently cut the infra-red beam. This generates digital signals that are processed electronically to produce analogue position and velocity signals relating to each actuator shaft.

3.3 Control System
Each actuator provides a means of applying forces, by varying an electric current supplied to each actuator motor. To derive adequate control of the forces applied, a closed loop control system is used to drive each actuator motor. This closed loop system is shown in Fig. 3. The input to the system is in the form of a force command, and this dictates what force is desired as an output from the actuators. The closed loop nature of each actuator control system ensures that any force error (i.e. the difference between the actual and desired force) is kept to a minimum. A block diagram representation of each actuator control system is shown in Fig. 4.

A commercial d.c. Power Amplifier is used to drive each actuator motor, operating in current command mode. The forward loop gain, \(G_v\), is scaled to give a current loop bandwidth of approximately 200 Hz. Each actuator motor is of the permanent magnet printed armature type possessing very low armature inertia - this is represented in a fairly conventional manner. The ball screw within each actuator body which effectively performs the task of a gearbox may be represented by a constant gain in the forward path of the control system.

Representing the characteristics of a patient connected to an actuator presents difficulties. Very little exists at present in the way of relevant published material to describe the dynamic nature of the human body. Consequently, the effect of a body connected to the system is represented as a simple concentrated mass. This has proved to be adequate, at least for preliminary studies. A spring component is included to allow for stiffness in the force transducer, and linkage from an actuator to a patient.

The components of the analogue control circuitry may also be identified in the block diagram. Using the signal from the force transducer the analogue controller closes the force loop, and generates a force-error signal that forms a current command for the Power amplifier. Provision is made to inject into the system static force, spring, and damping commands. A static force command will cause an actuator to apply a set force upon a patient, regardless of the patient's position or movements. A spring force command generates supportive forces, increasing as a patient leans away from a central (upright) position. In this way the actuator may be controlled to appear like a spring. Similarly, damping forces will oppose rapid changes in position made by a patient, and higher speeds of body measurement will be opposed. Static force, spring and damping levels may be independently varied, allowing the matching of the actuator characteristics to be generated. When connected to a patient, an actuator may be used to apply static input forces, or provide supportive forces in the form of spring and damping characteristics.

3.4 Actuator Configuration
In order to apply forces and support a patient in two dimensions, the actuators are mounted at right angles to each other. Each is mounted on a sturdy framework so that all forces are centred upon a patient's centre of gravity, being approximately at the height of the iliac crests. The actuator and framework configuration is shown in Fig. 5. Each actuator has a trunnion mounting, supported in a free gimbal arrangement that allows an actuator limited freedom to swing vertically and horizontally. This ensures that any movements made by a patient when connected to the system are accommodated. Mounting the actuators at right angles allows a force vector in any (horizontal) direction to be generated. In this way a patient may be subjected to a force, or receive support, from any direction.

A harness couples each actuator to a patient. Padded interior sections are of rigid plastazote, and are contoured to fit roughly around the iliac crests. The plastazote ensures a patient's comfort, and maintains a rigid connection between a patient and the actuators. Two rose joints on the harness allow it to be quickly connected to the actuators by means of spring clips. The rose joints take up any twisting that a patient may perform, as well as accommodating the changing geometry of the actuators when moving.
Fig 4 Control block diagram

Fig 5 The experimental arrangement

Fig 6 Position transducer circuitry
The actuator frames are positioned such that, during a trial, the patient stands centrally on a force measuring platform, giving information relating to the force reaction at the patient's feet.

3.5 Electronic Circuitry

For each actuator, the control system incorporates both analogue and digital circuitry. The analogue circuitry implements the control function described in Section 3.3. The digital circuitry constantly monitors the position of each actuator shaft and handles safety checks, as well as enabling or disabling the power amplifiers housed in the lower cabinet.

The position of an actuator is monitored by the digital circuit shown in Fig. 6. The position transducer in an actuator produces two channels of pulses as an actuator moves. These signals are processed and generate up/down signals which are fed to a 13-bit digital counter. The resultant binary count is converted to an analogue signal which represents the linear displacement of an actuator shaft, and hence the position of any patient connected to the system.

Digital circuitry also monitors the state of the system and performs safety checks for equipment failure or operator error (see Fig.7). Any potentially hazardous errors disable the motor power amplifier, causing an actuator to stop moving. As an actuator is inherently stiff when not powered, patient safety is ensured in the event of a system failure.

4. Computer System

An Apple IIe Microcomputer is used to control and monitor the system. Fig.4 shows the interaction between the computer and the rest of the system. The processing power of the computer makes it possible to sample, analyse, and display data. It may also be made to control each actuator by generating force command signals.

4.1 Data collection

Signals from the Apple Computer are sent and received by using a commercial interfacing package. This may be used to transmit up to 16 analogue signals, and receive and monitor a further 16 input channels. The architecture of the computer and interface is configured such that the monitoring of an input signal merely involves examining a particular memory location. Similarly an analogue output signal may be generated by writing to another memory location.

The floor mounted force measuring platform generates four signals of interest, one signal from each of the load cells spaced symmetrically within the plate. These four signals are processed to generate the X and Y components of the centre of pressure. (The converting equation is derived simply from the geometry of the force platform.)

The computer samples the following 8 input channels at 32 Hz: the 2 load cells to give the forces exerted via the harness, 2 actuator displacements to give the position of the C of G of the patient's body, and 4 signals from the force plate relating to the patient's centre of pressure. During a trial of sixty seconds duration, the computer stores over fifteen thousand data samples temporarily in RAM memory. At the successful conclusion of a patient trial, the data relating to centre of pressure is processed, and the resultant data is stored as a named file on floppy disc.

4.2 Force Command

The software developed for the system allows the operator to control each actuator by designing and saving a force command file. The file consists of a mathematical force profile for each actuator, in the form of a sequence of numeric samples. During a patient trial, these samples are sent to each analogue controller circuit, and appear as forces exerted by each actuator. This happens at the same frequency as data sampling occurs. Typical force command files are shown in Fig.8. Each actuator may be controlled independently so as to apply from zero to 100N of force, either pushing or pulling. Sinusoidal force command files may be created by specifying a magnitude and frequency for each actuator. Other force command files are created by using the graphics capability on the Apple computer. The resulting profiles are processed to generate a sequence of numeric samples, subsequently being stored as a single command file on floppy disc. Having the facility to create command profiles in this way allows a particular force profile to be retrieved from disc and used in many different patient trials. This ensures that for each trial the same forces and timings are always maintained. Such a technique generates very consistent test conditions, and ensures that results that from using the same force profile on different patients may be validly compared.

With the computer sampling data from the system, and controlling each actuator, operator intervention during a patient trial is minimal. Force, spring and damping controls on the analogue controller remain active whilst standard force command files are used during a patient trial, with varying levels of supporting force. Various combinations of spring and damping forces may be employed, and the benefit of supportive forces assessed.

4.3 Software

The software system is menu driven from floppy disc. All programs are in structured Basic, with a small number of Assembly language routines. A structure diagram for the software system is shown in Fig.9. The software system may be broadly split into three sub-systems: creation and storage of force command files, capture and storage of data generated during patient trials, and display and analysis of data.
Chapter 12 Project Publications

Fig 7 Digital circuitry used to monitor system

Fig 8 Examples of force command files

Fig 9 Software diagram
Control spreads outwards from the main menu until a specific task is accomplished, after which control returns to the menu. Such a top-down programming approach facilitates potential software modifications, and also allows additional routines that may be required in future. Specifically, other display techniques may be developed, and accommodated within the existing system, without the need for extensive software modification.

4.4 Presentation Techniques
The display techniques detailed below represent an attempt to extract and display some of the better documented and well known stability parameters, along with some display methods not widely adopted. Each display technique may be used to highlight, in a comparative manner, particular features of stability during successive patient trials. Data is displayed either in Cartesian or profile form. Displays in Cartesian form require only scaling to transform each data file into a meaningful form, whereas displays of the profile type require a small amount of mathematical processing.

Signals from the system may be displayed in Cartesian form, with time as the major axis. A typical display is shown in Fig. 10, in which three plots are overlayed for comparison. Although the displays are non-mathematical, they may be used to visually compare the performance of different patients, or the same patient over a period of time. In addition, trends occurring over the sixty second period may be noticed. For example, a patient may become noticeably less stable towards the end of a trial, perhaps owing to fatigue.

Data relating to a patient's centre of pressure may also be displayed in the form of a force plate map. This is simply a graphical representation of a plan view of the Kistler force place, with centre of pressure position being superimposed. It may be used as a real-time active display i.e. the display alters as the patient sways, or a cumulative display, where all centre of pressure positions over a sixty second period are displayed. A typical force plate map is shown in Fig. 11 relating to patient trials involving no force commands. This map may be used in determining the magnitude of a patient's sway and will indicate whether a patient has a weakness in handling forces from any specific direction. Details relating to maximum sway magnitudes are expressed in the form of an enclosing rectangle, with the exact magnitudes being given.

Further processing of the centre of pressure and centre of gravity data allows a number of histograms to be generated. Typical histograms are shown in Fig. 12 corresponding to tests involving people with good and bad stability. During a trial a patient will sway; a high level of stability generates a highly compact displacement profile, while high time percentage values. Poorer stability causes the traces to broaden and the maximum time percentage reduces, implying that greater and more frequent excursions take place. If a patient leans to one side through a trial, causing his 'normal' centre of pressure position to be displaced, the displacement profile is shifted to the left or right of the vertical time axis. Time percentages remain unaffected. Hence valid comparisons may be made between able and less able stroke victims, since their differing weight distributions do not affect the important parameters of the displacement profile display.

With the position of a patient's centre of pressure constantly sampled, it is possible to generate histograms of velocity and speed. These can give a further indication of the patient's stability. Other parameters, such as the total distance traversed by the position of the centre of pressure and the average speed of sway, are also available to quantify postural stability.

All of the profiles so far discussed have used data taken from a patient's centre of pressure movements. Positional information from both actuators may be used to produce three similar profiles relating to harness movements. Since the harness attaches around the iliac crests, the four profiles approximately relate to a patient's centre of gravity. Centre of gravity variations may therefore also be examined by using profiles, locus length and mean speed data.

The use of the profile displays, real-time displays, force plate maps and Cartesian displays, gives a reasonably broad method of investigating and analysing results from patient trials. The method of storing incoming data allows other display techniques to be developed in the future. By using the displays already developed, it is intended that specific characteristics of stability may be identified. This may only be done by examining and analysing large quantities of data from many patient trials, so that form of correlation between results may be developed, making it possible to quantify a patient's relative stability.

5. CONCLUSION
The engineering requirements to support the concept of applying external forces in order to enhance postural stability have been thoroughly investigated. It has been shown that advanced control engineering techniques can be successfully used to provide the highly controllable forces necessary, and the effectiveness of the system of actuators which has been developed has been proved in a number of trials with both able-bodied and hemiplegic subjects. Novel techniques for quantifying and displaying factors relating to stability in posture have been demonstrated, and these are giving valuable information by which the effect the stabilising forces is being assessed.

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