DETECTION OF GAIT EVENTS USING A GYROSCOPE SENSOR IN FES DROP-FOOT CORRECTION

by

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Abstract

Electrical stimulation is a technique used to produce muscle contraction for a range of purposes, for example: exercise, cardiac pacemakers and diaphragm movement. One common use of this technique is functional electrical stimulation (FES), which is used to provide functional movement to otherwise paralysed individuals. An example is sit to stand transfers and walking for paraplegics. In 1961 Liberson pioneered the use of FES to correct drop-foot in hemiplegics. Today, FES has become routine therapy in a number of hospitals for many patients suffering from drop-foot. However, little has changed regarding the basic design of the sensor used to trigger stimulation as the subject walks.

In this study, a novel piezoelectric gyroscope-based sensor was evaluated for use within a FES drop-foot correction system. As a pilot study, three such sensors were mounted on each lower limb segment to measure joint angle. Software was developed in LabVIEW to provide a PC system to collect sensor data and calculate joint angle. Although the sensor proved unsuitable for this particular application, it did show potential for use as an FES drop-foot sensor. A second short study, involving fifteen able-bodied and ten hemiplegic subjects, demonstrated further this potential as the results compared well with similar measurements from other apparatus. The developed system used LabVIEW software for user input and graphical output, and ‘C’ software to calculate whether the foot was in contact with the ground (this could be used for stimulation timing).

In order to facilitate further evaluation, a portable data-logger was developed, capable of collecting measurements from the gyroscope sensor along with a pair of established foot sensors currently used in FES drop-foot systems. A third method was also used, which used seven cameras to track the position of reflective spheres mounted at various points on the subject’s body. The data-logger was also capable of performing the real-time calculations to determine whether the foot was in contact with the ground. Five able-bodied subjects were involved in this study. Results suggested that the gyroscope sensor was more accurate in determining foot contact with the ground than established sensors, when used by able-bodied subjects. However, in a similar study carried out with four hemiplegic patients, results were less conclusive (the camera system could not be used). Further work, including
improvements to the current system and the need for additional evaluation, is discussed in the final chapter.

This work was conducted between October 1997 and April 2001. The thesis was submitted in April 2003.
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1.1 Introduction

Walking is a means for humans to move their bodies from one place to another without assistance. While walking is not such an efficient mode of transportation as the wheel, it may be adapted to suit a variety of terrains, including rough ground and staircases.

The complex process of walking is made possible through a combination of balance, posture and limb movement. The precise muscle contractions required are coordinated by the two major divisions of the nervous system: the motor (efferent) division and the sensory (afferent) division. Normal motor control is not possible without sensory information from the body, e.g. mechanoreceptors (of which there are two types: tactile receptors providing information regarding touch, pressure and vibration, and proprioceptors monitoring the positions of joints and muscles).

Damage to the nervous system can affect both motor and sensory divisions to varying degrees, and may result in paralysis. The paralysis may affect any number of limbs: the most common combinations are paraplegia (affects both lower limbs) and hemiplegia (paralysis of the arm and leg on one side only; the trunk may also be affected). Less frequently occurring types of paralysis are quadriplegia (paralysis of all four limbs) and monoplegia (affecting one limb only). A complete spinal cord injury (SCI) can cause total loss of motor control and sensory information from the toes up to a point on the body dictated by the location of the lesion. Paraplegia usually results from damage to the lower sections of the spinal cord; quadriplegia occurs when the damage is in the upper sections.

An incomplete lesion causes only partial paralysis and sensory input loss. A cerebrovascular accident (CVA or stroke) usually causes partial loss of motor control and sensory information on one side only (hemiplegia), some or all of which may return during the recovery period. In many cases there is a greater return of upper limb
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function compared to that of the lower limb, resulting in a permanent lower limb disability. Some patients are able to walk with this disability, but it is usually the case that either the foot on the affected side is dragged along the ground, or the entire leg is swung away from the body to clear the ground. Multiple sclerosis (MS) often begins with stroke-like symptoms, but the loss of motor control and sensory information can slowly increase throughout the sufferer’s life.

Standing and walking often become difficult or impossible in many cases of stroke, MS and SCI (both complete and incomplete). Standing is important for humans psychologically (particularly when addressing others) and for blood circulation and proper function of many organs Kralj et al. (1990). Other activities, such as chair to bed transfers and washing one’s body can also become difficult or impossible for the sufferer. The majority of the 100,000 people per year in the UK who had suffered their first stroke experienced a reduced quality of life (Taylor et al., 1998).

This project was directed at the improvement in the walking of individuals who had suffered a stroke or incomplete SCI, or MS. Incidence of stroke is far higher than incomplete SCI or MS. Such improvement is possible using different methods, such as physiotherapy, drugs and orthoses. Another method is the electrical stimulation of nerves to produce muscle contraction. During walking, the muscle(s) controlled in this way are required to contract only during certain periods. These periods are determined using sensor(s) mounted on (or near) the foot or leg. In this project a novel sensor, based on a gyroscope for the detection of such periods, was developed.

1.2 Overview of Lower Limb Movement

There are three joints associated with the lower limbs: the hip, the knee and the ankle. Each joint permits different movements; the knee essentially permits only flexion and extension (some very limited movement is also possible within other planes). The hip joints permit flexion and extension, and are also capable of rotating, adducting and abducting the thigh. The ankle joint permits dorsiflexion (upwards movement of the
foot), plantar flexion (downwards movement), inversion and eversion and adduction of the foot. Figure 1.1 shows adduction, abduction, inversion and eversion movements.

![Figure 1.1 Description of Abduction, Adduction, Inversion and Eversion](image)

Movement of the hip and knee joint involve the use of many muscles, some of which are contracted only to modulate the actions of others. For example, the Biceps Femoris of the thigh may be contracted to flex the knee joint, but at the same time this action will cause hip extension. If knee flexion only is required, hip flexors may be used to counteract any hip extension and stabilise this joint. Table 1.1 shows the major muscles involved in moving the foot and the action of each one.

In table 1.1 only one muscle is shown that produces (significant) dorsiflexion of the foot: the tibialis anterior (the extensor digitorum longus and extensor hallucis longus also produce a minor dorsiflexion movement). Three muscles each adduct and invert the foot, two of which also produce plantar flexion (the Gastrocnemius is also capable of flexing the knee joint). The inverters, everters and adductors all serve to stabilise the foot during gait (preventing undesired rotation). Two types of muscle contraction are used here: concentric and eccentric. In concentric contraction, muscles are shortened in length and bones are pulled closer together to control joint movement when muscle
tension is greater than the external force, e.g. to push away from gravity). In eccentric contraction, muscles are lengthened and bones are pushed further apart to control joint movement when muscle tension is less than the external force, e.g. to decelerate the body after the foot strikes the ground during running.

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<tr>
<td>Gastrocnemius</td>
<td>Plantar flexes, Inverts and Adducts Foot</td>
</tr>
<tr>
<td>Peroneus Brevis</td>
<td>Everts Foot</td>
</tr>
<tr>
<td>Peroneus Longus</td>
<td>Everts and Plantar flexes Foot</td>
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<tr>
<td>Soleus</td>
<td>Plantar flexes, Inverts and Adducts Foot</td>
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<td>Tibialis Posterior</td>
<td>Inverts and Adducts Foot</td>
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*Table 1.1* Muscles that Evert, Invert, Adduct, Dorsiflex and Plantar flex the Foot

Figure 1.2 shows a simplified view of the muscles that plantar flex and dorsiflex the foot, and how contraction of the plantar flexors and dorsiflexors cause downward and upward rotation of the foot respectively. The muscles are antagonistic pairs: one muscle opposes the other (such combinations exist throughout the body).
1.3 Gait

A person’s style of walking - gait - is very unique, yet at the same time many common components are shared with that of every other walking person. Gait may be divided into two phases: the swing phase and the stance phase. Each leg moves between the stance phase and the swing phase during walking (in healthy individuals, each leg performs very similar motions but at different times). Throughout the stance phase the leg is used to support the body while the contralateral leg swings forwards. During the swing phase the leg swings forward in preparation to advance the body while the contralateral leg supports the body.

While it is true that during the time that one leg is in the swing phase the other will be in the stance phase, the opposite is not true. When one leg is in the stance phase, there is a short period of time in which the other leg is also in the stance phase (double stance period). This is to allow the role of supporting the body to pass from one leg to the other. Once support of the body has been transferred, one leg may enter swing phase. Figure 1.3 shows how each phase of gait overlaps.
During the stance phase, there are four events characterised by heel and toe contact with the ground: heel contact (HC), foot flat (FF), heel rise (HR) and toe off (TO). The stance phase begins when the heel first makes contact with the ground at HC, and ends when the toe leaves the ground at TO. Throughout the majority of the stance phase both the heel and the toe are in contact with ground (between FF and HR). Figure 1.4 shows the complete stride of one leg.

During the swing phase, the hip and knee flex, and the foot is dorsiflexed. These joint movements effectively reduce the length of the leg allowing it to be moved forwards. If the effective leg length was not shortened, the foot would drag along the ground and possibly lead to tripping. Dorsiflexion also continues into early stance phase to prevent the foot from slapping against the ground immediately after heel contact (footslap).

One role of the plantar flexors is to lift the heel just before the end of the stance phase and push away from the ground to further advance the contralateral swinging limb. This is a relatively strong, fast movement, and would be produced mainly by the gastrocnemius muscle (a concentric contraction). The plantar flexors are also applied for a between mid and end stance to control the advancing rotation of the limb about the stationary foot. This activity takes place over a longer period, but does not require such intense activity - the soleus muscle is mainly responsible for this movement (an eccentric contraction).
1.4 Gait and Hemiplegia

Hemiplegia may be caused by a stroke, MS or an incomplete SCI (inflammations, tumours and injury of the head are also causes). Although some function can return after a period of time, permanent disabilities usually remain, including reduced hand function and difficulty flexing the hip and knee joint during the swing phase. Hemiplegic gait is usually characterised by poor weight bearing (stance phase), and ineffective hip and knee flexion and a lack of ankle dorsiflexion (swing phase). This is caused by a combination of muscle weakness and spasticity, and leaves the sufferer with a reduced ability to support the body and to lift the foot from the ground.

Without being able to adequately flex the hip and knee and dorsiflex the ankle, the effective leg length cannot be reduced during the swing phase. This results in either the foot being dragged along the ground or a compensatory gait style being employed, such as circumduction (the leg is swung outwards away from the body to clear the ground). This condition is known as drop-foot.

It is common for spasticity to be present in the plantar flexor muscles (the muscle is constantly in a contracted state) of hemiplegics. This contraction pulls the foot into a
continuous state of plantar flexion, which a weak dorsiflexing muscle is not able to correct. In some cases there is a complete lack of activity in any of the ankle muscles; in this case the foot will be flaccid.

An ankle-foot orthosis (AFO) may be used to hold the foot in a neutral position (not in dorsiflexion or plantar flexion). The AFO is constructed from a thermoplastic material, such as polypropylene, and is custom fabricated and fitted to the patient (Chu and Reddy, 1995). The orthosis extends from below the knee to the end of the foot. Many users of the AFO find that it is uncomfortable, and this can affect the patient’s ability to walk and lead to reluctance to continue using the device. An AFO may be used to hold the foot at the required angle in the case of flaccid (very weak) or high tone (in permanent contraction) muscles. Since ankle joint movement is required to reduce muscle tone, use of the AFO (which prevents any ankle movement) can impede improvement.

Increased joint movement through muscle tone relief can also be achieved using botox injections to produce short-term (6 months) paralysis of the muscle. Once the injection has been administered, the patient would require physiotherapy for the duration of the botox effect; walking would be possible only whilst using an AFO. Physiotherapy is a technique that can be used to restore joint movement (this may also provide muscle strengthening). However, the patient usually only benefits from this whilst actually working with the physiotherapist, making progress slow. It is possible for patients to receive inadequate physiotherapy if there is a lack of skilled personnel and if the hospital does not wish to meet all of the costs involved.

1.5 Functional Electrical Stimulation to Correct Hemiplegic Gait

The application of an electric current to produce muscle contraction resulting in functional limb movement is called functional electrical stimulation (FES). Current is usually applied to the innervating nerve using electrodes. One use of FES is as an electrical orthosis to correct hemiplegic gait. By applying current (indirectly) to the nerves innervating the muscles which flex the hip and knee joints and dorsiflex the
ankle joint, the foot will be lifted out of contact with the ground during the swing phase. In some cases, the current is used only to contract the tibialis anterior and dorsiflex the foot, which is often sufficient to correct gait. Whichever muscles are stimulated, a sensor is required to ensure that the stimulation is only applied during the stance phase (although this may actually begin at heel rise and terminate after heel contact). The most common choice of sensor is the footswitch, which is placed under the heel and switches electrical stimulation on whenever the heel is not in contact with the ground. A number of limitations of this sensor led researchers to consider the use of other sensors (see section 2.3.2).

The aim of this project was to investigate the replacement of the footswitch with a novel gyroscope-based sensor in FES drop-foot correction. A comparison of the two sensors' performance was made, along with the measurements from a third system. In chapter 2 literature regarding FES, hemiplegic gait and sensors for drop-foot correction FES systems is discussed. Chapter 3 details a pilot study in which the use of three gyroscope sensors to measure lower limb joint angles (as an alternative gait analysis system) is investigated. The use of the gyroscope sensor and a desktop PC to detect gait events (such as heel contact and heel rise) in normal and pathological gait is discussed in chapter 4. In chapter 5 the development of a portable system to perform the same role as the desktop PC is detailed, and this system is evaluated in chapter 6. Conclusions and recommended further work are presented in chapters 7 and 8.

1.6 Research Hypothesis

The hypothesis for this study is divided into two sections:

1. The gyroscope sensor (combined with a supporting microcontroller) can be connected to a stimulator intended for use with footswitches. The sensor can also be donned and doffed with similar ease to that of footswitches.

2. The gyroscope sensor offers similar or improved stimulation timing and reliability compared to footswitches when used with an FES drop-foot correction system.
Chapter 2
Literature Review

2.1 Introduction
The purpose of this chapter is to provide background literature for the project. Before discussing functional electrical stimulation (FES), the physiology of skeletal muscle contraction is described. The operation of FES and its applications are reviewed next, including a more detailed review of drop-foot correction systems. Following this, existing sensors for drop-foot correction systems are appraised, and finally the operation of the gyroscope sensor and its use in such systems is discussed. At the end of the chapter the hypothesis and objectives are outlined.

2.2 Functional Electrical Stimulation
Electrical stimulation is the application of an electric current to human nervous tissue, in order to produce a muscle contraction (although in some rare cases the current is applied directly to the muscles themselves). If the muscle contraction(s) is coordinated in such a way that a functional limb movement is produced (e.g. standing up) then the use of stimulation becomes functional (functional electrical stimulation –FES). FES has a wide range of applications including diaphragm, bowel and bladder movement, cardiac pacemakers and limb movement for paraplegics and hemiplegics. Stimulation of the lower limbs may allow paraplegics to stand (from a seated position and return again), and walk (for a limited distance only) using some form of external assistance. In hemiplegia, stimulation can be used to increase hand function, as well as provide improved gait.

2.2.1 FES and Muscle Contraction
There are three types of muscle within the human body: cardiac muscle, smooth muscle and skeletal muscle. Cardiac muscle is found only within the heart, whereas smooth muscle is found in almost every organ, regulating either blood flow or movement within
the digestive and urinary systems. Cardiac muscle and smooth muscle are controlled automatically; skeletal muscle control may be either voluntary or involuntary.

Skeletal muscle is required to perform the following functions: produce skeletal movement, maintain posture and body position, support soft tissues (protecting and bearing the weight of visceral organs), guard entrances and exits (the openings of the digestive and urinary tracts) and maintain body temperature (some of the energy required for muscle contraction is converted into heat). When used with skeletal muscle, FES is concerned with the generation of skeletal movement.

Before discussing the use of FES to produce muscle contractions, the physiology of the nervous system and skeletal muscle is presented in the following text.

### 2.2.1.1 Physiology of Voluntary Skeletal Muscle Contraction

The sequence of events leading to voluntary skeletal muscle contraction begins either as an external stimulus causing an involuntary reflex action (e.g. the patellar reflex), or in the brain as a voluntary motor command (e.g. standing up). Motor control is initiated in the nervous system, which is divided into the central nervous system (CNS) and the peripheral nervous system (PNS). In the case of the external stimulus, an involuntary motor command originates in the spinal cord region of the CNS without involving the brain. In both cases the PNS is required to transmit the command from the CNS.

The word *somatic* is used when referring to the parts of the body responsible for sensory input; its meaning is “of, relating to or affecting the body”. In reference to the parts of the body dealing with automatic (i.e. involuntary) motor control, the word *autonomic* is used, meaning: “acting or occurring involuntarily”. Figure 2.1 shows how motor commands are passed from the CNS to skeletal muscle, via the two functional divisions of the nervous system: the efferent and the afferent divisions. The efferent division is responsible for the transmission of motor commands from the CNS, and the afferent division transmits sensory information to the CNS. The diagram also shows how muscle contraction is regulated using a feedback loop providing sensory information which is passed to the CNS.
The afferent division receives information from the somatic sensory receptors. These receptors exist throughout the body and are classified according to the stimulus that excites them:

- Nociceptors - respond to a variety of stimuli usually associated with skin damage, and cause the sensation of pain.
- Thermoreceptors - respond to changes in temperature.
- Mechanoreceptors - are stimulated or inhibited by physical distortion, contact or pressure on their cell membranes.
- Chemoreceptors - monitor the chemical composition of body fluids and respond to the presence of specific molecules.

Figure 2.1 Skeletal Muscle Contraction within the Nervous System (Martini FH, 1995)
The efferent division consists of the somatic nervous system (SNS) and the autonomic nervous system (ANS). The ANS provides automatic control of smooth muscle and cardiac muscle, and the SNS controls the contraction of skeletal muscles. The remainder of this section will describe the function of the SNS.

In figure 2.2 a diagrammatic view of CNS tissue is shown. The top half of the picture represents grey matter, which is coloured as such because of the large number of neuron cell bodies present. On the lower half of the picture is white matter, containing the axons that transmit information to the muscles. Many of these axons are myelinated, and it is the myelin that causes white matter to be white. The myelin, a sheath primarily consisting of lipids, increases the speed at which the nerve signals (action potentials) are propagated along the axon. Figure 2.4 shows a nerve fibre with its myelin sheaths and nodes of Ranvier (the gaps between the sheath). Grey and white matter are clearly visible in the transverse view of the spinal cord shown in figure 2.3.
In figure 2.3 a peripheral nerve is shown extending from the spinal cord. There are many nerves of this type in the body connecting the spinal cord to skeletal muscles. Each peripheral nerve contains (along with blood vessels) a number of fasicles, which are a collection of either efferent (motor fibres conveying signals from the spinal cord to muscles) or afferent (sensory fibres conveying signals from sensory receptors to the spinal cord). In figure 2.4 the transverse view of a nerve fibre is shown, and figure 2.5 shows how peripheral nerves are organised.
Information is passed from the CNS to skeletal muscles via the peripheral nerves in the form of action potentials. An action potential is a chemical change that begins at the cell body of a neuron, and is conducted along the length of the axon until it reaches the neuromuscular junction between the axon and a muscle fibre. Action potentials propagate along the nerve fibre by setting up current loops between each node of Ranvier, as shown in figure 2.6. Before activation, the section of nerve tissue is said to be in a resting state and the potential across the cell membrane (the edge of the cell) is about -70mV. At the first activation (brought about by a previous action potential), current leaving the nerve fibre causes a depolarisation (membrane potential is -60mV) which in turn initiates a new current loop (membrane potential rises to +30mV). This new loop causes the next activation (and the membrane potential falls to -90mV before returning to -70mV), where current again leaves the nerve starting the next current loop and so on. Each node repolarises after it has depolarised (taking around 3ms). Figure 2.7 shows an action potential.
The neurons responsible for transmitting motor commands to skeletal muscle are called motor neurons. These neurons originate in grey matter, and at the end of the neuron is a neuromuscular junction through which it is able to communicate with muscles. Figure 2.8 shows the relationship between the nerve axon and the muscle it controls.
A narrow space called the synaptic cleft separates the synaptic knob from the muscle fibre. When the action potential reaches the synaptic knob, a chemical neurotransmitter is released which causes changes to occur in the muscle's receiving surface. This surface is called a motor end plate, and when the neurotransmitter is received at this surface an action potential is formed producing muscle contraction.

In figure 2.8 the section of muscle shown is a muscle fibre. Figure 2.9 shows how a muscle fibre relates to the skeletal muscle that it resides within. Typically, each skeletal muscle will contain thousands of fibres controlled by a much smaller number of motor neurons. Although most motor neurons control thousands of muscle fibres (e.g. in leg muscles), there are some that control just one or two fibres (e.g. in eye muscles). The fewer the number of fibres controlled by each neuron (or the smaller the size of the motor unit), the finer the control over the muscle will be. Figure 2.10 shows motor units within a skeletal muscle.

Muscle contraction is controlled by the action potentials propagated along the length of neurons. Upon reaching the muscle fibre, each action potential produces a short sequence of events that result in a twitch. The twitch may be divided into three phases: the latent period (no contraction occurs), the contraction phase (fibre tension rises to a
peak) and the relaxation phase (tension returns to resting levels). The duration of the entire twitch – stimulation, contraction and relaxation – can vary from 10 to 100ms depending upon the muscle type.

![Figure 2.9 Organisation of Skeletal Muscle (Martini FH, 1995)](image)

In figure 2.10 three motor neurons are shown extending from the spinal cord to correspondingly coloured muscle fibres. All of the muscle fibres controlled by a single motor neuron are known collectively as a motor unit. Muscle fibres representing one motor unit are distributed evenly throughout each muscle, allowing constant tension to be produced despite the fact that some motor units within the muscle are not active. When the motor units that were first activated become fatigued, units that were inactive are able to become active, maintaining the required muscle tension and allowing other units to rest. If prolonged muscle tension is required, motor units are able to cycle...
between contraction and relaxation in order to delay the onset of fatigue (although at maximum contraction strength this is no longer possible).

There are three types of skeletal muscle fibre: Type I, Type II A and Type IIB, with contraction times that are slow, fast and very fast respectively. Type I muscles are coloured red and have a high resistance to fatigue; Type II A muscles are also red and have an intermediate level of resistance to fatigue; Type II B muscles are white in colour and have a low resistance to fatigue. Most skeletal muscles contain fast fibres, which are relatively large in diameter and are able to contract in around 10ms or less following stimulation. These fibres produce the most powerful contractions but fatigue the most quickly. Slow fibres are about half the diameter of fast fibres and take around three times as long to contract after stimulation. However, these fibres have a superior oxygen supply enabling them to fatigue less quickly than fast fibres. Intermediate fibres have properties that lie between those of fast and slow fibres.

**Figure 2.10** Innervation of Motor Units within Skeletal Muscle (Martini FH, 1995)
Since motor neurons are only capable switching fibres on and off, each muscle fibre within a skeletal muscle produces a similar amount of tension each time it is stimulated (depending on level of fatigue and type of muscle). Muscle tension is controlled by the number of fibres stimulated, and also by the frequency at which each fibre is stimulated. The smooth increase in muscular tension produced by increasing the number of active motor units is called spatial recruitment; an increase produced by raising the frequency of fibre stimulation is known as temporal recruitment. If a high tension is required, recruitment will be greatest (or fastest) but this will lead to fatigue in a shorter time.

Muscle fibres do not reach their maximum tension immediately after one twitch. Since the duration of a twitch is greater than that of an action potential, the contractions produced by successive action potentials can summate. Therefore, if a second action potential occurs before the relaxation phase of the first has ended, the second contraction will be added to the force exerted when the second action potential arrived (instead of beginning at zero). After successive action potentials, the contraction force exerted by the muscle fibre will reach its maximum. When the relaxation phase is completely eliminated (i.e. the next action potential arrives exactly at the end of the preceding one) a state known as complete tetanus is achieved. The effect of both incomplete and complete tetanus is shown in figures 2.11a and 2.11b respectively (each arrow represents the arrival of an action potential).

The time during which a nerve cannot be stimulated to produce another action potential is termed its absolute refractory period (ARP). A nerve fibre cannot fire a second action potential until it is just outside its ARP, therefore the maximum frequency of its firing is a little less than the reciprocal of this period. For example, if the ARP is measured as 1ms, the fibre can fire at just under 1000 times per second. A relative refractory period occurs immediately after this, during which the nerve will respond to a stronger-than-usual stimulus.
2.2.1.2 Neuromuscular Electrical Stimulation

An electric current applied to nervous tissue is capable of producing action potentials causing skeletal muscle contraction. The current is usually transmitted from the generating device (the stimulator) via thin electrical wires and electrodes, all of which may be implanted within the body. If the stimulator system is to be used externally, then stimulation may be either transcutaneous (using skin surface electrodes) or percutaneous (using electrodes that pass through the skin). Transcutaneous stimulation is in most cases the preferred choice, as this does not involve penetrating the skin and may be
more easily administered by the user. Also, implantable systems require surgery (at much greater cost) and may lead to complications arising from infection (although this risk is becoming smaller). Transcutaneous stimulation, however, requires higher current levels as the electrodes are not in direct contact with the nervous tissue.

The level of influence that electrical stimulation has over muscles is greatly dependent upon the position of the electrodes. If implantation is chosen, the optimal site need be located only once since the electrode position will not change. If skin surface electrodes (and needle electrodes) are used, then the optimal site must be located each time the electrodes are removed (usually once a day).

A motor point is defined as the most electrically excitable area of the muscle, and represents the greatest concentration of nerve endings. Motor points are located on the skin over the muscle and are approximately the area in which the nerve enters the muscle belly. This area is known as the neuromuscular junction or zone of innervation. For effective stimulation, electrodes must be placed at or near either the motor point of the muscle or on the supplying nerve. When using skin surface electrodes to stimulate the motor point, only more superficial muscles may be considered (higher electrical currents may be used to stimulate deeper muscles, but this will cause most other muscles in the vicinity to contract and may be intolerably painful for the user). Correct location of the motor point will produce a stronger muscle contraction, therefore the level of current required for a given movement is reduced. As increased current levels lead to greater sensation for the user, accurate location of the motor point is important. When a nerve is stimulated or depolarised, all of the muscles connected to it will contract. Stimulation of pain fibres within the nerve may also cause a reflex action to be triggered. Accurate location is necessary to maximise the response and minimise the level of current required and therefore the pain experienced by the user. Figure 2.12 shows the simplified flow of current between two electrodes attached to the skin.
In figure 2.12 electrical current can be seen flowing from the indifferent (positive) electrode through body tissues to the active (negative) electrode. The nerve fibre is shown surrounded by sections of myelin sheaths. Figure 2.12 shows a number of general pathways (numbered 1, 2 and 3) that the current can take to reach the active electrode:

1. Through the extracellular fluid only
2. Through the extracellular fluid and along the nerve fibre
3. Through the extracellular fluid, crossing the nerve fibre only momentarily

Figure 2.12 shows that not all of the current flowing between the two electrodes reach the nerve fibre, i.e. there is some loss. Therefore, only current flowing along paths 2 and 3 will contribute towards action potential generation, as shown in figure 2.13. At point ‘B’, the negative field around the active electrode will attract positive ions, reducing the positive charge on the outside of the nerve cell. The decrease of positive ions outside the
bundle allows the negatively charged ions to fall away from the cell wall (or membrane) deeper into the cell. Thus the potential difference across the cell membrane is reduced (depolarisation) and it becomes closer to the threshold of action potential generation. At point ‘A’, the positive field produces the opposite effect, and the potential difference between the inside and outside of the cell is increased (hyperpolarisation). If the current intensity is high enough this depolarization/hyperpolarisation state of the cell leads to the generation of an action potential.

![Diagram of Current Flow Between Two Electrodes Attached to the Skin]

Figure 2.13 Diagram of Current Flow Between Two Electrodes Attached to the Skin

Neuromuscular electrical stimulation differs from normal muscle contraction in the method used to recruit fibres. Electrical stimulation recruits motorneurons in the reverse order to that in normal physiological contraction. In physiological muscle contractions, the small motorneurons innervating slow-fatiguing muscle fibres are usually the first to contract. This will then be followed by the larger, more powerful collections of fibres which fatigue more quickly. However, in the case of electrical stimulation these fast-
fatigue and more powerful fibres are recruited first (at lower levels of stimulus intensity). The slower-fatiguing fibres are only recruited once the stimulus intensity is increased (Baker et al., 1993). Also, electrical stimulation does not allow recruited fibres to rest and this also leads to the occurrence of faster fatigue.

Electrical stimulation of nerves can produce two components: nerve stimulation of the muscles that are innervated by that nerve, and an automatic reflex action produced by stimulation of the sensory fibres within the nerve. Reflexes range from the simple stretch reflex of skeletal muscle (this provides a method of automatically re-adjusting muscle length when an external stimulus increases muscle length) to the flexion withdrawal reflex - a coordinated action involving several muscles. One example of a flexion withdrawal reflex is the result of stepping on a pin. Pain receptors in the foot generate a signal that is relayed to the spinal cord via pain fibres. In the spinal cord motor neurons are activated to produce a contraction of specific muscles in the limb. Hip and knee flexion, and ankle dorsiflexion result in the foot being lifted from the ground. Electrical stimulation of the peroneal nerve also causes pain fibres to send a signal to the spinal cord, which in turn produces the same reflex response. The degree of response is, however, subject to the particular physiology of the patient and can vary considerably.

Although electrical stimulation may be achieved using a simple sinusoid or rectangular-shaped wave, a more complex waveform is usually favoured. Figure 2.14 shows some example waveforms for electrical stimulation.
The sinusoid waveform (a) is the most simple of the four stimulation waveforms shown in figure 2.14, but the rectangular-shaped waveform (b) is preferred since the occurrence of nerve accommodation is reduced. Accommodation occurs when the stimulation amplitude’s rate of change is too low, causing the threshold of muscle response to increase. Another effect of stimulation is habituation, and this occurs when a muscle becomes less responsive to the applied electrical stimulation (within one session). Habituation is caused by a tendency of some neurons to require either a stronger nerve signal or a longer recharge period before it can fire again, if it has been triggered recently. If habituation occurs, the required stimulation intensity is greater leading to increased stimulation of sensory fibres. A person using stimulation may, however, become able to tolerate higher levels of sensation after several sessions simply because they have become used to it.
The monophasic waveform (b) is far from ideal, since charge may travel in one direction only. The result of this is a net transfer of ions into the body, which can cause tissue damage and electrode deterioration (Baker et al., 1993). The symmetrical biphasic waveform (c) allows the charge to travel in both directions, thus avoiding a net transfer of ions. Waveform (d) is also charge balanced, but is assymmetrical (little or no stimulation occurs during the negative phase of this waveform). Despite being charge balanced, even stimulation devices using assymmetrical waveforms have been reported to cause minor skin irritation (Burridge et al., 1997). Waveforms (c) and (d) are both preferred, with (d) producing the least chemical reactions between electrode and body and (c) being slightly more specific in terms of the muscle(s) recruited (Baker et al., 1993).

Parameters that determine the muscle contraction force are the frequency, current amplitude and the pulse width of stimulation. The frequency determines the summation of twitches; increasing the frequency results in a greater frequency of twitches and therefore a stronger muscle contraction. The current amplitude and pulse width determine the spatial recruitment of nerves - greater numbers of nerves are recruited when the value of these parameters is increased. Typical values for neuromuscular electrical stimulation are shown in table 2.1.

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Typical Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Current Intensity</td>
<td>20 - 50mA</td>
</tr>
<tr>
<td>Frequency</td>
<td>10 - 40Hz</td>
</tr>
<tr>
<td>Pulse Width</td>
<td>100 - 300µs</td>
</tr>
</tbody>
</table>

Table 2.1 Typical Values for Neuromuscular Electrical Stimulation (Baker et al., 1993)
2.2.2 Applications of FES

The most common patient groups that are able to benefit from FES are:

- Spinal Cord Injury (SCI)
- Multiple Sclerosis (MS)
- Cerebral Palsy (CP)
- Traumatic Brain Injury (TBI)
- Cerebrovascular Accident (CVA)

SCI and TBI patients may suffer from complete loss of movement in the lower limbs (and in some cases the upper limbs also). FES has been used to enable such people to stand, sit and walk with the aid of a mechanical structure, such as a walking frame or crutches (Bajd et al., 1981; Ewins et al., 1988; Philips, 1989; Kayaga et al., 1995) However, this application is not as developed as drop-foot correction. The need for more complex stimulation systems (multi-channel instead of single-channel and closed-loop instead of open-loop) has prevented this application from becoming more widespread.

The cognitive state of each patient should be assessed when considering their suitability for FES. Many SCI and MS (and some stroke) patients do not suffer from the impaired cognition that can affect stroke and TBI patients. CP patients can often be young children, and for similar reasons this group may not be appropriate. The ability that the patient has to use the equipment must also be considered.

In addition to directly stimulated limb movements, electrical stimulation is used as a therapeutic aid. One example is the stimulation of denervated muscle (Woodcock et al., 1998). Although FES is used with patients whose peripheral nervous system is intact, it is possible to produce muscle contractions in denervated muscle. A muscle that has had its nerve supply severed is said to be denervated. In this case, the muscle is stimulated directly and the stimulation current must be of greater intensity and duration. Pulse widths of 10 to 30ms are often required at a maximum frequency of 20Hz. Stimulation
of denervated muscle may be therapeutic (to avoid muscle atrophy and fibre degeneration, to increase blood flow and tissue perfusion and to simply improve the appearance of the limb) or functional (Woodcock, 1999). Another example is the non-assisted improvement seen in patients who otherwise use electrical stimulation. Some studies have found that hemiplegic patients using FES continue to show improvement even after the stimulator system has been removed – known as a carry-over effect (Bogataj et al., 1993; Daly et al., 1996; Taylor et al., 1998). Electrical stimulation may also be used therapeutically to improve the condition of muscles (the strength of contractions and the endurance), and to increase the range of motion of a joint (Bremner et al., 1992; Smith, 1990).

After experiencing a CVA (stroke), patients who have been able to recover some mobility may benefit from FES. Stimulation to correct drop-foot gait, a condition that affects many hemiplegic patients, is by far the most common use of FES. Indeed, the treatment has become routine practice in Salisbury District Hospital, U.K (Burridge et al., 1995; Swain et al., 1997; Taylor et al., 1998). Drop-foot can occur during the early stages of MS, and electrical stimulation may also be used to improve gait in this case. CP patients who walk with raised heels may also benefit from FES. Correction of drop-foot using FES (the subject of this study) is discussed in more detail in section 2.3.2.

2.3 FES and Hemiplegia

A stroke is a sudden disturbance of brain function caused by a vascular disorder causing disability (or death) within 24 hours (Rose and Capideo, 1981). First incidence of stroke is approximately 100,000 per year, of which over 80% survive (Taylor et al., 1998). Approximately 75% of the survivors experience a reduced quality of life and it is estimated that about 12,000 suffer from a dropped-foot condition. In addition to this figure are patients from other groups such as MS (approximately 2500 per year, MS Society, 2003).

The resulting disability is usually hemiplegia, although it is possible (but less common) that a stroke will affect both sides of the body. The word hemiplegia means complete or
Literature Review

partial paralysis of one half of the body, either with or without sensory loss (Hemianasthesia). The disability is most commonly the result of a stroke, but tumour, craniocerebral injury or inflammation of the spinal cord are other possible causes. The paralysis, which may include the face, can lead to reduced hand function, poor mobility, a loss of balance and control of limbs, and a slumped posture (Thompson and Morgan, 1990).

2.3.1 Hemiplegia

In a comparison with normal gait, the most apparent feature of hemiplegic gait is that it is slow and asymmetric (Roth et al., 1997). The hemiplegic patient exhibits a “preference for bearing weight on the non-hemiplegic limb because of weakness in the hemiplegic limb, resulting in significant asymmetry. Stance phase on the hemiplegic limb is short and abrupt. It does not prepare the body for forward progression”.

A shorter stance phase duration, decreased weight bearing and increased swing time for the paretic limb is reported by Craik and Oatis (1995 - conversely, the unaffected limb has increased stance time and decreased step length). The division of hemiplegic gait into three major abnormalities was also discussed: abnormal base of support (the foot), abnormal limb stability, and lack of limb clearance. The foot is often in the equinovarus position (inversion and heel raised), with toes in flexion. Limb instability is said to be the result of knee flexion occurring during early stance phase, along with a weakness in the quadriceps (it was reported that this could also be due to the over-lengthening of the Achilles tendon). Finally, lack of limb clearance during the swing phase is said to be caused by inadequate hip flexion, knee flexion and ankle dorsiflexion.

2.3.1.1 Spasticity

Along with muscle weakness in the affected limb, hemiplegia can cause spasticity to develop. In its simplest form spasticity is an increase in tone (a contracture), where tone is defined as “the resistance that is felt when a joint is passively moved by an outside examiner” (Wade et al., 1985). Martini (1995) defined spasticity as “a condition characterised by hesitant, jerky voluntary movements, increased muscle tone, and
hyperactive stretch reflexes”. Premature triggering of the stretch reflex produces the contracture, leading to joint angular stiffness. An analysis of the EMG produced shows that trigger is either clonic (5-8 Hz signal causing muscle spasm) or sustained, with the response being dependant upon the rate of stretch (Rose and Gamble, 1994).

When a hemiplegic patient with spasticity attempts to walk, there is usually a lack of ability to flex joints in the co-ordinated manor required. Savinelli et al. (1977) stated that movement using specific muscles rely on rapid relaxation of their antagonists so they can move quickly and with ease rather than depend on force. Spasticity of the antagonistic muscles limit the range of movement of a joint, especially when combined with a weakness in the agonist.

In order to avoid the development of contractures, Savinelli et al. (1978) recommended passively moving affected joints, and using splints and orthotic equipment to maintain proper alignment. Any rapid movement which could trigger the stretch reflex should be avoided. The use of electrical stimulation to stimulate the antagonist group was also suggested to reduce spasticity (by stretching the agonist muscle) and strengthen muscles.

2.3.1.2 Synergies

Synergies, as defined by Rose and Gamble (1994), are mass flexor or extensor patterns which are used when selective motor control is inadequate. During the stance phase of hemiplegic gait an extensor synergy is present – the hip and knee extensors and the ankle plantar flexors are activated simultaneously. If there is a large amount of muscle strength then stance phase progression is obstructed; if muscles are weak then stability is poor during this phase.

2.3.1.3 Motor Recovery After Stroke

According to Rose and Gamble (1994), muscle weakness (due to a lack of control) that is the primary limiting factor in walking; spasticity, contractures and synergistic patterns have not yet developed. A dorsiflexion weakness causes a drop-foot deviation during
the swing phase, and trunk and pelvic rotation is required to circumduct the limb as a result of weak hip and knee flexors (Rose and Gamble, 1994).

Motor recovery was divided into six stages by Brandstater et al. (1983), as shown in table 2.2. Initially no voluntary movement is possible, but once synergistic patterns develop there is improvement until the patient is able to perform isolated movements.

After the early recovery period, hemiplegic patients usually suffer from equinus (excessive plantar flexion) and a stiff knee (Knutson and Richards, 1979; Rose and Gamble, 1994; Craik and Oastis, 1995). More energy is required to swing the paretic limb, and despite ipsilateral hip and trunk movements and contralateral limb compensatory motions, toe clearance is not always achieved. Inappropriate toe flexion or extension can also leads to an abnormal base of support, which can interfere with weight bearing and cause significant pain.

The level of disability remaining after initial recovery from stroke may be divided into four (Winters et al., 1987; Whittle, 1991). Table 2.3 shows the four levels of severity, which range from drop-foot only to drop-foot combined with a contracture of the plantar flexors (i.e. presence of muscle tone in the gastrocnemius and/or soleus muscles), a stiff knee and reduced hip motion.
STAGE | LEVEL OF RECOVERY
--- | ---
1 | No voluntary movement of affected limb detectable
2 | Weak basic limb synergies initiated and spasticity is developing. Patients unable to walk alone until Stage 3 is reached
3 | Voluntary initiation of limb synergies of sufficient degree to show substantial joint movements. Spasticity is often marked.
4 | Movements outside limb synergies present and spasticity may be less
5 | Selective control of movements outside synergic patterns with further reduction of spasticity
6 | Patient can perform isolated joint movements freely and in a well coordinated manner

Table 2.2 Six Stages of Motor Recovery (Brandstater et al., 1983)

<table>
<thead>
<tr>
<th>Level</th>
<th>Severity</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>Drop-foot only</td>
</tr>
<tr>
<td>2</td>
<td>As above and contracture of plantar flexors</td>
</tr>
<tr>
<td>3</td>
<td>As above and stiff knee</td>
</tr>
<tr>
<td>4</td>
<td>As above and reduced hip motion</td>
</tr>
</tbody>
</table>

Table 2.3 Levels of Severity of Hemiplegics after the Recovery Period (Winters et al., 1987; Whittle, 1991)

2.3.1.4 Rehabilitation

Restoration of upper limb and lower limb function are both important following a stroke. However, Rose and Capideo (1981) stated that “Following an acute stroke, walking is the single most important activity and arm recovery should be considered a bonus at this stage”. While one might argue that arm recovery is in fact more important than walking, the latter function is very important.
It is the role of the physiotherapist to help rehabilitate the stroke patient, and a large part of the re-education process occurs during the time that the two people are together. There are of course other activities that should be practised before walking is attempted, including wheelchair-to-bed and wheelchair-to-toilet transfers. It is also very important to minimise the progression of spasticity by carefully positioning the patient at all times. “Spasticity develops insidiously a few days after the initial complete paralysis of the muscles of the affected side of the body” (Turnbull and Bell, 1985). Since walking can only take place after standing-up, wheelchair-to-standing and bed-to-standing transfers must also be practised first.

Recovery was also divided into stages by Bobath (1990). The three stages are:

- Initial flaccid stage
- Stage of spasticity
- Stage of relative recovery

The first stage occurs soon after the stroke takes place, and can remain for several weeks. The patient cannot move the affected side and is often completely unaware of the limbs. At this stage the patient is encouraged to perform routine non-functional movement of the affected limbs, such as flexion of the leg at all joints. Spasticity begins to develop during the first stage and usually increases as the patient’s level of activity improves throughout the first 18 months. The suggested treatment during the second stage is to begin to perform sitting, standing and walking.

Patients who reach the third stage of relative recovery will be those who were not severely affected at the beginning and who have made a good and spontaneous recovery, or who have performed well in their treatment. These patients are often able to walk unaided (even without a stick) and can use the affected arm for support and to hold an object in the hand. Spasticity will be greatly reduced in this stage, although problems may remain involving more complex tasks such as those requiring independent finger movement. Also, gait may still be asymmetrical and less efficient because of a lack of adequate dorsiflexion, and possibly lack of adequate knee and hip flexion.
A novel system to aid the rehabilitation of stroke patients was created by Colborne et al. (1990). A system that produced real-time visual and audible feedback to provide information about hemiplegic gait was designed and evaluated. Knee and ankle joint angle (measured using a goniometer) and EMG feedback (from the Vastus Lateralis muscle) were used to display target levels on a computer screen. A red light indicated when (during gait) a particular activity should take place, and an audible note was produced when the subject was successful.

Thirteen able-bodied subjects took part in the test, which involved achieving targets such as dorsiflexion at 10° above the subject's normal angle (measured over four sessions of five walks each containing five strides). Another target was the attainment of normal peak dorsiflexion at a time 10% earlier or later in the stride time than normal. To achieve the EMG target, subjects were required to produce twice the average level of normal peak activity, between gait events such as heel contact and foot flat. The difference between the targets set and what the subjects achieved was identified as an error. It was found that the maximum knee and ankle joint errors were approximately 7°, and the time error was approximately 2%. The maximum EMG time error was also found to be approximately 2% (values were comparable with joint angle and time error), but the activity target error was much greater at approximately 20%. The system was not used with hemiplegic subjects, but success of normal subjects in modifying the amplitude and timing of movement using feedback suggests that the system might be useful with pathological gait.

A comparison between conventional therapy and FES in stroke rehabilitation was conducted by Bogataj et al. (1993). Twenty hemiplegics were randomly placed into two groups, each containing an equal distribution of males and females. The groups had an equal average time since the onset of the stroke. One group received conventional therapy, including physical exercises, Bobath training (see previous text in this section and Bobath, 1990), massage, thermal therapy, biofeedback therapy and gait training. The other group received multichannel FES therapy, with stimulation applied to the peroneal nerve (for ankle dorsiflexion), to the soleus muscle (for plantar flexion), to the quadriceps (for knee extension), to the hamstring muscles (for knee flexion), to the
gluteus maximus muscle (for hip extension), and to the triceps brachii muscle (for reciprocal arm swing during the swing phase of the ipsilateral leg).

The study revealed that the FES therapy was able to produce greater improvement (in terms of stride time, length and velocity) than the conventional therapy in early stages of gait in hemiplegic patients. It was suggested that multichannel FES treatment be used as an integral part of the rehabilitation programme.

2.3.2 Electrical Stimulation to Correct Hemiplegic Gait

Electrical stimulation of the peroneal nerve can produce both nerve stimulation of the innervated muscles and a flexion withdrawal reflex. The muscles which are innervated by the peroneal nerve, along with their actions, are shown in table 2.4.

Typical electrode placement for peroneal nerve stimulation is shown in figure 2.15. The active electrode is placed over the peroneal nerve, just below the head of the fibula. The indifferent (or inactive) electrode is placed about 50mm below and slightly forward of the active electrode near to the motor point of the tibialis anterior.

<table>
<thead>
<tr>
<th>Muscle</th>
<th>Action</th>
</tr>
</thead>
<tbody>
<tr>
<td>Biceps Femoris of the Hamstrings</td>
<td>Flexes Knee, Extends and Adducts Thigh</td>
</tr>
<tr>
<td>Fibularis Peroneus (Brevis and Longus)</td>
<td>Everts Foot</td>
</tr>
<tr>
<td>Tibialis Anterior</td>
<td>Dorsiflexes Foot</td>
</tr>
<tr>
<td>Extensor (Digitorum and Hallucis) Longus</td>
<td>Extends Toes</td>
</tr>
<tr>
<td>Fibularis Tertius</td>
<td>Dorsiflexes Foot</td>
</tr>
</tbody>
</table>

Table 2.4 Muscles Innervated by the peroneal nerve
It is possible to adjust the response of stimulation by repositioning either or both of the stimulation electrodes. Eversion and inversion can in some cases be reduced, and increased knee flexion may also be possible. It is, however, possible that peroneal nerve stimulation does not produce a reflex response at all due to variations in the physiology of the sensory nervous system. If this is the case, the electrodes can be moved closer to the tibialis anterior muscle’s motor point to improve the response of this muscle. It is of course also possible to electrically stimulate the motor points of the hip and knee flexors directly, but this would require a greater number of electrodes and therefore a longer set-up time.

Electrical stimulation to correct hemiplegic gait was first used in 1961 (Liberson et al., 1961). Stimulation of the peroneal nerve was synchronised with heel events using a footswitch mounted under the heel. In figure 2.16 the active and inactive electrodes, the stimulator and the footswitch are shown. The footswitch is closed when the foot is in contact with the ground, causing the electrical current to flow through the shunt resistor shown in the stimulator unit. When the foot leaves the ground the switch is opened,
removing the shunt resistor circuit and driving the current through the electrodes and the human body.

Figure 2.16 The General Arrangement of the FES Drop-Foot Correction System
LIBERSON ET AL., 1961

The timing of electrical stimulation used by Liberson’s stimulator is simple but effective: stimulation current is switched on the moment the heel leaves the ground, and off again when the heel returns. A more complex timing strategy, however, can be more effective. For example, when the heel returns to the ground at the end of the swing phase of normal gait, foot flat (FF) occurs only after a delay. Without the delay, the foot
returns to the ground abruptly (footslap) and the gait is less efficient. Also, if
dorsiflexion occurred too early, i.e. at heel rise (HR) – (as might happen if electrical
stimulation was applied at heel rise), progression of the contralateral limb may be
compromised. Although a short delay occurs after the onset of stimulation, this might
not be sufficient; a suitable delay after the detection of heel rise but before the onset of
stimulation may allow the heel to rise first.

Ramping the stimulation amplitude up or down can minimise foot-slap, avoids the
possible induction of a reflex contraction in the antagonist muscle and reduces the
sensation during off-to-on. Two stimulation envelope strategies, Adaptive Timing and
Fixed Timing, are shown in Figure 2.17. When using adaptive timing, the stimulation
begins ramping down at heel contact (HC). With fixed timing, however, the stimulation
will start to ramp down only after a pre-set time. The time delay occurring before
ramping down does not vary with walking speed, and this delay begins when heel rise is
detected.

After the work of Liberson et al. in 1961, many other researchers also used electrical
stimulation to correct hemiplegic gait (Takebe et al., 1975; Waters et al., 1975; Lee and
Johnston, 1976; Brandell et al. 1985; Cozean et al., 1988; Buurke et al., 1990; Kljajic et
al., 1992; Bogataj et al., 1993; Kralj et al., 1993; Taylor et al., 1995; Granat et al.,
1996; Popovic et al., 1998). Methods of stimulation included: single channel peroneal
nerve (Buurke, Kljajic, Granat, Kralj, Popovic, Takebe, Waters, Lee - three different
sites of stimulation on the limb and the foot were evaluated); dual channel ankle
dorsiflexion (swing phase) and plantar flexion (at the end of stance phase, Cozean,
Taylor) and multichannel ankle dorsiflexion, knee and hip flexion (swing phase), and
knee and hip extension (stance phase) and ankle plantar flexion (at the end of stance
phase, Brandell, Stanic).

Taylor et al. (1995) used a stimulator with two channels of stimulation which could be
used in three ways: bilateral drop-foot correction, drop-foot correction with active push-
off and drop-foot with quadriceps stimulation for weight bearing during the stance
phase of gait. To achieve active push-off, a second footswitch was placed under the first
Metatarsal head switching stimulation of the plantar flexor muscles on at toe contact, and off either at heel rise or toe off (TO).

Figure 2.17  Two Stimulation Strategies for Drop-Foot Correction
Although multichannel stimulation provides selective control over articulations for movement during both the swing and the stance phase, single channel peroneal nerve stimulation is the most popular choice for regular use because it usually offers improved movement with a relatively low set-up time. In some cases researchers found that a number of subjects did not wish to continue using electrical stimulation due to the sensation or the length time taken to position electrodes (Takebe et al., 1975, Granat et al., 1996). In another case subjects were unable to continue because of cognitive difficulties (Cozean et al., 1988).

Other studies have evaluated the efficacy of FES (Maxwell et al., 1995; Burridge et al. 1997). Using walking speed and the Physiological Cost Index (PCI) as indicators, it was collectively found that thirteen of seventeen patients showed significant improvement.

Kralj et al. (1993) reported on electrical stimulation techniques used by patients in Slovenia. Of approximately 1840 cases of stroke, 1150 cases of MS and 35 cases of SCI injuries per year, up to 63% of annual cases were reported to be candidates for an FES based therapeutic locomotion rehabilitation program. Of this group, 60% received single-channel stimulation to correct drop-foot, 30% received dual or three channel stimulation and 10% were involved in four, six or eight channel stimulation. These figures demonstrate the increasing use of drop-foot correction using FES and also the fact that single channel stimulation is most popular.

A ‘carry over-effect’ (continued improvement in gait after stimulation is removed) was observed in some studies (Waters et al., 1975, Kljajic et al. 1992, Taylor et al., 1998). Taylor calculated the effect by comparing each patient’s gait three times over a 10m course initially and after four and a half months (a single channel peroneal nerve stimulator was used by the patient throughout the period). At the end of the period, a 14% increase in walking speed and a 19% decrease in PCI in 111 patients was observed (in the same study, 21 MS patients did not demonstrate any carry-over effect).
2.3.2.1 Implantable Electrodes

Some researchers have reported that patients using FES have decided not to continue with the treatment due to the time taken to locate the correct electrode position (Takebe et al., 1975, Granat et al., 1996, Taylor et al., 1998). To overcome this problem implantable stimulators (including implantable electrodes) have been successfully designed and evaluated by researchers, including Waters et al., 1975, Tahtinen et al., 1992; Rozman et al., 1995; Tomsic et al., 1995; Bugbee et al., 1997; Gider et al., 1997; Haugland, 1997). Waters mentioned that the use of implantable stimulators also eliminates the sensation of pain. Implanted electrodes were also used to record afferent neural signals (Yoshida and Hurtch, 1996) and Upshaw and Sinkjaer, 1997). This method removes the requirement for a footswitch (or other sensor to be used) since gait events are determined from the neural signal. However, there is a higher cost (for the necessary surgery), a risk of infection and poorer event detection accuracy (Upshaw reported that 85% of heel contacts were detected over 1,100 strides) currently associated with this method.

2.4 Sensors for the Detection of Gait Events

In order to determine electrical stimulation timing in drop-foot correction, a sensor is used to detect the gait events. The force sensitive resistor (FSR), or footswitch, is the sensor that is currently used in the majority of drop-foot correction systems. However, limitations of this type of sensor led to the evaluation of other devices.

2.4.1 Footswitches

Types of footswitch include the use of conductive rubber (Minns, 1982) a simple switch design (Ross and Ashman, 1987), and Force Sensitive Resistors (Interlink Electronics, 546 Flynn Road, Camarillo, CA 93012, USA). FSRs are the most common type of footswitch design, and (unlike the other designs) they have been developed to be reliable over periods of several months (FSRs form part of the Odstock Drop-Foot Stimulator, the use of which was reported by Taylor et al. 1998).
Force sensitive resistors (FSRs) are polymer thick film devices exhibiting a decrease in resistance with an increase in the force applied to the active surface. The FSR manufactured by Interlink Electronics consists of three layers: a flexible substrate with printed semi-conductor, a spacer adhesive, and a flexible substrate with printed interdigitaling electrodes.

An FSR that is used for drop-foot correction is often known as a footswitch (despite the fact that it is not actually a switch at all). When the footswitch is placed inside the shoe and under the heel, there is a change in force (and hence change in resistance) during gait that may allow a suitable measuring device to detect when the heel is on and off the ground, and hence determine gait events.

Footswitches are considered to be a good compromise between low cost, ease of donning and doffing and suitability of measurement method. Under ideal conditions, such as a normal gait style, footwear of good fitting and positioning of the sensor, the footswitches have the potential to detect gait events (HC, FF, HR and TO) reliably. However, a hemiplegic subject will often walk with inversion or eversion, and patients do not always position sensors as accurately as trained professionals. A system that operates well for a patient wearing one pair of shoes may not be as reliable when using another pair (Dai et al., 1996; Ott et al., 1998).

Footswitches deteriorate throughout use due to the relatively large forces exerted upon them during gait. Experience has shown that the lifetime of a footswitch used for drop-foot correction ranges from approximately 3 to 9 months, and the cost of replacement must therefore be met at least once every year (in addition, the connecting lead can break resulting in failure).

In some cases, patients suffer from ankle plantar flexion tone resulting in little or no heel contact with the ground (Ott et al., 1998). In this case, the footswitch must either be used under the toe or under the contralateral foot. If the footswitch is placed under the toe, electrical stimulation may be switched on too late (if the stimulation amplitude is ramped up then this may sometime after occur after toe off, instead of after heel rise).
and also switched off too late (this will take place after foot flat instead of after heel contact). If the affected leg foot contact is particularly poor, leading to inadequate compression of the footswitch, the sensor may be placed under the contralateral foot. If the footswitch is placed under this foot, stimulation timing will be compromised further as gait events will be nothing more than an indication of what the affected limb is doing during gait.

When traversing staircases, it is often the case that the toe only is placed in contact with each tread. This is because tread length is usually less than foot length. If a heel switch is used to determine stimulation timing, there will be no heel contact during this time. This results in failure of the system to provide assistance, and the stimulator becomes less useful for staircase traverses (Takebe et al., 1975; Ott et al., 1998).

Aejaz (1997) collected footswitch data from three normals and one hemiplegic subject (who had suffered a CVA 3 and a half years previous and had been using a single channel stimulator for 2 years). In one of the normal subjects and the hemiplegic subject, footswitch activation was present in the swing phase. This was thought to be due to a) type of footwear, b) position of footswitches in the shoe and 3) toe-clawing by the hemiplegic subject.

2.4.2 Other Sensors

Since the first use of footswitches by Liberson et al. in 1961, researchers have attempted to use other sensors to overcome some of the problems discussed in section 2.3.1.

2.4.2.1 Hand Switches

Ott et al. (1998) compared footswitches with the use of a simple hand-switch to trigger drop-foot correction using FES. The hand-switch, which is often mounted inside the handle of a crutch, allows the patient voluntary control over the timing of stimulation. Two cases (both incomplete SCI patients) used a hand-switch to trigger FES, with a footswitch placed under the affected foot from which measurements were taken only. The heel footswitch did not trigger stimulation in 10 of 17 strides and 20 of 32 strides.
In both cases, the heel was prematurely unloaded which would have caused poor stimulation timing. The hand switch was pressed, on average 1.1 and 0.4 seconds after heel rise and released between 0.1 and 0.2 seconds after initial foot contact. In four strides the hand-switch was not used correctly leading to inadequate stimulation.

One must question the practicality of relying on the patient to control stimulation timing. The requirement of pressing the button at the appropriate time is an additional burden for the hemiplegic patient who already has difficulty walking. It is likely that minor distractions would considerably affect stimulation timing judgement, and that many elderly patients would find the operation difficult.

2.4.2.2 Accelerometers

Willemsen et al. (1990) used four accelerometers mounted on the shank to detect heel contact and toe off. The sensors were mounted onto a harness which was then attached around the shank. Ankle joint acceleration was determined from the shank accelerations, and a detection algorithm was used to find the required gait events. The system was evaluated on four normal subjects and four hemiplegic subjects. It was found that some events were detected early, and more significantly that the controller often became out of phase with the gait cycle and some events were not detected. While detecting events too early may not limit the use of the system in drop-foot correction, false detection and lack of detection may greatly reduce the benefit of the stimulation system.

2.4.2.3 Hall Effect Velocity Transducer

The angular velocity of the knee joint was measured (as part of an FES system) developed by Heath et al. in 1995. As the output of the sensor crossed a threshold, stimulation of the peroneal nerve was switched on or off accordingly. It was believed that the system would be capable of detecting gait events (toe off and heel contact) because the angular velocity of the knee joint is known to vary predictably and repeatably throughout gait. Tests with eight able-bodied subjects found that the system triggered reliably in all cases, but the system had yet to be tested with pathological gait.
Literature Review

It is possible that in the case of pathological gait, the knee joint angular velocity may not vary with adequate predictability and repeatability.

2.4.2.4 Tilt Sensors

A single tilt sensor (a magneto resistive sensor, able to measure ±50° and withstand a maximum shock of 50g), was used by Dai et al. (1996) to control stimulation timing for stimulation of the peroneal nerve. The system switched stimulation on and off as the measured shank angle crossed a pre-set threshold. Six subjects, who had suffered either a stroke or an incomplete spinal cord lesion, were used for system testing. Although specific details were not presented, it was reported that initial trials with stroke and SCI subjects showed that tilt sensors can be replaced by foot switches to control FES in preventing drop-foot. However, it was also stated that subjects who had a limited swinging movement of the lower leg produced errors in the detection of step intention. A high number of hemiplegic individuals exhibit limited swinging movement of the lower leg, and this may reduce the number of patients able to use the system.

2.4.2.5 EMG

An assessment of EMG measurement for use in a closed loop drop-foot correction FES system was made by Kershaw et al. (1995). Surface EMG was collected from the tibialis anterior muscle of a healthy subject and also from three subjects with disabling conditions – mild MS, severe MS and peripheral peroneal nerve injury. Closed loop stimulation was expected to produce finely graded contractions and response to muscle fatigue. Recorded signals showed that very low activity occurred between FF and HR; higher bursts of activity occurred just before HC, and also some time after HR (although this was always some time after HR - maximum 379 ± 235 ms). During the swing phase, EMG activity appeared to decline to the same low level as recorded during the stance phase, before reaching the peak just before HC. The difference between the low level activity signal and the peaks was significantly reduced in the mild MS and peripheral peroneal nerve injured cases compared to the healthy subject, and was worst in the case of severe MS (no details were given). Detecting events from the measured
signals may prove to be difficult given the low difference between the required signal (the peaks) and the low-level activity signal. Also, routine use of EMG electrodes would require reasonably accurate positioning and a longer total equipment donning and doffing time.

2.4.2.6 Closed-Loop Ultrasonic Device

Micheal (1996) designed and tested a sensor which used ultrasound to determine the distance between the foot and the ground. Two sensors were placed on the sole of the foot, one at the heel and the other on the toe. Since each one measured the distance between itself and the ground, the inclination of the foot could also be measured. However, the use of footswitches was recommended when the sensor was used with carpets and other floor surfaces with poor reflectivity. A drop-foot system requiring multiple sensors may be undesirable due to the time required to don and doff.

2.4.2.7 Optical Fibre Based Goniometer

An optical fibre with good sensitivity to bending motions was designed by Peasgood et al. (1997). It was intended that the device (measuring knee joint angle) would be used with other sensors and a stimulator to correct drop foot. A plastic optical fibre was used with a light emitting diode at one end and a photo diode at the other. The fibre exhibited a bending loss proportional to the degree of curvature. The device operated linearly within the range of 20° to 80°, but demonstrated hysteresis (no values were reported). As mentioned in section 2.4.2.3, it is likely that with pathological gait the knee joint angular velocity will not vary predictably and repeatably, although knee joint angle measurement in conjunction with other sensor measurements may provide acceptable accuracy in determining gait events. However, as mentioned in section 2.3.2.6 a practical drop-foot system using many sensors may be undesirable.

2.4.2.8 Machine Learning Techniques

Machine learning techniques were used to determine optimum sensor combinations (Tong and Granat, 1998). The simulation of 22 virtual kinematic sensors was achieved
using 3 dimensional data collected from a motion analysis system (VICON). The sensors comprised of hip and knee joint goniometers, and thigh, shank and crutch mounted accelerometers, inclinometers and gyroscopes. Real sensors were also used: four footswitches were placed under the heel, big toe and the heads of the first and fifth metatarsal heads and six strain gauges were mounted on the crutch tip. Two incomplete SCI subjects with drop-foot (who both used crutches and single-channel peroneal nerve stimulation) were used to evaluate the system, which involved establishing the optimum three-sensor set for each subject (three sensors were considered to be acceptable for regular donning and doffing).

A strain gauge on the right crutch, an FSR under the right heel and a goniometer on the right hip were considered to be the ideal sensor combination for the first subject. The accuracy was measured by comparing gait event times with those measured using the VICON system at intervals throughout the six-month period. After six months the accuracy of the three-sensor combination was found to be 91%, compared to an accuracy of only 75% for the footswitch. The optimum sensor set found for the second subject was an inclinometer and a gyroscope mounted on the anterior aspect of the right shank and an FSR under the right toe. After four months the accuracy of this combination of sensors was found to be 94% compared to only 71% footswitch accuracy.

Machine learning techniques have also been investigated by other researchers (Kirkwood et al., 1989, Andrews et al., 1995, Wang et al., 1997, Sepulveda et al., 1998, Jonic et al., 1999). While it may be useful to determine ideal sets of sensors for each individual, the use of many sensors may not be ideal due to donning and doffing time for each patient). However, the use of machine learning to provide improved event detection from the same sensor (e.g. artificial neural networks) is worthy of further investigation.
2.4.2.9 Other Sensors: Discussion

Since Liberson's pioneering work in 1961, footswitches have remained the most popular choice for drop-foot correction systems. However, there is evidence to support the belief that an improved sensor could reduce or eliminate the disadvantages discussed in section 2.3.1, whilst retaining most or all of the benefits of footswitches. The use of other sensors to detect gait events has been discussed in section 2.4.2, but it is clear that a practical alternative to the footswitch remains to be developed. This is due to the combined requirement for adequate accuracy and straightforward donning and doffing.

In this study a vibratory gyroscope sensor (the Murata ENC-05E) is evaluated as a sensor for drop-foot correction systems. The following sections discuss the operation of the vibratory gyroscope, the current use of such sensors for FES and the reasons for its selection in this study. At the beginning of this study, only one published research article relating to this particular sensor application was located by the author (Heyn et al. 1993). In this study, measurements during normal gait were made using both shank and thigh mounted gyroscopes (the manufacturer and part number were not stated), and a VICON motion system. In a comparison of the measured shank angle using both methods, a correlation coefficient of 0.995 was found - suggesting an excellent relationship between the two methods.

2.5 The Vibratory Gyroscope

The vibratory gyroscope is a velocity sensor containing piezoelectric elements and may be found in navigational equipment and camcorders (for image stabilization). Although the vibratory gyroscope does not offer the same performance as the more traditional spinning device, it does offer advantages relating to price, size and weight. The operation of all gyroscope sensors is based upon the Coriolis principle.
2.5.1 The Coriolis Principle

The Coriolis principle involves the generation of a force composed of two separate physical effects. The simplest possible motion in which this appears is a rotating disc containing a slot in which an object may slide, as shown in figure 2.18a.

![Figure 2.18a Demonstration of the Coriolis Principle](image)

Let the disc shown in figure 2.18a rotate with a constant angular velocity $\omega = \dot{\theta}$, and the particle A move inside the slot with a constant speed of $V_{rel} = \dot{x}$ relative to the slot. Under these conditions, the velocity of A will have two components: $\ddot{x}$ (the linear movement relative to the slot) and $x\omega$ (the rotational movement due to the disc motion).

If the disc rotates by an angle $d\theta$, then the $x - y$ axes rotate with the disc through an angle $d\theta$ to $a - b$, as shown in figure 2.18b. The velocity increment due to the change in direction of $V_{rel}$ (due to the disc rotation) is $\dot{x} d\theta$, and the velocity increment due to the change in magnitude of $x\omega$ (due to the increase in the distance $x$) is $\omega \, dx$. Dividing each increment by $dt$ ($\dot{x} d\theta$ becomes $\ddot{x} \omega$ and $\omega \, dx$ becomes $\omega \ddot{x}$) and adding produces the sum $\omega \ddot{x} + \ddot{x} \omega = 2 \ddot{x} \omega$, which is the Coriolis acceleration.
The direction of the Coriolis acceleration is always normal to $V_{rel}$, in the $y$-direction normal to the slot. Given that the Coriolis acceleration is:

$$a = 2. \dot{x}$$

and that:

$$F = ma$$

The Coriolis force may be written as follows:

$$F = 2m \dot{x} \omega$$

Given that the mass $M$ of the particle A is constant, and that ideally the velocity $\dot{x}$ is also constant, the force $F$ will be dependant only upon the angular velocity $\omega$. However, since it is not actually possible for the particle to continue travelling in one direction for
any length of time, A must instead move backwards and forwards through the slot and the velocity sampled when the distance between the centre of the disc and A is half the radius.

2.5.2 The Vibratory Gyroscope

There are two types of vibratory gyroscope: the tuning-fork and the bar type. Although the two types are physically different, the principle of operation is the same. Figure 2.19 shows the tuning-fork type and the bar type vibratory gyroscope.

The gyroscope sensors shown in figure 2.19 contain two detection elements (the second element is hidden in the bar type) driven by a third oscillating element. All elements are piezoelectric, and the driving element is set to oscillate at a pre-determined frequency.

Figure 2.19 The Tuning Fork and Bar Vibratory Gyroscope
When the gyroscope is rotated about the Z plane, a Coriolis force $F$ is produced and transmitted to the detection elements:

$$F = 2mv\Omega_0$$

To measure the angular velocity $\Omega_0$, the mass $m$ and the velocity $v$ of the detection element should ideally remain constant. While the mass of the object is constant, the velocity is not since the detection element will oscillate with the driving element (since they are physically linked). However, by sampling only the peak of the velocity amplitude the value appears to be constant. The Coriolis force $F$ therefore becomes dependant only upon the angular velocity, $\Omega_0$.

An example of the bar type design shown in figure 2.19 is the Murata ENC-05A gyroscope (dimensions: 21.5mm x 8.5mm x 7.1mm). This sensor has one driving element and two detecting elements. By using two detecting elements mounted 60° to each other, the Coriolis force is measured by both elements. The detecting elements respond inversely to each other, so that the difference between the two is proportional to the angular velocity. The advantage of this arrangement is that noise components are canceled out by the subtraction. Figure 2.20 shows the arrangement of the driving element (at the bottom) and the two detection elements, and the operation of noise cancellation and rotation.

![Figure 2.20 The Murata Gyroscope Sensor: Noise Cancellation and Rotation](image)
2.5.3 The Gyroscope Sensor in FES Systems

The gyroscope sensor has been used in proposed FES systems for improvement in hand grasp (Tong, 1999), sit-to-stand (and stand-to-sit) transitions, quiet standing and stepping (Williamson and Andrews, 1997) and drop-foot correction (Popovic et al., 1998).

Tong used the gyroscope sensor (the Murata ENC-0D) to detect five normal subject’s intention to initiate FES assisted hand grasp (for incomplete SCI patients). The sensor was mounted on the upper arm, and pre-defined movements were made to trigger an On/Off instruction. Normal activities did not appear to cause unintentional triggering.

Clusters of sensors, consisting of two accelerometers and one gyroscope sensor (the Murata ENV05), mounted on the trunk and each thigh and shank to measure lower limb joint angles and segment inclinations were developed by Williamson. The clusters were designed to be small and placed in areas that were unobtrusive to the users. The derived signals were said to be of sufficient quality for control of FES.

A gyroscope sensor (the Murata ENC-05A) and three footswitches was used by Popovic to determine gait events. Experiments performed with ten normal subjects and ten disabled subjects (no details given) showed that heel contact, foot flat, heel rise and toe off were measured with 99% accuracy (no details were provided regarding the method of confirmation). Subjects that were trained to use the sensor achieved better results than subjects using the sensor for the first time (this statement was not explained). The angular velocity signal from the gyroscope was integrated to provide foot inclination information.

2.6 Discussion

This chapter has cited a number of publications regarding the use of electrical stimulation for drop-foot correction. The requirement for a sensor or sensors for the detection of gait events to control stimulation timing has also been discussed. Initially the footswitch was the sensor of choice, and this is still true today due to a compromise
between cost, and a number of advantages of the sensor. However, disadvantages of footswitches have led researchers to consider other sensors, most of which have taken measurements from the knee joint or the shank.

Considerations that should be made in choosing a sensor for drop-foot correction FES systems are:

- The ease of donning and doffing (multiple sensors may require a longer set-up time)
- The accuracy of the sensor (poor accuracy leads to poor stimulation timing)
- The cost of the sensor (if the FES system is to be used by many patients)

The footswitch is considered to offer simple donning and doffing (the sensor or sensors are usually mounted inside a removable shoe insole), and the cost is acceptable (under £25 for one). However, footswitch accuracy can vary from acceptable to very poor, depending upon factors such as the patient’s gait, footwear and terrain. The short lifetime of the sensor can lead to greater inconvenience for the patient (if a footswitch fails, there may be a delay before a replacement arrives) and also the cost.

Of the other sensors considered, some involve a lengthy donning period (multiple accelerometers, EMG electrodes) or expensive surgery (natural sensors). In the majority of cases, the knee joint angle or the shank inclination was calculated to indicate the occurrence of gait events. While changes in the knee angle or shank inclination may be used to infer foot contact with reasonable accuracy in normal gait, clinical observations suggest that such movements are much less predictable in pathological gait (e.g. circumduction or stiff knee gait). If the detection of events is determined using the contralateral limb, or the measurement of trunk or upper limb movement, then accuracy will decrease leading to poor stimulation timing.

The gyroscope sensor has been considered for use in FES systems by a number of researchers. Popovic et al. (1998) has used the gyroscope sensor with three footswitches to detect gait events for the correction of drop-foot. The system performed with an accuracy of 99%, although the requirement for three footswitches to be used increases...
the cost and the donning/doffing time. The sensor system proposed may be unacceptable for multiple patient use due to these reasons.

As a novel sensor drop-foot FES, the Murata ENC-05E gyroscope sensor was evaluated as a sole gait event detection sensor in FES drop-foot correction. The gyroscope-based sensor was chosen for evaluation as a result of the findings of the Heyn et al. (1993) study (see page 48). This particular sensor was selected for its small size, low weight (less than 20g) and high sensitivity to movement (less than 1°/sec). Four gait events (HC, FF, HR, TO) were detected to allow comparison with both the heel and toe footwitches, and to potentially allow the sensor to control a one or two channel FES drop-foot correction system (see section 2.3.2).

The cost of the sensor is less than £25, which is comparable with that of one footswitch (£22.40 - Salisbury District Hospital, Medical Physics and Biomedical Engineering - note that the cost of two footswitches is £44.80). Realistically speaking, the cost of the gyroscope sensor should be increased by just a few pounds due to the requirement for supporting electronics to allow the sensor to be connected to a standard stimulator (one designed to operate with a footswitch). Finally, the lifetime of a footswitch can typically be as low as 6 months; it is expected that the gyroscope sensor will not malfunction before several years of operation (although the connecting lead can suffer breakage in the same way as that of the footswitch).

2.6.2 Project Objectives

1. To perform a literature review of current drop-foot FES devices.
2. To evaluate the gyroscope sensor (Murata ENC05E) within a pilot study, involving the development of a lower limb joint angle measurement system which could be used for simple gait analysis purposes
3. To design and build a device that can be used to calibrate the gyroscope sensor (Murata ENC05E).
4. To develop a real-time gait event detection PC based system to determine whether the gyroscope sensor is suitable for use within a drop-foot correction system.
5. To produce a portable data logger/real-time gait events detection system, allowing comparisons to be made between the gyroscope sensor and the established footswitch sensors when used by normal and hemiplegic patients.
Chapter 3
Calibration and Initial Use of the Gyroscope Sensor: Measurement of Lower Limb Joint Angles

3.1 Introduction

The Murata ENC-05E gyroscope sensor was selected for evaluation as a sensor within a FES drop-foot correction system (see section 2.4.2.9). In order to provide a relatively simple study for evaluation purposes, three sensor units were initially used to measure the angular velocity of the thigh, shank and foot. Since joint angles could easily be obtained through integration, the system could be used for general gait analysis purposes, as well as allowing the assessment of velocity and displacement signals for use within the FES system.

Three previously assembled sensor units and a sensor adapter had been provided by GEC-Marconi. Each sensor unit, shown in figure 3.1, was housed in box of dimensions 50mm x 30mm x 20mm (weighing approximately 70g). Each sensor was based around the Murata ENC-05E piezoelectric Vibrating Gyroscope IC, which measured angular velocity through a single axis. The IC produced a voltage proportional to the angular velocity measured (1.1mV/°/sec). A lead of length 10m was also supplied, allowing remote connection to a computer. The sensor adapter contained three sockets providing connection for three sensor units. The adapter unit also served as a power supply for the sensors, its own requirements being a +5V and a ±12V supply provided by the computer. Data collection software was written using LabVIEW (National Instruments); this was chosen because it offers a relatively simple and flexible approach to analogue data acquisition when used with a National Instruments data acquisition PC card. For this reason the PC-LPM-16 data acquisition card was also used, offering 16 analog inputs (measuring between ±5V) with a 12-bit ADC. The INA101HP (Burr-Brown) instrumentation amplifier IC was used to provided a gain of 6.88, allowing the system to measure a maximum angular velocity of 668°/sec. A low pass filter with a 3db point at 31Hz was constructed using a simple one pole RC filter. This removed unwanted
noise while retaining gait information. The circuit for each sensor unit is shown in figure 3.2.

Figure 3.1 Assembled Gyroscope Sensor Components (board shown is 30mm x 20mm)

Figure 3.2 The Gyroscope Sensor Circuit

Each sensor had a voltage offset of ±100mV when the angular velocity was zero. The offset was prone to drift with changes in temperature (see section 2.5) but since the equipment was to experience little change in temperature (±1°C), only a simple self-
calibration routine was required. This routine was performed at the beginning of any measurement, and required the subject to remain still for 2 seconds whilst the software collected 100 samples of the offset and subtracted an average of these values from subsequent measurements.

Two other studies have also used gyroscope sensors to take measurements during gait; Heyn et al. (1993) compared measurements in normal gait made by a VICON motion system, and two gyroscopes (the manufacturer and part number were not stated), one mounted on the thigh and the other on the shank. A correlation coefficients of 0.995 (VICON compared with gyroscope sensors - shank angle) was obtained. Tong et al. (1999) also compared measurements made by a VICON system and two gyroscopes (also the ENC 05E) - again one mounted on the thigh and the other on the shank. Both normal gait and the gait of a subject with an incomplete SCI were examined. Correlation coefficients of 0.94 (shank angular velocity), 0.91 (thigh angular velocity), 0.92 (shank angle), 0.90 (thigh angle) and 0.93 (knee joint angle) were obtained. These correlation values suggest that there is an excellent relationship between the VICON and the gyroscope measurements. In both studies sensors were placed only on the thigh and the shank – no ankle angle measurement was made.

3.2 Sensor Calibration

Before the sensors were used, a calibration study was conducted to test the relationship between angular velocity and voltage output for each sensor. Murata specifications for the gyroscope IC stated that the maximum measurable velocity was 90°/sec, despite observations suggesting that this figure might be higher. Also, no information was provided regarding its low-speed sensitivity. Calibration of the sensors was achieved using an adapted vinyl record turntable, incorporating a variable speed motor (0 – 0.90 rev/sec), a step-up gearbox (providing a ratio of 2.42:1) and an optical sensor unit (with a resolution of 0.72°/sec). With a gyroscope sensor mounted on the platter of the turntable, the device was capable of spinning the sensor at a maximum speed of 785°/sec. The sensor’s cable was wound around a central pulley as the sensor was spun, limiting the data collection time to 3 seconds at the maximum speed (this time was
Calibration and Initial Use of the Gyroscope Sensor

greater for slower speeds). Although the motor could run unloaded at any speed from 0.90 rev/sec to zero, platter speeds below 60°/sec were not possible due to a lack of low-speed motor torque. However, for lower speeds another gearbox could be attached, offering a step-down ratio of 100:1. This provided platter rotation speeds from 6 to 0.5°/sec.

Speeds of 0.5, 2, 6, 60, 300 and 600°/sec were chosen (the 3 lower speeds were easy to achieve using the step-down gearbox) to represent a reasonable cross-section of speeds between 0 and 600°/sec. The data from the optical and gyroscope sensors were collected at 100Hz using LabVIEW software. This frequency provided a minimum of 200 data points at the maximum speed (due to the time taken for the cable to be wrapped around the pulley, and also the time taken for the system to reach the desired speed). Figure 3.3 shows the collected data at all six speeds, and figure 3.4 shows the low speed results only.

The graphs suggest that there is a linear relationship between the optical sensor and the gyro sensors, although the relationship is slightly different for each sensor. The output of gyroscope 2 matched most closely the output from the optical sensor; the outputs of gyroscopes 3 and 1 may be normalised to gyroscope 2 by using factors of 1.020 and 0.6422 respectively. Specification tolerances would account for these differences.
Figure 3.3 Graph of Gyro Sensor Measurement Against Optical Sensor Measurement

Figure 3.4 Graph of Gyro Sensor Measurement Against Optical Sensor Measurement (low speeds only)
3.3 Procedure

Sensors were placed on the anterior aspect of the thigh and the shank, and just above the metatarsals of the foot, as shown in figure 3.5. Initially, the sensor units were attached using the ‘Velcro’ straps that were built into the units. However, the attachment method was changed (as a temporary measure only) to electrical tape in order to avoid any relative movement of the sensor compared to the movement of the limb segment. Although it was necessary to ensure that the orientation of the sensor was such that the measurement axis was in line with the rotation of the limb segments, the alignment was approximate only.

![Figure 3.5 Sensor Locations on the Limb for Joint Angle Measurement](image)

The primary function of the measurement software, written in LabVIEW, was to collect and process data from each of the sensors and present it in graphical format. Data was collected from the three sensor channels each at 100Hz (the majority of frequency components associated with gait are below 30Hz). Processing involved calculating the differences between both the hip and the shank velocity and the shank and the foot velocity to determine the velocity at the knee and the ankle respectively (the hip angle was calculated using the thigh sensor alone, since the trunk was assumed to remain reasonably vertical during gait). Each derived joint velocity sample was then integrated using the equation:

$$Y(i) = Y(i-1) + \frac{1}{6} [X(i-1) + 4X(i) + X(i+1)] \, dt$$
Where $X$ is the angular velocity and $Y$ is the derived angle (LabVIEW, National Instruments). The resulting joint angles were then plotted on an angle/time graph.

### 3.4 Initial Results

The system was used initially to measure the gait of three able-bodied male adults with a normal and a simulated pathological gait (circumduction of the measured limb). Figure 3.6 shows one such measurement for normal gait, with the joint angle (degrees) shown on the y-axis, and time (seconds) shown on the x-axis. No joint angle convention has been used, thus the relative hip, knee and ankle angles only are shown.

A visual comparison of kinematic data from this and other studies (Heyn et al., 1993; Tong et al., 1999) confirmed that in most cases, the data obtained was approximately as expected for normal gait. In some cases of normal gait, however, signal drift occurred when using the gyroscope sensors. This drift was more apparent during measurement involving simulated pathological gait, and an example of this drift is shown in figure 3.7. The graph shows some drift occurring in the knee angle, and, to a much greater degree, in the ankle angle. This was clearly not a true representation of the actual joint angles.

It was found that the source of drift was an inappropriate sensor response to shock (a high acceleration - several hundred g - occurring when the sensor is stopped very suddenly). This was demonstrated by placing the sensor in collision with a heavy, fixed object such as a desk; after this occurred the sensor displayed an offset despite being still. The output voltage (proportional to angular velocity) is momentarily inaccurate at this time, and in the angle domain this error appears as a change in the initial start position.
Calibration and Initial Use of the Gyroscope Sensor

Figure 3.6 Graph of Hip, Knee and Ankle Angles during Normal Gait

Figure 3.7 Graph of Hip, Knee and Ankle Angles during Simulated Circumduction Gait
3.5 Drift Compensation

In an attempt to remove the signal drift shown in figure 3.7, a high-pass filter (cut-off frequency 0.3Hz) was first used, but appeared to have no effect on the error. A modification was then made to the software. Since the drift appeared to be constant throughout each gait cycle (taken as the time between one heel strike and the next) but varied across several gait cycles, an algorithm was developed to apply individual drift compensation factors to each gait cycle. Figure 3.8 shows the measured angular velocity (the unprocessed signal) of the shank and the foot. To automatically detect heel strike, the unprocessed signal from the foot was used. The swing phase was detected by counting 15 consecutive values that were greater than 0.5V (large anticlockwise foot rotation). Approximately at the end of swing phase the anticlockwise foot rotation ends. It is here that the measured angular velocity (and therefore the output voltage) will change from positive to negative. Therefore, a positive to negative zero crossing was used to detect heel contact.

After each heel strike had been detected, the acquired data was divided into groups of one gait cycle each. Before integration occurred, half the difference between the first and last velocity sample in the first group was subtracted from each of the velocity samples within that group:

\[ V_{\text{Compensated}}^{i} = V_{i} - \frac{V_{f} - V_{l}}{2} \]

Where \( V_{i} \) is the current velocity sample, \( V_{l} \) is the last sample in the gait cycle and \( V_{f} \) is the first.
The same operation was then repeated on the second group, and so on. Thus, if a cycle had no drift, the difference between the first and last value would be zero (the area bounded by the plot would also be zero), and each value would remain unaltered. However, if a cycle had experienced a positive drift, resulting in the last value being greater than the first value (and hence the area bounded by the plot would be positive), the compensation factor to be applied would be positive (making the area bounded by the graph zero as before). This method assumed that the drift within each gait cycle was constant.

3.6 Results After Modification

After the drift compensation modification was applied to the software, the system was used again to measure lower limb joint angles during gait. This time the results were compared with the same angles measured simultaneously using a MacReflex marker detection system (MDS). The MacReflex system used five cameras to track the position of a number of reflective spheres each mounted at various positions on the body. Data
was collected at 60Hz, and the start time for measurement using the two systems was synchronised by way of a control signal from the MacReflex system to the gyroscope system. The subject (a male adult) was asked to walk (with the gyroscope sensors attached) through the calibrated volume of the MacReflex system with a normal gait and a simulated pathological gait as before. The subject performed 19 walks with normal gait, and 13 walks with simulated pathological gait (6 footslap and 7 circumduction). Figure 3.9 shows the measurement taken by the MacReflex system during two strides of the measured limb. The same measurement using the gyroscope sensors can be seen in figure 3.10. Figure 3.11 shows the same data, but with one graph laid over the other for comparison. In each graph the MacReflex plot can be identified as that which is least smooth. HC and TO were determined using a ground reaction force walkway.
Figure 3.10 Graph of Lower Limb Joint Angles Measured using the Gyroscope Sensors

Figure 3.11 Graph of Lower Limb Joint Angles Measured using the MacReflex and the Gyroscope Sensors (Normal Gait - 1)
Figure 3.12 shows a second comparison of joint angles during normal gait; the subject did not wear shoes during the measurement. In figure 3.13 data collected during a simulated footslap style of gait is shown; again the subject did not wear shoes during the measurement. The measured joint angles during an alternative simulated pathological gait are shown in figure 3.14. The subject, who was asked to circumduct the measured limb, did not wear shoes during the measurement.

**Figure 3.12** Graph of Lower Limb Joint Angles Measured using the MacReflex and the Gyroscope Sensors (Normal Gait - 2)
Calibration and Initial Use of the Gyroscope Sensor

Figure 3.13 Graph of Lower Limb Joint Angles Measured using a MacReflex System and Gyroscope Sensors (Simulated Pathological Gait -1)

Figure 3.14 Graph of Lower Limb Joint Angles Measured using a MacReflex System and Gyroscope Sensors (Simulated Pathological Gait -2)
Table 3.1 shows the calculated correlation coefficients for all 19 walks. Correlation coefficients were calculated using Excel (Microsoft Office 97) using a technique comparable with that used in other similar studies.

<table>
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<th>Walk</th>
<th>Normal or Pathological</th>
<th>Shoes Worn?</th>
<th>Hip Averages</th>
<th>Knee Averages</th>
<th>Ankle Averages</th>
<th>Ankle Averages</th>
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Table 3.1 Correlation Coefficients of measurements of Lower Limb Joint Angles taken using a MacReflex System and Gyroscope Sensors
In the table it can be seen that for normal gait, the average correlation for the hip and knee is very high, but still noticeably above that of the ankle measurements. When comparing normal gait with shoes on and shoes off, it is clear that the average correlation for all measurements is lower when shoes are not worn, but this difference is small. The table shows that the average correlation coefficient is distinctly higher for the hip measurements in simulated pathological gait, followed by the knee angle.

3.5 Conclusions

Agreement between the measured data from the gyroscope sensors and the MacReflex system was strongest during normal gait, and in this case correlation coefficient values have been found to be similar to those found in other studies (Heyn et al., 1993; Tong et al., 1999). The graphs shown in figures 3.11 and 3.12 were typical for comparisons of this kind, and both graphs show that there was a lower correlation between measurements of the ankle joint. Simulated pathological gait measurements found less agreement in the hip joint angle measurements, becoming progressively worse in the knee and ankle joint angles. The lowest correlation coefficient value of 0.35 was found for the ankle joint angle in one circumduction gait style measurement made without shoes.

There are a number of factors that reduce the degree of agreement between the two measurements. The first contributing factor to measurement error is a change in the zero offset value, which occurred when the sensor experienced shock (a high acceleration of several hundred g). The sensor was exposed to such a shock during gait, resulting in small changes to the zero offset (the software measured the zero offset at the start of each walk, and the resulting value was then subtracted from the angular velocity measurements). The compensation modification appeared to eliminate drift in normal gait, but in simulated pathological gait this error remained. This can be seen in figures 3.13: the gyroscope ankle joint angle plot begins above the MacReflex plot on the left hand-side of the graph, but drops lower towards the right hand-side. A high-pass filter did not remove this error.
A second contributing factor is a compression of the measurement axis caused by the sensor axis not being exactly aligned with the sagittal plane. This error results in attenuation of the measured signal. Misalignment can be caused either by poor sensor positioning on the limb (this would produce a constant measurement error), or by limb rotation during gait (in this case the magnitude of error would vary). It is likely that in most cases poor sensor positioning would lead to small errors (±1 or 2%), but in pathological gait limb rotation in the transverse plane could cause significantly greater errors.

Finally, the degree of agreement between the two sets of data for a given walk could have been reduced by the possibility of errors in the MacReflex measurement. This would be caused by any markers being momentarily out of view to some or even all of the cameras, resulting in the generation of inaccurate interpolated data by MacReflex. Some obvious features occurring as a result of this error can be seen in figure 3.14: the hip and knee joint angle plots show positive and negative peaks respectively.

It has been shown in this study the gyroscope system is not suited to the measurement of lower limb joint angles during pathological gait, due to errors that occurred as a result of integration. These errors arise from a failure of the calculated displacement signal to return to its original position, during sensor movements that are similar to those found in simple harmonic motion. This in turn occurs whenever the sensor experiences shock (a high acceleration) when it is stopped very suddenly, as a result of the foot striking the ground during walking. Since the angular velocity signal is produced without signal processing, however, it did not suffer the same integration related errors. It was therefore considered further for use in the control of stimulation timing. In the next chapter the angular velocity signal is evaluated as a means of detecting gait events.
Chapter 4
The Detection of Gait Events Using a Vibratory Gyroscope: A Preliminary Study Using a Desktop PC

4.1 Introduction

The development of software to provide the detection of gait events using a gyroscope sensor is described in this chapter. In chapter 2 various sensors for use in drop-foot FES were described, and a gyroscope sensor was considered as part of a novel system. The sensor was first used to measure lower limb joint angles, and the hardware development for this study was described in chapter 3. This hardware was also used in this preliminary study. The software, which was designed to run on a desktop PC, was written in order to demonstrate the gyroscope’s ability to detect four gait events: heel contact (HC), foot flat (FF), heel rise (HR) and toe off (TO). Once a basic algorithm was developed with three subjects, the software was refined in order to offer gait event detection with a larger range of subjects. After the examination of 15 subjects with normal gait, data was collected from 10 hemiplegic subjects. Initially a ground reaction force (GRF) walkway was also used to determine gait events, allowing comparison with the gyroscope detection. However, since a GRF walkway is only capable of measuring HC and TO, in some cases two force sensitive resistors (FSRs) placed at the heel and toe were used instead.

4.2 Sensor Anatomical Position

The sensors were connected to a standard PC (90MHz) via an amplifier and a National Instruments data acquisition (DAQ) card (see chapter 3). Software was written using LabVIEW (National Instruments).

Alternative sensor sites for electrical drop-foot orthoses that have been considered by other researchers include the shank (measurement of inclination or acceleration, Dai et al., 1996; Willemsen et al., 1990) and the knee joint (measurement of angle, Heath et al., 1995). One reason for choosing these locations is that the sensor would be conveniently close to the electrodes site (see section 2.2.2), thereby allowing the sensor and electrode
wires to run together. With these positions the stimulator itself could also be situated nearby (instead of at the waist), thus avoiding lengthy wires. In order to determine whether the foot is on or off the ground during gait, features of the shank angle or acceleration, or of the knee joint angle have been used as indicators. These features have been shown to be concurrent with gait events (see section 2.3.2), but often do so less reliably than foot mounted sensors.

Figure 4.1 Location of Gyroscope Sensors on the Three Limb Segments of the Leg

To demonstrate greater reliability in detecting all four events using the foot mounted gyroscope sensor, three mounting positions on the leg were considered: the anterior aspect of the thigh, shank, and just above the metatarsals of the foot (as shown in figure 4.1). After consideration of the angular velocity measured at each location over 20 strides, it was decided that the foot was the best location due to the number of gait events features that were observed in the recorded signal. Figures 4.2 to 4.4 show examples of the recorded signal from each position. The graphs show the four gait events: HC, FF, HR and TO; HC and TO were measured using a GRF walkway, and two footswitches were used to determine the occurrence of FF and HR.
Figure 4.2 Graph of the Angular Velocity Measured at the Thigh During Normal Gait

Figure 4.3 Graph of the Angular Velocity Measured at the Shank During Normal Gait
Examining the 3 plots of angular velocity in figures 4.2 to 4.4, it can be seen that only the foot plot (figure 4.4) has features (zero crossings and constant near-zero periods) at the four gait events: HC, FF, HR and TO. The shank plot (figure 4.3) shows features (zero crossings) at HC and TO, and the thigh (figure 4.2) has features (also zero crossings) at HC and FF. It is clear that the foot sensor features are the most discernable, and this combined with the useful near-zero period between FF and HR (during this period there can be no doubt as to which phase of gait the walker is in) led to the decision to use the foot sensor position as shown in figure 4.5
4.3 Software Development

The software served two main purposes: to collect and process the sensor data at the required sampling frequency and so determine the occurrence of gait events, and to display the resulting data in a convenient graphical format. Processing for gait event detection was achieved in real-time, in order to assess processor speed requirements for future portable systems (see chapter 5). A flowchart for the algorithm used in the LabVIEW software is shown in figure 4.6.

4.3.1 LabVIEW Gait Events Detection Algorithm

The rule-based algorithm used to detect gait events is shown as a flow chart in figure 4.6. Also shown (for reference) is a simplified angular velocity signal measured over one gait cycle by the gyroscope sensor. The values in the boxes are voltages which are relative to the current measured sensor voltage.

At the beginning of a gait cycle (or algorithm cycle), the positive peak that occurs during swing phase is detected and all other variables are reset. The peak is the result of an anti-clockwise swing of the foot (as viewed in the sagittal plane), which is measured by the sensor as a positive voltage (the peak value was around 3V). Towards the end of the swing phase the foot swing velocity decreases and approaches zero. The zero crossing is detected next by the software, and it is during this time that the foot swing direction changes. A few milliseconds later the heel strikes the ground, and this is detected when the sensor output voltage reaches a preset negative value (0.2V).
The Detection of Gait Events Using a Vibratory Gyroscope

Figure 4.6 Flowchart of the Algorithm used in the LabVIEW Detection Software
(values in boxes are voltages relative to the current sensor output voltage)

Next, the first negative peak (the peak value was between $-2.5V$ to $-3.5V$) is detected. This occurs when the foot is rotating in a clockwise direction, and is achieved by comparing the current measured value with previous values; when the current value
becomes greater than previous values (for three consecutive samples) it is assumed that
the negative peak has occurred. At the end of the negative peak (or clockwise foot
swing) the foot swing velocity again approaches zero. It is at this time that toe contact
occurs (or foot flat), and is detected when the voltage reaches a very small value (-
0.5V). After toe contact the foot is stationary until heel rise, which causes a second
negative peak to occur (during the second clockwise rotation of the foot). The algorithm
used a value of \(-0.5\)V as a threshold to detect heel rise, after which the second negative
peak (again between \(-2.5\)V to \(-3.5\)V), was detected. Finally, a threshold of \(-0.5\)V was
used to detect the velocity approaching zero, when toe off occurs. Figure 4.7 shows the
direction of rotation of the foot during a single gait cycle.

**Figure 4.7** The Direction of Foot Rotation During One Gait Cycle
4.3.2 LabVIEW Gait Events Detection Results

Data was collected from seven able-bodied subjects (age range 24–46) over three walks. The computer also collected data from a GRF walkway, which consisted of two platforms (one for each foot) of length 3.3m, allowing three traverses of the instrumented foot to be measured. Four gait events, HC, FF, HR and TO were detected by the software and compared with the data from the appropriate platform. It should be noted that the platform could only be used to determine HC and TO. Detection using the gyroscope sensor was achieved in every walk, and an acceptable timing correlation of ±30ms was found when comparing the time between HC and TO detection, and the same time as measured by the GRF walkway. For a stimulation frequency of 40Hz (period 25ms), a delay of 30ms would lead to only one pulse being lost. Figure 4.8 shows the angular velocity, gait events detection and GRF walkway measurement (for the same foot) collected by the computer over two strides. For the detection data, high values (5V) indicate that either the heel or toe is in contact with the ground.

Figure 4.8 Graph of Angular Velocity, Vertical Force and Gait Events Detection During Normal Gait
Data was also collected from one subject who had hemiplegia (the cause of the disability was a stroke that had occurred nine months before). The subject had regained mobility and had been using single channel electrical stimulation (peroneal nerve) for 3 months. After three walks without stimulation and two with, it was found that events detection (HC, FF, HR and TO) was only possible with electrical stimulation applied (detection was then achieved in every case). Detected events were compared with those measured by two footswitches (heel and toe). This was attributed to changes in the foot movement pattern, particularly the lack of a large negative peak around the time of heel contact. If initial foot contact is not with the heel only (as is the case with normal gait) but with the entire foot, then this first negative peak will not occur. Figure 4.9 shows two strides of hemiplegic gait without stimulation. Gyroscope sensor detection times were compared to the start of rises and end of the falls of the footswitch plots. It can be seen from this graph that events measured by the footswitch data (the high values again indicate that the foot was in contact with the ground) did not coincide with those detected by the gyroscope system. In figure 4.10 the same data is displayed, but this time electrical stimulation was used. The graph shows that in this case an acceptable difference of ±15ms existed between the footswitch data and the gyroscope system.
Figure 4.9 Graph of Angular Velocity, Footswitch Data and Gait Events Detection During Hemiplegic Gait

Figure 4.10 Graph of Angular Velocity, Footswitch Data and Gait Events Detection During Hemiplegic Gait
After the trial program was developed, it was decided that although LabVIEW proved to be an excellent choice for the graphical data analysis, it did not lend itself to the kind of real-time sequential programming required for data processing. This was due to a combination of a number of factors:

- algorithm debugging and modification were found to be difficult
- concerns regarding processing speed (such as that on a Pentium 90MHz)
- the requirement to ultimately write software (using assembly language - also a text-based language) that would execute on a microcontroller IC.

4.3.3 ‘C’ Within LabVIEW – General Operation

In order to improve the speed of the detection algorithm and the ease with which the software could be modified and debugged, the gait events detection part of the software was re-written using the C (Microsoft Visual C++) platform. A simple input/output program was set up in LabVIEW to allow data to be sent from a user-friendly control panel to the C code via a Code Interface Module, which is an integral part of LabVIEW allowing the use of algorithms written in other languages. Figure 4.11 Shows the LabVIEW code used to interface to the C code, and figure 4.12 shows the user interface panel. The C code is shown in section 9.2.

In figure 4.11, the data items shown on the left hand side of the Code Interface Module are provided for the user to control the timing of gait events. The same variables also appear in figure 4.12, the user control panel, and all values are entered before the program is started. The Sampling Time is the amount of data - in seconds - that the program will collect. On pressing the ‘button’ Measure Offset, the program will measure the gyroscope offset one hundred times, take the average and write that value to a file. The stored value is then subtracted from any subsequent sensor measurements during a particular session, in order to remove the offset. The other button, Data Source: Gyro/File, selects the location of the data to be processed. When this button is pressed, data is taken from a file containing previously stored data, instead of sampling the sensors connected to the computer.
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Figure 4.11 LabVIEW Code Allowing Data to be Passed to and From The C Code Interface Module

Figure 4.12 The User Interface Panel

The six numerical control windows shown on the right hand-side of figure 4.12 are used to control the timing of events detection. HC, FF, HR and TO. The values indicate the
delay (from the onset of detection) the assertion of each phase of Stance Phase. Heel On and Toe On select the minimum time between HC and HR or FF and TO respectively. This is useful to adjust stimulation timing for each individual, due to minor differences in gait styles. A HC delay of 10ms was used with two subjects to achieve the minimum difference in timing compared with the GRF walkway. This could be used if the subject's foot moved between FF and HR.

On the right hand-side of figure 4.10, seven variables are shown emerging from the Code Interface Module. These are: the gyroscope sensor signal, the heel and toe contact detection signals, data from two footswitches (if connected) and data from two GRF platforms (again, if connected). The data is then passed to a global variable called Global Data, which is used by an analysis subroutine (also written in LabVIEW) that is opened automatically once the sampling time has expired (or all the data has been read from file). The variable named “Measure Offset” can be seen in the top left corner of figure 4.11, and is shown connected not only to the Code Interface Module, but to the analysis subroutine, labeled Open Anal. This routine produces a graph of all the data, and provides other functions, such as a zoom in/out function and the facility to turn individual plots on and off. The data can also be saved to file from this routine.

The gyroscope sensor is sampled from within the C routine, at a frequency of 100 samples per second (as before). The collected (unprocessed) data is also passed directly to the LabVIEW program for graphical analysis. Data from two footswitches (and a GRF walkway if required) is also collected and passed directly to the graphical analysis program. After each sample is taken from all the connected devices (gyroscope sensor, footswitches etc.), the program enters the detection algorithm and the determination of an event is based upon current and previous angular velocities.

4.3.4 ‘C’ Within LabVIEW – Gait Events Detection Algorithm

Initially, the detection algorithm was written to operate in an identical manner to that of the LabVIEW program (see figure 4.6). The algorithm then underwent a number of refinements which led to a final version, which was then re-written in assembly
language for use with the data-logger (see chapter 5). Development was facilitated by
the fact that the C program was easier to read (given that the program operated
sequentially), and also the program’s ability to re-process data that had been collected
previously (after a particular algorithm modification had been applied).

The detection algorithm was refined to improve its robustness during use with
pathological gait. Simple inspection of foot movement (i.e. determining whether the
foot was moving or stationary) ensured that detection was switched on or off should the
algorithm fail, or initial foot contact was made with the entire foot instead of the heel
only. Also, a second method of determining midstance (by simply detecting zero foot
movement after HC has occurred) ensured that reaching this point did not depend upon
the detection of FF. Upon returning to the swing phase, both the heel and toe are set to
‘off’ (i.e. off the ground) should HR or TO detection fail.

The general operation of the ‘C’ program is shown in figure 4.13, and the detection
algorithm is represented by the ‘Determine Gait Events’ process. It can be seen from
figure 4.13 that this process is performed repeatedly until every data item has been
processed. As with the LabVIEW algorithm, the measured angular velocity plot for one
gait cycle (see figure 4.6) is broken up into a number of features, such as the first
positive peak, the first zero crossing (positive to negative), the first negative peak etc.
Upon entering the algorithm, a program position is determined (either an existing one or
a new one) which corresponds with a particular feature.

Although the user will in most cases choose to measure the sensor offset before using
the system, this is not necessary if the measurement has been performed within a few
hours. The sensor has not been found to have produced a detectable offset in a period of
five hours.
Figure 4.13 General Operation of the ‘C’ Data Collection and Processing Routine
The refined gait events algorithm is depicted in more detail in figures 4.14a and 4.14b. The first time the program enters the algorithm, all variables are initialised (set to zero). Use of global variables (i.e. variables that are declared in the main program, not in the function itself) prevents values from being reset each time the program returns to the main program. A series of IF statements regarding the current measured angular velocity then follows. The first IF statement examines whether the angular velocity is greater than $-0.1V$; if this condition is found to be TRUE then Toe Off is detected. The second IF statement operates in a similar way: if the angular velocity is less than $-0.2V$ then HR is detected. The next three IF statements detect Midstance, FF and HC (in that order). Finally, the last IF statement is concerned with the detection of swing phase. When the order of events (TO, HR, Midstance, FF, HC and swing phase) is considered, it is apparent that the sequence of events is chronologically in reverse. However, detection actually occurs in the correct order because each event is dependent upon the previous (chronologically) event having been detected. Thus, detection of TO is not possible without first having detected HR, which is then dependant upon Midstance being detected, and so on.

This method of sequentially detecting events that have been listed in reverse prevents the detection of two events during one pass. For example, if HR was listed in the program before TO, detection of HR would be immediately followed by TO detection. Another feature incorporated into the algorithm is the resetting of variables only upon the detection of swing phase (i.e. once every gait cycle). This avoids a second detection of any event, e.g. HC would only occur once after the swing phase, and would be prevented from occurring again until after the next swing phase.
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Figure 4.14a General Operation of the ‘C’ Events Detection Algorithm
Figure 4.14b General Operation of the 'C' Events Detection Algorithm
On the right hand-side of figure 4.12 the control parameters for detection timing delay can be seen (see section 4.3.3). These are used to either delay the detection of an event (HC, FF, HR and TO), or to set a minimum time that either the toe or heel are said to be in contact with the ground. These two features are achieved by adding a count function to each IF statement. If a particular condition is found to be true, a count variable is incremented instead of detecting an event. On successive returns to the algorithm, the same conditions provoke further incrementations of that particular count variable. The number that each count must reach before an event is detected is set in one of the windows shown in figure 4.12; by default these are set to zero, which effectively causes the event to be detected immediately after the IF statement is executed.

4.4 Results (‘C’ Within LabVIEW)

The software’s ability to re-process data after it has been processed facilitated algorithm refinement. After the initial C program was developed, data was collected from seven stroke subjects, one MS and two incomplete SCI subjects. Subjects performed two walks without stimulation followed by two walks with. With the data from these walks stored on computer, it was possible to examine the effects of any modifications, until an optimum algorithm was reached (i.e. an algorithm that detected all events correctly with each saved file). No adjustment of the control parameters were made (they were all left at zero).

Table 4.1 shows the maximum timing difference between the gyroscope sensor and heel and toe footswitches. Footswitches were used for validation since no other method was available, although it is of course true that these sensors cannot be considered to be an accurate method of measuring gait events. All subjects used single channel stimulation of the peroneal nerve except subject A, who used a second channel to stimulate the hamstrings muscles.
<table>
<thead>
<tr>
<th>Subject</th>
<th>Age</th>
<th>Pathology</th>
<th>Maximum Difference</th>
<th>Comments</th>
</tr>
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<tr>
<td>A</td>
<td>58</td>
<td>MS</td>
<td>30ms</td>
<td>MS: 25 Months Using FES 6 months</td>
</tr>
<tr>
<td>B</td>
<td>55</td>
<td>Stroke</td>
<td>80ms</td>
<td>Stroke: 12 Months Using FES 3 months</td>
</tr>
<tr>
<td>C</td>
<td>71</td>
<td>Stroke</td>
<td>30ms</td>
<td>Stroke: 14 months Using FES 6 months</td>
</tr>
<tr>
<td>D</td>
<td>53</td>
<td>Incomplete SCI</td>
<td>-</td>
<td>SCI: 10 months Using FES 3 months</td>
</tr>
<tr>
<td>E</td>
<td>73</td>
<td>Stroke</td>
<td>40ms</td>
<td>Stroke: 11 months Using FES 3 months</td>
</tr>
<tr>
<td>F</td>
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<td>80ms</td>
<td>Stroke: 15 months Using FES 6 months</td>
</tr>
<tr>
<td>G</td>
<td>42</td>
<td>Incomplete SCI</td>
<td>-</td>
<td>SCI: 4 years Using FES 6 months</td>
</tr>
<tr>
<td>H</td>
<td>69</td>
<td>Stroke</td>
<td>100ms</td>
<td>Stroke: 14 months Using FES 6 months</td>
</tr>
<tr>
<td>I</td>
<td>57</td>
<td>Stroke</td>
<td>90ms</td>
<td>Stroke: 15 months Using FES 3 months</td>
</tr>
<tr>
<td>J</td>
<td>66</td>
<td>Stroke</td>
<td>40ms</td>
<td>Stroke: 11 months Using FES 3 months</td>
</tr>
</tbody>
</table>

Table 4.1 Table of Maximum Timing Differences (all events) between the Gyroscope Sensor and Heel and Toe Footswitches (subjects with pathological gait)

Figure 4.15 shows a graph of angular velocity, heel and toe footswitches and gait events detection for subject ‘A’ during walking without stimulation. As before, the gyroscope...
sensor voltage data is in proportion with the measured angular velocity, and high values of detection data (5V) indicate that either the heel or toe is in contact with the ground. High values of footswitch data also indicate that the foot is in contact with the ground.

Figure 4.15 Graph of Angular Velocity, Heel and Toe Footswitches and Gait Events Detection During Hemiplegic Gait (A) No FES (x is Time [10xms], y is Volts [V])

Footswitch and heel events detection data from the graph in figure 4.15 are also shown in figure 4.16; a similar graph for toe data is shown in figure 4.17. It can be seen from the graphs that there is an acceptable difference between events measured by the two types of sensor (over all events, the greatest difference was 100ms, and the smallest was 15ms). In figure 4.18 the same subject is walking but this time using dual channel stimulation.
Figure 4.16 Graph of Heel Footswitch and Heel Gait Events Detection During Hemiplegic Gait (A) No FES (x is Time [10xms], y is Volts [V])

Figure 4.17 Graph of Toe Footswitch and Toe Gait Events Detection During Hemiplegic Gait (A) No FES (x is Time [10xms], y is Volts [V])
Differences between walking with and without FES are apparent when examining the angular velocity data from the gyroscope sensor. For comparison, the gyroscope sensor data collected from subject ‘A’ during walking without and with FES are shown in figures 4.19 and 4.20 respectively. The most striking difference between the two graphs is that the positive peak (the swing phase) is more pronounced in the walk with FES, i.e. the peak is of greater amplitude and the area bound by the curve is greater. This may be interpreted as an improved swing phase, since it is closer to what is observed in normal gait (see figure 4.4).

Another difference between the two graphs is the maximum (normalised) peak amplitude and duration of the first negative peak, which occurs immediately after the swing phase. This peak is caused by the clockwise rotation of the foot between HC and FF, and is an indication of the quality of the heel strike. At first strike, contact may have been solely with the heel (as in normal gait); it may have been initially with the heel but immediately afterwards with the remainder of the foot (as in footslap), or finally initial
contact with the entire foot may have occurred (no HC). In these cases it can be seen that the maximum (normalised) peak amplitude is greater in the ‘without FES’ graph (figure 4.19), indicating that the foot impact with the ground is higher (in most cases a high amplitude, high frequency positive peak also occurs immediately after the negative wave, indicating relatively high shock).

There is insufficient information regarding whether heel only or whole foot contact occurred, although from the data it may be stated with some confidence that, if the heel did make contact first, then foot flat occurred directly afterwards, i.e. footslap occurred. From figure 4.20 (the ‘with FES’ graph) it is clear from the negative peak (of longer duration compared to that of the ‘no FES graph’) immediately after HC that some clockwise rotation of the foot occurred after HC, an indication of a more normal gait. Gyroscope event detection in figures 4.15 and 4.18 (walking without and with FES respectively) show that there is a greater delay between HC and FF when FES is used.

**Figure 4.19** Graph of Angular Velocity During Hemiplegic Gait (A) No FES (x is Time [10xms], y is Volts [V])
There is little difference between the two graphs when comparing the second negative peak, which is caused by the clockwise rotation of the foot between HR and TO. FES appears to have made no (or little) difference to the subject’s ability to push off (this concurs with the fact that stimulation of the gastrocnemius was not used in this case).

In figure 4.20 the heel footswitch data is of low amplitude (this most likely to be due to a combination of the footswitch position and the subject’s gait). In figures 4.21 and 4.22 gait events and footswitch data (with no FES) are shown for subjects ‘C’ and ‘J’ respectively.

It can be seen from figures 4.19 and 4.22 that the angular velocity collected from different patients varies considerably in terms of amplitude and shape. However, the basic pattern is still similar to that shown in figure 4.6.
Figure 4.21 Graph of Heel and Toe Footswitches and Gait Events Detection During Hemiplegic Gait (C) No FES (x is Time [10xms], y is Volts [V])

Figure 4.22 Graph of Angular Velocity, Heel and Toe Footswitches and Gait Events Detection During Hemiplegic Gait (J) No FES (x is Time [10xms], y is Volts [V])
4.5 Incomplete SCI Injured patients

The data collected from the two incomplete SCI subjects presented a different set of results. Due to clonus muscle activity, relatively high frequency oscillations were observed in the angular velocity measurements (these oscillations were present in when walking both with and without FES). These oscillations were unexpected and proved difficult to accommodate within the detection system. Figure 4.23 shows a graph of angular velocity and gait events detection measured while subject ‘D’ walked (the incomplete SCI was due to a left frontal parietal intracerebral haemotoma brought about by a head injury).

In figure 4.23 angular velocity oscillations are shown to be occurring during (and possibly after) swing phase. Although in this case the rapid movement during swing did not appear to cause any problems, detection of HR and TO occurred too early due to foot movement between FF and HR (data from footswitches were not collected from these subjects).

![Figure 4.23 Graph of Angular Velocity and Gait Events Detection During Hemiplegic Gait (D) No FES (x is Time [10xms], y is Volts [V])](image-url)
The clonus activity is particularly visible during the third stride (the third positive peak from the left) in figure 4.23. Figure 4.24 shows a similar graph for subject ‘G’ (whose injury was the result of a tumour removal from the L4 region of the spinal cord); it can be seen that during the second phase of gait the detection algorithm did not operate correctly. Indeed, it is unlikely that swing phase ended as soon as the heel detection would suggest (plots of the heel and toe switches indicate that this is the case), and so it was decided that the algorithm would require more extensive modification if the system were to cater for patients with clonus. Patients with clonus were not considered further in this study due to the amount of time that would be required to develop suitable software.

Figure 4.24 Graph of Angular Velocity and Gait Events Detection During Hemiplegic Gait (G) No FES (x is Time [10xms], y is Volts [V])
4.6 Conclusions

Gait events had been detected by the gyroscope sensor and compared with either a GRF walkway or heel and toe footswitches. Results for normal subjects showed that the time differences fell within acceptable limits: maximum difference: 30ms. For pathological gait, the maximum time difference of 100ms may not be acceptable, if the footswitches were considered to be an accurate means of comparison. However, since this was not the case (a more accurate method for measurement of gait events was now required), it remained uncertain as to whether a true maximum stimulation timing delay of 100ms existed. It is possible that footswitch-triggered stimulation occurs too early or too late, but that is acceptable to the individual using the FES system.

A more extensive study was now required involving different terrains, walking at different speeds and greater numbers of walks. Also, an alternative method of determining gait events was necessary (particularly as the GRF walkway is only capable of determining HC and TO) for validation purposes.

The data collected during incomplete SCI gait showed that the current system was not suitable for subjects with this pathology. However, with extensive software modification it is likely that the system could be used with such subjects. Although low-pass filtering is the obvious choice for the higher frequency phases, large processing delays might occur, which would be unacceptable in a real time system. Also, since the next stage of design was to involve a simple microcontroller (within a portable system), the limited processing ability meant that it would have been difficult or impossible to implement such a filter. It was therefore decided that subjects with clonus would not be considered further in this study. However, adaptation of the software to allow inclusion of this group was by no means considered to be impossible.

In order to evaluate the gyroscope sensor more extensively, longer walks involving different terrains were required. If the system were to be used to control the timing of FES, suitable switching outputs would be required, and in the long term long leads attached to stationary computer equipment were obviously impractical. Also, as observed in chapter 3, it is possible that the lead attaching the sensors (and the subject)
to the desktop computer would have had an effect on the subject's gait. The lead would also obscure any reflective markers mounted on the body if the system were to be compared with a marker detection system. It was therefore necessary at this stage to design a portable system that would be capable of performing all of the functions carried out by the desktop computer. The design, construction and evaluation of such a system is discussed in the following chapters.
Chapter 5
The Development of a Portable Gait Events Detection System with Data-Logging Facilities

5.0 Introduction

For reasons discussed at the end of chapter 4, the next stage of development was to design and build a small portable computer system. The new system was designed to execute similar software as before, and also store collected data on a transferable card for uploading onto a PC. Since a more extensive evaluation was required in order to conclude this study, the portable unit provided a means to less restricted data collection.

5.1 System Requirements

The purpose of the new system was to provide a portable means of executing detection software identical to that described in chapter 4, along with the facility to store data that could be later transported onto a PC. By portable it is meant that unit should be small and light enough to be worn by patients. The data was to be collected from the gyroscope sensor and two footswitches, and this would be stored along with the heel and toe contact detection data. A version of the gait events detection software was to run on this system.

In order to sample from the sensor and footswitches, at least one analogue to digital convertor (ADC) was required. A wire link was necessary to transfer timing control parameters (six numbers - discussed in section 4.3.3) from a PC to the unit. This would allow clinical staff to make adjustments whilst the unit is attached to a patient. Table 5.1 shows the final specifications for the portable unit.
**Table 5.1** Table of Final Specifications for the Portable Data-Logging Unit

Examination of portable data-loggers currently for sale on the market revealed that although many were capable of collecting data from several channels at 100Hz or more, none could also execute custom-made software for real time data processing. Below are some examples of commercially available portable dataloggers.
1) XR440 Pocket Data Logger, Pace Scientific Inc., 542-6 Williamson Road, Mooresville, NC 28117 USA. $499 Standard Model
   - Each input will read resistance, 0-5V dc or a contact closure.
   - Adjustable resolution of 12, 10 or 8 bits (lowering resolution increases reading storage capacity, see specifications).
   - 1-200Hz sampling rate
   - Communicates with PC via RS232 connection
   - Stores 129,024 samples (4.3 minutes at 500 samples/sec)

2) MA2290-8 Universal Portable Data Logger, Clark 10 Brent Drive Hudson, MA 01749
   - 5 Input Channels
   - 100,000 Measured Value Memory (520Kb)
   - A/D Converter: 16-bit Multi-Slope Integrating
   - Measurement Rate: 2.5 or 10 Measurements/Second
   - Communicates with PC via RS232 connection

3) P-DAS8, Midé Technology Corporation 200 Boston Ave Suite 2500 Medford, MA 02155
   - 8 or 1 channels
   - 1 kHz (8 channels) or 8 kHz (1 channel) sample rate
   - 16 bit resolution
   - Uses Compact Flash memory card

Two of these dataloggers have features that are required for this study, such as an acceptable sample rate (at least 100Hz), adequate resolution (at least 8 bits providing 256 levels) and multiple sensor input (3 sensors required). Device number 2 sampled at a maximum rate of only 10Hz (and had only 520kb of memory) and was therefore clearly unsuitable. Device one could collect only a little over 4 minute’s worth of data at 100Hz and 5 channels, and used a relatively slow RS22 link to transmit data to a PC.
Only device 3, with 8 channels sampled at 1kHz and a compact flash memory card could therefore be considered viable, although 40 minutes of data would require a card of over 100Mb (all 8 channels are sampled at the high 1kHz).

There is one other limitation that stands in the way of all of these cards being truly viable for this study, however. Not one datalogger on the market appears to be capable of running real-time data processing software - another important requirement. Although this would not be significantly difficult for manufacturers to produce, it is clearly not in great demand and would be highly expensive if one were to be ordered for this study. It is for this reason that the decision was taken to build an 'in-house' datalogger.

5.2 Hardware

In figure 5.1 a basic block diagram of the system is shown (a component diagram and a circuit diagram is shown in figures 5.2 and 5.3 respectively). The processing unit is a PIC16C73A (Arizona Microchip Technology Ltd) microcontroller. The IC can be operated at 4MHz; this was considered to be fast enough to process the detection algorithm and maintain a sampling frequency of at least 100Hz (the highest possible rate of 164Hz was used), and contains a large amount of program memory space (192 bytes). It contains five ADCs and may be connected directly to a RS232 line driver IC via an on-board synchronous serial bus. Three of the five ADCs are used to sample the gyroscope sensor and two footswitches, avoiding the need for a multiplexer (required if only one ADC was available). The IC was considered to be a good compromise between physical size and number of pins, and the low power consumption of the unit was also a consideration.

The weight of the unit was also an important consideration, and it therefore decided that a single 9V PP3 battery was to be used as this was the heaviest component. Low-power LEDs are used in order to keep current consumption low, as the battery was required to have a life of at least one hour. An RS232 line driver IC with a 'power down' low-current mode is also used.
The Development of a Portable Gait Events Detection System

The unit provides two sets of output: five channels of data to be written to a transferable RAM card, and two digital outputs for controlling two stimulator channels. Two RAM options were available: the PCMCIA card and the smaller and lighter Smart Media card (card dimensions are 80x52x5.5mm and 45x37x0.76mm respectively). A PCMCIA (2Mb) card was chosen because technical data regarding wiring and operating this device was more readily available.

A MAX667 (Maxim Integrated Products, Inc.) voltage regulator supplies +5V from the +9V battery at a maximum current of 250mA. Several components also require a -5V supply, and this is generated using a MAX7660 (Also Maxim) voltage converter. A green LED indicates that the unit is on.

As well as the central processing unit (master PIC), a second PIC16C73A was used as a 21 bit counter (Slave PIC). The purpose of the counter was to sequentially address the 2MB memory card. Although this was a very under-demanding role for such a microcontroller, a 21 bit counter was not available commercially. The Slave PIC receives a count pulse generated by the master PIC and sent from pin 2 (A0 – this pulse also served as A0 for the memory addressing). The only other control line from the master PIC (pin 6) is the Slave PIC reset, that allowed the master PIC to control Slave PIC counting and reset it at any time. The clock circuitry for each PIC was based around a 3.579MHz crystal oscillator (this frequency was used with the master PIC to enable it to use an RS232 baud rate of 9600). A reset circuit was also included for the master PIC.
Figure 5.1 Basic Block Diagram of the Portable Data-Logging Unit

After the digitisation of the gyroscope sensor signal by the master PIC (for processing), a digital to analogue (DAC) was available to provide analogue signal feedback. The DAC, an AD7528 (Analog Devices) also provided a means of performing a system test (the digitised signal from the gyroscope sensor or either footswitch could be converted back to an analogue signal and examined on an oscilloscope).

No changes were made to the circuit that amplifies the gyroscope sensor signal. As used previously in the desktop PC system (see chapter 3), the INA101HP (Burr-Brown) instrumentation amplifier IC was used, this time to provide a gain of 6.88. The signal was filtered via a low pass filter, consisting of a simple one pole RC filter with a 3dB
point at 31Hz. The signal produced by the DAC was buffered using an operational amplifier on an LM342N (National Semiconductor) IC; another amplifier from the same IC was used to add the gyroscope sensor signal to an offset voltage (using a summing amplifier with a gain of 1). When the unit was set to calibrate mode, a suitable offset was generated which placed the sensor input at the ADC to 2.5V (this centred the input between the minimum and maximum ADC input values (0 to +5V). Figure 5.2 shows a component diagram of the circuit, and in figure 5.3 the actual circuit can be seen. In table 5.2 a list of components and the cost of each is shown.

A MAX205 RS232 line driver IC was used for serial communication between the unit and a PC. A control line from the master PIC (pin 11) was used to power-down the IC, placing it into a low current state (less than 1μA).

The unit was capable of driving a stimulator via two control lines that run from the master PIC (pins 12 and 16) to an external socket, each via a transistor which served as a buffer.

Port B of the master PIC is used as an 8 bit data bus, and writes data to the DAC and the memory card.
Figure 5.2 Component Diagram of the Portable Gait Detection/Data-Logging Unit
Figure 5.3 Circuit Diagram of the Portable Gait Detection/Data-Logging Unit
## The Development of a Portable Gait Events Detection System

<table>
<thead>
<tr>
<th>Component</th>
<th>Type</th>
<th>Quantity</th>
<th>Price (Each)</th>
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<td>PIC</td>
<td>PIC16C73</td>
<td>2</td>
<td>£12.68</td>
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<tr>
<td>A to D Convertor</td>
<td>AD7528</td>
<td>1</td>
<td>£6.88</td>
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<td>Instrumentation Amp.</td>
<td>INA101HP</td>
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<tr>
<td>Op. Amp.</td>
<td>LM324N</td>
<td>1</td>
<td>£2.82</td>
</tr>
<tr>
<td>5V Regulator</td>
<td>MAX667</td>
<td>1</td>
<td>£3.28</td>
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<tr>
<td>5V Invertor</td>
<td>MAX7660</td>
<td>1</td>
<td>£2.05</td>
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<tr>
<td>RS232 Line Driver</td>
<td>MAX205</td>
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<td>£8.83</td>
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<td>Crystal</td>
<td>3.579MHz</td>
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<td>£5.49</td>
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<td>Transistor</td>
<td>NPN</td>
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<td></td>
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<td>4p</td>
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<td>10K</td>
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<td>4p</td>
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<td></td>
<td>100K</td>
<td>6</td>
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<td>Capacitor</td>
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<td>1µF</td>
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<td>1</td>
<td>46p</td>
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<td></td>
<td>22pF</td>
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<td>9p</td>
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<td>PCMCIA Card Reader</td>
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<td>SPST Switch</td>
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<td></td>
<td>24 Pin DIL</td>
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<td>28p</td>
</tr>
<tr>
<td></td>
<td>20 Pin DIL</td>
<td>1</td>
<td>28p</td>
</tr>
<tr>
<td></td>
<td>14 Pin DIL</td>
<td>2</td>
<td>28p</td>
</tr>
<tr>
<td></td>
<td>8 Pin DIL</td>
<td>2</td>
<td>28p</td>
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<td>Socket</td>
<td>3.5mm 2 Way</td>
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<td>3 Pin VERO Connector</td>
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<td>5 Pin VERO Connector</td>
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<tr>
<td></td>
<td>PCMCIA Connector</td>
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<td>£3.77</td>
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<tr>
<td>Extras</td>
<td>Cables/Other Connectors</td>
<td></td>
<td>£10</td>
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</tbody>
</table>

Table 5.2 Components used in the Data Logger
5.2.1 Control Switches

The unit has a reset push-to-make switch that not only resets the master PIC, but actually serves as a master reset for the entire unit. There is also a latching pause button which temporarily suspends data collection and allows another operation mode to be selected. Another push-to-make switch is used to start data collection; it is also used to select Calibrate mode if the unit is paused. Along with the data collection and calibration modes, there is also a Serial Communications mode. This mode is selected from Pause mode, by switching the rocker switch (this also disables the pause and start buttons and instead connects the transmit and receive lines from the MAX205 IC to the master PIC). The unit returns to pause mode on completion of calibration or serial communications (the rocker switch must also be returned to its original position). Table 5.3 shows how each mode is selected, and also indicated by the red status LED.

<table>
<thead>
<tr>
<th>Mode</th>
<th>Red LED</th>
<th>Push Switch (Black)</th>
<th>Push Switch (Red)</th>
<th>Latching Switch</th>
<th>Rocker Switch</th>
</tr>
</thead>
<tbody>
<tr>
<td>Reset</td>
<td></td>
<td>Pushed</td>
<td>Open</td>
<td>Down</td>
<td></td>
</tr>
<tr>
<td>Data Collection (Normal)</td>
<td>On</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Pause</td>
<td>Flash</td>
<td></td>
<td></td>
<td>Closed</td>
<td>Down</td>
</tr>
<tr>
<td>Change Mode (from Pause Mode)</td>
<td>Flash (Short On, Long Off)</td>
<td>Pushed</td>
<td>Open</td>
<td>Down</td>
<td></td>
</tr>
<tr>
<td>Calibrate (from Change Mode)</td>
<td>On (Momentarily, then return to Pause Mode)</td>
<td>Pushed</td>
<td>Closed</td>
<td>Down</td>
<td></td>
</tr>
<tr>
<td>Serial Communications (from Change Mode)</td>
<td>On</td>
<td></td>
<td></td>
<td></td>
<td>Up</td>
</tr>
<tr>
<td>Memory Card Full</td>
<td>Fast Flash Rate</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

Table 5.3 Mode Selection and Indication
The red status LED indicates which mode the unit is currently in, either by being on or off, or by flashing. When the unit is first switched on the LED is off until the data collection mode is entered. The LED will remain on until the memory card is full, at which time the indicator will flash at a relatively fast rate. If the pause button is pressed at any time the LED will flash at a very slow rate until the button is pressed again.

### 5.2.2 The PCMCIA Memory Card

The memory card has 2MB of RAM and has its own battery power supply to avoid memory loss when the card is removed from the reader/writer. The card is a Personal Computer Memory Card International Association (PCMCIA) card, and a suitable socket is provided within the gait detection/data-logging unit. After data has been written to the card, a PC card reader/writer may be used (along with appropriate software) to transfer the data onto a PC. A Card Genie (CardWise, Caversham, Berks, UK) PCMCIA card reader was used for this purpose, and this connects to a standard PC via the parallel port.

Along with the previously mentioned 8 bit address bus and 21 bit address bus, several control lines are required to allow the master PIC to control the memory card in order to write data to it. The three lines used are: Card Enable 1 (CE1), Write Enable (WE) and Output Enable (OE) – section 5.3.2.5 discusses these in more detail.

The software used to read data from the PCMCIA card and convert it into a form suitable for Excel (Microsoft) was written using C (Microsoft Visual C++). The program reads the contents of the memory card, and writes this to a file to be stored on the PC's hard disk. While the data is being written, the control characters `\n' and `\t' are inserted into the data at appropriate times in order to allow Microsoft Excel to read the file (see section 9.1). For simplicity, the existing master PIC software writes only to the first byte at each memory location on the card (the full 2Mb was not required at this stage) and so the card reader software reads only every other byte.
5.3 Software

There are two programs used by the gait detection/data-logging unit, one executing on the master PIC and one on the Slave PIC. The PIC software is coded in assembly language, and the program on the master PIC may be sub-divided into two parts: the main program and the events detection routine. One other piece of software, written in ‘C’, allows data to be read from the memory card to the PC. This program also re-writes the data into a suitable form (text) for MS Excel.

5.3.1 Slave PIC Software

The Slave PIC is used only as a 21 bit counter, which sequentially addresses the memory card. Once variables have been initialised and registers set, the program enters a ‘do nothing’ loop whilst it waits for the first interrupt. Upon entering the interrupt service routine (ISR), a clock pulse (generated by the master PIC) is sampled on pin 7. When a rising edge is detected, the counter is updated each time until all 21 bits have been set (the master PIC program also contains a counter which determines when this condition has been reached – see section 5.3.1). When the ISR is finished, the program returns to the ‘do nothing’ loop until the next interrupt occurs. Figure 5.4 shows a flowchart of the Slave PIC program.

In order to detect rising edges of the count pulse, the Slave PIC program stores both the current and the previous sample values. If, upon entering the interrupt handling routine, the previous value is found to be clear and the current value set, then a rising edge is detected and the program will increment the address bus count.
Figure 5.4a Slave PIC Software
Figure 5.4b Slave PIC Software
5.3.2 Master PIC Software

Despite being physically identical to the Slave PIC, the master PIC's role within the unit is far more significant and therefore complex. It is the program executing within this PIC that determines the function of the entire unit at any time; when the user presses the master reset button the actual consequence is that this program is reset.

5.3.2.1 Overview

The program consists of a number of subroutines, the main (and largest) subroutine being the ADC ISR that in turn calls a number of smaller ones. As with the Slave PIC software, the program first enters a 'do nothing' loop whilst it waits for the first interrupt. Any interrupt will cause the program to enter the ISR routine, but if it is found that it was not an ADC interrupt that caused this, the program will immediately return to the loop. When an ADC interrupt occurs the ADC ISR is executed, during which time one of the external sensors is sampled. Each time the ISR executes, one of the three sensors - the gyroscope sensor and the two footswitches - is sampled in turn. If it is the turn of the gyroscope sensor then the gait events detection algorithm is executed (see sections 4.3.1 and 4.3.4).

When all 3 channels have been sampled and the detection algorithm has processed the gyroscope sample, all current samples and heel/toe detection data are written to the PCMCIA memory card. Before this is done a control signal increments the count on the slave PIC (used as a counter) to ensure that current data are written at the next memory card address. Although each card address holds one word, only the first byte is written to for simplicity. This effectively reduces the memory available to 1Mb, but this was deemed to be initially acceptable.

If the communications mode is selected during ADC interrupt handling, then a flag will be set which, upon returning to the 'do nothing' loop, will disable ADC interrupts and enable serial communication interrupts. When the first interrupt occurs, the program will enter another ISR to communicate with the host PC. At the end of this ISR the
program will disable serial communication interrupts and re-enable ADC interrupts again. Figure 5.5 shows a flowchart of the master PIC Software (the main program).

Figure 5.5a Overview of Master PIC Software (Main)
Figure 5.5b Overview of Master PIC Software (Main)
5.3.2.2 Pause and CheckButtons Subroutines

The Pause subroutine is called whenever the unit is in data collection mode and the Pause button is pressed. The Pause button is a latching switch, i.e. if pushed once it will latch closed, and a second push will release the latch and the switch will be open again.
The status of the switch is inspected every time the ADC interrupt routine is executed; if it is found to be closed, the Pause subroutine is executed.

The purpose of the Pause subroutine is to suspend data collection, and to allow the user to access the Calibrate and Serial Communications modes. In order to do this, the routine itself calls another subroutine called CheckButtons to determine whether either the Start button has been pressed (to select a ‘change mode’ condition), or the pause button has been unlatched (to exit pause mode). If the start button has been pressed, then a flag (the ‘ChangeEvent’ flag) is set which, upon returning to the Pause subroutine, will determine the desired mode from the position of the toggle switch. If the toggle switch position is changed (within 10 seconds of returning to the Pause subroutine), then the unit will go into Serial Communications mode (changing the state of the switch also connects the master PIC to the RS232 line driver IC – the same pins are otherwise connected to the Start and Pause switches). If there is no change from the usual position of this switch, the unit will simply wait for 10 seconds and enter Calibration mode. During the ten seconds that the unit is waiting in the ‘change mode’ state, the LED will flash at a faster rate to indicate that the unit is waiting.

Figures 5.6 shows the Pause subroutine, and CheckButtons subroutine is shown in figure 5.7.
Figure 5.6 The *Pause* Subroutine
Figure 5.7 The CheckButtons Subroutine
5.3.2.3 Serial Communications Mode

Figure 5.8 shows a flowchart of the subroutine responsible for setting serial communications on the master PIC. The routine is necessary for selecting the required operating parameters for serial communications between the system and a standard PC: enabling serial communications interrupts and requesting a 9600 baud data transfer rate. The routine cannot be executed until the ADC interrupts are first disabled; in figure 5.6 this is shown to occur directly after the toggle switch position is changed.

In figure 5.9 a flowchart of the serial communications routine is shown. After the timing control parameters (six numbers) are sent from the PC, the unit sends them back to the PC for error checking (the sent values are compared with the original numbers).

![Flowchart of Serial Communications Set Up Subroutine](image)

Figure 5.8 The Serial Communications Set Up Subroutine
The Development of a Portable Gait Events Detection System

Figure 5.9a The Serial Communications Subroutine
Figure 5.9b The Serial Communications Subroutine
5.3.2.4 Calibration Mode

Along with serial communications, Calibration mode may be selected when the unit is in Pause mode. Calibrate mode should be selected by the user before any data is collected, as the purpose of this mode is to determine the additional offset voltage required by the gyroscope sensor in order for data to be collected properly. Since the ADC measures between 0 and 5V only, any value that is less than 0V (the amplified gyroscope signal varies between –2V and 2V) will be ignored. Any voltage generated by the DAC may be added to the gyroscope signal, and in doing so effectively increase the sensor output range to 0.5 to 4.5V.

The DAC value required to raise the gyroscope sensor signal to a suitable range for the ADC is found using the Calibration subroutine shown in figure 5.10. A measured value of 2.5V would be represented by the DAC as 128, since it has 256 (0 – 255) levels of quantisation. An additional offset value of 2.5V would bring the sensor output into the correct range for the ADC, but may not place the output at exactly zero when the sensor is not in motion (i.e. at zero °/second). This is due to the small offset voltage (±100mV) that the sensor has at its output, which only varies significantly (see section 3.2) over time (several hours) and with temperature (for each °C above or below 32.5°C the change in output will be ±0.31%). If the sensor input to the ADC is not ‘zero’ (or 2.5V) then the detection algorithm will not operate correctly, as it uses values (such as 128 for zero) to determine whether the angular velocity has crossed a threshold.

When the unit is first switched on, it is possible that the sensor’s own offset would have changed from its previous value, measured during the last time the unit was in operation. The new offset value must therefore be determined and eliminated before the detection algorithm is executed.
The Development of a Portable Gait Events Detection System

Figure 5.10 The *Calibrate* Subroutine
5.3.2.5 The Write Subroutine

After each item of data is either collected from external sensors or derived from the detection algorithm, that item is then written into the PCMCIA card for storage and transfer to a PC. Three data control lines, Card Enable 1 (CE1), Write Enable (WE) and Output Enable (OE) are required to control the memory card for reading and writing operations. Although reading is not actually required during normal unit operations, it is nevertheless useful to have this facility when executing test programs.

Of the three data control lines, only the CE line is required to change state during write operations. In order to write data to the card, the line must first be held low for at least 140ns. At all other times the line should be in a high state. WE and OE are held permanently low and high respectively. Should a read operation be required, all three lines must be held low throughout.

In figure 5.11 a flowchart of the write operation subroutine can be seen. Each delay shown is generated using the same subroutine, which simply performs several count operations taking in total approximately 10ms. Although the delay is larger than necessary, it is also used elsewhere in the program and there was no requirement to provide shorter delays.
Figure 5.11 The Write Subroutine
5.4 Summary

The design of a portable gait events detection system with data-logging facilities has been discussed in this chapter. The device collects data from the gyroscope sensor and one heel and one toe footswitch, and real-time processing is performed on the gyroscope signal to produce heel and toe ground contact information. Data is collected and processed at 164Hz, and stored on a 2MB PCMCIA card that can then be removed and its contents transferred to a PC. The processing algorithm operates using an identical method to the algorithm described in chapter 4.

In figure 5.12 a picture of the portable gait events detection system is shown (size: 110mm x 70mm x 20mm; weight: approximately 500g). When the unit was to be carried by a subject, a pouch with a shoulder strap was used. The unit was used to collect the data presented in chapter 6 (with the exception of marker detection system data) from five normal subjects and four subjects with pathological gait.
Chapter 6
Evaluation of the Portable Gait Events Detection System

6.0 Introduction
This chapter details the evaluation of the portable gait events detection system described in chapter 5. Before the results are presented and discussed, the procedure for normal and hemiplegic data collection is discussed.

Before construction of the portable gait events detection system, an initial assessment of the detection algorithm and gyroscope sensor had already been made using a desktop PC connected to a foot-mounted sensor (see chapter 4). However, this assessment was limited in that only one type of terrain - a smooth, level surface - had been used by each subject. Whilst this is a common surface encountered by any walking person, slopes and staircases are also commonplace.

Another limitation of the assessment was in the method of communication between the sensor and the processing unit (the computer). The sensor was attached to the PC via a long lead, and it is reasonable to expect this to have had some effect on each subject and therefore their gait. Thus the development of the portable system served two main purposes: to allow the algorithm and sensor to be tested over a greater variety of terrain (both inside and outside the laboratory), and to eliminate any effect that the lead might have on the subject’s gait. A ready-made notebook or handheld computer was also considered, but it was decided that this could be too heavy and bulky and did not offer any advantage in terms of progress towards a complete practical and portable solution.

6.1 Procedure
The evaluation of the portable unit followed the same pattern as that of the desktop PC, in that data was collected from subjects with normal gait before data collection from hemiplegic subjects took place. However, more extensive testing was achieved using a greater number of simultaneous measurement methods and a more varied terrain.
Evaluation of the Portable Gait Events Detection System

Subjects with Normal Gait

Data was collected from five subjects with normal gait from within the laboratory only. This was to allow data to be collected from a marker detection system (MDS), along with data from the gyroscope sensor and two footswitches. The MDS (Qualisys) consists of seven Proreflex cameras connected to a PC, and uses QTrac software to determine the three dimensional position of passive infrared markers placed on the body. The system resolution is ± 0.2mm within a 1m³ volume (this value will increase as the volume increases), and the accuracy is 1:60,000 within the field of view for each camera.

For this evaluation, one marker was placed on the lateral side of the heel (calcaneous bone) and another placed on the lateral side of the fifth metatarsal. These positions were chosen in order to minimise obscurity, and also to reflect the commonly used positions of the footswitches. A third marker at the lateral ankle (malleolus) formed a triangle, and once data from the three markers had been collected this triangle was used to determine the time at which gait events (HC, FF, HR and TO) occurred.

A visual inspection method was used to determine the occurrence of gait events from the MDS. The method used was repeatable over several gait cycles, and as such could have been implemented as a computer program. Although such a program would allow data to be processed more quickly, it was decided that the development time would outweigh any time saving offered unless very large amounts of data were to be collected.

Each gait event was determined by examining the horizontal (sagittal plane) and vertical velocities of the heel and toe markers (the third marker was used if the heel marker became momentarily obscured). At each event, the horizontal and vertical velocities of each marker would either decrease to zero or increase from zero in a logical, repeatable fashion. For example, at HC the horizontal velocities of both markers would decrease to zero, and only at TO would these velocities leave zero again (ignoring any small horizontal velocities caused by the foot rotating between HC and TO). At HC the
vertical velocity of the heel marker would also become zero, but the same velocity of the toe marker would not reach zero until FF. The vertical velocity of the heel marker would leave zero at HR, but the toe marker would not move until TO. Table 6.1 summarises the velocity at the heel and toe during the four gait events. An arrow indicates that the event was determined just before or just after the velocity was zero, i.e. the velocity was almost zero but still moving.

<table>
<thead>
<tr>
<th>Gait Event</th>
<th>Heel marker H. Velocity</th>
<th>Toe marker H. Velocity</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>V. Velocity</td>
<td>V. Velocity</td>
</tr>
<tr>
<td>HC</td>
<td>→ 0</td>
<td>→ 0</td>
</tr>
<tr>
<td>FF</td>
<td>0</td>
<td>→ 0</td>
</tr>
<tr>
<td>HR</td>
<td>0</td>
<td>0</td>
</tr>
<tr>
<td>TO</td>
<td>0</td>
<td>Not Used</td>
</tr>
<tr>
<td></td>
<td>0</td>
<td>0</td>
</tr>
</tbody>
</table>

Table 6.1 Vertical and Horizontal Velocities of Heel and Toe Markers at Gait Events

The gyroscope sensor was placed on the foot above the metatarsals (as before). Data were collected using the portable unit from this sensor, as well as from one heel and one toe footswitch. The sampling frequencies used by the portable unit and the PC collecting data from the MDS were matched (164Hz), and a synchronisation signal was sent from the marker system (from the moment it began collecting data) to the portable system. This signal was recorded by the portable unit and later used to synchronize the data from the two systems.

The timing of gait events as determined by the footswitches was considered to be at the beginning (initial heel or toe contact) and end (final heel or toe contact) of voltage changes. Whilst other methods could have been used instead to determine the occurrence of events (such as the crossing of a predetermined threshold), the rate of change is the most common method used by drop foot correction stimulators.

Table 6.2 shows details of all the subjects with normal gait.

<table>
<thead>
<tr>
<th>Subject</th>
<th>1</th>
<th>2</th>
<th>3</th>
<th>4</th>
<th>5</th>
</tr>
</thead>
<tbody>
<tr>
<td>Age</td>
<td>21</td>
<td>29</td>
<td>34</td>
<td>49</td>
<td>56</td>
</tr>
<tr>
<td>Sex</td>
<td>Male</td>
<td>Male</td>
<td>Male</td>
<td>Male</td>
<td>Male</td>
</tr>
</tbody>
</table>

Table 6.2 Details of Subjects with Normal Gait
6.1.1 Subjects with Hemiplegic Gait

Data collection from four subjects with hemiplegic gait took place in Salisbury District Hospital. Measurements were to be taken from both inside and outside (patients were to be asked to walk along a path to a nearby carpark and back), but this was not possible due to poor weather conditions. Instead, patients walked through various rooms and corridors inside the hospital, and two of the patients also ascended a 4m ramp and a seven-step staircase. The ramp was constructed using two sections of ½ inch plywood to form the walking surface, supported by two eight by four inch beams running underneath. The ramp was raised at one end by resting the walkway upon a ready-made wooden ‘horse’. The steps of the staircase (30cm long and 15cm high) were constructed from ½ inch plywood, and these rested upon ¾ inch plywood sections forming the sides of the staircase. Figure 6.1 shows the staircase used in this study. Data was collected from the gyroscope sensor and two footswitches only, although a video camera was also used throughout to provide a visual record of each walk.

The patient’s own stimulator was used during some of the walks. Since this required a second footswitch to be placed under the heel (all patients used stimulation triggered by a heel switch), a dual footswitch was constructed. The dual footswitch had dimensions (most importantly thickness) almost identical to the standard footswitch, allowing the patient to walk without any discomfort.
Evaluation of the Portable Gait Events Detection System

Table 6.3 Details of Subjects with Hemiplegic Gait

<table>
<thead>
<tr>
<th>Subject</th>
<th>Age</th>
<th>Sex</th>
<th>Pathology</th>
<th>Affected Side</th>
<th>Diagnosis Date</th>
<th>Using FES Since</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>64</td>
<td>Female</td>
<td>CVA</td>
<td>Left</td>
<td>Dec-92</td>
<td>Sep-00</td>
</tr>
<tr>
<td></td>
<td>55</td>
<td>Female</td>
<td>MS</td>
<td>Left</td>
<td>Feb-97</td>
<td>Nov-99</td>
</tr>
<tr>
<td></td>
<td>63</td>
<td>Male</td>
<td>CVA</td>
<td>Right</td>
<td>Oct-97</td>
<td>Aug-00</td>
</tr>
<tr>
<td></td>
<td>29</td>
<td>Female</td>
<td>MS</td>
<td>Right</td>
<td>Jan-97</td>
<td>May-00</td>
</tr>
</tbody>
</table>

Figure 6.1 The Staircase
6.1.2 Protocol: Subjects with Normal Gait

Equipment used:

Portable unit (data logger)
2 x standard footswitch (heel and toe)
gyroscope sensor
2 x Velcro strap
Shoulder pouch (holds unit at the waist)
PCMCIA memory card, card reader and software
PC
Marker detection system
Ramp and staircase

Method (as written at the time of study):

Each subject to be fitted with two footswitches (heel and toe) and the gyroscope sensor. The data logger (carried in the shoulder pouch) will be used to collect data. Each subject will be asked walk twice (slow and normal speed) over three different terrains (a smooth level surface, a 4m ramp inclined at approximately 8° and a staircase with seven 150mm high steps.) The position of three markers placed on the foot will be recorded using the MDS. Each walk is to be captured on video camera.

1) Explain the nature of the study to the subject
2) Explain the walking routine to the subject
3) Reset the portable unit (data logger)
4) Calibrate the portable unit
5) Start the video camera
6) Start collecting data
7) Start the marker detection system
8) Ask the subject to begin walking
9) When the subject reaches the end location, stop the video camera
10) Stop the data logger
11) Upload data from the card to the PC
12) Ensure that the marker detection system has been ‘seen’ all markers throughout
13) Repeat steps 3 to 12 for the next walk with stimulator switched on.

### 6.1.3 Protocol: Subjects with Hemiplegic Gait

*Equipment required:*

- Portable unit (data logger)
- 1 x standard footswitch (toe) and 1 x dual footswitch (heel)
- Gyroscope sensor
- 2 x Velcro strap
- Shoulder pouch (holds unit at the waist)
- PCMCIA memory card, card reader and software
- PC
- Subject’s own stimulator

*Method (as written at the time of study):*

Each subject to be fitted with two footswitches (toe and dual heel) and the gyroscope sensor. The data logger (carried in the shoulder pouch) will be used to collect data. Each subject will be asked to walk from the laboratory to the carpark and back twice – the first time without stimulation, and the second time with stimulation (a suitable rest period should take place between the two). Each walk is to be captured on video camera.

Poor weather conditions prompted a change in the route described. Instead of walking to the car park, subjects 1 and 2 were asked instead to walk along a level corridor, which included a 3m incline of approximately 4 degrees and a flight of stairs (which were ascended) at the end. Subjects 3 and 4 were already displaying signs of fatigue before any measurements had been taken (due to their routine assessment having been already carried out by clinicians), and therefore they were only asked to walk a short distance along a level corridor.
Evaluation of the Portable Gait Events Detection System

1) Explain the nature of the study to the subject (subject is seated)
2) Explain the walking routine to the subject
3) Reset the portable unit (data logger)
4) Calibrate the portable unit
5) Ask the subject to stand
6) Start the video camera
7) Start collecting data
8) Ask the subject to begin walking (walk with the subject)
9) When the subject returns to the start location, stop the video camera
10) Stop the data logger (subject is seated)
11) Upload data from the card to the PC
12) Repeat steps 3 to 11 for the second walk with stimulator switched on.

6.2 Results: Normal Subjects

Collected data was divided into three groups for comparison: Gyroscope sensor and marker detection data, gyroscope sensor and footswitch data, and marker detection and footswitch data. Within each group, data are presented in table (6.4 to 6.12) and graph (6.2 to 6.10) format for each gait event occurring during each walk. Each table displays the average difference (normal, absolute and standard deviation – SD) and the range of difference between the time of events for each measurement method. The number of events missed by the gyroscope sensor is also presented. Events that were missed were FF and HR detection during walking down the staircase; this occurred due to variations in the angular velocity signal on these occasions.

Each graph compares measurements of the same event (e.g. FF, first stride during normal speed on level surface) made using two of the three different methods: gyroscope sensor, MDS and footswitches. Events measured by the two different methods that occur within 10ms of each other are grouped together and displayed in one column (0,10). The next column shows events that occurred within 10 to 20 ms of each other (10,20), and so on. The height of each column represents the number of events (as a percentage of the total number) that fall into each category.
The tables and graphs indicate a closer relationship between two measurement methods when values in the tables are lower (and fewer events are missed). In this study the measurements taken by the MDS are considered to be the most accurate (due to the high measurement resolution and accuracy of the system – see section 6.1.1). Since the accuracy of footswitches is considered acceptable for use with FES, a gyroscope sensor/MDS difference that is lower than the corresponding footswitch/MDS difference indicates that the gyroscope sensor was more accurate in that instance.
### 6.2.1 Tables of Gyroscope Sensor – Marker Detection System Comparison

<table>
<thead>
<tr>
<th>Walk</th>
<th>Event</th>
<th>HC</th>
<th>FF</th>
<th>HR</th>
<th>TO</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Level Surface</strong></td>
<td>Average Difference</td>
<td>-1</td>
<td>17</td>
<td>11</td>
<td>50</td>
</tr>
<tr>
<td></td>
<td>Av. Absolute Difference</td>
<td>36</td>
<td>21</td>
<td>36</td>
<td>54</td>
</tr>
<tr>
<td></td>
<td>Range of Difference</td>
<td>6 to 113</td>
<td>-6 to 60</td>
<td>0 to 107</td>
<td>6 to 125</td>
</tr>
<tr>
<td></td>
<td>Standard Deviation</td>
<td>46</td>
<td>20</td>
<td>49</td>
<td>47</td>
</tr>
<tr>
<td></td>
<td>Events Missed</td>
<td>None</td>
<td>None</td>
<td>None</td>
<td>None</td>
</tr>
<tr>
<td><strong>Normal Speed</strong></td>
<td>Average Difference</td>
<td>-13</td>
<td>27</td>
<td>-4</td>
<td>61</td>
</tr>
<tr>
<td></td>
<td>Av. Absolute Difference</td>
<td>21</td>
<td>29</td>
<td>23</td>
<td>61</td>
</tr>
<tr>
<td></td>
<td>Range of Difference</td>
<td>-36 to 30</td>
<td>0 to 83</td>
<td>0 to 42</td>
<td>36 to 120</td>
</tr>
<tr>
<td></td>
<td>Standard Deviation</td>
<td>20</td>
<td>26</td>
<td>27</td>
<td>24</td>
</tr>
<tr>
<td></td>
<td>Events Missed</td>
<td>None</td>
<td>None</td>
<td>None</td>
<td>None</td>
</tr>
<tr>
<td><strong>Ramp (up)</strong></td>
<td>Average Difference</td>
<td>10</td>
<td>24</td>
<td>104</td>
<td>63</td>
</tr>
<tr>
<td></td>
<td>Av. Absolute Difference</td>
<td>18</td>
<td>24</td>
<td>106</td>
<td>63</td>
</tr>
<tr>
<td></td>
<td>Range of Difference</td>
<td>-18 to 42</td>
<td>0 to 54</td>
<td>-12 to 386</td>
<td>6 to 101</td>
</tr>
<tr>
<td></td>
<td>Standard Deviation</td>
<td>20</td>
<td>15</td>
<td>137</td>
<td>31</td>
</tr>
<tr>
<td></td>
<td>Events Missed</td>
<td>None</td>
<td>None</td>
<td>None</td>
<td>None</td>
</tr>
<tr>
<td><strong>Ramp (down)</strong></td>
<td>Average Difference</td>
<td>-22</td>
<td>35</td>
<td>-42</td>
<td>62</td>
</tr>
<tr>
<td></td>
<td>Av. Absolute Difference</td>
<td>33</td>
<td>21</td>
<td>89</td>
<td>64</td>
</tr>
<tr>
<td></td>
<td>Range of Difference</td>
<td>-83 to 30</td>
<td>-30 to 83</td>
<td>-440 to 83</td>
<td>-12 to 119</td>
</tr>
<tr>
<td></td>
<td>Standard Deviation</td>
<td>33</td>
<td>27</td>
<td>153</td>
<td>45</td>
</tr>
<tr>
<td></td>
<td>Events Missed</td>
<td>None</td>
<td>2 (/10)</td>
<td>None</td>
<td>None</td>
</tr>
</tbody>
</table>

*Table 6.4 Table of Gyroscope Sensor – MDS Comparison Data (ms)*
### Table 6.5 Table of Gyroscope Sensor – MDS Comparison Data (ms)

<table>
<thead>
<tr>
<th>Walk</th>
<th>Event</th>
<th>HC</th>
<th>FF</th>
<th>HR</th>
<th>TO</th>
</tr>
</thead>
<tbody>
<tr>
<td>Ramp (up)</td>
<td>Average Difference</td>
<td>-11</td>
<td>22</td>
<td>26</td>
<td>83</td>
</tr>
<tr>
<td></td>
<td>Av. Absolute Difference</td>
<td>26</td>
<td>35</td>
<td>39</td>
<td>83</td>
</tr>
<tr>
<td>Normal Speed</td>
<td>Range of Difference</td>
<td>-48 to 36</td>
<td>-36 to 95</td>
<td>-36 to 101</td>
<td>6 to 196</td>
</tr>
<tr>
<td></td>
<td>Standard Deviation</td>
<td>29</td>
<td>43</td>
<td>48</td>
<td>67</td>
</tr>
<tr>
<td></td>
<td>Events Missed</td>
<td>None</td>
<td>None</td>
<td>None</td>
<td>None</td>
</tr>
<tr>
<td>Ramp (down)</td>
<td>Average Difference</td>
<td>-21</td>
<td>42</td>
<td>13</td>
<td>67</td>
</tr>
<tr>
<td></td>
<td>Av. Absolute Difference</td>
<td>24</td>
<td>42</td>
<td>37</td>
<td>67</td>
</tr>
<tr>
<td>Normal Speed</td>
<td>Range of Difference</td>
<td>-42 to 12</td>
<td>0 to 113</td>
<td>-48 to 60</td>
<td>18 to 137</td>
</tr>
<tr>
<td></td>
<td>Standard Deviation</td>
<td>19</td>
<td>44</td>
<td>39</td>
<td>45</td>
</tr>
<tr>
<td></td>
<td>Events Missed</td>
<td>None</td>
<td>None</td>
<td>None</td>
<td>None</td>
</tr>
<tr>
<td>Staircase (up)</td>
<td>Average Difference</td>
<td>-10</td>
<td>33</td>
<td>82</td>
<td>151</td>
</tr>
<tr>
<td></td>
<td>Av. Absolute Difference</td>
<td>40</td>
<td>46</td>
<td>127</td>
<td>151</td>
</tr>
<tr>
<td>Slow Speed</td>
<td>Range of Difference</td>
<td>-83 to 107</td>
<td>-67 to 107</td>
<td>-149 to 411</td>
<td>54 to 214</td>
</tr>
<tr>
<td></td>
<td>Standard Deviation</td>
<td>51</td>
<td>53</td>
<td>21</td>
<td>54</td>
</tr>
<tr>
<td></td>
<td>Events Missed</td>
<td>None</td>
<td>None</td>
<td>None</td>
<td>None</td>
</tr>
<tr>
<td>Staircase (down)</td>
<td>Average Difference</td>
<td>-105</td>
<td>-</td>
<td>-</td>
<td>5</td>
</tr>
<tr>
<td></td>
<td>Av. Absolute Difference</td>
<td>182</td>
<td>-</td>
<td>-</td>
<td>30</td>
</tr>
<tr>
<td>Slow Speed</td>
<td>Range of Difference</td>
<td>-381 to 214</td>
<td>-</td>
<td>-</td>
<td>-42 to 77</td>
</tr>
<tr>
<td></td>
<td>Standard Deviation</td>
<td>190</td>
<td>-</td>
<td>-</td>
<td>38</td>
</tr>
<tr>
<td></td>
<td>Events Missed</td>
<td>None</td>
<td>14 (All)</td>
<td>14 (All)</td>
<td>None</td>
</tr>
</tbody>
</table>
Evaluation of the Portable Gait Events Detection System

<table>
<thead>
<tr>
<th>Walk</th>
<th>Event</th>
<th>HC</th>
<th>FF</th>
<th>HR</th>
<th>TO</th>
</tr>
</thead>
<tbody>
<tr>
<td>Staircase</td>
<td>Average Difference</td>
<td>17</td>
<td>41</td>
<td>138</td>
<td>164</td>
</tr>
<tr>
<td>(up)</td>
<td>Av. Absolute Difference</td>
<td>41</td>
<td>48</td>
<td>144</td>
<td>164</td>
</tr>
<tr>
<td>Normal</td>
<td>Range of Difference</td>
<td>-36 to 202</td>
<td>-42 to 244</td>
<td>-18 to 452</td>
<td>-18 to 452</td>
</tr>
<tr>
<td>Speed</td>
<td>Standard Deviation</td>
<td>72</td>
<td>78</td>
<td>179</td>
<td>47</td>
</tr>
<tr>
<td>Events Missed</td>
<td>None</td>
<td>None</td>
<td>None</td>
<td>None</td>
<td>None</td>
</tr>
</tbody>
</table>

| Staircase  | Average Difference   | -90 | -  | -  | 2   |
| (down)     | Av. Absolute Difference | 117 | -  | -  | 48  |
| Normal     | Range of Difference  | -256 to 208 | -  | -  | -119 to 77 |
| Speed      | Standard Deviation   | 110 | -  | -  | 64  |
| Events Missed | None | 15 (All) | 15 (All) | None | None |

Table 6.6 Table of Gyroscope Sensor – MDS Comparison Data (ms)

6.2.2 Graphs of Gyroscope Sensor – Marker Detection System Comparison

Figure 6.2 Graph of Range of Differences in HC Event Detection Times During Walking on a Level Surface at Slow Speed
Figure 6.3 Graph of Range of Differences of Gait Events During Walking on a Level Surface at a Self Selected Speed
Figure 6.4 Graph of Range of Differences of Gait Events During Walking on Various Surfaces
### 6.2.3 Tables of Gyroscope Sensor – Footswitches Comparison

<table>
<thead>
<tr>
<th>Walk</th>
<th>Event</th>
<th>HC</th>
<th>FF</th>
<th>HR</th>
<th>TO</th>
</tr>
</thead>
<tbody>
<tr>
<td>Level Surface</td>
<td>Average Difference</td>
<td>-50</td>
<td>-73</td>
<td>37</td>
<td>130</td>
</tr>
<tr>
<td>Level Surface</td>
<td>Av. Absolute Difference</td>
<td>50</td>
<td>96</td>
<td>115</td>
<td>130</td>
</tr>
<tr>
<td>Slow Speed</td>
<td>Range of Difference</td>
<td>-84 to -6</td>
<td>-483 to 36</td>
<td>-149 to 274</td>
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<th>TO</th>
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**Table 6.7 Table of Gyroscope Sensor – Footswitches Comparison Data (ms)**
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<th>TO</th>
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Table 6.8 Table of Gyroscope Sensor – Footswitches Comparison Data (ms)
## Evaluation of the Portable Gait Events Detection System

### Table 6.9 Table of Gyroscope Sensor – Footswitches Comparison Data (ms)

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<th>HR</th>
<th>TO</th>
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<td>89</td>
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### 6.2.4 Graphs of Gyroscope Sensor – Footswitches Comparison

**Figure 6.5** Graph of Range of Differences in HC Event Detection Times During Walking on a Level Surface at Slow Speed
Evaluation of the Portable Gait Events Detection System

**Figure 6.6** Graph of Range of Differences of Gait Events During Walking on a Level Surface at a Self Selected Speed
Figure 6.7 Graph of Range of Differences of Gait Events During Walking on Various Surfaces
### 6.2.5 Tables of Footswitches - Marker Detection System Comparison

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**Table 6.10** Table of Gyroscope Sensor – MDS Comparison Data (ms)
### Table 6.11 Table of Gyroscope Sensor – MDS Comparison Data (ms)

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### Table 6.12 Table of Gyroscope Sensor – Footswitches Comparison Data (ms)

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<td>Standard Deviation</td>
<td>67</td>
<td>75</td>
<td>133</td>
<td>76</td>
</tr>
<tr>
<td>Events Missed</td>
<td>None</td>
<td>None</td>
<td>None</td>
<td>None</td>
<td>None</td>
</tr>
</tbody>
</table>

6.2.6 Graphs of Footswitches - Marker Detection System Comparison During Walking on Various Surfaces at Different Speeds

![Graph of Range of Differences in HC Event Detection Times During Walking on a Level Surface at Slow Speed](image)

**Figure 6.8** Graph of Range of Differences in HC Event Detection Times During Walking on a Level Surface at Slow Speed
Figure 6.9 Graph of Range of Differences of Gait Events During Walking on a Level Surface at a Self Selected Speed
**Figure 6.10** Graph of Range of Differences of Gait Events During Walking on Various Surfaces
6.2.7 Summary of Results for Subjects with Normal Gait

The data in the tables shows that only the gyroscope sensor missed events, although in almost every case this was limited to FF and HR when using the staircase (the other case was FF being missed twice whilst one subject was descending the ramp at slow speed). Since the staircase step length used was 300mm, all walkers placed their entire foot – both heel and toe - upon the step when ascending and descending. Therefore the heel and toe footswitches were both compressed and therefore did not miss events. Generally speaking, however, the foot makes contact with each step with heel and toe at around the same time (unlike normal walking on level surfaces), and therefore it is reasonably acceptable to ignore FF and HR in this case. It should be pointed out that many staircases have shorter step lengths that do not accommodate the entire foot, either toe only (ascending) or heel only (descending) foot contact. This presents a problem to single footswitch systems (see section 2.3.1).

The tables also display the range for each event, and for the gyroscope sensor/MDS comparison there is a general trend for this to be lowest during walking on the level surface and greatest on the staircase, with the ramp between the two. Figures 6.11 and 6.12 show the difference range for all events. On the ordinate is the difference range for each individual event in ms (e.g. for slow walking HC on the level surface, the data presented represents an average of three strides for each subject). Along the abscissa, each gait event (HC, FF, HR and TO) is shown for every walk.

In figure 6.11, (gyroscope/MDS difference) it can be seen that the difference ranges are clearly greatest for HR during walking up and down the ramp, and HR and whilst walking up the staircase and HC whilst walking down. HR, TO (up) and HC (down) are differences that are also significantly high for the staircase. A linear trendline (calculated using the ‘least squares’ method – Microsoft Excel) is also shown, and this increases in value on the left-hand-side from around 100ms (level surface) to 250ms on the right (stairs). In figure 6.12 the ranges are generally much greater. A similar trendline is shown, increasing in value from just under 300ms to about 310ms. A comparison of the two figures indicates that the range of differences is lower between
Evaluation of the Portable Gait Events Detection System

the gyroscope sensor and the MDS than those between the gyroscope sensor and the footswitches.

![Graph of Gyroscope Sensor/MDS Ranges of Differences](image)

**Figure 6.11** Bar Chart of Gyroscopic Sensor/MDS Difference Ranges For All Events

![Graph of Gyroscope Sensor/Footswitches Ranges of Differences](image)

**Figure 6.12** Bar Chart of Gyroscopic Sensor/Footswitches Difference Ranges For All Events
It is possible that a high range of differences could be the result of a small number of outliers with the rest of the data very close to zero. In order to observe the distribution of differences, figures 6.1 to 6.9 were plotted, and these show the distribution across the range of a particular walk (e.g. ascending the ramp at slow speed). Amongst the gyroscope/MDS and footswitches/MDS comparisons, outliers are clearly present in figures 6.3 and 6.6. In all three comparisons, it is clear that the number of bars increases as the reader observes each graph from walking on the level surface, through walking on the ramp to ascending and descending the staircase. This indicates that, generally speaking, the distribution of differences increased from the level surface through the ramp to the staircase data.

In order to make an overall comparison of the gyroscope and the footswitch measurement methods, the average of the absolute differences were taken. Although the precise time that each event occurred is unknown, the times given by the MDS were taken as the closest to the actual event time, with the measurements from the other two methods seen as ‘competing’ to be as close as possible to the MDS value. Table 6.13 shows these values for the differences in measurement between the gyroscope sensor and the MDS, and the footswitches and the MDS. On the right-hand side of the table the same values are also shown for the differences between the gyroscope sensor and the footswitches. Although it is the two differences involving the MDS that are most relevant, it is useful to present the gyroscope sensor/FSR differences for comparison with the data obtained in the hemiplegic study in which no MDS was used.

<table>
<thead>
<tr>
<th></th>
<th>Gyro – Marker</th>
<th>FSR – Marker</th>
<th>Gyro – FSR</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Average Absolute Differences</strong></td>
<td>63</td>
<td>82</td>
<td>108</td>
</tr>
<tr>
<td><strong>Standard Deviation of Absolute Differences</strong></td>
<td>78</td>
<td>88</td>
<td>99</td>
</tr>
</tbody>
</table>

Table 6.13 Average and Standard Deviation of Absolute Differences for All Events (ms)
In table 6.13 it can be seen the gyroscope sensor/MDS average difference is lower than the footswitches/MDS average difference, an indication that the gyroscope sensor may be as accurate (or more accurate) than the footswitches. The SD is also lowest for the gyroscope sensor/MDS differences indicating that the FSR data is spread across a greater range (although all SD values are relatively very high – greater than the averages themselves).

Figure 6.10 shows that measurement differences are greatest when walking on the staircase. When the values collected on the level surface and the ramp are considered only, the average and SD for the difference between the gyroscope sensor and the MDS are reduced, while the FSR/MDS average difference is also relatively unchanged. These new values are shown in table 6.14. Figure 6.10 also shows that measurement differences are lowest when walking on level ground. In table 6.14 the average and SD are shown for the difference measurements taken only on the level surface (first row). Again it can be seen that the average and SD for the difference between the gyroscope sensor and the MDS are further reduced, and the FSR/MDS average difference is still relatively unchanged when these results only are considered. Averages for both walking up and down are shown for the ramp and the staircase.

Table 6.14 shows the averages and SDs for several combinations of terrain. Also included within this table are the average and SD values for HC and HR events measured on the level surface only. These values are included to allow comparison with the heel footswitch, providing an indication how the gyroscope sensor would perform as a replacement of the heel switch in a single channel drop foot stimulator.

Table 6.14 indicates that the gyroscope sensor has outperformed the footswitches in every presented case except the staircase. In two cases (level surface: HC and HR) both the gyroscope sensor/MDS difference average and SD are lower than 30ms, and also in two cases (level surface: all events and HC only) both the gyroscope sensor/MDS difference average and SD are considerably lower (approx. 40ms). The footswitches performed relatively well in the ‘HC only’ comparison, and therefore the gyroscope sensor improvement in this case was only marginal.
<table>
<thead>
<tr>
<th>Terrain</th>
<th>Event(s)</th>
<th>Parameter</th>
<th>Gyro – Marker</th>
<th>FSR – Marker</th>
<th>Gyro – FSR</th>
</tr>
</thead>
<tbody>
<tr>
<td>Level</td>
<td>HC FF</td>
<td>Average</td>
<td>35</td>
<td>71</td>
<td>83</td>
</tr>
<tr>
<td></td>
<td>HR TO</td>
<td>SD</td>
<td>29</td>
<td>76</td>
<td>73</td>
</tr>
<tr>
<td>Ramp (Up and Down)</td>
<td>HC FF</td>
<td>Average</td>
<td>49</td>
<td>85</td>
<td>93</td>
</tr>
<tr>
<td></td>
<td>HR TO</td>
<td>SD</td>
<td>59</td>
<td>91</td>
<td>87</td>
</tr>
<tr>
<td>Stairs (Up and Down)</td>
<td>HC TO</td>
<td>Average</td>
<td>94</td>
<td>87</td>
<td>138</td>
</tr>
<tr>
<td></td>
<td></td>
<td>SD</td>
<td>101</td>
<td>92</td>
<td>115</td>
</tr>
<tr>
<td>Level and Ramp (Up and Down)</td>
<td>HC FF</td>
<td>Average</td>
<td>43</td>
<td>79</td>
<td>89</td>
</tr>
<tr>
<td></td>
<td>HR TO</td>
<td>SD</td>
<td>49</td>
<td>85</td>
<td>81</td>
</tr>
<tr>
<td>Level, Ramp and Stairs (Up and Down)</td>
<td>HC FF</td>
<td>Average</td>
<td>63</td>
<td>82</td>
<td>108</td>
</tr>
<tr>
<td></td>
<td>HR TO</td>
<td>SD</td>
<td>78</td>
<td>88</td>
<td>99</td>
</tr>
<tr>
<td>Level</td>
<td>HC</td>
<td>Average</td>
<td>28</td>
<td>37</td>
<td>38</td>
</tr>
<tr>
<td></td>
<td></td>
<td>SD</td>
<td>21</td>
<td>42</td>
<td>21</td>
</tr>
<tr>
<td>Level</td>
<td>HR</td>
<td>Average</td>
<td>29</td>
<td>80</td>
<td>99</td>
</tr>
<tr>
<td></td>
<td></td>
<td>SD</td>
<td>27</td>
<td>43</td>
<td>59</td>
</tr>
</tbody>
</table>

**Table 6.14** Average and Standard Deviation of Absolute Differences for Various Terrain and Events (ms)

In figure 6.12 all measured differences (gyroscope sensor – MDS, footswitches – MDS and gyroscope sensor – footswitches) on the ordinate have been plotted against their corresponding distribution centiles on the abscissa. The graph shows that differences for the footswitches – MDS plot are consistently greater than the differences for the gyroscope sensor – MDS plot. Since lower values on the ordinate are preferable (i.e. a
lower difference between one of the sensors and the MDS) the graph is another indication that the gyroscope sensor has performed at least as well as the footswitches.

**Figure 6.13** Graph of Distribution Centiles Against Measurement Differences
6.3 Results: Hemiplegic Subjects

As with the data from normal subjects, average differences (including absolute values), ranges, SD and the number of events missed are presented in tables (6.15 and 6.16). The data is divided by type of terrain and whether with or without stimulation. Some events were not detected by the either or both sensors (shown as ‘events missed’ and ‘sensor in error’ in the tables). Since patients three and four complained that they were tired before measurements were taken, data was also divided into patient groups: one and two and three and four. Graphs of differences for each gait event, either with or without electrical stimulation are also shown (6.14 and 6.15).

6.3.1 Tables of Gyroscope Sensor - Footswitches Comparison

<table>
<thead>
<tr>
<th>Walk</th>
<th>Event</th>
<th>HC</th>
<th>FF</th>
<th>HR</th>
<th>TO</th>
</tr>
</thead>
<tbody>
<tr>
<td>Level Surface</td>
<td>Average Difference</td>
<td>-40</td>
<td>-3</td>
<td>89</td>
<td>98</td>
</tr>
<tr>
<td></td>
<td>Av. Absolute Difference</td>
<td>40</td>
<td>16</td>
<td>113</td>
<td>98</td>
</tr>
<tr>
<td>Subjects 1 and 2</td>
<td>Range of Difference</td>
<td>-79 to 12</td>
<td>-79 to 24</td>
<td>-140 to 219</td>
<td>67 to 189</td>
</tr>
<tr>
<td>No Stimulation</td>
<td>Standard Deviation</td>
<td>21</td>
<td>22</td>
<td>90</td>
<td>23</td>
</tr>
<tr>
<td>Events Missed</td>
<td>None</td>
<td>None</td>
<td>None</td>
<td>None</td>
<td>None</td>
</tr>
<tr>
<td>Level Surface</td>
<td>Average Difference</td>
<td>22</td>
<td>-</td>
<td>188</td>
<td>-</td>
</tr>
<tr>
<td></td>
<td>Av. Absolute Difference</td>
<td>54</td>
<td>-</td>
<td>188</td>
<td>-</td>
</tr>
<tr>
<td>Subjects 3 and 4</td>
<td>Range of Difference</td>
<td>-103 to 110</td>
<td>-</td>
<td>48 to 274</td>
<td>-</td>
</tr>
<tr>
<td>No Stimulation</td>
<td>Standard Deviation</td>
<td>61</td>
<td>-</td>
<td>57</td>
<td>-</td>
</tr>
<tr>
<td>Events Missed</td>
<td>None</td>
<td>49 (All)</td>
<td>30/95</td>
<td>49 (All)</td>
<td></td>
</tr>
<tr>
<td>Sensor in Error</td>
<td>Both</td>
<td>Both</td>
<td>FSR</td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

Table 6.15 Table of Gyroscope Sensor – Footswitches Comparison Data (ms)
<table>
<thead>
<tr>
<th>Level Surface</th>
<th>Average Difference</th>
<th>-31</th>
<th>46</th>
<th>54</th>
<th>103</th>
</tr>
</thead>
<tbody>
<tr>
<td>Av. Absolute Difference</td>
<td>31</td>
<td>54</td>
<td>151</td>
<td>103</td>
<td></td>
</tr>
<tr>
<td>Range of Difference</td>
<td>-79 to 0</td>
<td>-61 to 165</td>
<td>-293 to 232</td>
<td>73 to 201</td>
<td></td>
</tr>
<tr>
<td>Standard Deviation</td>
<td>22</td>
<td>54</td>
<td>153</td>
<td>28</td>
<td></td>
</tr>
<tr>
<td>Events Missed</td>
<td>None</td>
<td>None</td>
<td>None</td>
<td>None</td>
<td></td>
</tr>
</tbody>
</table>

| Subjects 1 and 2 | Average Difference | -13 | - | 157 | -212 |
| Av. Absolute Difference | 15 | - | 166 | 217 |
| Range of Difference | -42 to 6 | - | -293 to 232 | -512 to 43 |
| Standard Deviation | 13 | - | 86 | 184 |
| Events Missed | 22/82 | 17 (All) | 22/82 | 17/82 |
| Sensor in Error FSR Both Both FSR |

| Staircase (Up and Down) | Average Difference | 3 | 255 | -14 | 276 |
| Av. Absolute Difference | 89 | 255 | 69 | 276 |
| Range of Difference | -85 to 244 | 98 to 409 | -348 to 159 | 91 to 664 |
| Standard Deviation | 119 | 117 | 57 | 165 |
| Events Missed | None | None | None | None |

Table 6.16 Table of Gyroscope Sensor – Footswitches Comparison Data (ms)
6.3.2 Graphs of Gyroscope Sensor - Footswitches Comparison

Figure 6.14 Graph of Range of Differences of Gait Events During Hemiplegic Walking on a Level Surface
Evaluation of the Portable Gait Events Detection System

Figure 6.15 Graph of Range of Differences of Gait Events During Hemiplegic Walking on a Level Surface
6.3.3 Summary of Results for Subjects with Hemiplegic Gait

The tables show that, while the gyroscope sensor missed events (FF and HR) only whilst ascending the staircase, both types of sensor missed events in patients three and four. The worst case was FF, which was missed by both sensors in subject three and by the footswitches only in subject four. HR was missed by both sensors when subject three walked with stimulation, and by the gyroscope sensor only when without. There appears to be no trend regarding the ranges of difference when comparing walking with and without the stimulator.

Figures 6.13 to 6.14 show the distribution of differences across the ranges for each gait event. As with the tables, the graphs do not appear to display any obvious trends in the data. Walking with and without the stimulator appears to make no overall difference to the sensor differences.

As before, the average of all sensor differences (i.e. all gait events from all subjects) was calculated and presented. Table 6.12 shows the average difference (and SD) for all data measured with and without electrical stimulation. The table shows that the average difference (and SD) is lower for unassisted (no electrical stimulation) gait. It is not known why electrical stimulation would cause the difference to be greater.

<table>
<thead>
<tr>
<th>Parameter</th>
<th>No Stimulation</th>
<th>With Stimulation</th>
</tr>
</thead>
<tbody>
<tr>
<td>Average</td>
<td>82</td>
<td>111</td>
</tr>
<tr>
<td>SD</td>
<td>73</td>
<td>109</td>
</tr>
</tbody>
</table>

Table 6.17 Average and Standard Deviation of Absolute Differences For All Events (ms)
In table 6.13 the averages and SDs are presented for several combinations of terrain. Average and SD values for HC and HR events measured on the flat surface are also shown to allow comparison with the heel footswitch.

<table>
<thead>
<tr>
<th>Terrain</th>
<th>Event(s)</th>
<th>Parameter</th>
<th>No Stimulation</th>
<th>With Stimulation</th>
</tr>
</thead>
<tbody>
<tr>
<td>Flat (&amp; Ramp)</td>
<td>HC FF</td>
<td>Average</td>
<td>73</td>
<td>100</td>
</tr>
<tr>
<td></td>
<td>HR TO</td>
<td>SD</td>
<td>59</td>
<td>91</td>
</tr>
<tr>
<td>Stairs</td>
<td>HC TO</td>
<td>Average</td>
<td>192</td>
<td>173</td>
</tr>
<tr>
<td></td>
<td></td>
<td>SD</td>
<td>126</td>
<td>165</td>
</tr>
<tr>
<td>Flat (&amp; Ramp) and Stairs</td>
<td>HC FF</td>
<td>Average</td>
<td>82</td>
<td>111</td>
</tr>
<tr>
<td></td>
<td>HR TO</td>
<td>SD</td>
<td>73</td>
<td>109</td>
</tr>
<tr>
<td>Flat (&amp; Ramp)</td>
<td>HC</td>
<td>Average</td>
<td>48</td>
<td>26</td>
</tr>
<tr>
<td></td>
<td></td>
<td>SD</td>
<td>29</td>
<td>20</td>
</tr>
<tr>
<td>Flat (&amp; Ramp)</td>
<td>HR</td>
<td>Average</td>
<td>135</td>
<td>156</td>
</tr>
<tr>
<td></td>
<td></td>
<td>SD</td>
<td>66</td>
<td>57</td>
</tr>
</tbody>
</table>

Table 6.18 Average and Standard Deviation of Absolute Differences for Various Terrain and Events (ms)

Comparing the values measured with and without electrical stimulation in table 6.13, it appears that there is not any overall effect of the use of electrical stimulation (table 6.12 indicated that sensor measurements were closer without stimulation). Averages and SDs are generally greater than those measured during able-bodied gait. This is to be expected as the preliminary study showed that the same differences were greater. The greatest difference between the two sensor measurements is observed when walking on the staircase (as is the case with normal gait), and this may be attributed to the gyroscope sensor’s apparent poorer accuracy with this terrain.
Comparing average differences measured for HC and HR, there is a considerably greater difference in the HR data. In the summary table for normal subjects (table 6.11), the gyroscope sensor/MDS difference for HC is only 9 ms less than of a similar value to that of the footswitch/MDS difference (28 and 37 ms respectively). However, the same HR differences are not similar values (29 and 80 ms). This larger difference suggests that one of the sensors is less accurate than the other (assuming that one of the sensors has an accuracy in detecting this event that is reasonably close to the actual occurrence (which for this study was taken to be the MDS time). Since (in normal gait measurements) the MDS values were closer to the gyroscope sensor values than the footswitch values, it is possible that the large sensor difference in hemiplegic gait HR was due to the greater inaccuracy of the footswitch in detecting this event.
Chapter 7
Summary and Conclusions

7.1 Introduction

FES has been used to correct drop-foot for several decades, with the most commonly used system being single channel. Single channel stimulation does not always address specific problems related to drop-foot gait, namely lack of knee flexion and calf muscle spasticity. However, single channel stimulation is far less complicated to set up and use, and in many cases offers considerable improvement to patients’ gait.

Footswitches have become well established as gait event detecting sensors in drop-foot correction systems. Limitations of these sensors, however, have led to researchers evaluating other sensors for augmentation or replacement. A gyroscope sensor was used in this study to determine gait events, and these measurements were compared with the similar measurements made by heel and toe footswitches, and a marker detection system (MDS). In order to assess the performance of the gyroscope sensor, software was first developed on a desktop PC and later re-written on a custom built portable unit.

7.2 The Sensor

A gyroscope sensor was developed that was small and light enough to be inconspicuously mounted on top of the shoe. The total cost of the sensor package was approximately £25, which is comparable to a footswitch. It is envisaged that the lifetime of the sensor would be several years greater than that of a footswitch. Precise sensor positioning on the shoe was not required, so long as the orientation was correct and there was no movement relative to the shoe.

7.3 Desktop PC System

A desktop PC system was developed to collect data from the gyroscope sensor, two footswitches and a ground force reaction (GRF) platform. The gyroscope sensor and
footswitches were connected to the PC via a 10m cable. Software was used to control
data collection and also to process the gyroscope output in real time.

Data was initially collected from seven normal subjects who walked on a level surface
only. The system detected the gait events HC, FF, HR and TO of all subjects, and in
comparison with measurements from a GRF platform the HC and TO difference was a
maximum of ±25ms. The detection algorithm (originally written in LabVIEW) was
converted to 'C' as this text based language was more suited to the rule based detection
algorithm, and therefore enabled modifications to be made more easily and execution to
be performed more quickly. Data was then collected from ten hemiplegic subjects who
also walked on a level surface only: seven had suffered a stroke, one had MS and two
had incomplete SCI injuries. A maximum difference of ±100ms (average difference
32.4ms) was observed in comparison with measurements from heel and toe
footswitches.

7.4 Portable System

A portable system was developed to perform all of the functions that the desktop PC
system was capable of (with the exception of data collection from the GRF platform).
The unit is small and lightweight enough to be worn by the subject in a shoulder pouch
about the waist, and thin 1m cables connect it to the gyroscope sensor and heel and toe
footswitches. Data is sampled at 164Hz, and is written to a 2M PCMCIA card providing
over 40 minutes of sampling time. The card may then be connected to a PC for data
analysis using a card reader and dedicated software. Power is provided by a 9V PP3
battery that can sustain the unit for over 1 hour. The system is microcontroller based,
and is capable of executing an events detection algorithm. An RS232 communication
link is provided to allow control parameters (used to delay the onset of event detection)
to be downloaded from a PC.

The portable system has two digital outputs allowing connection to a stimulator unit,
such as the Compustim 10B (Micheal, 1996). However, the system was not used with a
stimulator in this study as it was not deemed necessary during this stage of evaluation.
Conclusions

Without the data-logging facilities, the electronic components necessary for event detection may be housed within a stimulator unit.

Incorrectly detected events could result in electrical stimulation being applied at the wrong moment, which would be potentially dangerous for the patient (e.g. during the stance phase). Some safety features were incorporated into the software algorithm to attempt to avoid this: detection of a particular event was dependant upon a number of conditions – if the initial condition was met but subsequent conditions were not, the stimulation could be cancelled in a very short space of time minimizing its effects.

The system was used to collect data from five normal subjects who walked at normal and slow speeds on a level surface, a ramp inclined at approximately 8° and a staircase with seven steps each with a 150mm tread depth. Data was compared with simultaneously collected heel and toe footswitch data and data collected by the MDS. Averages of all the collected data and of data collected from each terrain were calculated. Of all the collected data, the gyroscope data was found to have the lowest average difference with the MDS: 63ms (SD 78ms); the average difference between the footswitches and the MDS was 82ms (SD 88ms). For the data collected on the level surface only, the gyroscope sensor/MDS average difference was 35ms (SD 29ms) and the footswitches/MDS difference was 71ms (SD 76ms). Since the MDS measurement method was considered to be more accurate than the other methods, these results indicate that the gyroscope detected gait events with greater accuracy than the footswitches.

Data was also collected from four hemiplegic subjects: two had suffered a stroke and two had MS. The subjects were asked to walk (with and without FES) along a level surface whilst data was collected from the gyroscope sensor and heel and toe footswitches. The first two subjects (one stroke and one MS) were able to walk further, and therefore were also asked to walk up and down a slope and a staircase. Of all the data collected, the average difference between the two measurement methods was found to be 82ms (SD 73ms) without stimulation and 111ms (SD 109ms) with stimulation. However, the same comparison made without including the staircase data revealed that
the HC differences were only 48ms (SD 29ms) without stimulation and 26ms (SD 20ms) with stimulation. HR differences were greater: 135ms (SD 66ms) without stimulation and 156 (SD 57) with stimulation, although there is evidence to suggest that these greater differences may be due to footswitch inaccuracy (see section 6.3.3).

7.5 Research Hypothesis

The research hypothesis (which is divided into two sections) was first stated at the end of chapter 1:

1. The gyroscope sensor (combined with a supporting microcontroller) can be connected to a stimulator intended for use with footswitches. The sensor can also be donned and doffed with similar ease to that of footswitches.

2. The gyroscope sensor offers similar or improved stimulation timing and reliability compared to footswitches when used with an FES drop-foot correction system.

Although the sensor system was not used with a stimulator, the output was designed to be similar to that of a footswitch (i.e. high (+5V) and low (0V) voltages corresponding to the heel/toe on or off the ground (this can be changed by way of a simple software modification). The Velcro strap used to fasten the gyroscope sensor to the foot provided a fast and simple method of attachment (comparable to that of a footswitch), but was not cosmetically appealing.

The second half of the hypothesis is discussed in the following sections.

7.6 Observations

The final evaluation of the gyroscope sensor may be divided into two sections: (a) measurements taken from normal subjects and (b) measurements taken from hemiplegic subjects. Following analysis of the measurements, observations were made which are listed in the proceeding text.
Conclusions

Section (a):

- The gyroscope sensor detected gait events measured on the level surface and the ramp with greater accuracy than the footswitches. The footswitches detected events measured on the staircase with marginally greater accuracy.

- Only the gyroscope sensor left gait events undetected. FF was undetected on two occasions whilst one subject was descending the ramp at slow walking speed. All FF and HR events were missed while traversing the staircase, although in this case only two events usually occur within each stride: foot contact and foot rise.

- The gyroscope sensor performed relatively badly when detecting HR at a slow walking speed on the ramp (up and down). This is most likely to be due to the unsuitability of particularly slow heel movement under these conditions when using the gyroscope sensor detection algorithm. In comparison with the desktop PC system, there was an increased level of noise present in the gyroscope sensor output (less than 1% of FSD). The source of the noise was the master PIC’s analogue to digital converter (ADC), leading to the presence of noise during the stance phase (no foot movement). A slight delay was therefore required at the end of the stance phase before the detection algorithm could be sure that the foot was moving and assert the HR state.

- The gyroscope sensor performed best when subjects were walking at their normal self-selected speed. Performance was best on the level surface, followed by the ramp. When using the system on the staircase, the heel and toe detection signals had to be combined (using a logical AND function) to produce foot contact and foot rise.

- The footswitches detected HR on the level surface with much greater inaccuracy than the gyroscope sensor (HC accuracy was similar for both sensors, although the gyroscope performed marginally better).
Conclusions

Section (b):

- Of all the collected data, the average difference between HC event measurements on
  the level surface (and ramped surface) was much less than the average difference
  between HR events.

- The overall average difference (and SD) between measurements on the level surface
  (and ramped surface) was lower for HC events with stimulation.

The average difference between sensor measurements was less when the subject
traversed the level surface (and ramped surface) than when using the staircase.

- As before, the gyroscope sensor missed all FF and HR events while traversing the
  staircase (although these events do not usually occur in this case). In patients three
  and four both sensors missed events: FF was missed completely by both sensors in
  subject three and by the footswitches in subject four. HR was missed completely by
  both sensors when subject three walked with stimulation and by the gyroscope only
  without. Both subjects three and four were showing signs of fatigue before data
  collection began, and were unable to walk as far as subjects one and two. Subject
  three walked with excessively pathological gait, and in many cases did not make
  heel contact with the ground (the foot remained in plantar flexion).

7.7 Discussion

Data collected from normal subjects by the gyroscope sensor, MDS and footswitches
suggests that the gyroscope is able to detect gait events with greater accuracy than the
footswitches. However, this study also found that the gyroscope missed (a very low
number of) events and on some occasions detected HR with poor accuracy.

Data collected from the hemiplegic subjects demonstrated that the gyroscope sensor and
the footswitches both missed a similar number of (different) events. In this test the
overall average timing difference between the two types of sensor was 82ms (without
stimulation) and 111ms (with stimulation). A difference of 100ms between sensor
detection and the actual event may be considered acceptable for slower (pathological)
gait. However without a third, more accurate method of measurement (such as the MDS) it is not possible to determine which sensor was more accurate. Footswitch replacement with the gyroscope sensor therefore requires further consideration. Such considerations are discussed in chapter 8.

The reduced difference in HC detection times observed in FES assisted hemiplegic gait (compared with unassisted gait) is an important consideration. While it is possible that electrical stimulation improved the detection accuracy of the footswitches or the gyroscope sensor, it is also possible that they were both improved or that one or both was made worse. One reasonable explanation for the reduced difference is that the stimulation improved dorsiflexion at HC thereby reducing the time difference between HC and the footswitch detection of this event.
Chapter 8
Further Work

8.1 Introduction

In this chapter recommendations for further improvement and evaluation of the gyroscope sensor are presented. Regardless of any improvements made, the evaluation of the existing system is not complete (no trials with the system controlling a stimulator took place). However, it is recommended that these improvements be carried out first.

8.2 Improvements to the Existing System

Before the system was evaluated using the ramp and staircase, implementation of the detection algorithm was based upon data collected from many subjects walking on the level surface. Whilst acceptable results were found when using the system with the level surface, the ramp and staircase traverses produced much greater inaccuracies. It is recommended that analysis of the causes of poor HR detection during slow ramp walking be conducted. Other ramp events were also detected with marginally greater error compared to level surface walking. However, the angular velocity signal obtained while walking on the ramp is similar to that of level ground, and it is therefore likely that the existing algorithm could be improved and a more accurate and robust detection system could be developed that is suited to both types of terrain.

When traversing the staircase, the time between HC and FF, and HR and TO is very small, i.e. the pairs of events occur at about the same time (from MDS data). It may therefore be considered acceptable for only two events to be used for stimulation while walking on the staircase. However, the angular velocity signal produced by staircase walking was dissimilar from that obtained on level ground (although it also had a highly repetitive pattern), and therefore the existing algorithm was not suitable for this type of terrain. Gyroscope sensor detection accuracy could be improved by collecting data from staircase walking and developing an alternative algorithm for this type of terrain,
selected either using a switch mounted on the stimulator unit or automatically using more sophisticated software.

In the majority of cases, the gyroscope sensor detected events later than the MDS. By modifying the algorithm to use signal features that occur earlier, the gyroscope system could detect events at the time of occurrence or even before. In this way a variable delay could be added to suit the gait of each patient.

Improved reduction of the electrical noise produced by the master PIC’s analogue to digital converter (ADC) may lead to better detection accuracy. Noise present after FF (when no foot movement takes place) requires that detection of HR must be delayed in order to ensure that the change in output voltage is due to the foot moving, and not noise. The noise (which was not present in the desktop system) could possibly be removed by using a separate, superior ADC, allowing the length of the delay to be decreased.

The portable unit used to process the gyroscope output is quite large (110mm x 70mm x 20mm). Without the data-logging facilities the required hardware would be small enough to be housed inside a stimulator unit. This would be necessary for the next stage of evaluation of the unit (after the software modifications have been tested), in order for the amount of equipment carried by the subject to be kept to a minimum.

The PC software used to read data from the PCMCIA card is functional but not user friendly. A user interface panel, written in Visual C (Microsoft) or LabVIEW (National Instruments) would allow this program to be used by others more intuitively.

The temperature drift mentioned in section 2.5 may adversely affect the system in extreme weather conditions, such as cold winter days (the temperature difference between a heated room and an icy pavement) and warm summer days (an air conditioned room and a hot outdoors). A solution to this problem would be to compensate for this using a temperature transducer.
The existing master PIC software used wrote only to the first byte of the word at each memory location within the PCMCIA card. Since only 1Mb of the full 2Mb was being used, writing to the full word would double the length of data collection time using the portable unit.

### 8.3 Further Evaluation

The gyroscope system was evaluated over short walks (the maximum was just less than one minute) and three different types of terrain. Further evaluation of the system requires longer walks (maximum system sampling time is 40 minutes) over more varied terrain (rough ground, slopes at other inclinations and more complex maneuvers such as turning).

The method of determining gait events from the MDS may be automated using computer software to monitor the marker movement. In this study, gait events were determined by human observation. Replacing this method with an automatic system would reduce any measurement method inconsistency and greatly reduce the length of time required by the process.

In order to completely evaluate a stimulator system, the patient is required to give their comments regarding their experience of FES assisted walking. Although the stimulator may appear to be offering benefit to the patient, it may be the case that patient thinks otherwise. Further system evaluation therefore requires that the stimulation output channels from the datalogger be connected to a patient stimulator. This may be achieved easily as the datalogger output is either high [+5V] or low [0V], and the stimulator will respond when the voltage crosses a threshold such as 2.5V.

### 8.4 A Shank Based Detection System

As discussed in chapter 4, alternative sites for the gyroscope sensor are the thigh and the shank. One significant advantage of using the shank position is the elimination of the requirement for connecting leads between the sensor and the stimulator. Although figures 4.3 and 4.4 show that gait events detection using the gyroscope at the shank is
more complex that at the foot, it is still possible to determine HC and TO (it may also be possible and therefore useful to estimate FF and HR). Such work has already commenced at the University of Surrey by Salim Ghoussayni - his work is based upon an algorithm developed in this thesis. The results thus far are encouraging.
Appendix A

Graphs of Gyroscope Sensor – Marker Detection System Comparison During Walking on a Level Surface at Slow Speed

**Figure 9.1** Graph of Range of Differences in HC Event Detection Times (Gyroscope Sensor – Marker Detection System) as a Percentage of Total Events

**Figure 9.2** Graph of Range of Differences in FF Event Detection Times (Gyroscope Sensor – Marker Detection System) as a Percentage of Total Events
Figure 9.3 Graph of Range of Differences in HR Event Detection Times (Gyroscope Sensor – Marker Detection System) as a Percentage of Total Events

Figure 9.4 Graph of Range of Differences in TO Event Detection Times (Gyroscope Sensor – Marker Detection System) as a Percentage of Total Events
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Graphs of Gyroscope Sensor – Marker Detection System Comparison During Walking on a Flat Surface at Normal Speed

**Figure 9.5** Graph of Range of Differences in HS Event Detection Times (Gyroscope Sensor – Marker Detection System) as a Percentage of Total Events

**Figure 9.6** Graph of Range of Differences in FF Event Detection Times (Gyroscope Sensor – Marker Detection System) as a Percentage of Total Events
Figure 9.7 Graph of Range of Differences in HR Event Detection Times (Gyroscope Sensor – Marker Detection System) as a Percentage of Total Events

Figure 9.8 Graph of Range of Differences in TO Event Detection Times (Gyroscope Sensor – Marker Detection System) as a Percentage of Total Events
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Graphs of Gyroscope Sensor – Marker Detection System Comparison During Walking on a Ramp at Slow Speed

**Figure 9.9** Graph of Range of Differences in HC Event Detection Times (Gyroscope Sensor – Marker Detection System) as a Percentage of Total Events

**Figure 9.10** Graph of Range of Differences in FF Event Detection Times (Gyroscope Sensor – Marker Detection System) as a Percentage of Total Events
Figure 9.11 Graph of Range of Differences in HR Event Detection Times (Gyroscope Sensor – Marker Detection System) as a Percentage of Total Events

Figure 9.12 Graph of Range of Differences in TO Event Detection Times (Gyroscope Sensor – Marker Detection System) as a Percentage of Total Events
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Graphs of Gyroscope Sensor – Marker Detection System Comparison During Walking on a Ramp at Normal Speed

**Figure 9.13** Graph of Range of Differences in HC Event Detection Times (Gyroscope Sensor – Marker Detection System) as a Percentage of Total Events

**Figure 9.14** Graph of Range of Differences in FF Event Detection Times (Gyroscope Sensor – Marker Detection System) as a Percentage of Total Events
Figure 9.15 Graph of Range of Differences in HR Event Detection Times (Gyroscope Sensor – Marker Detection System) as a Percentage of Total Events

Figure 9.16 Graph of Range of Differences in TO Event Detection Times (Gyroscope Sensor – Marker Detection System) as a Percentage of Total Events
Graphs of Gyroscope Sensor – Marker Detection System
Comparison During Walking on a Staircase at Slow Speed

Figure 9.17 Graph of Range of Differences in HC Event Detection Times (Gyroscope Sensor – Marker Detection System) as a Percentage of Total Events

Figure 9.18 Graph of Range of Differences in FF Event Detection Times (Gyroscope Sensor – Marker Detection System) as a Percentage of Total Events
**Figure 9.19** Graph of Range of Differences in HR Event Detection Times (Gyroscope Sensor – Marker Detection System) as a Percentage of Total Events

**Figure 9.20** Graph of Range of Differences in TO Event Detection Times (Gyroscope Sensor – Marker Detection System) as a Percentage of Total Events
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Graphs of Gyroscope Sensor – Marker Detection System Comparison During Walking on a Staircase at Normal Speed

**Figure 9.21** Graph of Range of Differences in HC Event Detection Times (Gyroscope Sensor – Marker Detection System) as a Percentage of Total Events

**Figure 9.22** Graph of Range of Differences in FF Event Detection Times (Gyroscope Sensor – Marker Detection System) as a Percentage of Total Events
Figure 9.23 Graph of Range of Differences in HR Event Detection Times (Gyroscope Sensor – Marker Detection System) as a Percentage of Total Events

Figure 9.24 Graph of Range of Differences in TO Event Detection Times (Gyroscope Sensor – Marker Detection System) as a Percentage of Total Events
Appendix A

Graphs of Gyroscope Sensor – Footswitches Comparison During Walking on a Flat Surface at Slow Speed

Figure 9.25 Graph of Range of Differences in HC Event Detection Times (Gyroscope Sensor - Footswitches) as a Percentage of Total Events

Figure 9.26 Graph of Range of Differences in FF Event Detection Times (Gyroscope Sensor - Footswitches) as a Percentage of Total Events
Figure 9.27 Graph of Range of Differences in HR Event Detection Times (Gyroscope Sensor - Footswitches) as a Percentage of Total Events

Figure 9.28 Graph of Range of Differences in TO Event Detection Times (Gyroscope Sensor - Footswitches) as a Percentage of Total Events
Graphs of Gyroscope Sensor – Footswitches Comparison During Walking on a Flat Surface at Normal Speed

**Figure 9.29** Graph of Range of Differences in HC Event Detection Times (Gyroscope Sensor - Footswitches) as a Percentage of Total Events

**Figure 9.30** Graph of Range of Differences in FF Event Detection Times (Gyroscope Sensor - Footswitches) as a Percentage of Total Events
Figure 9.31 Graph of Range of Differences in HC Event Detection Times (Gyroscope Sensor - Footswitches) as a Percentage of Total Events

Figure 9.32 Graph of Range of Differences in TO Event Detection Times (Gyroscope Sensor - Footswitches) as a Percentage of Total Events
Graphs of Gyroscope Sensor – Footswitches Comparison During Walking on a Ramp at Slow Speed

**Figure 9.33** Graph of Range of Differences in HC Event Detection Times (Gyroscope Sensor - Footswitches) as a Percentage of Total Events

**Figure 9.34** Graph of Range of Differences in FF Event Detection Times (Gyroscope Sensor - Footswitches) as a Percentage of Total Events
Figure 9.35 Graph of Range of Differences in HR Event Detection Times (Gyroscope Sensor - Footswitches) as a Percentage of Total Events

Figure 9.36 Graph of Range of Differences in TO Event Detection Times (Gyroscope Sensor - Footswitches) as a Percentage of Total Events
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Graphs of Gyroscope Sensor – Footswitches Comparison During Walking on a Ramp at Normal Speed

Figure 9.37 Graph of Range of Differences in HC Event Detection Times (Gyroscope Sensor - Footswitches) as a Percentage of Total Events

Figure 9.38 Graph of Range of Differences in FF Event Detection Times (Gyroscope Sensor - Footswitches) as a Percentage of Total Events
Figure 9.39 Graph of Range of Differences in HR Event Detection Times (Gyroscope Sensor - Footswitches) as a Percentage of Total Events

Figure 9.40 Graph of Range of Differences in TO Event Detection Times (Gyroscope Sensor - Footswitches) as a Percentage of Total Events
Graphs of Gyroscope Sensor – Footswitches Comparison During Walking on a Staircase at Slow Speed

**Figure 9.41** Graph of Range of Differences in HC Event Detection Times (Gyroscope Sensor - Footswitches) as a Percentage of Total Events

**Figure 9.42** Graph of Range of Differences in FF Event Detection Times (Gyroscope Sensor - Footswitches) as a Percentage of Total Events
Figure 9.43 Graph of Range of Differences in HR Event Detection Times (Gyroscope Sensor - Footswitches) as a Percentage of Total Events

Figure 9.44 Graph of Range of Differences in TO Event Detection Times (Gyroscope Sensor - Footswitches) as a Percentage of Total Events
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Graphs of Gyroscope Sensor – Footswitches Comparison During Walking on a Staircase at Normal Speed

Figure 9.45 Graph of Range of Differences in HC Event Detection Times (Gyroscope Sensor - Footswitches) as a Percentage of Total Events

Figure 9.46 Graph of Range of Differences in FF Event Detection Times (Gyroscope Sensor - Footswitches) as a Percentage of Total Events
Figure 9.47 Graph of Range of Differences in HR Event Detection Times (Gyroscope Sensor - Footswitches) as a Percentage of Total Events

Figure 9.48 Graph of Range of Differences in TO Event Detection Times (Gyroscope Sensor - Footswitches) as a Percentage of Total Events
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Graphs of Marker Detection System – Footswitches Comparison During Walking on a Flat Surface at Slow Speed

**Figure 9.49** Graph of Range of Differences in HC Event Detection Times (Marker Detection System - Footswitches) as a Percentage of Total Events

**Figure 9.50** Graph of Range of Differences in FF Event Detection Times (Marker Detection System - Footswitches) as a Percentage of Total Events
Figure 9.51 Graph of Range of Differences in HR Event Detection Times (Marker Detection System - Footswitches) as a Percentage of Total Events

Figure 9.52 Graph of Range of Differences in TO Event Detection Times (Marker Detection System - Footswitches) as a Percentage of Total Events
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Graphs of Marker Detection System – Footswitches Comparison During Walking on a Flat Surface at Normal Speed

Figure 9.53 Graph of Range of Differences in HC Event Detection Times (Marker Detection System - Footswitches) as a Percentage of Total Events

Figure 9.54 Graph of Range of Differences in FF Event Detection Times (Marker Detection System - Footswitches) as a Percentage of Total Events
Figure 9.55 Graph of Range of Differences in HR Event Detection Times (Marker Detection System - Footswitches) as a Percentage of Total Events

Figure 9.56 Graph of Range of Differences in TO Event Detection Times (Marker Detection System - Footswitches) as a Percentage of Total Events
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Graphs of Marker Detection System – Footswitches Comparison During Walking on a Ramp at Slow Speed

Figure 9.57 Graph of Range of Differences in HC Event Detection Times (Marker Detection System - Footswitches) as a Percentage of Total Events

Figure 9.58 Graph of Range of Differences in FF Event Detection Times (Marker Detection System - Footswitches) as a Percentage of Total Events
Figure 9.59 Graph of Range of Differences in HR Event Detection Times (Marker Detection System - Footswitches) as a Percentage of Total Events

Figure 9.60 Graph of Range of Differences in TO Event Detection Times (Marker Detection System - Footswitches) as a Percentage of Total Events
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Graphs of Marker Detection System – Footswitches Comparison During Walking on a Ramp at Normal Speed

Figure 9.61 Graph of Range of Differences in HC Event Detection Times (Marker Detection System - Footswitches) as a Percentage of Total Events

Figure 9.62 Graph of Range of Differences in FF Event Detection Times (Marker Detection System - Footswitches) as a Percentage of Total Events
Figure 9.63 Graph of Range of Differences in HR Event Detection Times (Marker Detection System - Footswitches) as a Percentage of Total Events

Figure 9.64 Graph of Range of Differences in TO Event Detection Times (Marker Detection System - Footswitches) as a Percentage of Total Events
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Graphs of Marker Detection System – Footswitches Comparison During Walking on a Staircase at Slow Speed

**Figure 9.65** Graph of Range of Differences in HC Event Detection Times (Marker Detection System Sensor - Footswitches) as a Percentage of Total Events

**Figure 9.66** Graph of Range of Differences in FF Event Detection Times (Marker Detection System Sensor - Footswitches) as a Percentage of Total Events
**Figure 9.67** Graph of Range of Differences in HR Event Detection Times (Marker Detection System Sensor - Footswitches) as a Percentage of Total Events

**Figure 9.68** Graph of Range of Differences in TO Event Detection Times (Marker Detection System Sensor - Footswitches) as a Percentage of Total Events
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Graphs of Marker Detection System - Footswitches Comparison During Walking on a Staircase at Normal Speed

**Figure 9.69** Graph of Range of Differences in HC Event Detection Times (Marker Detection System Sensor - Footswitches) as a Percentage of Total Events

**Figure 9.70** Graph of Range of Differences in HC Event Detection Times (Marker Detection System Sensor - Footswitches) as a Percentage of Total Events
Figure 9.71 Graph of Range of Differences in HR Event Detection Times (Marker Detection System Sensor - Footswitches) as a Percentage of Total Events

Figure 9.72 Graph of Range of Differences in TO Event Detection Times (Marker Detection System Sensor - Footswitches) as a Percentage of Total Events
Appendix B

'C' Program to Sample from ADC Channels and Apply Rule-Based Algorithm Running Within LabVIEW

```c
#include "extcode.h"
#include "nidaqex.h"
#include <stdio.h>
#include <stdlib.h>
#include <time.h>
#include <string.h>

#define numberOfSamplesInFile 100000  // Max 3 minutes

void *sampleAndStimulate (float, float);
void sleep(clock_t wait);
void readBuffer();
void resetArray();
void getOffset();
void createOffset();
double* getDataValuesFromFile(char fileName[100]);
void readFromArrayAndStimulate();
void analyseData(float voltage);

static i16 piBuffer[100000] = {0};
static f64 pdVoltBuffer[100000] = {0.0};

float64 *resultElmtp, *heelSwitchData, *toeSwitchData, *leftForcePlatform,
*rightForcePlatform,
int loopCount, loopCount1, loopCount2, loopCount3, loopCount4, dataSource;

double *HeelP, *ToeP, *startAddress, offset = 0;

typedef struct {
    int32 dimSize;
    float64 arg [1];
} TD1;
typedef TD1 **TD1Hdl;

CIN MgErr CINRun(TD1Hdl Velocity, TD1Hdl Heel, TD1Hdl Toe, TD1Hdl
HeelSwitchData, TD1Hdl ToeSwitchData, TD1Hdl LeftForcePlatform,
TD1Hdl RightForcePlatform,
LVBoolean *MEASURE_OFFSET, LVBoolean
*DATA_SOURCE_GYRO_FILE,
    LStrHandle var10, float64 *HC, float64 *FF, float64
*HR, float64 *TO,
```
float64 *SAMPLING_TIME_S_, float64 *HEEL_ON);

#define ParamNumber 0 /* The return parameter is parameter 0 */

CIN MgErr CINRun(TD1Hdl Velocity, TD1Hdl Heel, TD1Hdl Toe, TD1Hdl HeelSwitchData, TD1Hdl ToeSwitchData, TD1Hdl LeftForcePlatform, TD1Hdl RightForcePlatform, LVBoolean *MEASURE_OFFSET, LVBoolean *DATA_SOURCE_GYRO_FILE, LStrHandle varIO, float64 *HC, float64 *FF, float64 *HR, float64 *TO, float64 *SAMPLING_TIME_S_, float64 *HEEL_ON)
{
    char fileName[100];

    MgErr mgError=noErr;

    if (mgError = SetCINArraySize((UHandle)Velocity, ParamNumber, numberOfSamplesInFile))
        goto out;

    if (mgError = SetCINArraySize((UHandle)Heel, ParamNumber, numberOfSamplesInFile))
        goto out;

    if (mgError = SetCINArraySize((UHandle)Toe, ParamNumber, numberOfSamplesInFile))
        goto out;

    if (mgError = SetCINArraySize((UHandle)HeelSwitchData, ParamNumber, numberOfSamplesInFile))
        goto out;

    if (mgError = SetCINArraySize((UHandle)ToeSwitchData, ParamNumber, numberOfSamplesInFile))
        goto out;

    if (mgError = SetCINArraySize((UHandle)LeftForcePlatform, ParamNumber, numberOfSamplesInFile))
        goto out;

    if (mgError = SetCINArraySize((UHandle)RightForcePlatform, ParamNumber, numberOfSamplesInFile))
        goto out;

    startAddress = (double *) malloc(sizeof(double)*numberOfSamplesInFile);
    HeelP = (double *) malloc(sizeof(double)*numberOfSamplesInFile);
    ToeP = (double *) malloc(sizeof(double)*numberOfSamplesInFile);
(*Velocity)-dimSize = numberOfSamplesInFile;
(*Heel)-dimSize = numberOfSamplesInFile;
(*Toe)-dimSize = numberOfSamplesInFile;
(*HeelSwitchData)-dimSize = numberOfSamplesInFile;
(*ToeSwitchData)-dimSize = numberOfSamplesInFile;
(*LeftForcePlatform)-dimSize = numberOfSamplesInFile;
(*RightForcePlatform)-dimSize = numberOfSamplesInFile;

resultElmtp = (*Velocity)->arg1;
HeelP = (*Heel)->arg1;
ToeP = (*Toe)->arg1;
heelSwitchData = (*HeelSwitchData)->arg1;
toeSwitchData = (*ToeSwitchData)->arg1;
leftForcePlatform = (*LeftForcePlatform)->arg1;
rightForcePlatform = (*RightForcePlatform)->arg1;

P1P = HC;
P2P = FF;
P3P = HR;
P4P = TO;
P5P = HEEL_ON;

strcpy (fileName, (LStrBuf(*var10))); // *DATA_FILENAME ));

if (*DATA_SOURCE_GYRO_FILE == LVTRUE)
{
    dataSource = 1;
    startAddress = getDataValuesFromFile(fileName);
}
else
    dataSource = 0;

resetArray();

if (*MEASURE_OFFSET == LVTRUE) //If offset to be measured, make sample time
{
    *SAMPLING_TIME_S_ = 1; //equal to 1 second (100 samples)
    offset = 0;
    //Make offset = 0 incase offset to be remeasured
}
else
    getOffset();

if ( dataSource == 0)
Appendix B

```c
void *sampleAndStimulate(float sampleTime, float offset)
{
    i16 iStatus = 0;
i16 iRetVal = 0;
i16 iDevice = 1;
i16 iChan = 0;
i16 iGain = 1;
f64 dSampRate = 100.0;
u32 ulCount = 100000;
f64 dGainAdjust = 1.0;
f64 dOffset = 0.0;
i16 iUnits = 0;
i16 iSampTB = 1;
u16 uSampInt = 10;
i16 iScanTB = 0;
u16 uScanInt = 0;
i16 iDAQstopped = 0;
u32 ulRetrieved = 0;
i16 iNumMUXBrds = 0;
i16 iNumChans = 1;
static i16 piChanVect[2] = {1, 2};
static i16 piGainVect[2] = {1, 1};
i16 iIgnoreWarning = 0;
i16 iYieldON = 1;
    int SampleNumber;
}
```

```c
sampleAndStimulate(*SAMPLING_TIME_S_, offset);
//Sample

gyro at 100Hz for duration *TIME_TO_SAMPLE_S_
else
    readFromArrayAndStimulate();

    if (*MEASURE_OFFSET == LVTRUE) //If offset
to be measured,
        createOffset(); //create
    and save value to file

    out:
        return mgError;
    }

//*******************************************************************
****************************************
```
iStatus = DAQ_Rate(dSampRate, iUnits, &iSampTB, &uSampInt);

iRetVal = NIDAQErrorHandler(iStatus, "SCAN_Setup", iIgnoreWarning);

// Acquire data from multiple channels.

iStatus = Lab_ISCAN_Start(iDevice, iNumChans, 1, piBuffer, ulCount, iSampTB, uSampInt, uScanInt);

iRetVal = NIDAQErrorHandler(iStatus, "Lab_ISCAN_Start", iIgnoreWarning);

// Loop until all acquisition is complete. HINT: You can be doing other foreground tasks during this time.

iStatus = DAQ_Check(iDevice, &iDAQstopped, &ulRetrieved);

// Set up data acquisition from multiple channels
iRetVal = NIDAQYield(iYieldON);

SampleNumber = sampleTime * 100;  //Sample at 100Hz

sleep(100);  // Pauses for 10 ms (to give the buffer a chance to fill up)

for (loopCount=0; loopCount<SampleNumber; loopCount++)
{
    sleep(10);  //Pauses for 10 ms
    readBuffer();  //Reads data from buffer, scales it to give voltage,

    analyseData(resultElmtp[loopCount]);  //and subtracts the offset
}

iStatus = SCAN_Demux(piBuffer, ulCount, iNumChans, iNumMUXBrds);

iRetVal = NIDAQErrorHandler(iStatus, "SCAN_Demux", iIgnoreWarning);

iStatus = DAQ_VScale(iDevice, iChan, iGain, dGainAdjust, dOffset, ulCount, piBuffer, pdVoltBuffer);

// Stops scan
iRetVal = NIDAQErrorHandler(iStatus, "DAQ_VScale", ilgnoreWarning);

// CLEANUP - Don't check for errors on purpose.

iStatus = DAQ_Clear(iDevice);

//*******************************************************************
void sleep(clock_t wait) // Pauses for a specified number of milliseconds
{
    clock_t goal;
    goal = wait + clock();
    while( goal > clock() );
}

//*******************************************************************
void readBuffer()
{
    resultElmtp[loopCount] = piBuffer[loopCount];
    resultElmtp[loopCount] = ((resultElmtp[loopCount] - 47) / 410) * -1;
    resultElmtp[loopCount] = resultElmtp[loopCount] - offset;
}

//*******************************************************************
void resetArray() //Make all data array elements zero
{
    int loopCount;
    for (loopCount=0; loopCount<numberOfSamplesInFile; loopCount++)
    {
        resultElmtp[loopCount] = 0; //Reset array (all elements = 0)
    }
}
Appendix B

void getOffset() //Read previously measured offset from file
{
    FILE *stream = NULL;
    {
        stream = fopen("e:\Temp\OFFSET.TXT", "r");
        fscanf(stream, "%f", &offset);
        fclose (stream);
    }
}

void createOffset() //Create offset and save value to file
{
    int loopCount;
    FILE *stream = NULL;
    for (loopCount=0; loopCount<100; loopCount++)
    {
        offset += resultElmp[loopCount]; //sum all (100) measurements
    }
    offset = offset /100; //divide by 100
    stream = fopen("e:\Temp\OFFSET.TXT", "w"); //and save average to file
    fprintf(stream, "%f”, offset);
    fclose (stream);
}

double* getDataValuesFromFile(char name[100]) //Read previously measured velocity values from file
{
    char nameOfFile[100];
    FILE *stream;
    double d, *wholeArrayData;
    int fileNameLength, fileArrayCount, fileNameArray = 0;

    wholeArrayData = (double *) malloc(sizeof(double)*numberOfSamplesInFile); //Reserve memory for whole of
//input file data
strcpy (nameOfFile, name);
fileNameLength = strlen (nameOfFile);
//Length of string containing filename
for (fileNameArray = 0; fileNameArray<fileNameLength; fileNameArray++)
{
    if ((nameOfFile[fileNameArray-2] == 'x') && (nameOfFile[fileNameArray-
        1] == 'l')
        && (nameOfFile[fileNameArray] == 's'))
        //Search for 'xls'
        nameOfFile[fileNameArray+1] = '\0';
        //Add termination character
}

stream = fopen ( nameOfFile, "r" );
for (fileArrayCount=0; fileArrayCount<numberOfSamplesInFile; fileArrayCount++)
{
    fscanf(stream, "%Lf", &wholeArrayData[fileArrayCount]); //Read in all data to "wholeArrayData"
}
fclose (stream);

// stream = fopen ( "d:\CurrentLabviewFiles\FOOT DROP\Data2\test2.xls", "w" );
// fprintf(stream, "%s", nameOfFile);
// fclose (stream);
return &wholeArrayData[0];
    //Returns start address of data array
}

//*******************************************************************
void readFromArrayAndStimulate()
{
    int colThreeCount = 0, numberOfDataFileSamples = 0;

    loopCount = 0;
    loopCount1 = 0;
    loopCount2 = 0;
    loopCount3 = 0;
    loopCount4 = 0;

    for (numberOfDataFileSamples=0;
        numberOfDataFileSamples<numberOfSamplesInFile;
        numberOfDataFileSamples++)
Appendix B

```c
{
    colThreeCount++;
    // Count from 1-7
    if (colThreeCount == 3)
        // Selects column 3
        {
            resultElmtp[loopCount] = startAddress[numberOfDataFileSamples]; // Copy selected column to output array
            analyseData(resultElmtp[loopCount]); // Analyse data
            loopCount++;
            // Increment pointer
        }
    if (colThreeCount == 4)
        // Selects column 4
        {
            heelSwitchData[loopCount1] = startAddress[numberOfDataFileSamples]; // Copy selected column to output array
            loopCount1++;
            // Increment pointer
        }
    if (colThreeCount == 5)
        // Selects column 5
        {
            toeSwitchData[loopCount2] = startAddress[numberOfDataFileSamples]; // Copy selected column to output array
            loopCount2++;
            // Increment pointer
        }
    if (colThreeCount == 6)
        // Selects column 6
        {
            leftForcePlatform[loopCount1] = startAddress[numberOfDataFileSamples]; // Copy selected column to output array
            loopCount3++;
            // Increment pointer
        }
    if (colThreeCount == 7)
        // Selects column 7
        {
```
rightForcePlatform[loopCount2] = startAddress[numberOfFileSamples];  //Copy selected column to output array

loopCount4++;  //Increment pointer

if (colThreeCount == 7)  //Reset 1-7 counter
    colThreeCount = 0;

/**
***

void analyseData (float velocity)  //Voltage passed to this function

    //which is used to determine whether the
    //heel and toe are touching the ground

    // #define FALSE 0  Not necessary in MS C++ enviroment
    // #define TRUE 1  becuase TRUE and FALSE exist as keywords
{

    float static previousVelocity;

    unsigned static int logicHeelOnOff, logicToeOnOff, swingPhaseCount, swingPhaseDetected,
                 stancePhaseCount, midStancePhaseCount, midStancePhase,
                 midStancePhaseFound,
                 stancePhaseDetected, heelOffCount, toeOnCount, readyForToeOn,
                 ReadyForMidStance, footFlat, heelRise,
                 toeOffCount, lowPointReached, notSoLowPointReached, footStill,
                 heelOnTime, toeOnTime, detectionStarted;

    zero:
    if (loopCount == 0)
    {
        swingPhaseDetected = FALSE;
        swingPhaseCount = 0;
        stancePhaseDetected = FALSE;
        stancePhaseCount = 0;
        logicHeelOnOff = FALSE;
        logicToeOnOff = FALSE;
        heelOffCount = 0;
        midStancePhaseCount = 0;
    

midStancePhase = FALSE;
midStancePhaseFound = FALSE;
toeOnCount = 0;
toeOffCount = 0;
readyForToeOn = FALSE;
ReadyForMidStance = FALSE;
footFlat = FALSE;
heelRise = FALSE;
lowPointReached = TRUE;
previousVelocity = 0;
notSoLowPointReached = FALSE;
footStill = 0;
heelOnTime = 0;
toeOnTime = 0;
detectionStarted = FALSE;
}

goto jump;  //***********NEW*************

if (heelRise == TRUE)  //Toe Off
Detection
{
  if (velocity < -0.5)  //Velocity usually
goes this low
    lowPointReached = TRUE;
  if (velocity < -0.2)
    notSoLowPointReached = TRUE;  //If not, try this
one
  if (((velocity > (previousVelocity + 0.1)) && (lowPointReached ==
TRUE))
    || ((velocity > -0.1) && (notSoLowPointReached == TRUE)))

    /*LOWPOINTERACHED" Test velocity gradient (is it steep enough?)
    toeOffCount++;
  else
    toeOffCount = 0;
  if (toeOffCount > *P4P)  //**************
    /*Test
"ToeOffCount", or "NOTSOLOWPOINTERACHED" and velocity > -0.1
    logicToeOnOff = FALSE;  //Assert Toe Off
}

//***************

//if ((heelOnTime >= *P5P) && ((midStancePhase == TRUE) ||
if (((ReadyForMidStance == TRUE) && (velocity < -0.2)) //))
{

/*
if (velocity < -0.1) //Heel Rise
Detection
{
    stancePhaseDetected = FALSE; //Stance phase
finished
if (velocity < -0.1)
    heelOffCount++; if (velocity < -0.2) //Rewards for
lower velocities
    heelOffCount++; if (velocity < -0.3)
    heelOffCount++; //if (heelOffCount > *P3P)
    //***************
    
    logicHeelOnOff = FALSE; //Assert
Heel Off
    heelRise = TRUE; //Heel-
rise occurring
    midStancePhase = FALSE; //Mid-
Stance phase over
    ReadyForMidStance = FALSE; //Not ready for Mid-Stance
}
}

//*******************

if (ReadyForMidStance == TRUE) //Mid Stance
Detection
{
    if (velocity == 0) //Difference between this and the 'foot inactive'
    
        //detection is that this must occur between foot flat
        midStancePhaseCount++; //and heel
        rise (unless velocity goes less than -0.2)
        if (midStancePhaseCount > 20) //
        
            midStancePhase = TRUE;
            footFlat = TRUE;
            ReadyForMidStance = FALSE;
    
    }
}

//*******************
if (stancePhaseDetected == TRUE) //Foot Flat Detection
{
    if (velocity >= -0.1) //check for velocity going positive
    {
        toeOnCount++; //Foot Flat Detection
        if (toeOnCount > (1 + *P2P)) //change this value to delay foot flat detect
        {
            logicToeOnOff = TRUE;
            ReadyForMidStance = TRUE;
            stancePhaseDetected = FALSE;
        }
    }
    else toeOnCount = 0;
}

//*************************

if (swingPhaseDetected == TRUE) //Heel Strike Detection
{
    if (velocity <= 0) //Check for velocity going negative (was -0.1)
    {
        stancePhaseCount++; //Rewards for lower velocities
        if (velocity < -0.1)
            stancePhaseCount++; //Rewards for lower velocities
        if (velocity < -0.2)
            stancePhaseCount++; //Rewards for lower velocities
        if (velocity < -0.3)
            stancePhaseCount++; //Rewards for lower velocities
        if (velocity < -0.4)
            stancePhaseCount++; //Rewards for lower velocities
        if (stancePhaseCount > *P1P) //Increase value to delay heel strike detect
        {
            logicHeelOnOff = TRUE;
            stancePhaseDetected = TRUE;
            swingPhaseDetected = FALSE;
        }
    }
}

//*************************

jump: //*********NEW***********
// if (velocity == 0) //ORIGINAL
// Foot inactive (assert heel/toe On) Detection
if ((velocity < 0.25) && (velocity > -0.25))
footStill++;
else
footStill = 0;

//***********NEW************* (was not blanked out before)
// if (((velocity > 0.1) || (velocity < -0.1)) && (detectionStarted == FALSE))

if ((velocity > 0.25) || (velocity < -0.25))
{
    logicHeelOnOff = FALSE;
    logicToeOnOff = FALSE;
}

if (footStill > 10) //Must be stil for 20 counts
{
    logicHeelOnOff = TRUE;
    logicToeOnOff = TRUE;
}

if (velocity > 0.1) //Swing Phase Detection
{
    swingPhaseCount++; //Test whether velocity is positive for 20 counts
    if (velocity > 0.2) //Rewards
        swingPhaseCount++;
    if (velocity > 0.4)
        swingPhaseCount++;
    if (swingPhaseCount > 20)
    {
        stancePhaseDetected = FALSE; //Reset variables
        swingPhaseDetected = TRUE;
        stancePhaseCount = 0;
        midStancePhase = FALSE;
        midStancePhaseCount = 0;
        heelOffCount = 0;
        toeOnCount = 0;
        toeOffCount = 0;
        swingPhaseCount = 0;
        lowPointReached = FALSE;
    }
 notSoLowPointReached = FALSE;
logicToeOnOff = FALSE;
heelRise = FALSE;
logicHeelOnOff = FALSE;
footStill = 0;
detectionStarted = TRUE;

else
   swingPhaseCount = 0;

previousVelocity = velocity;

//end of rules section****************************

eleven:
if (logicHeelOnOff == TRUE)
{
    HeelP[loopCount] = 5.1;
    heelOnTime++;
}
else
{
    HeelP[loopCount] = 0;
    heelOnTime = 0;
}

if (logicToeOnOff == TRUE)
{
    ToeP[loopCount] = 5;
    toeOnTime++;
}
else
{
    ToeP[loopCount] = 0;
    toeOnTime = 0;
}
Assembly Language Program Running on Master PIC

LIST P=16C73, F=INHX8M

include "picreg.equ"

TEMP equ 20h ; Temporary storage variable
COUNT equ 21h
Velocity equ 22h
VAR1 equ 23h
VAR2 equ 24h
BYTE_A equ 25h
BYTE_B equ 26h
BYTE_C equ 27h
Current equ 28h
Count1equ 29h
Count2equ 2Ah
Count3equ 2Bh
TEMP2 equ 2Ch
CompRes equ 2Dh
heelRise equ 2Eh
heelOnTime equ 2Fh
logicHeelOnOff equ 30h
toeOnCount equ 31h
logicToeOnOff equ 32h
stancePhaseDetected equ 33h
previousVelocity equ 34h
lowPointReached equ 35h
notSoLowPointReached equ 36h
toeOffCount equ 37h
logicOffCount equ 38h
ReadyForMidStance equ 39h
MEMCYCF equ 3Ah
midStancePhase equ 3Bh
heelOffCount equ 3Ch
midStancePhaseCount equ 3Dh
footFlat equ 3Eh
stancePhaseCount equ 3Fh
swingPhaseDetected equ 40h
footStill equ 41h
detectionStarted equ 42h
swingPhaseCount equ 43h
finishFlag equ 44h
TEMP4 equ 45h
LEDOnTime equ 46h
LEDOffTime equ 47h
callibrateReady equ 48h
Appendix C

rs232Flag equ 49h
control1 equ 4Ah
control2 equ 4Bh
control3 equ 4Ch
control4 equ 4Dh
control5 equ 4Eh
whichControl equ 4Fh
numContToggle equ 50h
offset equ 51h
intCount equ 52h
VarToWrite equ 53h
HeelOnOff equ 54h
ToeOnOff equ 55h
long equ 56h
long2 equ 57h
readyToChangeMode equ 58h
VarToRead equ 59h
toeOnTime equ 5Ah
control6 equ 5Bh
offsetSum equ 5Ch
offsetSumLarge equ 5Dh
offsetSumCount equ 5Eh
shiftCount equ 5Fh
VeryLowPointReached equ 60h
firstTimeCal equ 61h
firstLowFound equ 62h
stairModeSelected equ 63h
TEMP3 equ 64h
stairModeCount equ 65h
stairHeelLift equ 66h
readyForNextSwing equ 67h
footFlatCount equ 68h

ADIF equ 6 ; A/D interrupt flag bit
GO equ 2 ; A/D GO/DONE flag bit
ADIE equ 6 ; A/D interrupt enable bit
PEIE equ 6 ; Peripheral int enable bit
GIE equ 7 ; Global int enable bit

ORG 0x00 ; Reset Vector
goto start

org 0x04 ; Interrupt Vector
goto service_int

org 0x10
Appendix C

start

bcf STATUS,RP0 ;Select register page 0
movlw 0FFh ;Set Port_B high
movwf PORT_B
bsf STATUS,RP0 ;Select register page 1
movlw b'11001011' ;Set Port_A as inputs, except bits 2 and 5 (all 1's)
movwf TRIS_A
clrf TRIS_B ;Set Port_B as outputs (all 0's)
movlw b'11000000' ;Set Port_C as outputs, except bits 6 and 7
movwf TRIS_C
bcf STATUS, RPO

clrf PORT_A
clrf PORT_B
clrf PORT_C
bcf PORT_A,4 ;Reset Slave PIC
call SetupDelay ;Reset Slave PIC
bsf PORT_A,4 ;Reset Slave PIC
clrf BYTE_B ;Reset all Variables
clrf BYTE_C
clrf BYTE_A
clrf Count1
clrf Count2
clrf Count3
clrf heelRise
clrf stancePhaseDetected
clrf swingPhaseDetected
clrf swingPhaseCount
clrf stancePhaseCount
bsf logicHeelOnOff,0
clrf HeelOnOff
bsf logicToeOnOff,0
clrf ToeOnOff
clrf heelOffCount
clrf midStancePhaseCount
clrf midStancePhase
clrf toeOnCount
clrf toeOffCount
clrf ReadyForMidStance
clrf footFlat
clrf heelRise
clrf lowPointReached
clrf previousVelocity
clrf notSoLowPointReached
clrf footStill
clrf detectionStarted
clrf shiftCount
clrf VeryLowPointReached
clrf firstTimeCal
clrf heelOnTime
clrf toeOnTime
clrf stairModeSelected
clrf stairHeelLift
bsf readyForNextSwing,0

clrf firstLowFound
clrf TEMP2
clrf calibrateReady
clrf numContToggle
clrf control1
clrf control2
clrf control3
clrf control4
clrf control5
clrf control6
clrf offset
clrf offsetSum
clrf offsetSumLarge
clrf offsetSumCount
clrf MEMCYCF
clrf readyToChangeMode
bsf PORT_C,0 ; No values to be loaded onto DAC

clrf COUNT ; and Put RS232 Driver to Sleep
clrf stairModeCount
clrf finishFlag
clrf rs232Flag
clrf intCount
bsf PORT_C,3 ; OE High (Output Enable off)
bcf PORT_C,4 ; WE Low (Write Enable on)
bcf PORT_C,5

call InitializeAD

movlw b'00000000'
movwf PORT_B ; Put A/D value onto PORT B
bcf PORT_C,0 ; Load Value onto DAC

call SetupDelay ; Load Value onto DAC
bsf PORT_C,0 ; Load Value onto DAC

call bigDelay

bsf PORT_C,3
; OE High (Output Enable off)
bcf PORT_C,4 ; WE Low (Write Enable on)
btfss PORT_C,7 ; No values to be loaded onto DAC

wait goto wait

bsf PORT_C,0 ; Load Value onto DAC

call bigDelay

wait2
Appendix C

```assembly
btfsc PORT_C,7
goto wait2

bsf PORT_A,2 ;Turn on LED
movlw b'1111111'
movwf TEMP3

wait4
call bigDelay
decfsz TEMP3
goto wait4

call SetupDelay ;Delay for Tad
bsf ADCON0,GO ;Start A/D conversion

loop
btfsc rs232Flag,0
goto comms
goto loop

*********************************************************************
**********
comms
bcf PORT_A,2 ;LED off
bcf PORT_C,0 ;Awaken RS232 driver
bsf STATUS,RPO

movlw b'00100000' ;Enable RCIF interrupt
movwf PIE1
movlw .22 ;9600 baud @4MHz
movwf SPBRG

movlw b'10100100' ;Async, High baud rate
movwf TXSTA

bcf STATUS,RPO

movlw b'10010000' ;Enable continious reception
movwf RCSTA

movlw b'11000000' ;Enable global interrupts
movwf INTCON
bcf rs232Flag,0
movlw b'11111111'
movwf TEMP4

keepLEDOn1
call bigDelay
```
call bigDelay
call bigDelay
call bigDelay
call bigDelay
call bigDelay
call bigDelay
call bigDelay
call bigDelay
decfsz TEMP4,1 ; Decrement Temp
goto keepLEDOn1

bsf PORT_A, 2 ; LED on
goto loop

;******************************************************************************
*************

IntVector
bcf readyToChangeMode, 0
movlw 06h ; Mask out unwanted bits
andwf RCSTA, W ; Check for errors
btfss STATUS, Z
  goto RcvError ; Found error, flag it

btfss PIR1, 5 ; Check for data ready
  retfie ; Some other interrupt, exit

btfsc numContToggle, 0
goto getValue
movfw RCREG ; Get input data
movwf whichControl
bsf numContToggle, 0
retfie

getValue
bcf numContToggle, 0

RX
movfw RCREG ; Get input data
btfsc whichControl, 0
  movwf control1
btfsc whichControl, 1
  movwf control2
btfsc whichControl, 2
  movwf control3
btfsc whichControl, 3
  movwf control4
btfsc whichControl, 4
  movwf control5
btfsc whichControl, 5
goto lastNumber
retfie

lastNumber
movwf control6
movfw control1
movwf TXREG
call bigDelay
movfw control2
movwf TXREG
call bigDelay
movfw control3
movwf TXREG
call bigDelay
movfw control4
movwf TXREG
call bigDelay
movfw control5
movwf TXREG
call bigDelay
movfw control6
movwf TXREG

noTX
btfsc PORT_C,7
goto noTX

john
call reinitialiseADC
bsf STATUS,RP0 ;Select register page 1
movlw B'00000100' ;RA0,RA1,RA3 as analog inputs (RA3 not RA4)
movwf ADCON1 ;
bsf PIE1,ADIE ;Enable A/D interrupt
bcf STATUS,RP0 ;Select register page 0
bsf INTCON,PEIE ;Enable peripheral ints
bsf INTCON,GIE ;Enable Global ints (comment was disable)

clrfr RCSTA

bsf PORT_C,0 ;No values to be loaded onto DAC
;and Put RS232 Driver to Sleep

retfie

RcvError
bcf RCSTA,4 ;Clear reciever status
bsf RCSTA,4
retfie
; Service A/D interrupt

; Check RS232 for data ready
btfsc PIR1,5 ; Check RS232 for data ready
  goto IntVector

; Make sure A/D interrupted
btfss PIR1, ADIF ; Make sure A/D interrupted
  retfie ; Not A/D, reenable, return

; Has 'pause' button been pressed?
btfsc calibrateReady, 0
  goto noPause

; Skip this instruction if not
btfsc PORT_C, 6 ; Has 'pause' button been pressed?
  goto pause ; Skip this instruction if not

noPause
btfsc PORT_C, 7 ; Check push-to-make switch
call stairMode

btfsc intCount, 0 ; Check bit 0 to see if set
  goto firstBitSet ; If set, goto...
goto notFirstBitSet ; If not set, goto...

firstBitSet
call getFSR1
  retfie

notFirstBitSet
btfsc intCount, 1 ; Check bit 1 to see if set
  goto secondBitSet ; If set, goto...
goto notSecondBitSet ; If not set, goto...

secondBitSet
call getFSR2
  retfie

notSecondBitSet
incf intCount

btfsc MEMCYCF, 0 ; Goto finish if memory full
  goto finish ; This test has been moved here
  ; after the 2 FSRs are read to
  ; ensure that they are always read
  ; first (as they are 'reads' 3 and 4)
; Did move ADRES to PORT_B here
movfw ADRES ; Get A/D value
movwf Velocity ; Velocity = current A/D value

btfsc calibrateReady,0
call setOffset

btfsc calibrateReady,0
call loadOffset

movlw b'11001001' ; Select RC osc, Analog inp 1
movwf ADCON0 ; for FSR1 input (next)
call reinitialiseADC

call SetupDelay
call SetupDelay
call SetupDelay
call SetupDelay

movfw Velocity
movwf VarToWrite
call write ; Write data to card and increment count
call bigDelay

btfss stairModeSelected,0
call analyse

btfsc stairModeSelected,0
call stairAnalyse

movfw HeelOnOff
movfw VarToWrite
call write ; Write data to card and increment count
call bigDelay

movfw ToeOnOff
movfw VarToWrite
call write ; Write data to card and increment count
call bigDelay

retfie

;**********************
;
getFSR1
movlw b'11011001' ; Select RC osc, Analog inp 3
movwf ADCON0 ; for FSR2 (next)
call initialiseADC
incf intCount
movfw ADRES ; Get A/D value
movwf VarToWrite ; Write FSR1 signal to 'VarToWrite'
call write ; Write data to card and increment count
call bigDelay
return

getFSR2
movlw b'11000001' ; Select RC osc, Analog inp 0
movwf ADCON0 ; for gyro input (next)
call initialiseADC
clrf intCount
movfw ADRES ; Get A/D value
movwf VarToWrite ; Write FSR2 signal to 'VarToWrite'
call write ; Write data to card and increment count
call bigDelay
return

stairMode
incf stairModeSelected

clear
btfsc PORT_C, 7 ; Check push-to-make switch
goto clear

movlw b'11111111'
movwf TEMP3

wait3
call bigDelay
decfsz TEMP3
goto wait3

return

loadOffset
movfw offset
movwf PORT_B

test
bcf PORT_C, 0 ; Load Value onto DAC

THIS WILL BE
CHANGED
call SetupDelay ;Load Value onto DAC WHEN WE
WISH TO USE
bsf PORT_C,0 ;Load Value onto DAC THE DAC TO
CORRECT ; THE OFFSET
  return

*******************************************************************
write
  bcf PORT_C,2 ;Pulse A0 Low
  call SetupDelay
  call SetupDelay
  call SetupDelay
  call SetupDelay

  movfw VarToWrite ;Move VarToWrite
  movwf PORT_B ;into PORTB
  call SetupDelay
  call SetupDelay

  bcf PORT_A,5 ;CE low (FOR WRITE)
  call SetupDelay
  call SetupDelay
  call SetupDelay
  call SetupDelay

  bsf PORT_A,5 ;CE high (WRITE END)
  call SetupDelay
  call SetupDelay
  call SetupDelay
  call SetupDelay

  bsf PORT_C,2 ;Pulse A0 High

bcf PIR1,ADIF ;Reset A/D int flag
bsf INTCON,PEIE ;Enable Peripheral ints
call SetupDelay ;Delay for 2xTad
call SetupDelay
bsf ADCONO,GO ;Start A/D conversion
return

*******************************************************************
call    counter    ;Update counter
return

**************************************************************************************
*******

setOffset
btfss  firstTimeCal, 0; Check to see if first time (to reset Velocity)
cclf    Velocity
bsf     firstTimeCal, 0; set 'firstTimeCal'
movfw  Velocity
movwf  VAR1    ;Move velocity into VAR1
movlw b'00111111'
movwf  VAR2    ;Move 0 (127) into VAR2
call    compare    ;Compare Velocity with '0'
btfsc  CompRes, 1; Test bit1 to see if VARs equal
goto    done
btfss  CompRes, 0; Test bit0 to see if VAR1 larger
goto    done
btfss  CompRes, 0; Test bit0 to see if VAR1 larger
goto    notDone
done
incf  offsetSumCount
movfw offsetSumCount
movwf  VAR1    ;Move offsetSumCount into VAR1
movlw b'10000000'
movwf VAR2    ;Move (128) into VAR2
call    compare    ;Compare offsetSumCount with '128'
btfss  CompRes, 0; Test bit0 to see if VAR1 larger
goto    allCounts
movfw  Velocity
addwf offsetSum
btfsc  STATUS, 0
incf  offsetSumLarge    ;Set bit 0 if add result is greater than 255
return

allCounts
incf   shiftCount    ;Increment shiftCount by 1
rrf    offsetSum    ;Shift bits in offsetSum
bcf    'offsetSum, 7 ;Make MSB zero (as LSB becomes MSB during shift)

shift
btfsc  offsetSumLarge, 0 ;Check offsetSumLarge
bsf    offsetSum, 7   ;If LSB is one, set MSB of offsetSum
rrf    offsetSumLarge    ;Shift bits in OffsetSumLarge
bcf    offsetSumLarge, 7 ;Clear MSB
movfw shiftCount    ;See if ShiftCount is 7 yet
movwf  VAR1    ;Move count into VAR1
Appendix C

```
movlw b'00000111'
movwf VAR2 ;Move (7) into VAR2
call compare ;Compare shiftCount with 7'
btfss CompRes,1 ;Test bit0 to see if VARs equal
goto allCounts

bcf callibrateReady,0
clrf readyToChangeMode
endcal

movlw b'10000000'
subwf offsetSum ;Subtract '128' from OffsetSum (to leave a value of 1-
5)

movfw offsetSum ;Add offset (around 125)
subwf offset
call loadOffset
clrf shiftCount
clrf firstTimeCal
return

notDone
btfss CompRes,0 ;Test bit0 to see if VAR1 larger
decf offset
btfsc CompRes,0 ;Test bit0 to see if VAR2 larger
incf offset
call loadOffset
return

******************************************************************************
*******

;InitializeAD, initializes and sets up the A/D hardware.
;Select ch0 to ch3 as analog inputs, RC clock, and read ch0.

InitializeAD

bsf STATUS,RP0 ;Select register page 1
movlw B'00000100' ;RAO,RA1,RA3 as analog inputs (RA3 not RA4)
movwf ADCON1 ;

bsf PIE1,ADIE ;Enable A/D interrupt
bcf STATUS,RP0 ;Select register page 0
movlw b'11000001' ;Select RC osc, Analog inp 0
movwf ADCON0

bcf PIR1,ADIF ;Clear A/D int flag
bsf INTCON,PEIE ;Enable peripheral ints
bsf INTCON,GIE ;Enable Global ints (comment was disable)
return
```
This routine is a software delay of 10\textmu S for the a/d setup. At 4\text{Mhz} clock, the loop takes 3\textmu S, so initialize TEMP with a value of 3 to give 9\textmu S, plus the move etc should result in a total time of > 10\textmu S.

\textbf{SetupDelay}
\begin{verbatim}
movlw .3 ;Load Temp with decimal 3
movwf TEMP

SD
decfsz TEMP ;Delay loop
goto SD
return

\\
\end{verbatim}

\textbf{counter}

\textbf{Increm}
\begin{verbatim}
incfsz Count1,1 ;increment least significant byte
goto CONT1

goto PortClnc ;but if zero increment second byte

CONT1
movfw Count1 ;move result into BYTE B
movwf BYTE_B

goto UpdatePrev

PortClnc
incfsz Count2,1 ;increment middle byte (i.e. first byte result was zero)
goto CONT2

goto PortAInc ;but if zero increment most significant bytes in PORT

A

CONT2
movfw Count2 ;move result into BYTE C
movwf BYTE_C

goto CONT1 ;go back to update BYTE B

PortAInc
\end{verbatim}
Appendix C

```assembly
incf Count3,1 ; increment most significant bits in BYTE A (i.e.
second byte result was zero)

movfw Count3
movwf BYTE_A ; move result into PORT A
btfsc BYTE_A,0 ; Should be '5'
bsf MEMCYCF,0
goto CONT2 ; go back to update PORT C

UpdatePrev

return

*******************************************************************
bigDelay
movlw b'11111111' ; Load Temp with 255
movwf TEMP

SD1
decfsz TEMP ; Delay loop
goto SD1

return

*******************************************************************
finish
bsf finishFlag,0
movlw 040h
movwf LEDonTime
movlw 040h
movwf LEDoffTime
goto finishJump

pause
movlw b'11111111'
movwf LEDonTime
movlw b'11111111'
movwf LEDoffTime

finishJump

LEDflash

repeat
bsf PORT_A,2 ; LED on
```
movfw LEDonTime ;Time for LED to be on
movwf TEMP4

LEDon
    call    bigDelay
    btfss  finishFlag, 0 ;Skip button checks, if program finished
    call    checkButtons
    decfsz TEMP4, 1 ;Decrement Temp
    goto    LEDon ;If not zero, goto 'LEDon'

bcf    PORT_A, 2 ;LED off
movfw LEDoffTime ;Time for LED to be off
movwf TEMP4

LEDoff
    call    bigDelay
    btfss  finishFlag, 0 ;Skip button checks, if program finished
    call    checkButtons
    decfsz TEMP4, 1 ;Decrement Temp
    goto    LEDoff

    goto    repeat

;*****************************************************************************
******

checkButtons

    btfss  PORT_C, 6 ;Check latching switch first
    retfie  ;If unlatched then return to main program

    btfss  PORT_C, 7 ;Check push-to-make switch
    return

cleared

    btfsc  PORT_C, 7 ;Check push-to-make switch has now cleared
    goto    cleared ;If not cleared then repeat

movlw  b'00001111'
movwf  long2

SD3

movlw  b'00011111'
movwf  long

movlw  b'00001111'
movwf  long

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Appendix C

SD2

call  bigDelay
bcf  PORT_A, 2
call  bigDelay
call  bigDelay
call  bigDelay
call  bigDelay
call  bigDelay
call  bigDelay
call  bigDelay
call  bigDelay
call  bigDelay
call  bigDelay
call  bigDelay

decfsz long ; Delay loop
goto  SD2

bsf  PORT_A, 2
call  bigDelay
call  bigDelay
call  bigDelay
call  bigDelay
call  bigDelay
call  bigDelay
call  bigDelay
decfsz long2 ; Delay loop
goto  SD3

btfsc PORT_C, 7
goto  rs232

goto  callibrate

return

*******************************************************************
***********
callibrate
    clrf offset
    clrf offsetSumCount
    bcf PORT_A, 2 ; LED on
    movlw b'11111111'
    movwf TEMP4
keepLEDOn
    call bigDelay
    call bigDelay
    call bigDelay
    call bigDelay
    call bigDelay
    call bigDelay
    call bigDelay
    decfsz TEMP4,1 ; Decrement Temp
    goto keepLEDOn
    bsf PORT_A, 2 ; LED off
    call bigDelay
    call bigDelay
    call bigDelay
    call bigDelay
    bsf calibrateReady,0 ; Set flag - tells us that unit
                      ; is in calibration mode
    retfie

**************

compare
    clrf CompRes
    movfw VAR1
    subwf VAR2,0
    btfsc STATUS, 2
    bsf CompRes, 1 ; Set bit 1 if equal
    btfsc STATUS, 0
    bsf CompRes, 0 ; Set bit 0 if VAR2 larger
    return

**************

analyse
    bsf PORT_A, 2 ; Turn on LED
seven

btfss heelRise,0 ; heelRise = TRUE?
goto six

movfw Velocity
movwf VAR1 ; Move velocity into VAR1
movlw b'01001110'
movwf VAR2 ; Move '-2' (77) into VAR2
call compare ; Compare Velocity with -0.5
btfsc CompRes,0 ; Test bit 0 to see if VAR2 larger
bsf VeryLowPointReached,0 ; VAR2 (-2) larger

movfw Velocity
movwf VAR1 ; Move velocity into VAR1
movlw b'01110011'
movwf VAR2 ; Move '-0.5' (115) into VAR2
call compare ; Compare Velocity with -0.5
btfsc CompRes,0 ; Test bit 0 to see if VAR2 larger
bsf lowPointReached,0 ; VAR2 (-0.5) larger

movfw Velocity
movwf VAR1 ; Move velocity into VAR1
movlw b'01111010'
movwf VAR2 ; Move '-0.2' (122) into VAR2
call compare ; Compare Velocity with -0.2
btfsc CompRes,0 ; Test bit 0 to see if VAR2 larger
bsf notSoLowPointReached,0 ; VAR2 (-0.2) larger

movfw Velocity
movwf VAR1 ; Move velocity into VAR1
movlw b'00000011' ; '0.1' (3)
addwf previousVelocity,0 ; Add '0.1' (3) to previousVelocity
movwf VAR2
call compare ; Compare Velocity with PreviousVelocity + 0.1
btfss CompRes,0 ; Test bit 0 to see if VAR1 bigger
goto Accept72

movfw Velocity
movwf VAR1 ; Move velocity into VAR1
movlw b'00000011' ; '0.1' (3)
addwf previousVelocity,0 ; Add '0.1' (3) to previousVelocity
movwf VAR2
call compare ; Compare Velocity with PreviousVelocity + 0.1
btfss CompRes,0 ; Test bit 0 to see if VAR1 bigger
goto Accept71 ; else do 'OR'

OR7
movfw Velocity
movwf VAR1      ;Move velocity into VAR1
movlw b'011111101'
movwf VAR2      ;Move -0.1 (125) into VAR2
call compare    ;Compare Velocity with -0.1
btfss CompRes,0 ;Test bit0 to see if VAR1 bigger
goto Accept73

goto else7      ;goto 'else'

Accept71
btfsc lowPointReached,0  ;test for 'LowPointReached' = TRUE?
goto Accept72
goto OR7

Accept72
incf toeOffCount,1       ;toeOffCount++
goto final7

Accept73
btfsc notSoLowPointReached,0  ;test for 'notSoLowPointReached' = TRUE?
goto Accept72
goto else7               ;goto 'else'

else7
clrf toeOffCount        ;ELSE' toeOffCount = 0

final7
movfw toeOffCount
movwf VAR1             ;Move toeOffCount into VAR1
movwf control4
movwf VAR2             ;Move 0 into VAR2
call compare           ;Compare toeOffCount with 0
btfss CompRes,0        ;Test bit0 to see if VAR1 larger (will be 0)
bcf logicOffCount,0    ;logicOffCount = FALSE

six
movfw heelOnTime
movwf VAR1            ;Move heelOnTime into VAR1
movfw control5
movwf VAR2            ;Move 'control5' into VAR2
call compare          ;Compare heelOnTime with 'control5'
btfss CompRes,0       ;Test bit0 to see if VAR1 larger
goto Accept60
btfsc CompRes,1       ;Test bit1 to see if VARs equal
goto Accept60

goto five
Appendix C

Accept60
  movfw footFlatCount
  movwf VAR1 ;Move toeOnTime into VAR1
  movfw control6
  movwf VAR2 ;Move 'control6' into VAR2
  call compare ;Compare heelOnTime with 'control5'
  btfss CompRes, 0 ;Test bit0 to see if VAR1 larger
  goto Accept61
  btfsc CompRes, 1 ;Test bit1 to see if VARs equal
  goto Accept61
  goto five

OR6
  btfsc ReadyForMidStance, 0
  goto Accept62
  goto clearFlag ;goto next segment (5)

Accept61
  btfsc midStancePhase, 0
  goto Accept625
  goto OR6

Accept62
  movfw Velocity
  movwf VAR1 ;Move velocity into VAR1
  movlw b'01111010' ;Move '-0.2' (122) into VAR2
  movwf VAR2
  call compare ;Compare Velocity with -0.2
  btfsc CompRes, 0 ;Test bit0 to see if VAR2 larger
  goto Accept63
  goto clearFlag ;goto next segment (5)

Accept625
  movfw Velocity
  movwf VAR1 ;Move velocity into VAR1
  movlw b'01111011' ;Move '-0.176' (123) into VAR2
  movwf VAR2
  call compare ;Compare Velocity with -0.176
  btfsc CompRes, 0 ;Test bit0 to see if VAR2 larger
  btfsc CompRes, 1 ;Test bit1 to see if VARs equal
  goto Accept63
  goto OR6

Accept63
  btfss firstLowFound, 0
  goto setFlag ;goto next segment (5)
  movfw Velocity
  movwf VAR1 ;Move velocity into VAR1
Appendix C

```
movlw b'01111101'
movwf VAR2 ; Move '0.1' (125) into VAR2
call compare ; Compare Velocity with -0.1
btfsc CompRes, 0 ; Test bit 0 to see if VAR2 larger
goto Accept64
goto five ; goto next segment (5)

Accept64

bcf stancePhaseDetected, 0 ; stancePhaseDetected = FALSE
movfw Velocity
movwf VAR1 ; Move velocity into VAR1
movlw b'01111101'
movwf VAR2 ; Move '0.1' (125) into VAR2
call compare ; Compare Velocity with -0.1
btfsc CompRes, 0 ; Test bit 0 to see if VAR2 larger
incf heelOffCount
movlw b'011111010' ; Leave velocity in VAR1
movwf VAR2 ; Move '0.2' (122) into VAR2
call compare ; Compare Velocity with -0.2
btfsc CompRes, 0 ; Test bit 0 to see if VAR2 larger
incf heelOffCount
movlw b'01111000' ; Leave velocity in VAR1
movwf VAR2 ; Move '0.3' (120) into VAR2
call compare ; Compare Velocity with -0.3
btfsc CompRes, 0 ; Test bit 0 to see if VAR2 larger
incf heelOffCount
movwf heelOffCount
movwf VAR1 ; Move heelOffCount into VAR1
movwf control3
movwf VAR2 ; Move 0 into VAR2
call compare ; Compare Velocity with 0
btfss CompRes, 0 ; Test bit 0 to see if VAR1 larger (will be 0)
goto Accept65
goto five

Accept65

bcf logicHeelOnOff, 0 ; Make FALSE
bsf heelRise, 0 ; Make TRUE
bcf midStancePhase, 0 ; Make FALSE
bcf ReadyForMidStance, 0 ; Make FALSE

clearFlag

bcf firstLowFound, 0
goto five

setFlag

bsf firstLowFound, 0 ; setFlag

five
```
Appendix C

btfsc ReadyForMidStance,0 ;readyForMidStance = TRUE?
goto Accept51
goto four

Accept51
movfw Velocity ;Move velocity into VAR1
movwf VAR1
movlw b'10000100'
movwf VAR2 ;Move '0.176' (13 2) into VAR2
call compare ;Compare Velocity with 0.176
btfsc ComnRes_0 ;Test bit0 to see if VAR2 larger
goto AND51
goto four

AND51
movfw Velocity ;Move velocity into VAR1
movwf VAR1
movlw b'01111011'
movwf VAR2 ;Move '-0.176' (123) into VAR2
call compare ;Compare Velocity with -0.1
btfss CompRes,0 ;Test bit0 to see if VAR1 larger
goto Accept52
goto four

Accept52
incf midStancePhaseCount
movfw midStancePhaseCount
movwf VAR1 ;Move midStancePhaseCount into VAR1
movlw b'00010100'
movwf VAR2 ;Move 20 into VAR2
call compare ;Compare midStancePhaseCount with 20
btfss CompRes,0 ;Test bit0 to see if VAR1 larger (will be 0)
goto Accept53
goto four

Accept53
bsf midStancePhase,0 ;midStancePhase = TRUE
bsf footFlat,0 ;footFlat = TRUE
bcf ReadyForMidStance,0 ;readyForMidStance = FALSE
four
btfsc stancePhaseDetected,0 ;stancePhaseDetected = TRUE?
goto Accept41
goto three

Accept41
movfw control6 ;Move velocity into VAR1
movwf VAR1
movlw b'00000000'

Appendix C

movwf VAR2 ;Move 0 into VAR2
call compare ;Compare Velocity with 0
btfss CompRes,1 ;Test bit 0 to see if VARs equal (=1)
goto Else4

movfw Velocity
movwf VAR1 ;Move Velocity into VAR1
movlw b'1111101' ;Move '-0.1' (125)
movwf VAR2 ;Move '0.1' (125)
call compare ;Compare Velocity with 0.1
btfss CompRes,0 ;Test bit 0 to see if VAR1 larger
goto Accept42
btfsc CompRes,1 ;Test bit 1 to see if VARs equal
goto Accept42
goto Else4

Accept42
incf toeOnCount
movfw toeOnCount
movwf VAR1 ;Move velocity into VAR1
movfw control2
movwf VAR2 ;Move '-0.1' (125) into VAR2
btfss CompRes,0 ;Test bit 0 to see if VAR1 larger
goto Accept43
goto three

Accept43
bsf logicToeOnOff,0 ;logicToeOnOff = TRUE
bsf ReadyForMidStance,0 ;readyForMidStance = TRUE
bcf stancePhaseDetected,0 ;stancePhaseDetected = FALSE
goto three

Else4
clrf toeOnCount

three
btfsc swingPhaseDetected,0 ;swingPhaseDetected = TRUE?
goto Accept31
goto two

Accept31
movfw Velocity
movwf VAR1 ;Move velocity into VAR1
movlw b'01111101' ;Move '-0.1' (125) into VAR2
call compare ;Compare Velocity with -0.1
btfsc CompRes,0 ;Test bit 0 to see if VAR2 larger
goto Accept32
btfsc CompRes,1 ;Test bit 1 to see if VARs equal
goto Accept32

Accept32

incf stancePhaseCount
movfw Velocity
movwf VAR1 ;Move velocity into VAR1
movlw b'01111101'
movwf VAR2 ;Move '0.1' (125) into VAR2
call compare ;Compare Velocity with -0.1
btfsc CompRes,0 ;Test bit0 to see if VAR2 larger
incf stancePhaseCount
movlw b'01111010' ;Leave velocity in VAR1
movwf VAR2 ;Move '0.2' (122) into VAR2
call compare ;Compare Velocity with -0.2
btfsc CompRes,0 ;Test bit0 to see if VAR2 larger
incf stancePhaseCount
movlw b'01111000' ;Leave velocity in VAR1
movwf VAR2 ;Move '0.3' (120) into VAR2
call compare ;Compare Velocity with -0.3
btfsc CompRes,0 ;Test bit0 to see if VAR2 larger
incf stancePhaseCount
movlw b'01110101' ;Leave velocity in VAR1
movwf VAR2 ;Move '0.4' (117) into VAR2
call compare ;Compare Velocity with -0.4
btfsc CompRes,0 ;Test bit0 to see if VAR2 larger
incf stancePhaseCount
movfw stancePhaseCount
movwf VAR1 ;Move heelOffCount into VAR1
movfw control
movfw VAR2 ;Move 0 into VAR2
call compare ;Compare Velocity with 0
btfss CompRes,0 ;Test bit0 to see if VAR1 larger (will be 0)
goto Accept33

go to two

Accept33

bsf logicHeelOnOff,0 ;Make TRUE
bsf stancePhaseDetected,0 ;Make TRUE
bcf swingPhaseDetected,0 ;Make FALSE

two

movfw Velocity
movwf VAR1 ;Move velocity into VAR1
movlw b'10000100'
movwf VAR2 ;Move 0.176' (132) into VAR2
call compare ;Compare Velocity with 0.176
btfsc CompRes,0 ;Test bit0 to see if VAR1 smaller
goto AND21 ;If Velocity < 0.176 goto AND section
goto Else2

AND21
movfw Velocity                    ;Move velocity into VAR1
movwf VAR1
movlw b'01111011'
movwf VAR2 ;Move '0.176' (123) into VAR2
call compare ;Compare Velocity with -0.1
btfss CompRes,0 ;Test bit0 to see if VAR2 larger
goto Accept21 ;If Velocity > -0.176 goto Accept 21
goto Else2

Accept21
incf footStill

goto Accept22

Else2
clrf footStill
;Move straight into Accept22

Accept22
movfw Velocity
movwf VAR1 ;Move velocity into VAR1
movlw b'10000101'
movwf VAR2 ;Move '0.216' (133) into VAR2
call compare ;Compare Velocity with 0.1
btfss CompRes,0 ;Test bit0 to see if VAR1 larger

goto Accept23 ;Move straight into OR

OR21
movfw Velocity
movwf VAR1 ;Move velocity into VAR1
movlw b'01111011'
movwf VAR2 ;Move '0.176' (123) into VAR2
call compare ;Compare Velocity with -0.1
btfsc CompRes,0 ;Test bit0 to see if VAR2 larger
goto Accept23
goto final2

Accept23
btfss detectionStarted,0

goto Accept24
goto final2

Accept24
bcf logicHeelOnOff,0
Appendix C

final2

bcf logicToeOnOff, 0

movfw footStill
movwf VAR1 ; Move footStill into VAR1
movlw b'00010100' ; Move 20 into VAR2
call compare ; Compare footStill with 20
btfss CompRes, 0 ; Test bit0 to see if VAR1 larger
goto Accept25
goto one

Accept25

bsf logicHeelOnOff, 0
bsf logicToeOnOff, 0

one

movfw Velocity
movwf VAR1 ; Move velocity into VAR1
movlw b'10000010' ; Move '0.1'(130) into VAR2
call compare ; Compare Velocity with 0.1
btfss CompRes, 0 ; Test bit0 to see if VAR1 larger
goto Accept11
goto else1

Accept11

incf swingPhaseCount
movfw Velocity
movwf VAR1 ; Move velocity into VAR1
movlw b'10001010' ; Move '0.2'(133) into VAR2
call compare ; Compare Velocity with 0.2
btfss CompRes, 0 ; Test bit0 to see if VAR1 larger
incf swingPhaseCount
movfw Velocity
movwf VAR1 ; Move velocity into VAR1
movlw b'10001010' ; Move '0.4'(138) into VAR2
call compare ; Compare Velocity with 0.4
btfss CompRes, 0 ; Test bit0 to see if VAR1 larger
incf swingPhaseCount
movfw swingPhaseCount
movfw VAR1 ; Move velocity into VAR1
movlw b'00010100' ; Move 20 into VAR2
call compare ; Compare Velocity with 20
btfss CompRes, 0 ; Test bit0 to see if VAR1 larger
Appendix C

goto Accept12
goto final1

Accept12
bcf stancePhaseDetected,0
bsf swingPhaseDetected,0
clrf stancePhaseCount
crf midStancePhase,0
clrf midStancePhaseCount
clrf heelOffCount
clrf toeOnCount
clrf toeOffCount
clrf swingPhaseCount
crf lowPointReached,0
crf notSoLowPointReached,0
crf logicToeOnOff,0
crf heelRise,0
crf logicHeelOnOff,0
clrf footStillsf detectionStarted,0
clrf ReadyForMidStance
incf stancePhaseCount ;Heel Strike
incf toeOnCount
incf heelOffCount
incf toeOffCount
goto final1

else

clrf swingPhaseCount

final1

movfw Velocity
movwf previousVelocity ;Move velocity into previousVelocity

zero

btfss logicHeelOnOff,0
goto AcceptHalf
btfss logicToeOnOff,0
goto AcceptHalf
incf footFlatCount

AcceptHalf

btfss logicHeelOnOff,0
goto Accept01
bsf PORT_C,1 ;Heel On
movlw b'11111111'
movwf HeelOnOff
incf heelOnTime
goto Accept02
Appendix C

Accept01
bcf PORT_C, 1 ; Heel Off
movlw b'00000000'
movwf HeelOnOff
cclf heelOnTime

Accept02
btfss logicToeOnOff, 0
goto Accept03
bsf PORT_C, 5 ; Toe On
movlw b'11111111'
movwf ToeOnOff
incf toeOnTime
goto Accept04

Accept03
bcf PORT_C, 5 ; Toe Off
movlw b'00000000'
movwf ToeOnOff
cclf toeOnTime

Accept04
return

%******************************************************************************************%

* stairAnalyse

return

%******************************************************************************************%

END
Appendix D

‘C’ Program to Read in Data from the PCMCIA Card Reader and Convert into Excel Format

#include <stdlib.h>
#include <string.h>
#include <stdio.h>
#include <process.h>

void main( void )
{
    int numread, i, j = 0, skip = 0, channel = 0, size;
    unsigned char *array, *output, *velocity, *heelprediction, *toeprediction,
        *heelFSR, *toeFSR;
    char name[20] = "temp";
    char ret[3] = " \n", ret2[3] = " \t";
    char extension[5] = ". dat";
    char extension2[5] = ". xls";
    char filename[30] = "c:\c\cdsutils\"
        FILE *pfile = NULL;

    array = ( unsigned char *)malloc(5000000 * sizeof ( unsigned char ));
    output = ( unsigned char *)malloc(50000000 * sizeof ( unsigned char ));

    system("cout temp. dat /1=100000");

    strcat(name, extension);
    strcat(filename, name);

    if( (pfile = fopen (filename, "rb")) != NULL )
    {
        numread = fread( array, 1, 5000000, pfile );
        printf( "Number read = %d \n", (numread/2) );
        for ( i = 0; i < numread; i++ )
        {
            if (skip == 0)
            {
                skip = 1;
                output[j] = array[i];
                j++;
            }
            else
                skip = 0;
        }
        fclose( pfile );
    }
else
    printf( "File could not be opened
\n" );
skip = 0;
strcpy (filename, "c:\pcmcia\newgaitdata\");
printf("\n
Enter the name of the file you wish to create\n\n" );
scanf("%s", &name);
strcat(name, extension2);
strcat(filename, name);
if( (pfile = fopen (filename, "wt")) != NULL )
{
    for ( i = 0; i < (numread*2); i++ )
    {
        if (channel == 0)
            fprintf( pfile, "%d", output[i] );
        if (channel == 1)
            fprintf( pfile, "%d", output[i] );
        if (channel == 2)
            fprintf( pfile, "%d", output[i] );
        if (channel == 3)
            fprintf( pfile, "%d", output[i] );
        if (channel == 4)
            fprintf( pfile, "%d", output[i] );
    if (channel != 4)
    {
        putc( ret2[0], pfile ); //Put in blanks for Excel
        putc( ret2[1], pfile );
        putc( ret2[2], pfile );
    }
    if (channel == 4)
    {
        putc( ret[0], pfile ); //Put in blanks for Excel
        putc( ret[1], pfile );
        putc( ret[2], pfile );
    }
    channel++;
    if (channel == 5)
        channel = 0;
}
fclose( pfile );
else
    printf( "File could not be opened\n" );
**SENSORS**

**PIEZOELECTRIC VIBRATING GYROSCOPE**

**ENC SERIES**

This ultra-small angular velocity sensor with Murata's unique triangular prism vibrating unit offers many excellent features.

**FEATURES**
- Small and lightweight
- Quick response
- Low cost, low power consumption

**APPLICATIONS**
- Video cameras
- Mouse, toys
- Movement detection

**DIMENSIONS: mm**

![Diagram of ENC-05EA/EB](image)

**SPECIFICATIONS**

<table>
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<tr>
<th>Characteristics</th>
<th>Symbol</th>
<th>Condition</th>
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<th>Std.</th>
<th>Max.</th>
<th>Unit</th>
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<td>Supply voltage</td>
<td>Vcc</td>
<td>at Vcc = 5.0 V</td>
<td>-4.75</td>
<td>-5.0</td>
<td>-5.25</td>
<td>VDC</td>
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<tr>
<td>Current consumption</td>
<td>Icc</td>
<td>at Vcc = 5.0 V</td>
<td>2.5</td>
<td>3.5</td>
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<tr>
<td>Comparative voltage</td>
<td>Vcom</td>
<td>at -5 to 75°C</td>
<td>+2.15</td>
<td>+2.3</td>
<td>+2.45</td>
<td>VDC</td>
</tr>
<tr>
<td>Static output (Rms)</td>
<td>V0</td>
<td>angular velocity = 0</td>
<td>+1.3</td>
<td>+2.3</td>
<td>+3.3</td>
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<td></td>
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<tr>
<td>Scale factor</td>
<td>Sf</td>
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<td>+10</td>
<td>%/S</td>
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<tr>
<td>Temperature coefficient of scale factor</td>
<td>Sf</td>
<td>Reference: Is at -5 to 75°C</td>
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<td></td>
<td></td>
<td>%/S</td>
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<tr>
<td>Resonance frequency</td>
<td>f0</td>
<td>at Vcc = 5.0 V</td>
<td>25.0</td>
<td></td>
<td></td>
<td>kHz</td>
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<td>26.5</td>
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<td>kHz</td>
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<td>Resonance frequency disparity</td>
<td>f0-fb</td>
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<td>Linearity</td>
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<td>in the resonance angular velocity range</td>
<td>-5</td>
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<td>Operating Temperature Range</td>
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<td></td>
<td>+75</td>
<td>°C</td>
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<td>-80</td>
<td></td>
<td>+85</td>
<td>°C</td>
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<td></td>
<td>-</td>
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<td>1.3</td>
<td>g</td>
</tr>
<tr>
<td>Dimension</td>
<td></td>
<td></td>
<td>21.5 x 8.1 x 7.0 mm</td>
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</tr>
</tbody>
</table>

All typical values.

Unless otherwise specified, ambient temperature To = 25 ± 3°C, Vcc = 5.0 V/DC. Use a sensor output load resistance of 500Ω or more. Comparative voltage (Vcom) should be grounded with capacitor of 47μF.

**PART NUMBERING SYSTEM**

![Diagram of part numbering system](image)

For more detailed information regarding this product line in North America, contact the North America Product Line Department. For more detailed information regarding this product line in Europe, see Catalog No. S428-1.


Bobath B "Adult Hemipleiga: Evaluation and Treatment" Butterworth Heinemann; 1990


Craik RL; Oatis CA; "Gait Analysis: Theory and Application"; Mosby; St. Louis; 1995


Measurements." 18th Annual International Conference Proceedings of the IEEE
engineering in Medicine and Biology Society.

Learning Techniques for Automatic Determination of Rules to Control

Kagaya, H., Shimada, Y., Ebata, K., Sato, M., Sato, K., Yukawa, T. and
Paraplegia Utilizing Functional Electrical-Stimulation." Archives of Physical
Medicine and Rehabilitation 76 (9): 876-881.

EMG activity of tibialis anterior during normal and abnormal gait.” 4th Biological
Engineering Society Symposium on Electrical Stimulation - Clinical Systems,
Glasgow.

of Gait Events: A Case Study using Inductive Learning Techniques." Journal of
Biomedical Engineering 11: 511-516.

Kljajic, M., Malezic, M., Acimovic, R., Vavken, E., Stanić, U., Pangrsic, B. and

Control in Gait of Hemiparetic Patients." Brain 102: 405-430.

Rehabilitation by Means of Functional Electrical Stimulation." Prosthetics and
Orthotics International 17: 107-114.


Martini FH; "Fundamentals of Anatomy & Physiology" 3rd Edition; Prentice Hall; New Jersey; 1995


Micheal P (1996) "Developments in Surface Electrical Orthosis for the Re-education of Hemiplegic Gait" PhD, University of Surrey


Rose J; Gamble JG; "Human Walking" 2nd Edition; Williams & Wilkins; Baltimore; 1994


Turnbull GI; Bell PA; "Maximising Mobility After Stroke"; Croom Helm; London; 1985


References


Woodcock A (1998) "Electrical Stimulation of Chronically Denervated Muscle" PhD, University of Surrey


Zahid A (1997) "Development of Control Algorithms for the Compustim 10B Electrical Stimulator: Foot Sensor Reliability and Speed Adaptation" MSc, University of Surrey