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PORTABLE INSHOE GAIT ANALYSIS DEVICE

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ABSTRACT

Gait analysers are devices or systems that quantify the planter pressures as they occur under the foot. The data obtained, be it in visual or digital format, assists the medical practitioner in the diagnosis of an abnormality of gait.

This thesis describes the development of a low-cost, portable, inshoe gait analysis device which overcomes all the limitations experienced by other systems currently in use.

The non-availability of robust, yet small pressure transducers, required the design and manufacture of these components. Capable of being embedded within an insole environment, it allows for the unobtrusive monitoring of localised planter pressures associated with all modes of gait. The transducer output, being analog and continuous in nature, allows for the production of a real-time pressure display - an important requirement for the therapeutic assessment and treatment of patients.

The flexibility and accuracy of this system surpasses that of other systems. Its configuration provides for operation in remote environments, allowing for gait measurements under specific ambulation conditions.

Use of a new transducer monitoring technique, in addition to a new video mixing technique, has enabled this device to compete with those systems making use of expensive processing and display equipment.

Two prototype gait analysis systems have been built and extensively tested under clinical conditions. The results obtained provide unique, hitherto unavailable data, which can now be used for a more precise classification of gait disorders. It is already clear that the data will provide the basis for more accurate diagnosis and therefore more appropriate treatment of a variety of gait disorders.

D E C L A R A T I O N

I declare that this thesis is my own, original work. It is being submitted for the degree of Doctor of Philosophy in Electrical Engineering at the University of Cape Town. It has not previously been submitted in this or any other form for a degree at any university.

Signed by candidate

(Name of Candidate)

3 day of SEPTEMBER 19 87

PREFACE

The clinical assessment of the weight-bearing foot is usually based on the subjective judgement of a physician or surgeon. Although anatomical abnormalities are often apparent at examination, the accurate diagnosis of an abnormality of function is more difficult, particularly if the malfunction manifests itself only under dynamic loading conditions. In addition, great emphasis has been placed on physical fitness achieved by jogging and running in recent years, which has led to a significant increase in the number of sports injuries, especially those to the foot.

Monitoring equipment is available for studying the biomechanics of the joints of the lower limbs, namely, the hips and knees. This has resulted in a better understanding of lower limb function. The biomechanics of foot function, however, are very difficult to analyse using the available monitoring systems, as it is difficult to isolate the plantar pressures occurring at specific anatomical sites during the stance phase of gait.

The work reported in this thesis was directed at the completion of the research requirements for a Ph.D degree at the University of Cape Town (UCT). In addition, the work was done with a particular engineering goal in mind, that being, to design and build an inshoe pressure-measuring device that would overcome the limitations imposed by existing systems.

The scope of this research was a) to design and manufacture suitable pressure transducers, b) to design and build the signal processing and display circuitry, c) to compile a software package that was user-friendly for the presentation of computed gait information and, d) to test this system extensively under medical supervision.

In detail, the thesis is structured as follows. Chapter 1 shows the advantages of inshoe pressure measurements and contains a review of those measuring systems in current use. Their limitations and restrictions provide guidelines for the design of a new instrumented insole gait analysis device.

Chapter 2 examines the various pressure transducer types as used in previous systems for incorporation with the new device. Their unsuitability results in the design and manufacture of small capacitive pressure-transducers, which makes use of a new type of dielectric configuration. This dielectric material is analytically shown to be suitable in withstanding the high plantar impact forces.

Chapter 3 describes the functional components of the new gait analysis device. Its flexibility becomes apparent by use of a modular construction technique.

Chapter 4 contains a detailed description of the hardware techniques used in the processing and display circuitry. A new method of measuring small capacitance values, independent of the distance to the measuring equipment, is described. In addition, a new state of the art method of synchronising two remotely-situated, crystal-referenced video sources is also introduced. A theoretical evaluation of this technique is included to highlight its practical implementation.

Chapter 5 describes the gait analysis software. The extensive nature of the computed gait parameters allows for its use in a variety of studies. The software, configured in a menu-driven format, is shown to be user-friendly.

Chapter 6 describes a number of tests conducted in evaluating the new gait analysis device. Conformity in results with that of other researchers verifies its accuracy and validity as a clinical assessment tool.

The thesis concludes with Chapter 7, which provides a summary of what was achieved together with the wide range of applications of the new gait analysis device. Proposed gait studies and further system improvements are also discussed.

The research would not have been possible without the assistance and support of Professor Manfred Reineck and Mr Jan Hesselink of UCT, and Professor Kit Vaughan, previously senior lecturer in the Department of Biomedical Engineering, UCT and now resident at Clemson University, United States of America.

The research was carried out in the laboratories of the Electrical and Electronic Department of UCT, with the performance and evaluation conducted in a laboratory in the Sports Science Centre of the UCT Medical School, under the supervision of Professor Tim Noakes, who is head of the centre.

The author was honoured at being awarded the BARNIB Design of the Year Award in the Medical Category (1985) for work completed on this proposed gait analysis device. The inspiration could not have been more timeous.

The author would like to thank the CSIR Foundation for Research and Development, for their financial assistance.

Lastly, the author is grateful to Miss Dawn v.d. Rheede for the effort and time spent in typing this thesis.

LIST OF ABBREVIATIONS

AC	Alternating current
ADC	Analog to digital converter
c-b	Common base
DAC	Digital to analog converter
DC	Direct current
DIP	Dual in-line package
DVM	Digital voltmeter
EDG	Electrodynogram
FSO	Full scale output
GF	Gauge factor
IC	Integrated circuit
IEC	International Electrotechnical Commission
LPF	Low Pass Filter
LSB	Least significant bit
op-amp	Operational amplifier
RAM	Random access memory
RF	Radio frequency
synch	Synchronisation
TV	Television
ULA	Uncommitted Logic Array
UCT	University of Cape Town
VCR	Video cassette recorder
VHF	Very high frequency
VMU	Video mixer unit

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CHAPTER ONE

INTRODUCTION

Attempts to measure the foot-to-ground interface pressures for diagnostic purposes date back to the nineteenth century. An early researcher in this field was Beely[1] (1882) who attempted to relate interface pressures to the depth of a foot impression in a thin bag filled with plaster-of-Paris. Later researchers, namely Elftman[2] (1934) and Morton[3] (1935), used the deformation properties of rubber projections on a walkpath mat to measure localised loads. A similar method was proposed by Arcan and Brull[4] (1976), where a set of circular interference fringes was used as an optical display of the localised loads.

By 1981, Lord[5] had noted that most of the studies dealing with quantitative estimates of the load distribution under the foot were primarily concerned with the location of the resultant centre of support, support forces and moments, particularly during walking. Although these results may have had certain research validity, localised forces occurring at the foot-to-shoe interface were still unknown.

At present, there are three types of gait analysis devices currently in use for the quantification of foot forces. Two of these are floor-mounted, capable of measuring the global and localised foot-to-ground interface pressures, whereas the third has the capability of measuring the foot-to-insole interface pressures. This latter type has many advantages in that it allows the lower extremity disability of human subjects to be studied for:-

a) detection of pressure points and zones, including ulceration caused by diabetes as reported by Stokes[6] ;

b) design effectiveness assessment of different types of corrective shoes prescribed for lower extremity orthopaedic disorders;
c) assessment of length as well as alignment of prosthesis; and
d) assessment of congenital dislocation of the hip or paralytic weakness of the gluteus maximus and gluteus minimus.

A number of gait-monitoring systems attempting to overcome the global measurement limitation have appeared in recent years. Most of these restrict measurements to a clinical environment and cannot accommodate modes of gait other than walking. It will be shown that the majority of limitations experienced by these systems can be related to the type of transducer used and its configuration at the plantar surface.

1.1 REVIEW OF INSHOE PRESSURE MEASURING SYSTEMS

The distribution of the dynamic loads under the foot can be measured in one of two ways - either by having the patient walk over a surface that is in some way pressure-sensitive or by introducing a thin transducer between the foot and the supporting surface. The former method enables the acquisition of gait data relating to one stance phase, and requires for the patient to place the designated foot in the proper way while walking over the pressure-sensitive platform. Due to the forced walking pattern, this can result in measurements which are not likely to be representative of the natural gait of the subject. Consecutive multiple-step recordings, containing information pertaining to the natural gait cycle, can however, be obtained by means of an appropriate, portable, under-foot transducer measuring system.

The predominant force acting on the foot during normal gait is that of the vertical component[7]. The behaviour of foot function can be readily assessed using this parameter. A number of gait-measuring systems, capable of measuring this force-component at the foot-shoe interface, will be reviewed.

1.1.1 STRAIN GAUGE PRESSURE TRANSDUCER SYSTEM - J.P.POLLARD et al.[8]

This system uses six Kulite™ vertical strain gauge pressure transducers, in conjunction with six customised semiconductor shear gauges. This resolves the discrete plantar pressures into their vertical, longitudinal and transverse components.

These transducers are attached to the plantar surface by means of double-sided adhesive tape. A 30 metre, 18-core cable, connects the transducers to a six-channel amplifier and ultra-violet recorder. A sample trace of the forces generated by a subject walking in conventional leather shoes is shown in Fig 1.1.

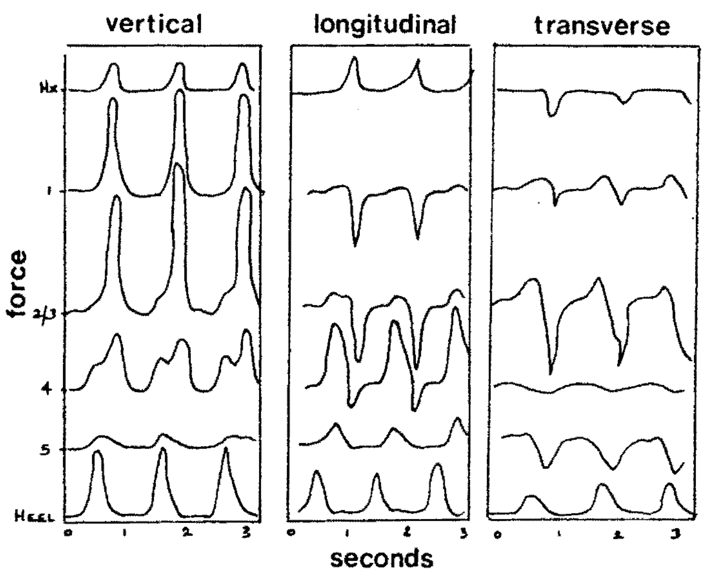


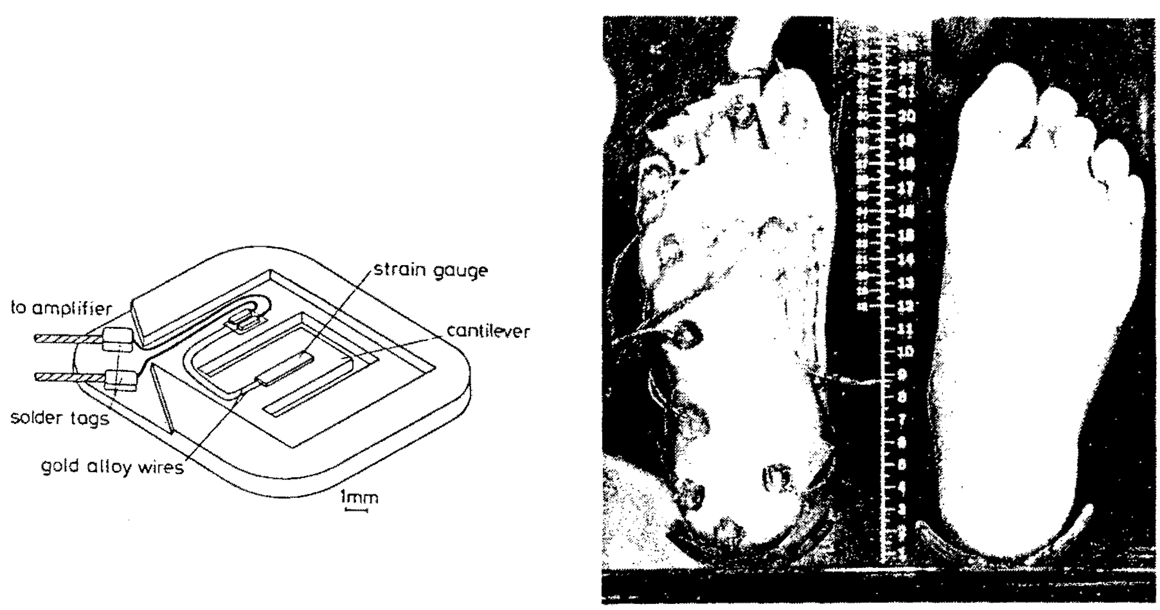
Fig 1.1 : Sample traces of the three inshoe force components[8]

Although this system is suitable for the clinical examination of the inshoe forces related to different types of footwear, the use of an umbilical cable prevents studies pertaining to the outdoor environment.

This system has a number of limitations, namely, the lack of computer backup which requires a subjective comparison of both inter/intrapatient gait traces; the method of transducer attachment which can affect the normal gait pattern; and the fragile nature of this transducer type which prevents the measurement of forces associated with running or jumping.

1.1.2 DISCRETE CANTILEVERED STRAIN GAUGE SYSTEM - R.W. SOAMES et al.[8]

This system uses sixteen semiconductor strain gauges (Kulite D-UCP-120-090), each of which are recessed in a beryllium-copper assembly (Fig 1.2 a). These are attached to the plantar surface (Fig 1.2 b) and, by virtue of their size, allow for a high pressure resolution of the plantar surface.



(a) (b)

Fig 1.2 : Cantilevered Strain Gauge and its Attachment[8]

The leads from each transducer are taken to a junction box strapped to the patient's waist, and from there via a multichannel umbilical cable to the transducer amplifiers and host PDP 11/10 computer.

The computer controls the digital sampling of the transducer outputs and stores this information prior to performing specific gait calculations. Many useful parameters relating to the gait cycle may be obtained, including localised impulse and force values. Gait information can also be displayed in a graphic format, as shown in Fig 1.3.

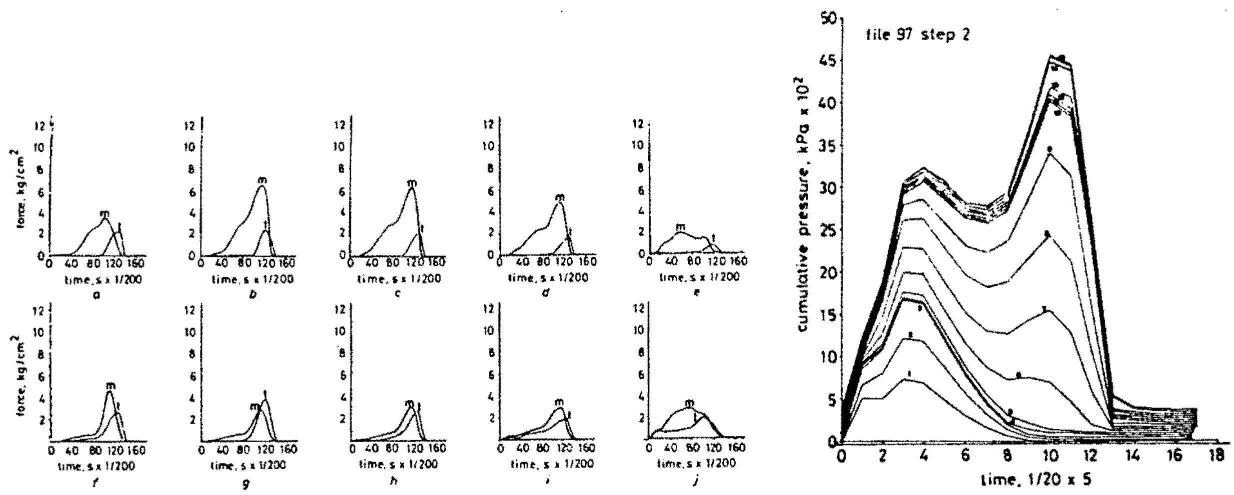


Fig 1.3 : Computer-generated force-time and cumulative pressure curves[8]

Similar limitations exist with this system as that of section 1.1.1. In addition, the high cost of the processing computer restricts measurements to a laboratory environment.

1.1.3 INSTRUMENTED SOLE PRESSURE MEASURING SYSTEM - R.B. FROST et al.[10]

In measuring the vertical forces occurring at the plantar surface during walking, six resistive strain gauges (Kyowa type KFC-1-C1-11, 120 ohm) are used in detecting the horizontal Poisson strains caused by vertical compressive forces on discs of rubber, which are embedded in a sole material (Fig 1.4).

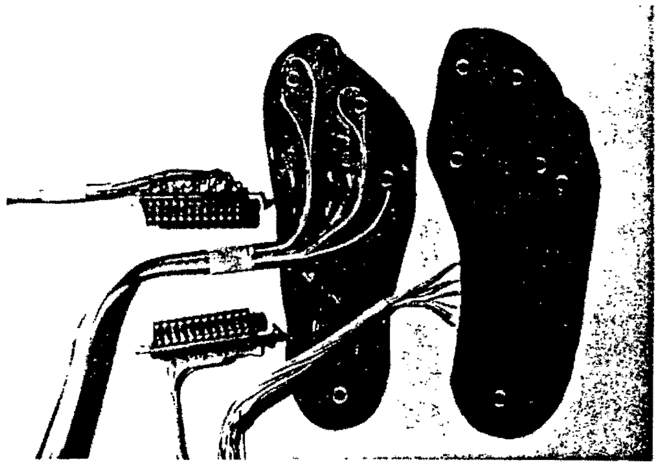


Fig 1.4 : Sole assembly containing strain gauge pressure transducers[10]

The embedded load cells are connected to strain gauge amplifiers by a long multi-core cable, which is held by the patient during examination. A six-channel chart recorder is used in obtaining the resulting pressure display (Fig 1.5).

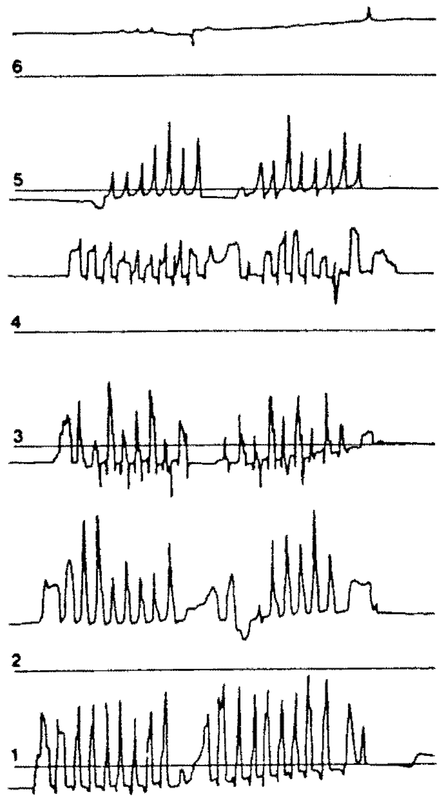


Fig 1.5 : Typical recorder chart of gait parameters[10]

Since local bending of the load cell can result in incoherent data, this sole must be worn on bare feet with measurements conducted on a smooth, flat surface.

Although this sole is particularly useful for recording peak forces, its inability to measure inshoe pressures and forces associated with gait modes other than that of walking severely restricts its use.

1.1.4 <u>THE PIEZOELECTRIC INSOLE MEASURING SYSTEM- P.R. CAVANAGH et al.</u>[11]

This device makes use of a flexible array of 499 lead zirconate titanate transducers (Fig 1.6), embedded in a silicone rubber insole, for measuring the dynamic inshoe plantar pressures.

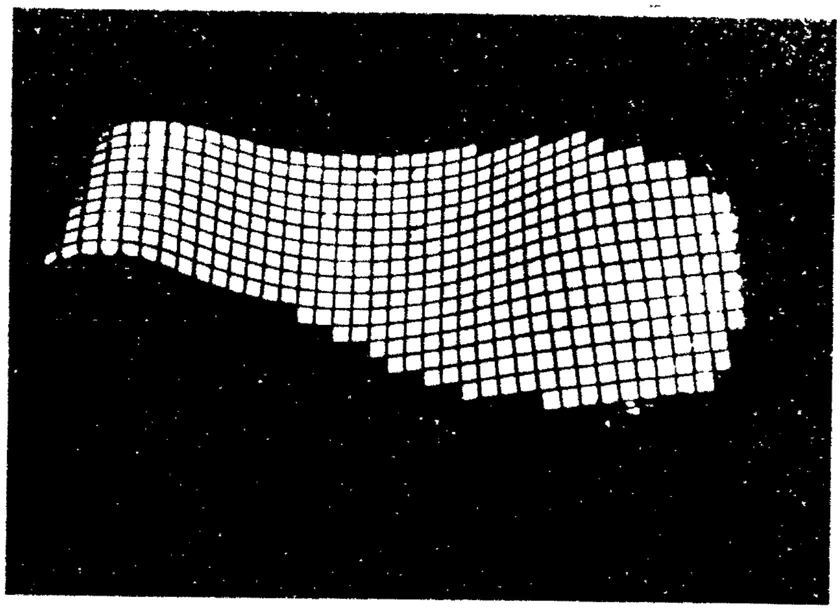


Fig 1.6 : Piezoelectric insole transducer array[11]

The charge amplifiers, which are associated with each transducer, are installed in a backpack unit which is worn by the patient. The signals from the amplifiers are multiplexed to a computer for immediate processing, or via a microcomputer onto disc for subsequent off-line processing (Fig 1.7).

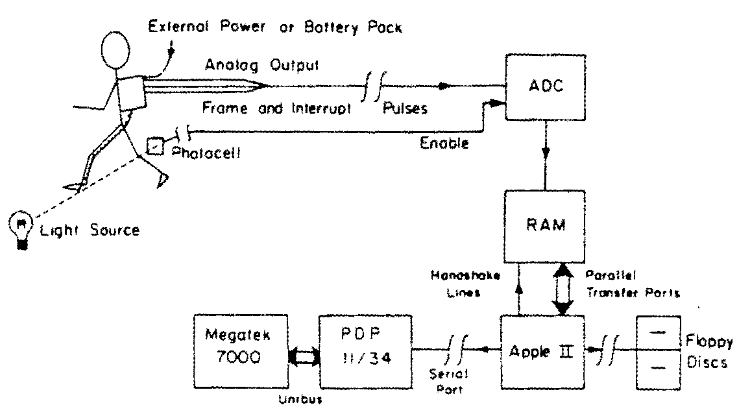


Fig 1.7 : Block diagram of piezoelectric insole system[11]

On completion of gait sampling, an elegant presentation of results is achieved using computer graphics (Fig 1.8).

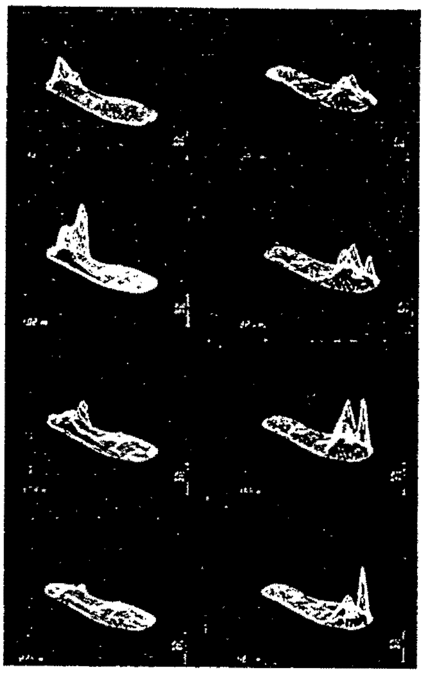


Fig 1.8 : Three-dimensional computer-generated gait display[11]

This device, by virtue of the transducer type used, is only capable of measuring the change in plartar pressures occurring during the stance phase of the gait cycle. The use of expensive display equipment and umbilical cable method of signal transfer restricts measurements to a clinical environment.

1.1.5 THE ELECTRODYNOGRAM (EDG) - S. LANGER et al.[12]

At present, this is the only inshoe gait analysis system which is available commercially. It makes use of customised, disposable strain gauges which, when attached to the plantar surface (Fig 1.9), allow for the measurement of localised plantar pressures. These transducers are covered with a sock, thus facilitating inshoe pressure measurements.

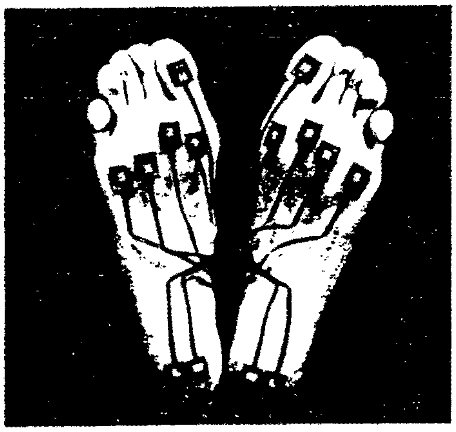


Fig 1.9 : EDG transducer attachment[12]

During ambulation, data is recorded for a limited time into the memory of a small microprocessor unit worn on the belt, and later downloaded to a host computer for analysis and display.

There are a number of factors which limit the attractiveness of the EDG as a universal diagnostic tool. These mainly refer to the transducer type used, the method of transducer attachment and sole digital option of gait sampling. Besides the cost factor of using disposable transducers, the discomfort that may be experienced by the adhesion of these transducers to the plantar surface could affect the behaviour of normal gait. As no real-time analog pressure display is possible, diagnostic and treatment interaction cannot be monitored effectively. Lastly, separate EDG processing units, together with their respective sensor types, are required for the walking and running modes of gait. This makes the entire system somewhat costly.

1.1.6 DISCRETE CAPACITIVE TRANSDUCER SYSTEM -H.J. HERMENS et al.[13]

In measuring the vertical reaction forces on both feet during walking, eight capacitive transducers, each with a surface area of 10 sq. cm, are attached to each sole of the patient's shoes. A long trailing cable connects these transducers to the processing circuitry and computer (Fig 1.10).

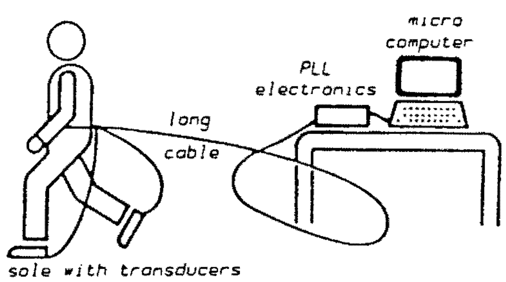


Fig 1.10 : Discrete capacitive gait analysis system[13]

The forces are measured during 20 seconds of walking with the data being stored in the RAM memory of the computer. Before and during a measurement, the measured forces are continuously displayed in bar graphs on the computer screen (Fig 1.11).

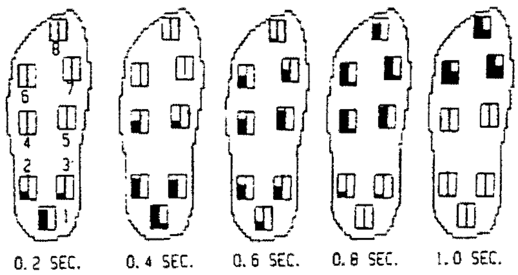


Fig. 1.11 : Computer-generated bar graph pressure display[13]

Computer analysis on the stored gait information allows the transducer forces and timing information to be computed and displayed in a descriptive and meaningful format.

A number of limitations exist with this system. The method of transducer attachment and use of an umbilical cable only allows for the measurement of walking-related forces, and restricts measurements to a clinical environment. In addition, the large transducer surface area does not allow for a high plantar pressure resolution.

1.2 THE OPTIMUM REQUIREMENTS FOR A UNIVERSAL GAIT ANALYSIS DEVICE

To overcome the limitations experienced by the gait analysis systems currently in use, and to allow for a more extensive study of the lower extremity function, it is necessary to design a new type of gait analysis device, suitable for a broad spectrum of users.

In compiling the specifications, this device should be :-

a) a low-cost, menu-driven, stand-alone system;
b) accurate in the reproduction and/or display of forces occurring at discrete plantar locations;
c) portable, suitable for measurements both in a clinical and outdoor environment;
d) easy to operate with both analog and digital capabilities;
e) capable of accommodating future hardware and software updates; and
f) designed in a manner that electrical circuitry used meets the International Electrotechnical Commission (IEC) standards for safety in medical equipment.

Before considering the type of pressure transducer to be used, it is desirable to examine briefly the plantar surface and, in particular, the pressure profile occurring during the normal gait cycle. This would then give an indication of the transducer size and dynamic capability required.

1.3 PLANTAR PRESSURES : THE QUANTISATION AND LOCALISATION

It has been established medically that the majority of plantar pressures develop under the heel and metatarsal heads. As a result, the regions to be monitored are identified. In order that the force sensors do not interfere with normal gait, it is preferable that they be embedded in an insole, which would be available in a variety of sizes.

The study of podiatric biomechanics, including clinical observations using various forms of slow-motion cinematography, slow-motion videotaping, graphic and electronic impressions, has enabled researchers to determine the flow of forces through the feet in both normal and abnormal patients. As a result, six areas have been identified which are deemed to provide sufficient data necessary for valid observations and diagnosis in patients.

These areas are located at the lateral and medial heel condyle, the second and fifth metatarsal heads, the first metatarsal head under both sesamoids, and the hallux-centre of the distal aspect of the proximal phalanx.

Since many sprinters and runners are forefoot strikers, two additional plantar areas are selected for measurement, these being at the third and fourth metatarsal heads. Besides providing a higher pressure resolution, information relating to the sequence of metatarsal head loading may be acquired.

With the transducer layout dictated by the anatomical structure of the foot, the transducer size becomes fairly well defined. It is considered best if all transducer sizes are the same while the thickness is kept to a minimum. The transducer layout, as used in this project is thus shown in Fig 1.12.

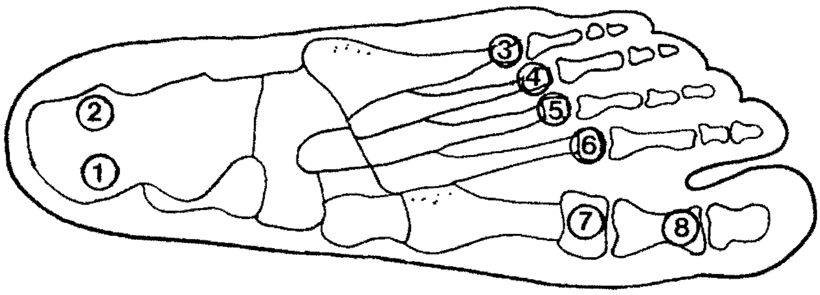


Fig 1.12 : Proposed insole transducer layout

In determining the transducer's dynamic range, tests conducted by Cavanagh and researchers at Nike Inc.[14] have shown that the maximum force on the foot after ground contact is as much as three times the body weight and that the acceleration transmitted to the leg can be ten times that of normal gravity.

1.4 TRANSDUCER SPECIFICATIONS

As a result of the above discussions, the following transducer specifications were devised and considered to meet most requirements :

a) dimensions : circular, diameter 14 mm
b) dynamic range : 0 - 25 kg
a) output : static and dynamic output related to vertical force component
d) frequency response : 50 Hz
e) minimum sensitivity : 40 grams.

In addition, the design should take account of the transducer's aging and temperature stability.

1.5 ADVANTAGES OF AN INSTRUMENTED INSOLE MEASURING SYSTEM

One of the critical parameters of measurement systems employing individual transducers is that of measurement repetition. The location of the transducers and their relative positioning is therefore of prime importance.

Unless discrete transducers are embedded is an insole, connecting wires run the risk of becoming detached when the foot is inserted or removed from the shoe. In addition, impulse loading which occurs during running or jumping can also cause these exposed wires to break.

The direct attachment of transducers to the plantar surface can cause patient discomfort and thus change the pattern of gait. This occurs particularly with patients suffering from advanced rheumatoid arthritis or excessive ulceration resulting from diabetes mellitus.

The use of a thin instrumented insole, therefore, can overcome many of the above problems, allowing for the gait pattern to be obtained in the most unobtrusive manner. Since the transducers in this way are placed to coincide with the sites of interest, repeatable measurements are assured.

Research conducted by Miller et al.[15] and Inman et al.[16] showed that variations in the angle of the metatarsal heads and the longitudinal axis of the foot remain relatively consistent under load-bearing conditions. This therefore allows for most feet of a given size to be studied with a single design of insole.

1.6 SUMMARY

A survey of the various inshoe gait analysis systems currently in use has been presented. As was shown, the umbilical cable method of monitoring the transducer outputs, in addition to the sophisticated computers required for the analysis of such data, restricts gait measurements to a clinical or laboratory environment.

The inability of the transducers, as used with these systems, to sustain the high impact forces associated with gait modes such as running or jumping without permanent deformation, restricts gait measurements to that of walking.

The optimum requirements for a universal gait analysis device were formulated to overcome the above limitations. The plantar pressures and their locations on the plantar surface dictate the transducer size and its dynamic performance requirements. Transducer specifications were devised and the advantages of using an instrumented insole instead of discrete transducers were discussed.

CHAPTER TWO

PRESSURE TRANSDUCERS

In this chapter, the various pressure transducers, suitable for measuring inshoe plantar pressures, are discussed with regard to their long term durability and ability to provide a real-time pressure display.

The unsuitability of commercially-available pressure transducers for use is an insole, necessitated the design and manufacture of small capacitive transducers capable of high pressure measurements. The material chosen as the dielectricum results from an extensive investigation into the properties of vulcanised rubber materials. A theoretical analysis of the dielectric behaviour is carried out and compared with results from laboratory testing.

2.1 PRINCIPLES UTILISED BY FORCE TRANSDUCERS

There are basically three types of transducers which can be used for the measurement of high impact forces. These may be classified as a) strain gauge, b) piezoelectric, and c) capacitive transducers.

2.1.1 STRAIN GAUGE TRANSDUCERS

A strain gauge pressure transducer operates on the principle that small dimensional changes in its material alter its electrical resistance. Two types of stain gauge are considered specifically, namely, etched-foil and semiconductor.

a) Etched-foil strain gauge

This pressure transducer is manufactured by techniques similar to those used in the production of printed circuits. Such a process readily lends itself to the production of shaped, special-purpose strain gauges, as shown in Fig 2.1.

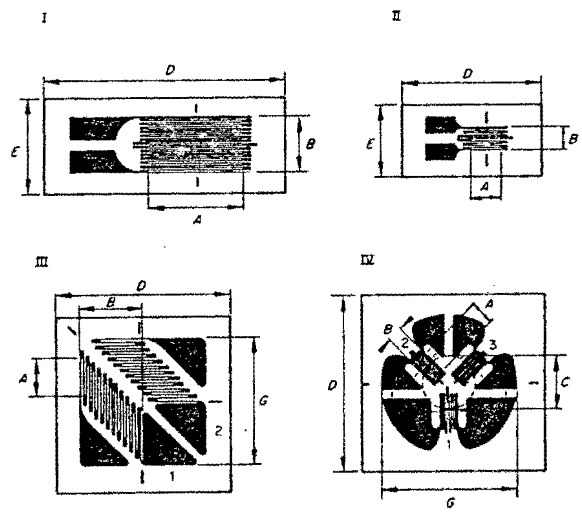


Fig 2.1 : Resistance foil gauges

Tensile and compressive forces, when applied to the strain gauge, cause the foil material to change its dimensions by an amount determined by the Young's modulus of that material. The resulting change in the foil cross-sectional area causes the overall transducer resistance to change and, providing that the elastic limit of the material is not exceeded, may be related to the applied load. For measurement purposes, a Wheatstone bridge arrangement is normally used for converting the resistance change to a voltage.

The sensitivity of a manufactured strain gauge is termed the gauge factor (GF). For an etched-foil type, it may be expressed as :

$$GF = (\Delta R/R)/\varepsilon = (\Delta R/R)/(\Delta L/L) \quad [17] \quad \ldots\ldots \; 2.1$$

where : $\Delta R/R$ = unit change in resistance

ε = unit strain

$\Delta L/L$ = unit change in length.

With this type of strain gauge, gauge factors in the region of 1.9 to 6 are attainable.

The transducer fabrication technique allows for the manufacture of stress-strain foil gauges (Fig 2.2). This would allow the applied force to be independently resolved in two planes.

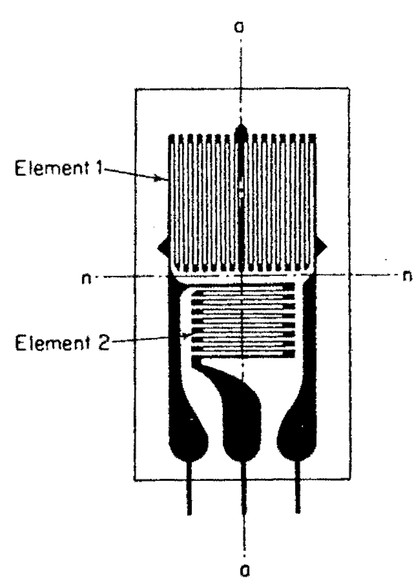


Fig 2.2 : Stress-strain foil gauge

If the gauge has appreciable dead resistance, it may be simultaneously sensitive to both extensile Poisson and bending strains, resulting in the recorded signals containing superimposed information[10,18]. This could seriously influence data interpretation.

Temperature variations in the gauge vicinity may cause the gauge factor to change and thus affect its performance. Although compensation would be possible by the inclusion of dummy gauges in the same vicinity, this could make the insoles bulky.

The transducer, by virtue of its construction, is unsuitable for use in the proposed insole system if gait modes other than normal ambulation are to be anticipated. The fragile nature of the etched resistance-material is unable to withstand the repetitive, high-impact forces associated with running or jumping without permanent damage or deformation.

b) Semiconductor strain gauge

The piezoresistive nature of the materials used to fabricate this device results in a resistance variation which is usually very much greater than that obtained in the foil equivalant. The GF for a semiconductor measuring longitudinal strain is given by,

$$GF = R/(R_0 \varepsilon) = 1 + 2\nu + Y\pi_e \quad [17] \qquad \ldots\ldots 2.2$$

where: R = change in resistance due to applied stress

R_0 = initial semiconductor resistance

ε = strain

ν = Poisson's ratio for the semiconductor

Y = Young's modulus

π_e = longitudinal piezoresistive coefficient.

Since the GF is dependent on resistance change due to a dimensional change $(1 + 2\nu)$, as well as a change of resistivity with strain $(Y\pi_e)$, gauge factors up to 200 may be attained.

Typical semiconductor strain gauge configurations are shown in Fig 2.3.

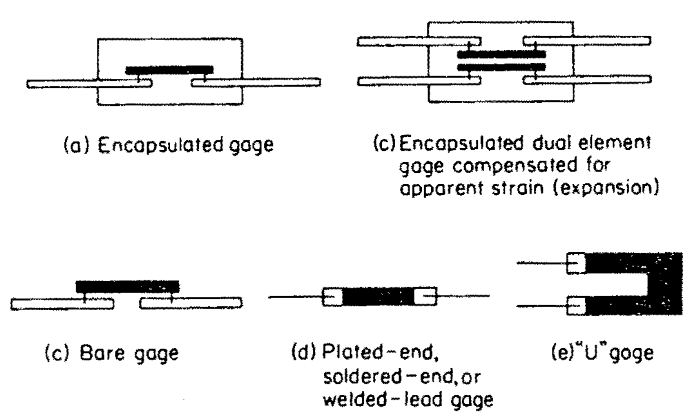


Fig 2.3 : Semiconductor strain gauges

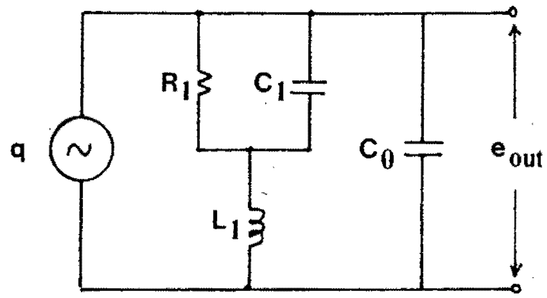
These devices, beside being inherently nonlinear, suffer from more pronounced temperature effects than those of the foil-type. The temperature coefficient of resistance of the semiconductors used as strain gauges is 60 to 100 times that of Constantan; the variation of GF with temperature is 3 to 5 times greater; and the Seebeck coefficient generated at connections can be 10 to 20 times larger.

Although temperature compensation is possible, the less robust nature of this transducer prevents its use with the proposed insole system.

2.1.2 PIEZOELECTRIC TRANSDUCERS

The operation of piezoelectric devices is based upon the fact that in certain crystalline materials, mechanical deformation of the crystal lattice results in a relative displacement of positive and negative charges within the material. This results in the generation of a voltage between opposite faces of the crystal which is proportional to the degree of distortion of the material.

Materials commonly used for such devices include quartz, Rochelle salt and barium titanate ceramics. The basic device is in effect a capacitor with the piezoelectric material as the dielectric between two electrodes. The equivalent circuit of this device is shown in Fig 2.4.



L_1 = Motional inductance
C_1 = Motional capacitance
R_1 = Motional resistance
C_0 = Shunt capacitance

Fig 2.4 : Equivalent circuit of a piezoelectric transducer

The equation that relates the transducer output voltage e_{out}, to the applied force F, can be expressed as

$$e_{out} = dF/C \quad [19] \qquad \text{...... 2.3}$$

where: d = charge sensitivity in coulombs per newton
C = capacitance in farad between the electrodes.

It is observed that a charge q, produced by a static deformation, produces across the device a voltage e_{out} whose magnitude is dependent upon the capacitance of the system. This voltage will only remain as long as the charge does not leak away via the measuring system connected across the transducer. The detector must therefore have a very high input impedance ($\sim 10^{14}$ -ohms) if a very low frequency response is required. A true zero frequency response, is however, not possible.

This transducer type is ideally suited for dynamic inshoe pressure measurements. Its inability to register static pressures excludes its use with the proposed insole system.

2.1.3 CAPACITIVE TRANSDUCERS

When force is exerted on a parallel plate capacitor containing a compressible dielectric material, the plate separation, as well as the dielectric constant change, which results in a change of capacitance. This can in turn be related to the applied force, and made available as an electrical output.

The equation that describes the dependence of the capacitance C on the plate area A, dielectric constant ε , and plate separation d, is

$$C = \int_A \frac{\varepsilon}{d} (p)\ dA \qquad \text{...... 2.4}$$

where: ε and d are functions of pressure p.

This type of transducer was extensively studied by Nicol et al.[20] who found its performance comparable with that of the Kistler force platform (Fig 2.5).

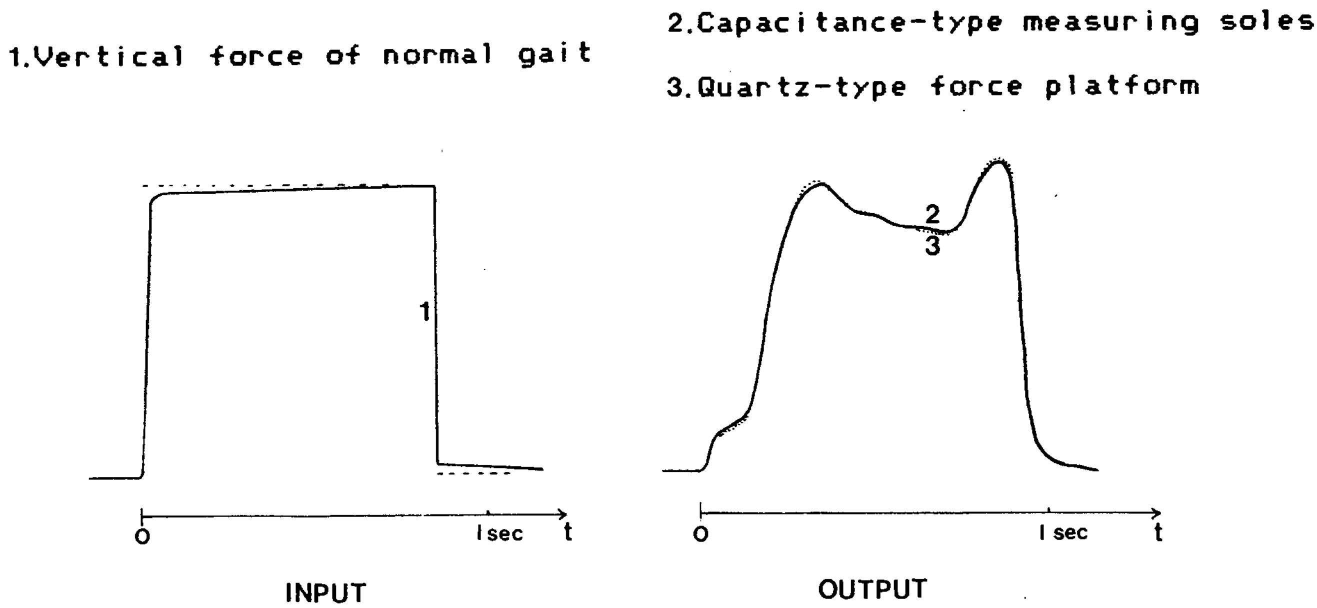


Fig 2.5 : Response of the Kistler-measuring platform and a capacitive pressure transducer[20]

During his investigation, Nicol[21] built capacitive pressure transducers that could be attached to the patient's shoes for the monitoring of their gait pressures. However, since they were of large surface area, they only provided limited gait information.

To obtain a higher plantar pressure resolution, the above method of gait measurement was dispensed with in favour of a flexible, multi-elemented capacitive mat configuration. To date, a modified version of this gait analysis device is still in use[22].

Although a high measurement accuracy is achieved, the accommodation of this measuring system within an insole environment is not possible.

2.2 THE DEVELOPMENT OF A NEW CAPACITIVE TRANSDUCER FOR HIGH PRESSURE MEASUREMENTS

The success obtained by Nicol[23] prompted an indepth investigation into the viability of rubber-filled capacitive transducers with a view to reducing the nonlinearities, hysteresis, tolerances and parasitic effects.

2.2.1 PROPERTIES REQUIRED OF RUBBER WHEN USED AS THE DIELECTRICUM FOR A CAPACITOR

The rubber material, when used as the dielectricum in a capacitor, must fulfil the following requirements. It must :-

a) be able to withstand continuous impact loading without permanent deformation;
b) operate exclusively in its rubbery-elastic region;
c) exhibit minimum creep as a function of static or dynamic loading;
d) possess a high rebound resilience;
e) possess a high electrical breakdown strength and low loss-factor;
f) exhibit long-term and temperature stability; and
g) be able to be shaped without deformation to its overall structure.

Examination of the properties of vulcanised natural and synthetic rubbers (Davey et al.[24]) show that natural rubber and neoprene are the most suitable for use as the dielectric material. Some of their general properties are shown in Table 2.1 for comparative purposes.

Table 2.1 : General properties of natural rubber and neoprene

	NATURAL RUBBER	NEOPRENE
Hardness range, °BS	30-100	40-90
Tensile strength (kg/cm^2)	> 210	210
Elasticity	high	medium
Loss factor: 20°C	0.08	0.09
Rebound resilience : cold	high	fairly high
Ageing resistance	medium	high
Dielectric constant, 25°C, 50Hz	2.8	7
Electrical resistance	10^{15} ohm-cm	10^{10} ohm-cm
Creep, %	2.5	8

In general, rubber in its bulk form can be considered incompressible. For this reason, homogeneous layers of the abovementioned rubber cannot be used alone as the dielectricum for the proposed capacitive transducers.

2.2.2 A NOVEL DIELECTRIC MATERIAL CONFIGURATION

Hermens et al.[13], in the design of their capacitive transducers using natural rubber as the dielectricum, used a layer of equally dimensioned rubber discs in covering the plate surface. Since the spatial distribution of these pads determine the transducer characteristics, the process of adhering these to the plate material is complicated if close tolerances are required. Due to the discrete nature of transducer manufacture, non-uniformity in transducer characteristics could be anticipated in miniaturising these transducers for use in the proposed insole system.

A new type of dielectric configuration was investigated with a view to simplifying the manufacturing process while satisfying the criteria of section 1.4. This would facilitate the mass production of pressure transducers, where each would possess the same capacitance to load characteristic.

The new dielectric material, shown in Fig 2.6, is manufactured from vulcanised neoprene, of hardness 50° BS.

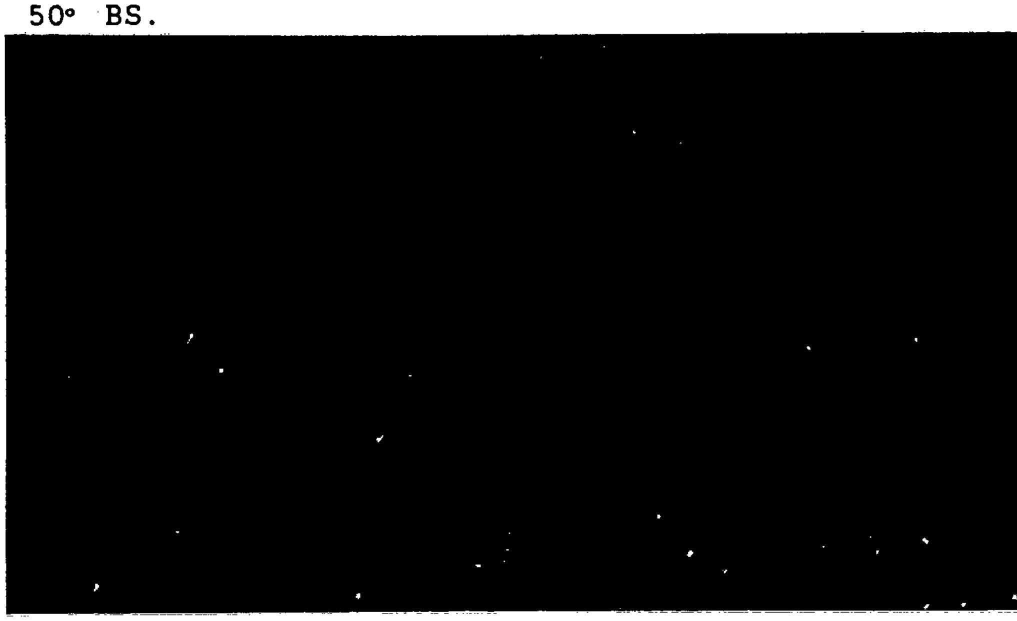
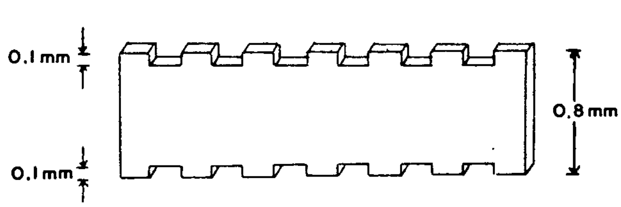


Fig 2.6 : Close-up view of the new dielectric material

A myriad of rectangular-like projections, of height 0.1 mm, covers each side of the central homogeneous layer, which has a uniform thickness of 0.6 mm. A cross-section of this material is schematically shown in Fig 2.7.



Projection dimensions :

Long side = l = 0.7147 mm
Short side = w = 0.5498 mm
Height = h = 0.1 mm

Fig 2.7 : Cross-section of the new dielectric material

The neoprene material, available in sheet form, may be shaped to correspond exactly with that of the plate material by using a punch.

2.2.3 ANALYSIS OF THE DIELECTRIC MATERIAL BEHAVIOUR

In modelling the behaviour of the neoprene dielectric, the following assumptions are made :-

a) only compressive loads are anticipated;
b) torsional and shear effects are zero;
c) compressive stress for each portion of the dielectric material can be computed independently;
d) loading frequency less than 100Hz; and
e) temperature kept constant for all calculations (25° C).

Payne[25] has shown that for the shapes of rubber commonly used under compression, and with bonded end faces, the compressive stress f is given by,

$$f = -G_r^* \ (\lambda - \lambda^{-2}).s \qquad \ldots\ldots \ 2.5$$

where : G_r^* = complex modulus of rigidity

λ = ratio of strained to unstrained length, the length being in the direction of deformation

s = shape function.

The shape function for rubber of circular cross-sectional area is given by,

$$s = 1 + B \ (d/h)^2 \qquad \ldots\ldots \ 2.6$$

where: $B = 0.103$ for $G_r^* = 8{,}45\ kg/cm^2$
d = diameter
h = height in direction of compression.

For the castellated outer dielectric region, the shape function is defined as

$$s = \frac{1.33 + 0.66\,\frac{w}{l} + C\left(\frac{w}{h}\right)^2}{1 + \frac{w}{l}} \qquad \ldots\ldots\ 2.7$$

where: $C = 0.225$ for $G_r^* = 8{,}45\ kg/cm^2$
l = long side of rectangular projection
w = short side of rectangular projection
h = height in direction of compression.

G_r defines the stiffness of the rubber. Being complex, it may be resolved into its in-phase and out-of-phase components, by

$$G_r = G_r' + j\,G_r'' \qquad \ldots\ldots\ 2.8$$

and

$$G_r^* = \left((G_r')^2 + (G_r'')^2\right)^{1/2} \qquad \ldots\ldots\ 2.9$$

G_r is affected by both deformation rate and temperature . The former may be best described by the composite curves of Fig 2.8, which depicts the variation of G_r', G_r'' and $\tan\delta$ (loss factor = G_r''/G_r') with frequency for a soft vulcanisate, similar to that of the dielectric material.

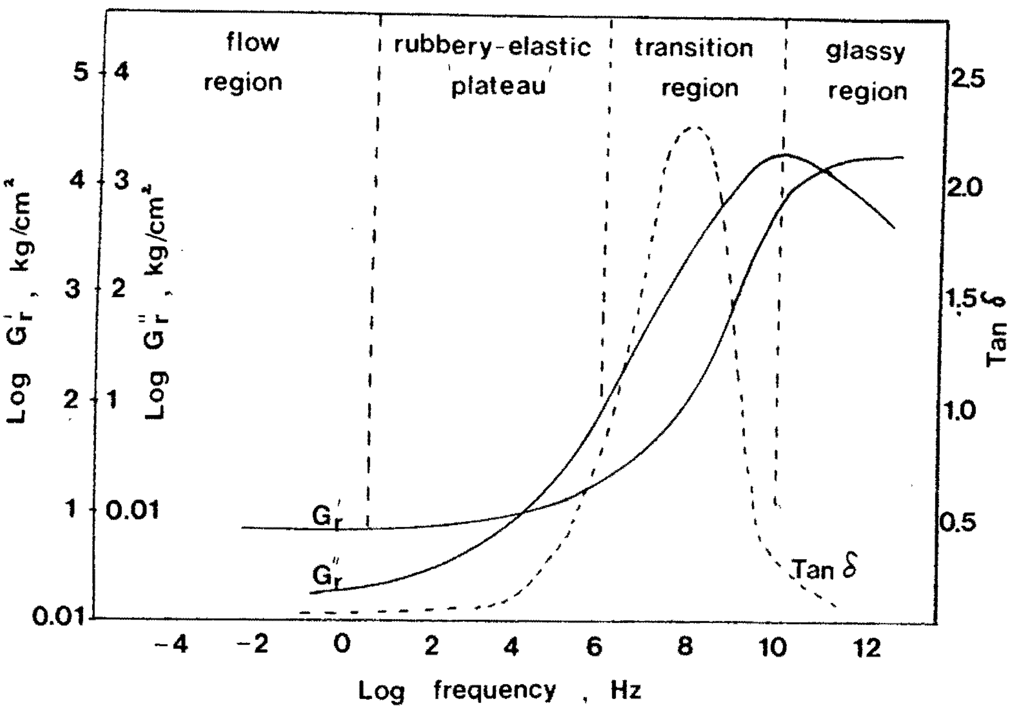


Fig 2.8 : Composite moduli–frequency curves for a soft vulcanisate (25°C)

In calculating G_r for various operating temperatures, the following conversion is used, namely,

$$G = G_r \frac{T \rho}{T_o \rho_o} \qquad \ldots\ldots 2.10$$

where : T = operating temperature in degrees Kelvin (°K)
T_o = reference temperature in °K
ρ, ρ_o = densities at T and T_o respectively.

Since the volume coefficient of thermal expansion is not greater than 0.0006/°C, the ρ terms can be omitted unless T and T_o differ widely.

In analysing the dielectric material, both static and dynamic loading conditions must be considered.

a) STATIC LOADING (25° C)

Consider the outer dielectric surface. Since the ratios (w/h) and (l/h) >> 1, no buckling will occur. This allows equation 2.5 to be used. The relative compressed height λ , can be determined from tables[24] by calculating ($\lambda - \lambda^{-2}$). Since $G_r'' \ll G_r'$ for static loading, then $G_r^* \sim G_r'$.

The application of a 250 N static load on a dielectric pad with a diameter of 14 mm results in $\lambda = 0.745$ for the castellated region. A similar calculation is performed on the central homogeneous layer of this pad. The effect of bulk compressibility becomes evident by λ here being equal to 0.985. This results from (d/h) >> 1 in the shaping function.

It is thus observed that compression occurs mainly in the castellated region of the dielectric material, with the central layer remaining relatively rigid. The overall relative height under compression is thus computed as 0.8725.

Since log $G_r^* = 0.93$ kg/cm² for the neoprene dielectricum, Fig 2.8 shows that compression is restricted to the rubbery-elastic region of the dielectric's characteristic. The loss factor, tan δ , which gives an indication of the energy input lost as heat, is observed to be negligible.

For an anticipated operating temperature range varying between 0 and 50° C, G_r^* varies from 7.75 to 9.16 kg/cm^2 (approx), according to equation 2.10. The effect on the overall compressed height is shown in Table 2.2.

Table 2.2 : Variation in relative compressed height of neoprene with temperature

Temp (°C)	0	25	50
G_r^* (kg/cm^2)	7.75	8.45	9.16
λ	0.867	0.873	0.880

b) DYNAMIC LOADING (25° C)

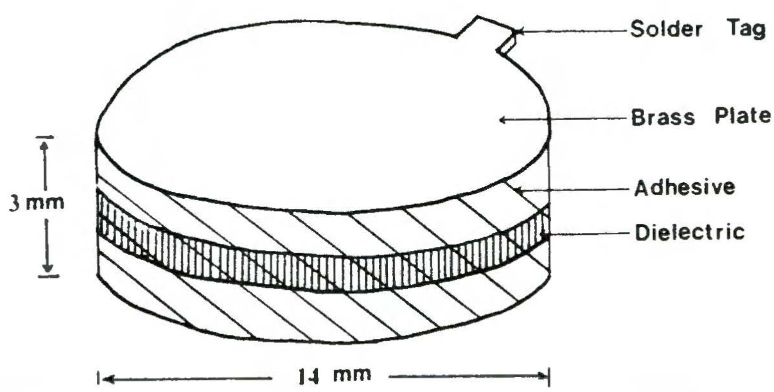
In determining whether the dielectric material is suitable for use in a capacitive pressure transducer, the effect of dynamic loading on G_r^* must also be investigated.

By observing Fig 2.8, G_r'' increases by 50% (approx) with G_r' remaining constant in response to a 100 Hz dynamic load. This results in a minimal increase in G_r^* (0.04%) since $G_r' \gg G_r''$. Thus, compression is restricted to the rubbery-elastic region of the moduli-frequency curve. In addition, the loss factor under these conditions is small, only becoming predominant at frequencies in excess of 1MHz.

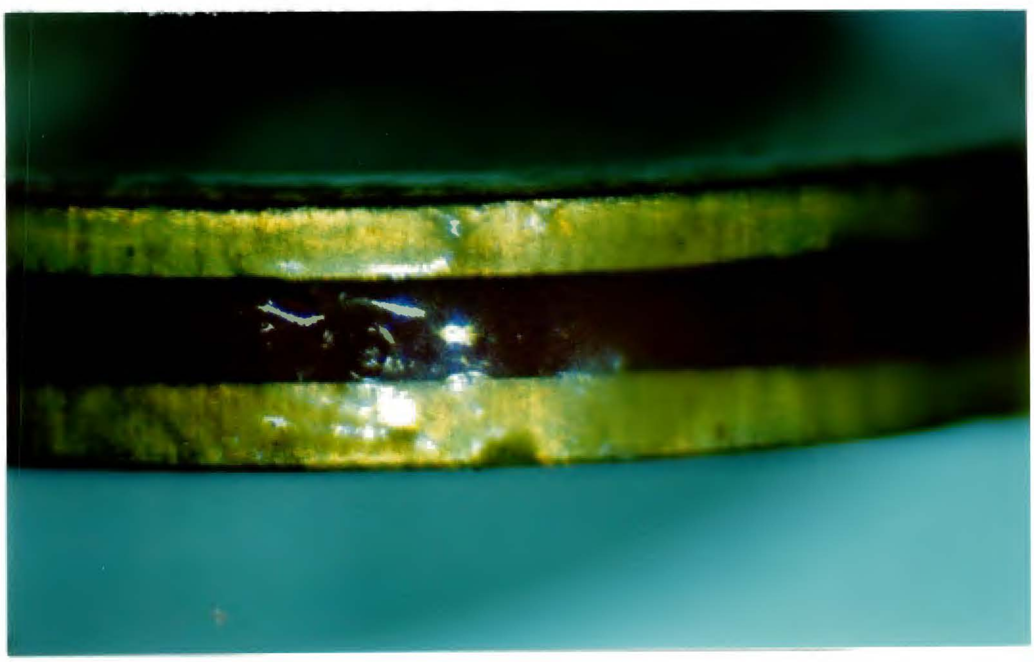
Creep, although varying logarithmically under static loading conditions, is considerably increased by mechanical vibrations, especially at ordinary temperatures. Since many external contributing factors exist, this effect will have to be measured under dynamic loading conditions.

2.3 THE MANUFACTURE OF A PROTOTYPE CAPACITIVE PRESSURE TRANSDUCER

A prototype version of the new pressure transducer is shown in Fig 2.9a with a side-view shown in Fig 2.9b. Prior to the mass production of these components, the prototype version must be tested for its conformation with theoretical predictions.



(a)



(b)

Fig 2.9 : Close-up view of the new pressure transducer

2.3.1 TRANSDUCER MATERIAL

The materials used in the construction of the transducers are described with reference to their particular properties.

a) Capacitor plate material

The plate material must be strong enough to withstand the continual impact forces experienced at the plantar surface during running or jumping without permanent deformation. The material must be conductive and uniformly even over its entire contact area. It has been found that brass is the most complient material for this application as it can easily be polished, and wires can be attached by way of soldering.

The circular brass plates are best produced by a metal punch as this can keep tolerances well within specifications. The plates are polished on a water-wheel using a fine grade of water paper (800-1000).

It is imperative for the plate edges to be perpendicular, as round edges can allow the adhesive used in the bonding process to creep and thus impair the polished conductive surfaces.

b) Dielectric material

The dielectric between the plates must be the same size as the plate itself. It is necessary for the plate punch to be used to produce the dielectric pads. Provided the surfaces are clean, good contact with the plate material is ensured.

c) Transducer adhesive

The bonding and flexibility of the adhesive used in keeping the sandwiched dielectric in place is very critical. The adhesive most suited for this application is Wondafix^{TM} (by Pratley). Being a blocked diisocyanate alloy, its high bond strength and limited flexibility restricts the plate movement to the vertical mode only, thus largely eliminating shear effects. Since it has the ability of film application, it contributes minimal additional creep to that of the dielectric.

2.3.2 TRANSDUCER MANUFACTURE

A specially constructed jig, suitable for use with a tensometer, is used to compress the transducer plates during the application of the adhesive. Besides preventing adhesive creep, it allows the plates to be perfectly aligned.

After adhesive application, the pressure is released and a curing period of 24 hours is required before the transducers may be removed from the jig and used.

2.4 TRANSDUCER ACCEPTANCE TESTS

For this transducer to suit the desired application, it is essential that it satisfies the criteria of section 1.4 in addition to conforming to theoretical predictions. A number of tests are thus conducted to determine the transducer's characteristics. These include an environmental test to determine its response to a varying temperature gradient. A capacitance to voltage converter is used for monitoring the transducer's output response.

a) Determination of the Transducer's Output Characteristic

The equation that best describes the transducer's output response is obtained by using a least-squares curve-fitting routine on data obtained by sequentially loading the prototype transducer.

Murdock[26] showed that if the response of a transducer conforms to that of an exponential function, then it may be expressed as:

$$v = a(1-e^{-bl}) \qquad \ldots\ldots\ 2.8$$

where : v = output voltage
l = load
a and b are constants.

It is also shown that :

$$a = \frac{\sum_i v_i\ (1 - e^{-b\ell_i})}{\sum_i (1 - e^{-b\ell_i})^2} \qquad \ldots\ldots\ 2.9$$

and

$$g(b) \sim \sum_i v_i\ \ell_i\ e^{-b\ell_i} \sum_i (1 - e^{-b\ell_i})^2 - \sum_i v_i\ (1 - e^{-b\ell_i}) \sum_i \ell_i\ e^{-b\ell_i}\ (1 - e^{-b\ell_i}) = 0 \qquad \ldots\ldots\ 2.10$$

Since equation 2.10 is nonlinear, an iterative solution is employed to find 'b' using Wegstein's method. After 'b' has been found, 'a' is obtained using equation 2.9.

The equation that best models the transducer's output response is thus calculated as :

$$v = 5.05(1 - e^{-l/90}) \text{ volts} \qquad \ldots\ldots\ 2.11$$

where : l = load in newtons.

The equation's closeness of fit to the transducer's actual response is shown graphically in Fig 2.10.

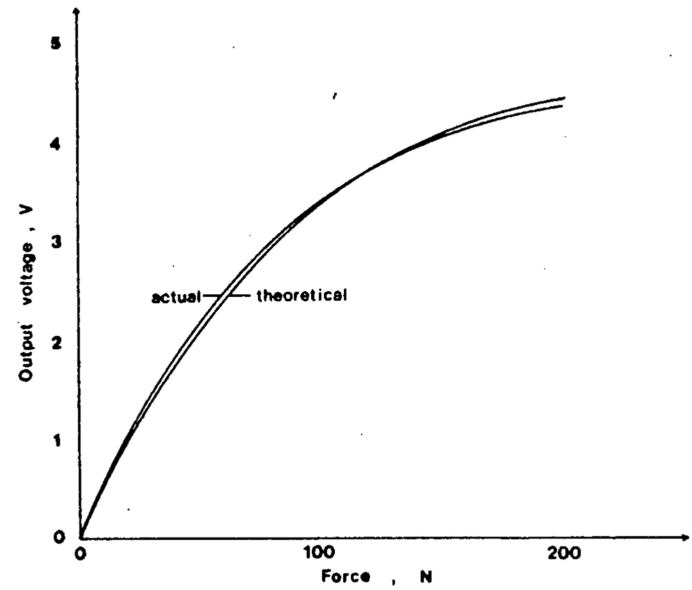


Fig 2.10 : Closeness of fit of modelled transducer output response

For a statically applied load of 250 N, the transducer capacitance was observed to vary from 5 to 12 pf.

The nonlinear output response results from the nonlinear characteristics of G_r^* (and hence ε) under mechanical stress.

b) <u>Measurement Repeatability</u>

The repeatability of the transducer's response is determined using data obtained from a number of consecutive static calibration cycles. Since repeatability is expressed as the maximum difference between the transducer's output related to any force in the range, it is calculated to be within ± 3% FSO (full scale output). The band of permissible deviations of output values from the reference curve under these conditions is defined by the shape of the static error band, shown in Fig 2.11.

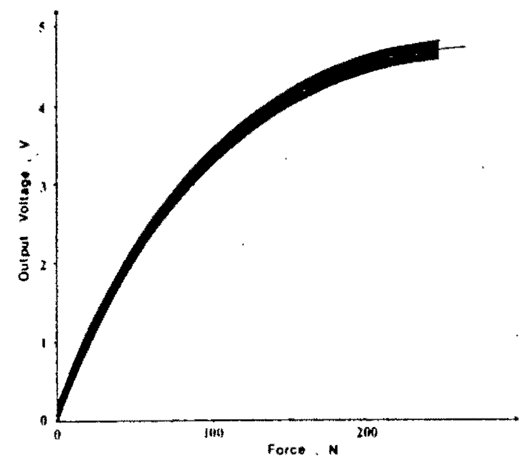


Fig. 2.11 : Transducer's static error band

The amount of creep contributed by the dielectric and adhesive under static loading conditions is shown to follow a characteristic law if the applied load is of short duration (less than 1 minute), and does not exceed 250 N. This effect manifests itself as a continuous variation in transducer output voltage and is given by

$$V = v + a \log t \quad [24] \qquad \ldots\ldots \; 2.12$$

where: V = time dependent output voltage

v = instantaneous transducer output voltage, given by equation 2.11

a = constant = 5.6×10^{-2}

t = time in seconds after load application.

Upon load release, the recovery of the dielectric and adhesive to its original shape is initially very rapid and then settles down to a roughly linear progression on a log-time scale.

Since this transducer is to be used mainly in the dynamic mode, it must sustain continuous impact loading without altering the system's accuracy. This property is determined by measuring the transducer's static output response immediately after repetitive shock excitation. The results obtained show the effect of impulse loading on system accuracy to be within $\pm$ 3.5% FSO. The additional output variation is contributed by the increase in creep, resulting from dynamic transducer loading.

c) <u>Response time</u>

The transducer's response time is determined by monitoring the rate of change in transducer output resulting from an isolated impulsive load. A digital storage oscilliscope is used to store this response (Fig 2.12).

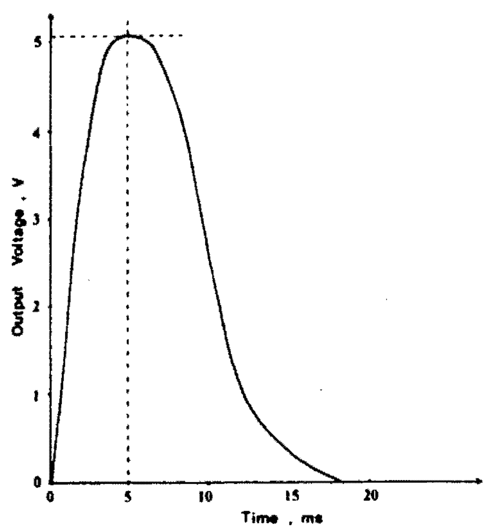


Fig 2.12 : Transducer's output response to an impulsive load

The transducer's response time (98% FSO) of 5 ms (approx) and associated decay time of 11 ms (approx) is within the specifications of section 1.4. This ensures the registration of high-frequency plantar pressures.

By virtue of the output returning to zero, no permanent variation in G_r^* takes place. This agrees with the theoretical predictions of section 2.2.3 (b).

d) Temperature effects

The effect of temperature on the transducer is determined by inserting it in an environmental chamber and monitoring its output response resulting from various controlled temperature gradients.

In its unstressed state, no variation in rest capacitance is observed for temperatures in the range 0 to 50° C. In verifying the variation in λ with temperature as calculated in section 2.2.3 (a), the transducer is removed from the chamber upon reaching the temperature extremities and statically loaded. The resulting response compares favourably with equation 2.11, with a maximum deviation of 4.1% in FSO.

Variation in plate spacing under static loading conditions is measured using a tensometer and micrometer. The application of a 250 N load results in a variation in plate spacing of 42 x 10^{-3} mm, thus differing from the predicted value by 6%. This discrepancy can be attributed to G_r^* not being corrected to the ambient measuring temperature.

e) Sensitivity

The minimum sensitivity is determined by incrementally loading the transducer with small loads and determining the minimum load required fo the output to register a change. This is observed to be 35 grams (approx).

The resolution on the other hand, is found to be better than 100 grams below 10 kg and above 10 kg, better than 200 grams (average).

All the measured transducer values conform to the desired specifications and correlate well with the theoretical predictions of section 2.2.3.

2.5 BATCH TRANSDUCER MANUFACTURE

The standardisation of transducer manufacture is an important aspect if they are all to conform to the desired specification.

In batch producing these transducers, ten unbonded transducers are stacked on each other with cardboard spacers separating them. After being placed in the tensometer jig and loaded with a 100 N force, a thin film of adhesive is applied to the transducer periphery (Fig 2.13).

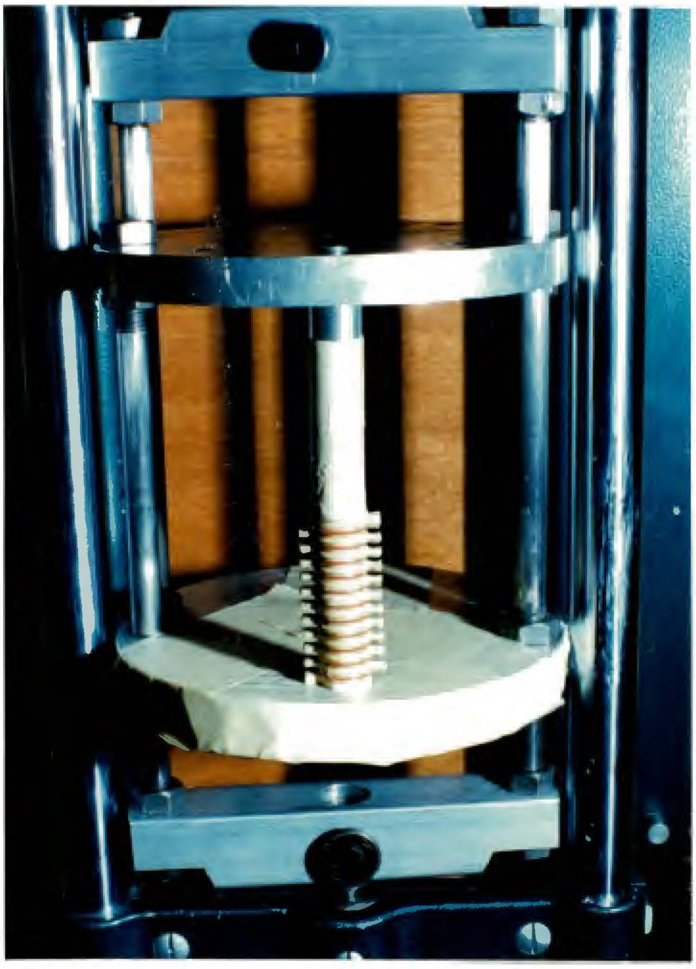


Fig 2.13 : Transducer bonding process

The pressure is subsequently released and the adhesive is allowed to cure for 24 hours.

After curing, the transducers are separated and subjected to a static calibration test. Approximately 10% of these transducers fail to conform to the desired specifications and are thus rejected.

2.6 INSOLE CONSTRUCTION

The accommodation of the needs of all patients, and the retention of a flexible gait-monitoring system, demand that a range of insole sizes be available. Sizes four to ten are chosen for the pilot study. Each have eight transducers strategically embedded to coincide with the predetermined plantar regions, as described in section 1.3.

In the evaluation of running, paediatric and neuromotor dysfunction, it becomes apparent that other selected segmental points may require monitoring. For such applications, a whole range of insoles incorporating different transducer configurations must be available.

The insole is made from 2 mm thick foam rubber. After being cut to a particular shoe size, the transducer locations are marked using an overlay containing information derived from X-rays of load-bearing feet of the same shoe size. These locations are then punched out, using the same punch as that used to produce the transducer plates and dielectric pads.

Signals are transmitted to and from the transducer sites by means of 2 mm diameter coaxial cables (Sühner RG178B/U), which are embedded in channels cut into the insole material. These are secured in position using a rubber adhesive. Good contact between the transducer and the plantar surface is ensured by having the transducer height more than that of the surrounding insole material.

The transducer connecting wires require special attention as stress applied to them can cause them to break. This problem is overcome by using copper brush-wire which is looped in a small cavity provided near the transducer contact points (Fig 2.14).

Fig 2.14 : Exposed insole showing transducer mounting

The insole is finally covered on both sides with a thin vinyl material so that the transducers are well protected. The insole in total weighs 150 grams (max) and is 3.5 mm thick. Tests have shown it to be most comfortable, allowing for the effective monitoring of all modes of gait.

2.7 SUMMARY

The various types of pressure transducers used in gait analysis systems currently in use were examined as to their suitability with the proposed insole system. Their unsuitability and lack of commercially-available transducers necessitated the design and manufacture of capacitive pressure transducers, capable of high pressure measurement.

The dielectric material as used for these capacitors was analysed and its performance modelled. Comparison with results obtained from prototype transducer testing showed good correlation.

The method of batch transducer manufacture under controlled conditions was described. This resulted in all the transducers exhibiting the same capacitance to load characteristic.

The manufacture of the insole was described. Use of a looping technique at the transducer contact points allows unrestricted insole movement. This permits the monitoring of fast modes of gait.

CHAPTER THREE

THE COMPONENTS OF THE NEW PORTABLE GAIT ANALYSIS DEVICE

In the design of a gait analysis device, a modular construction technique is considered to be most suitable. With this, performance characteristics can easily be checked allowing design change and improvements to be most conveniently effected. The size of the complete system should be kept as small as possible, thus requiring careful and economical layouts.

The usefulness and suitability of any gait analysis device is measured against the performance criteria in the field of application. Its acceptability by the user is greatly enhanced through ease of operation, as well as clear and understandable results. The provision of a real-time gait pressure display in addition to computer-assisted gait analysis could well give this device a clear competitive advantage.

3.1 FUNCTIONAL SYSTEM COMPONENTS

A low-cost gait analysis device statisfying the criteria of section 1.2 was designed and manufactured. A block diagram describing all the functional components is shown in Fig 3.1.

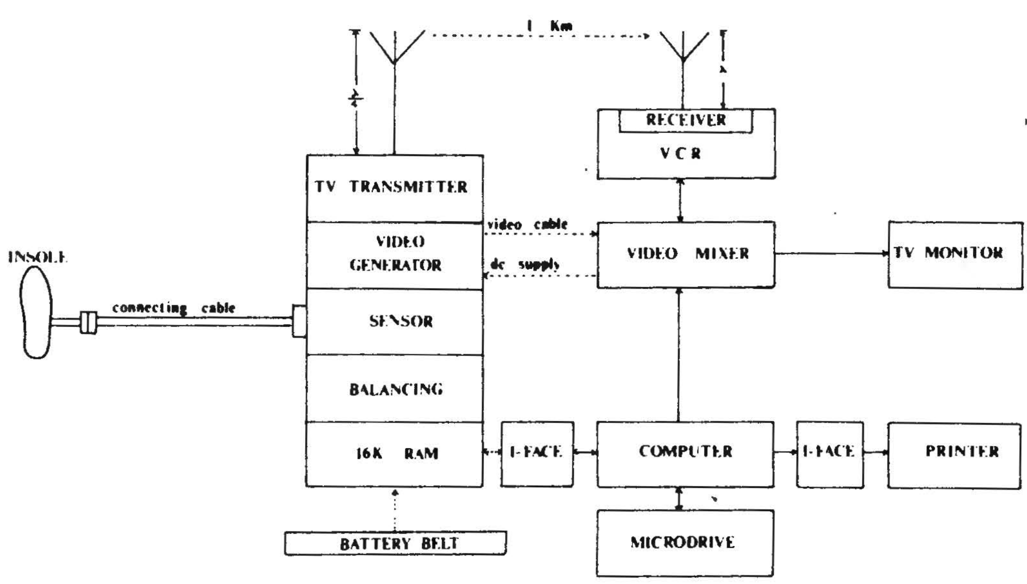


Fig 3.1 : Modular format of the functional components of the new gait analysis device

A functional description of each module comprising the new gait analysis device follows.

3.1.1 INSOLE AND CONNECTING CABLE

Prior to gait measurement, the insole is fitted into the patient's shoe and connected to the processing circuitry by a long flexible cable. Because of the symmetry normally expected between the left and the right foot, but more particularly of the plantar pressure regions, allows the insole to be used for both feet. An equalising insole of similar dimensions is placed in the other shoe to maintain a balanced condition.

Monitoring of an athlete's gait profile in a clinical environment is carried out on a variable speed treadmill. Since the processing circuitry is contained in a backpack unit capable of being worn by the patient, the availability of long insole connecting cables allow for the backpack to be placed alongside during gait evaluation. Use of such cables between the insole and backpack unit is made possible by the implementation of a new capacitance measuring technique, described in the following chapter.

3.1.2 BACKPACK PROCESSING UNIT

The processing circuitry of the gait analysis device is contained in a backpack unit shown in Fig 3.2. In its present design, this unit measures 230 x 120 x 95 mm, and has a total mass of 1800 grams. Since the memory and transmitter modules are not always required, they may be detached, thus reducing the mass by 860 grams.

Fig 3.2 : Backpack unit

The functions performed by the backpack unit are as follows :-

a) converts the transducer capacitance values into discrete voltages;
b) balances each transducer channel prior to gait measurement;
c) encodes the transducer voltages into a video signal for display on a television (TV) monitor;
d) permits wireless gait information transmission on a TV link; and
e) stores 16K of gait information for computer analysis.

Power to the backpack unit can be supplied from a battery belt worn by the patient or alternatively supplied with an umbilical cable when remote measurements are not required.

3.1.3 VIDEO MIXER UNIT (VMU)

A video mixer is required for the simultaneous display of the video information pertaining to the gait analysis device and the computer. The techniques used by those devices currently available are sophisticated in nature resulting in a high unit-cost.

A new type of video mixing technique was thus developed for use with the gait analysis device. This resulted in the manufacture of a low-cost VMU (Fig 3.3) that permitted a software-generated pressure overlay to be superimposed with that of the incoming gait information (Fig 3.4).

Fig 3.3 : New-type, low-cost VMU

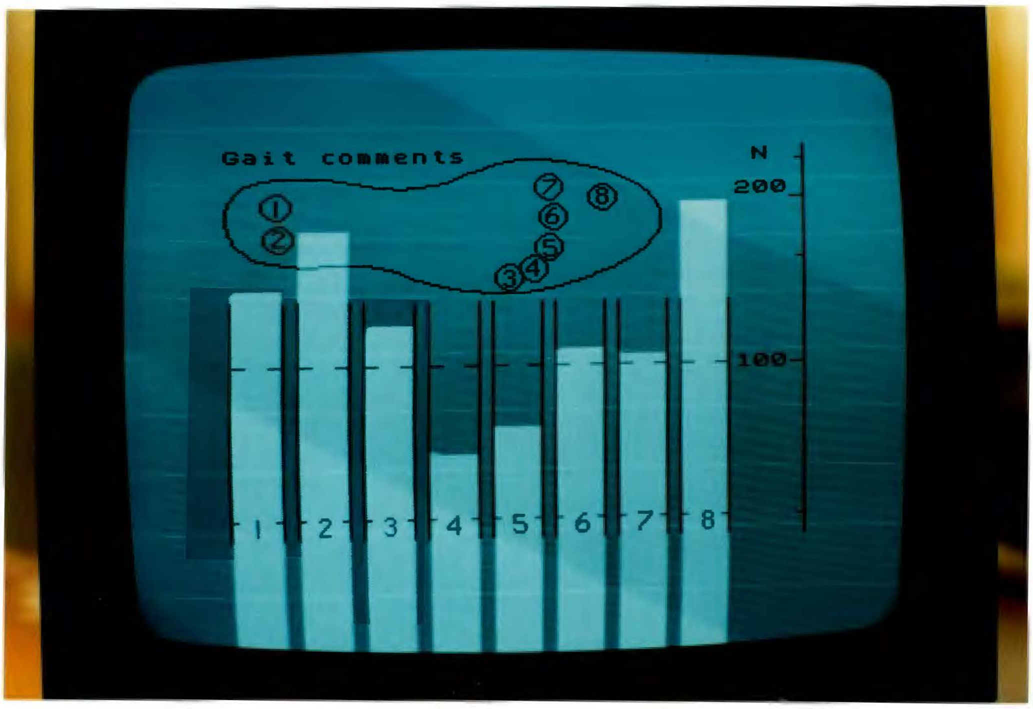


Fig 3.4 : Superimposed computer-generated pressure-graticule

The circuitry used in the design of this novel video mixer is dealt with in detail in the following chapter.

For convenience, the power source for the backpack unit, when used in a clinical environment, resides in the VMU.

3.1.4 COMPUTER AND PERIPHERALS

A Sinclair Spectrum™ (48K) home computer is used with this system. In addition to ensuring portability, system costs are kept to a minimum. All software used with this system is configured in a menu-driven format and is stored on a microdrive cassette.

The computer is used for analysing the gait information stored during digital gait sampling. A STAR Gemini 10X printer, utilising a TASMAN Type B printer interface, is used for producing the hardcopy printouts of the computed gait measures.

In addition to providing a software-generated pressure graticule, the computer may also be used for entering patient particulars and past medical history for storage purposes.

3.1.5 VIDEO CASSETTE RECORDER (VCR)

A VCR may be used to record the combined video signal relating to the incoming gait information and the computer for analysis as well as storage purposes. This allows the gait pattern to be examined under real-time or slow-motion conditions.

Since the VCR is used to demodulate the incoming RF information, it must be able to accommodate frequencies in the BAND I region. A video output socket must be available on this machine to allow this signal to be input to the VMU.

A National NV7000 BA VCR is used with this system. In addition to satisfying the above requirements, its freeze-frame facility has proved to be most useful.

There are many factors which can affect the pattern of gait, for example, shoe apparel, walking surface, inserts, etc. - factors which must be considered for the diagnosis of an abnormality of gait. A facility thus exists that allows comments and remarks to be added to the recorded video display.

3.2 DEVICE CONFIGURATION

The new gait analysis device can operate in both the clinical and outdoor environment. In the clinical environment, the backpack may be placed alongside the treadmill (Fig 3.5), or worn by the patient in the outdoor environment (Fig 3.6).

Fig 3.5 : Remotely situated backpack unit for clinical gait evaluation

Fig 3.6 : Backpack configuration for remote pressure measurements

The complete trolley-mounted gait analysis device as used in the clinical environment is shown in Fig 3.7.

Fig 3.7 : Complete gait analysis system

Applications where only a visual gait assessment is required, allow for a number of peripherals and modules to be dispensed with. Although affording a huge cost saving, no computer-aided gait evaluation is possible.

3.3 SUMMARY

The various components comprising the new gait analysis device have been presented together with a description of their function. Its modular format allows for its configuration in a number of ways, depending on the measuring requirements.

The implementation of a novel video mixing technique allows for a comprehensive real-time pressure display to be produced.

This facilitates in the accurate diagnosis of a foot/gait disorder. Optimal computer analysis makes provision for a hardcopy printout of gait-related measurements.

The low cost and portability afforded by this new gait analysis device make it unique for the measurement of static and dynamic inshoe plantar pressures.

CHAPTER FOUR

HARDWARE TECHNIQUES

A number of new design techniques have been incorporated in the manufacture of the new gait analysis device. These include a method for the remote measurement of small capacitance values, in addition to a new technique for the mixing of two crystal-referenced video signals.

Each module of the gait analysis device will be treated separately. Where necessary, an evaluation of the circuitry used will be given. For a comprehensive set of circuit diagrams, refer to Appendix A.

4.1 INSOLE

The insole, containing eight equally-dimensioned transducers, may be schematically represented as shown in Fig 4.1.

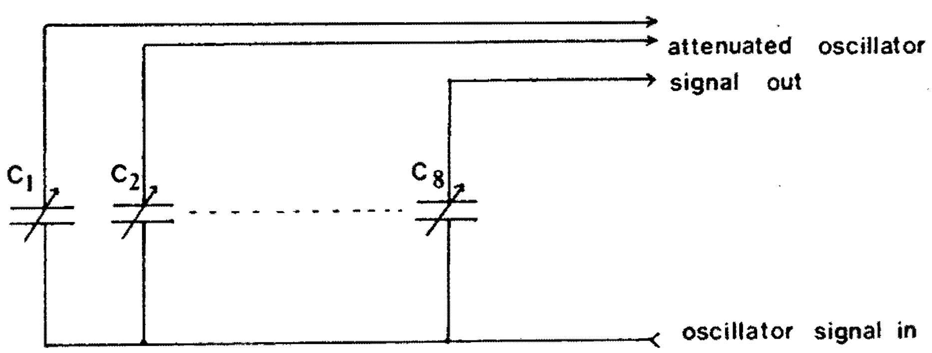


Fig 4.1 : Schematic representation of the new instrumented insole

Since the transducer capacitance values are low, varying from 7 to 12 pf under load, the effect of ground proximity on the transducer outputs must be evaluated. Here, two cases are considered - a) oscillator side of transducer flush with the ground-plane, and b) transducer take-off point flush with the ground-plane. For both these cases, it is assumed that all the transducers are stressed simultaneously and that the ground-plane is perfectly conductive.

a) Oscillator-side flush with ground-plane

This condition may be represented as shown in Fig 4.2.

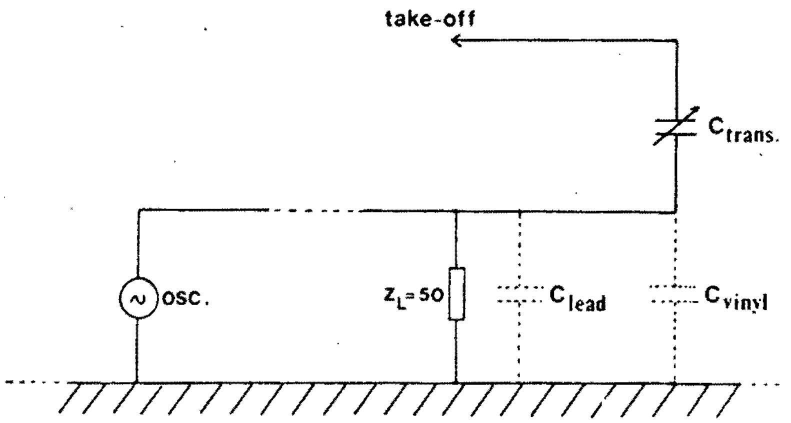


Fig 4.2 : Transducer representation when oscillator-side flush with ground-plane

Under maximum loading conditions, the combined leakage capacitance contributed by C_{lead} and C_{vinyl} ~ 5 pf. By virtue of these being in parallel with the inserted 50 - ohm load impedance Z_L, negligible transducer signal attenuation occurs.

b) Take-off side flush with ground-plane

This condition may be represented as shown in Fig 4.3.

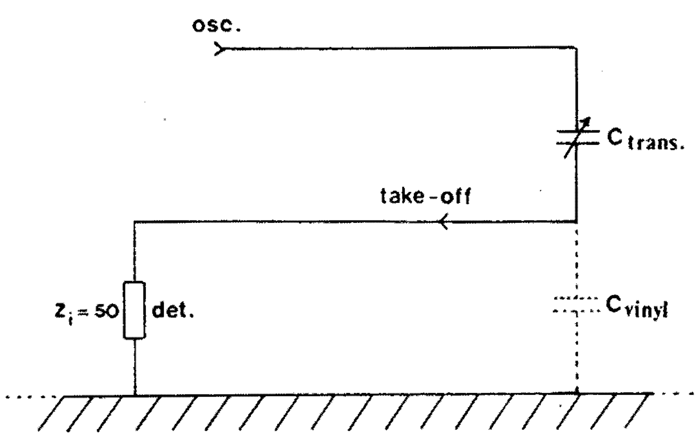


Fig 4.3 : Transducer representation when take-off point flush with ground-plane

By uniformly stressing the insole parallel to the ground-plane, $C_{vinyl} \sim 3$ pf. By virtue of this high impedance being in parallel with that of the detector's input impedance ($Z_i = 50$ - ohms), results in negligible attenuation of the transducer's output signal.

4.2 INSOLE-CONNECTING CABLE

Two methods are essentially used in the measurement of small capacitance values. Either a high-frequency alternating current (AC) bridge is used, or alternatively, the capacitor can be made part of a resonant circuit. In both these cases, the capacitance of the interconnecting leads or cables must be taken into account.

In allowing for variable lengths of insole-connecting cables to be used without affecting the system accuracy, a new technique is introduced for the monitoring of small capacitance values when situated a distance from the measuring equipment. This technique requires that signal transfer to and from the transducer site be implemented effected using suitably matched and terminated transmission lines. A schematic of this measurement technique is shown in Fig 4.4.

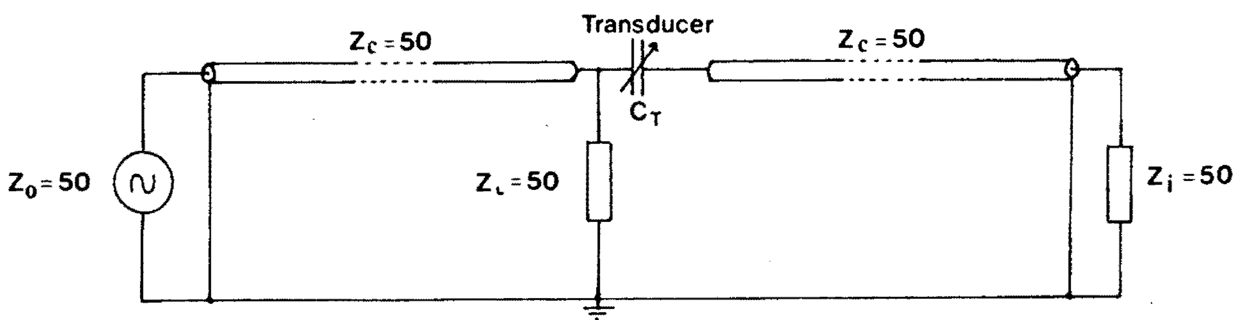


Fig 4.4 : New capacitance measurement technique

To prevent the effects of radio frequency (RF) electromagnetic and electrostatic interference, the screens of the above coaxial cables are connected at ground potential.

Transducer capacitance variations are detected as variations in oscillator signal amplitude at the detector input. In ensuring maximum signal transfer on the line, load impedances Z_L and Z_i are made equal to Z_c, the characteristic line impedance. This results in the voltage reflection coefficient Γ being zero as no reflected waves exist on the line. This is verified by the relation,

$$\Gamma = (Z_L - Z_c)/(Z_L + Z_c)^{27} \qquad \ldots\ldots 4.1$$

Z_L may be configured from a simple 50 - ohm resistor whereas a suitably designed common-base (c-b) stage must be used for Z_i. This would allow the attenuated transducer signal to be amplified with the minimum ~~amount~~ number of components.

It is imperative for the transducer lead wires to be as short as possible. This would then allow the coaxial transmission cables to be situated in the immediate transducer vicinity.

Since short lengths of coaxial cables are used (< 5 m), their resistance losses may be neglected. Thus, no signal degradation or loss of sensitivity is encountered when using connecting cables of variable lengths.

4.3 BACKPACK UNIT

A block diagram of the modules performing the various functions within the backpack unit are shown in Fig 4.5.

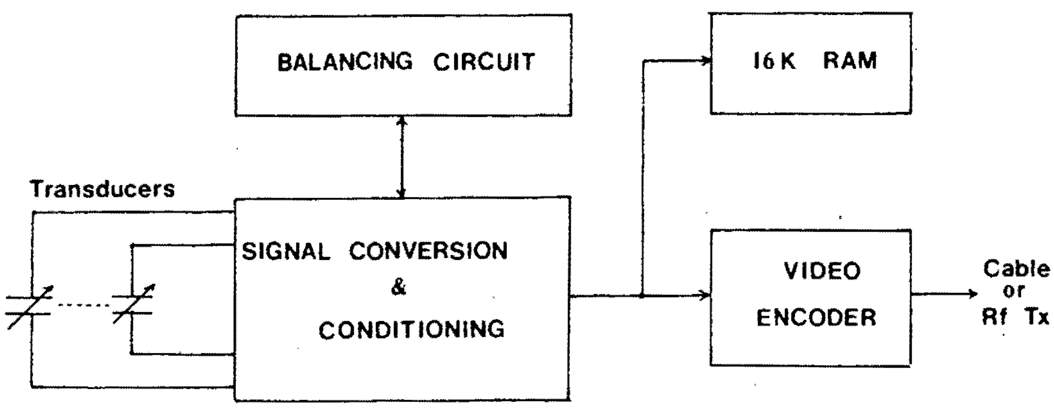


Fig 4.5 : Backpack module configuration

All power supplies within this unit are regulated. The batteries used in the battery-belt are of the lead-acid type, (YUASA NP 1.2-6) because of their relatively short charge-cycle.

4.3.1 SENSOR MODULE

This module detects variations in capacitance values and converts them into discrete voltages, suitable for input to the video encoder and memory module. The signal conversion and conditioning taking place in this module is shown for one channel in Fig 4.6.

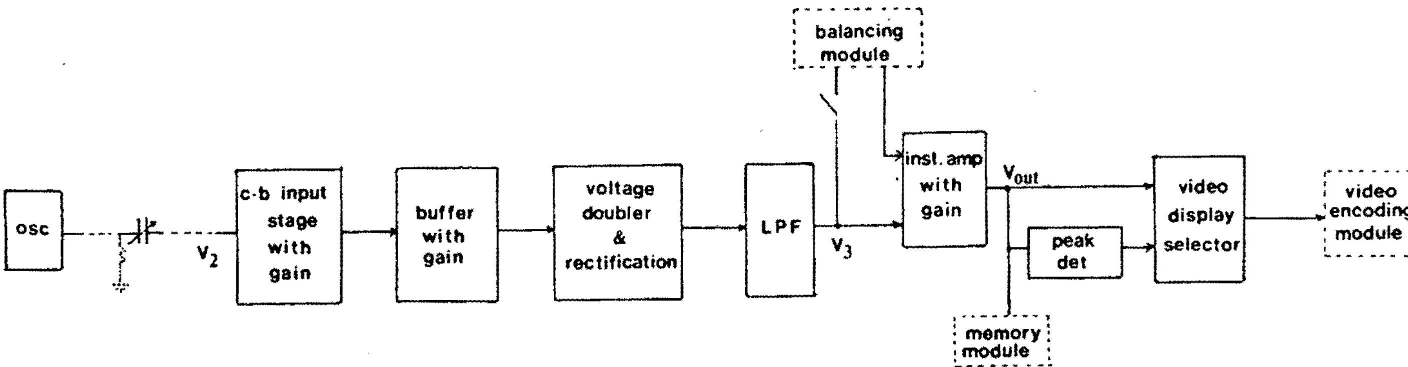


Fig 4.6 : Signal conversion and conditioning for one transducer channel

A circuit analysis of this module follows.

a) Oscillator

A sine-wave oscillator is requried for supplying each transducer with an AC voltage. The transducer, acting simply as a coupling element between the oscillator and detection circuit, attenuates this signal depending on the load exerted on it. The transducers Its impedance is governed by the equation,

$$Z_T = 1/(\omega C_T) \qquad \ldots\ldots 4.2$$

where: $\omega = 2\pi f$ = constant

C_T = load-dependent transducer capacitance.

Stability in oscillator frequency and amplitude is essential for accurate gait pressure measurements. This is best achieved by use of a crystal-controlled oscillator, followed by a bootstrapped emitter-follower for providing the isolation. A 5.5 V_{PP}, 250 kHz oscillator signal is thus derived.

The maximum variation in oscillator load impedance resulting from transducer loading is calculated as 0.74%. Since the current sourcing capability of the oscillator is far greater than that required by all the transducers (410 μA), no "current-hogging" phenomena between the transducers is experienced.

b) <u>Determination of c-b input voltage, v_2</u>

Oscillator signal attenuation resulting from the high transducer impedance is counteracted by use of a high gain c-b input detector stage. The equation relating the detector input voltage v_2 to the variable transducer impedance Z_T, is derived with the aid of Fig 4.7.

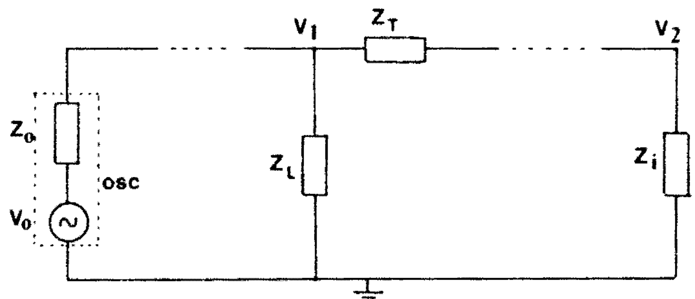


Fig 4.7 : Equivalent circuit for detector input-voltage derivation

Voltage at node 2 = v_2 = $(Z_i / (Z_i + Z_T))\ v_1$ 4.3

Since v_1 = 2.72 volts (peak-to-peak) and $53k \leqslant |Z_T| \leqslant 91$ k-ohms for $0 \leqslant F \leqslant 250$ N, results in a maximum voltage variation in v_2 of 1 mv (approx). The gain contributed by the c-b input stage amplifies this voltage variation by 307 (approx).

In increasing the input voltage variations still further, an amplifying buffer stage in the form of an inverting operational amplifier (op-amp) is used.

c) <u>Determination of instrument amplifier input voltage, v_3</u>

Digital circuitry as used in effecting channel balancing, video encoding and data storage, requires direct current (DC) voltages as its inputs. The detector's output signal must therefore be rectified.

A voltage doubling rectification stage, configured with germanium diodes for minimising the insertion loss, is used, followed by a passive low pass filter (LPF) for reducing any remaining ripple.

The equation that relates v_3 to v_2 is shown as,

$$v_3 = ((A_{c-b} . A_b) v_2 - 2\ v_d).DR \qquad ...... 4.4.$$

where : A_{c-b} = voltage gain of c-b stage (~ 307)
A_b = voltage gain of buffer stage (~ 3.9)
v_d = diode forward volt drop (~ 0.2)
DR = division ratio of LPF. (~0.99)

For maximum transducer loading, v_3 can vary from 1.29 volts (no-load) to 2.5 volts (full-load).

The video encoder, as used with this gait analysis device, requires as its input, transducer dependent voltages ranging from 0 to +5 volts. v_3 must therefore be modified to ensure its compatibility.

d) <u>Derivation of V_{out}</u>

A method for allowing variations in v_3 to register as variations in a 0 to +5 volt range is shown in Fig 4.8.

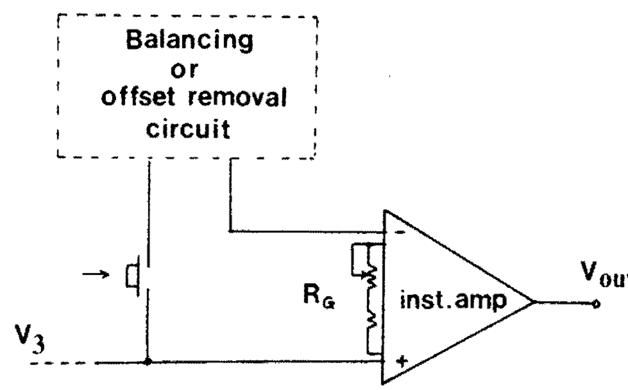


Fig 4.8 : Voltage translation circuit

The instrumentation amplifier output voltage V_{out}, is related to its input voltages V^+ and V^- by the following equation,

$$V_{out} = A_{vcl} . (V^+ - V^-) \text{ volts} \qquad ...... 4.5$$

where : A_{vcl} = closed loop voltage gain.

By ensuring $V^+ = V^-$ prior to transducer loading, and by correctly choosing A_{vcl}, V_{out} can be made to vary from 0 to +5 volts in response to maximum transducer loading.

V^+, being equal to v_3, was shown to have a maximum voltage variation of 1.21 volts. In allowing the transducer's characteristic to be represented by equation 2.11, it is necessary for V_{out} to register as 4.74 volts in response to a 250 N load. This can be achieved if A_{vcl} = 3.92.

For the LH0036 instrument amplifier, A_{vcl} may be expressed as,

A_{vcl} = 1 + $(50k/R_G)$ [28] 4.6

where : R_G = resistance between pins 4 and 7.

In allowing V_{out} to be accurately set, R_G is configured so that $15k \leqslant R_G \leqslant 18k3$ - ohms.

e) <u>Video display selector</u>

During certain modes of gait, V_{out} changes rapidly as a result of pressure variations on the plantar surface. Since the values of V_{out} are encoded for display on a TV monitor, a quantitative evaluation of the resulting pressure display concerning the peak values is difficult and necessitates the inclusion of a peak-hold facility. Use of an externally activated electronic switch allows the display to be frozen for 5 second intervals after which time it returns to the dynamic mode. The circuit for controlling the video display in schematically shown in Fig 4.9.

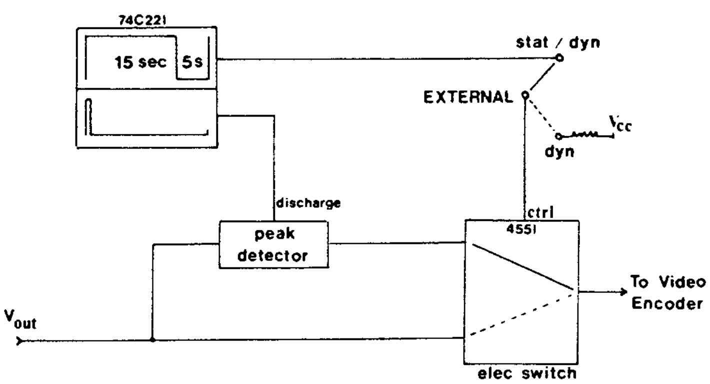


Fig 4.9 : Schematic of video display-freeze circuit

4.3.2 CHANNEL-BALANCING MODULE

Channel balancing is accomplished by pressing an externally mounted switch. The voltages pertaining to the unloaded transducer states are digitally sampled and stored in a random access memory (RAM) integrated circuit (IC). These are then reconverted back to their analog format for use with the op-amp configuration of Fig 4.8. A schematic of this circuit is shown in Fig 4.10.

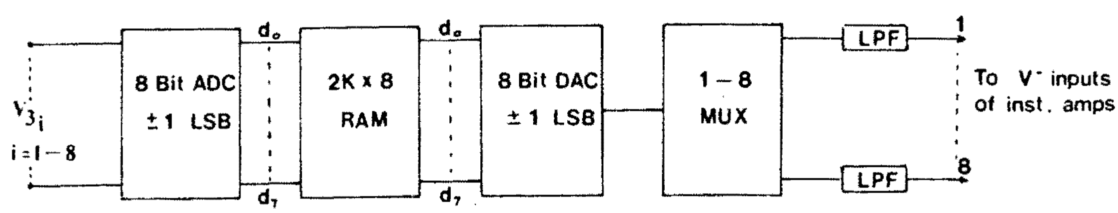


Fig 4.10 : Schematic of channel-balancing circuit

The accuracy of V_{out} is dependent on the conversion accuracy of the analog-to-digital converter (ADC) and digital-to-analog converter (DAC) as used in the above circuit. Use of 8 bit devices, each having 1 least significant bit (LSB) accuracies, allows the error in output voltage, $V_{E\ out}$, to be expressed as,

$$V_{E\ out} = \pm A_{vcl}\,(LSB_{ADC} + LSB_{DAC}) . \qquad \ldots\ldots\ 4.7$$

Since $LSB_{ADC} = LSB_{DAC} = 11.72$ mV, $V_{E\ out} = \pm\ 0.09$ volts.

$V_{E\ out}$ contributes to the overall system-measuring accuracy and is only relevant when localised forces in excess of 200 N are being registered.

4.3.3 VIDEO ENCODING MODULE

A modified version of the video encoding circuit of R. de Jong[29] (Appendix B), is used in providing the visual pressure display for this gait analysis device.

The encoder provides a video signal, which, when connected to a TV monitor, produces a vertical display of pressure bars, with the height of each providing an accurate representation of the pressure applied.

A schematic of the modified encoder circuit is shown in Fig 4.11.

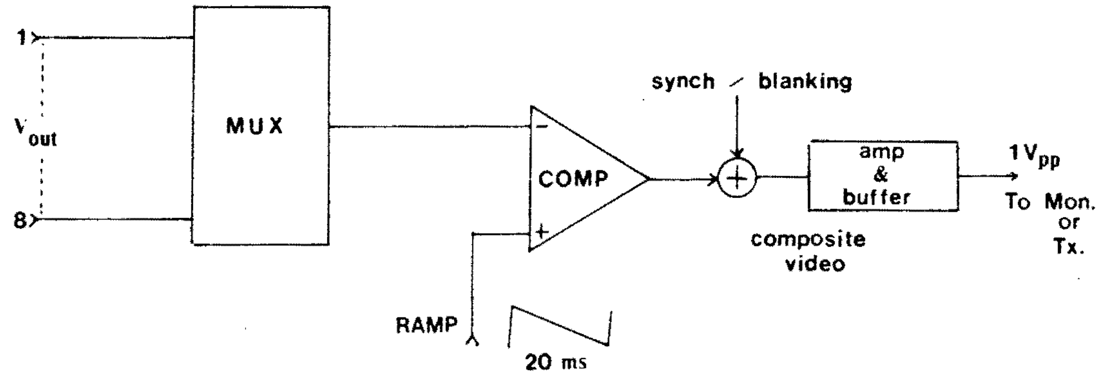


Fig 4.11 : Schematic of the video encoder module

A linear ramp voltage is used as the comparitor's reference to ensure a linear video display of the gait pressures. This reference voltage is compared with each transducer input voltage in turn, thus causing the comparitor to generate a signal which turns the TV tube gun on or off. However, this depends on whether it is scanning across a bar or not. These signals are subsequently mixed with the horizontal and vertical synchronisation (synch) pulses to form the composite video signal.

For the video encoder's output signal to be compatible with most TV monitors, it must have an amplitude of 1 V_{PP} (0,7 V signal : 0.3V synch). This necessitates the inclusion of an amplification stage and buffer. For remote pressure measurements, this signal may be input to a small TV transmitter.

4.3.4 TV TRANSMITTER MODULE

Instead of a long trailing video cable, a 1 Watt TV transmitter weighing no more than 250 grams may be mounted on the backpack for remote pressure measurements. Operating in the VHF Band I (47-68 MHz) region, this offers ultimate measurement flexibility. A safe range of up to 1 km has been obtained using a matched whip-antenna for transmission and a three-element yagi for base-station reception.

4.3.5 16K GAIT STORAGE MODULE

In addition to V_{out} being used in producing a visual pressure display, it may also be sampled and stored in a RAM module, available as an option for those practitioners who require computer-aided gait analysis.

The module is self-contained, and is attached to the backpack by four press-studs which, when pressed home, will allow the socket and plug to be engaged for the electrical connection.

This memory module is schematically shown in Fig 4.12.

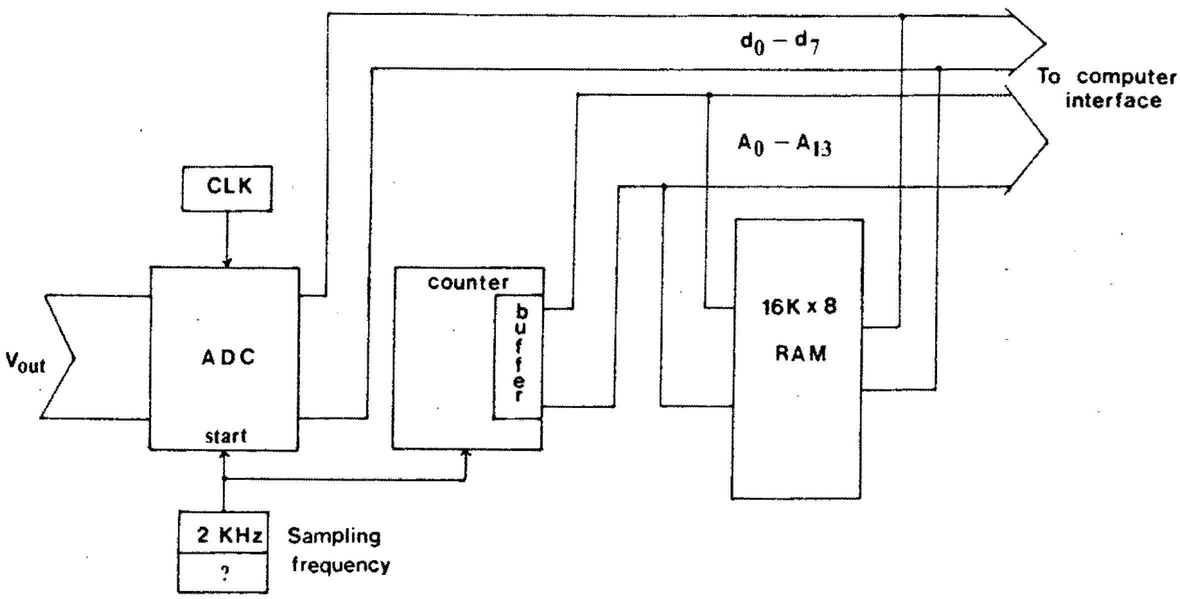


Fig 4.12 : Schematic of 16K memory module

The memory module uses an 8 bit ADC (ADC0809) for data encoding and can sequentially store 16K of gait information. A fixed sampling frequency of 2 kHz has been used in studies so far; however, a variable frequency option is available with a maximum sampling rate of 10 kHz.

The 2 kHz sampling frequency ensures a high resolution in stored gait information for all modes of gait, since each transducer is sampled at 4 ms intervals.

Sampling and storage of gait information is initiated by a hand-held push-button. In ensuring the capture of natural gait cycles, a time delay function in the form of a monostable is simultaneously activated, which inhibits the ADC clock for a few seconds.

On completion of gait storage, the memory bus is TRI-STATED, thus allowing it to be connected to that of any computer for data downloading.

4.4 VIDEO MIXER UNIT : A NEW DESIGN

If the video signals pertaining to the gait analysis device and computer are to be simultaneously displayed on a TV monitor, they must both be of the same phase and frequency. The conventional method used in mixing two remotely-situated, independently-generated video signals, makes use of a digital framestore technique, shown in Fig 4.13.

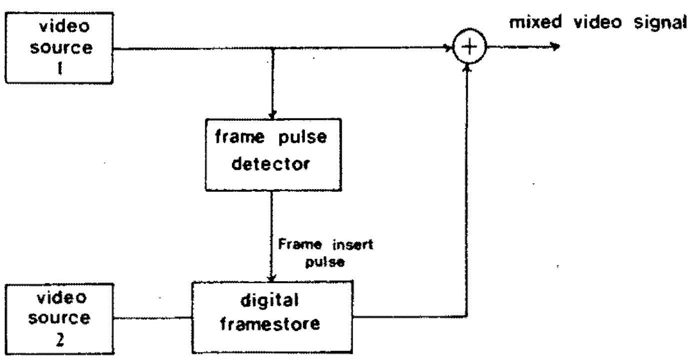


Fig 4.13 : Framestore video mixing technique

The high probability that the phase relationship of these two video signals will not be the same for mixing purposes, requires the continuous digital storage of one video signal on a frame-by-frame basis. A frame pulse detector, referenced to the other video signal, is used for controlling the mixing of these two signals.

This video mixing technique, although extensively used, is complex and results in a high VMU-production cost. In satisfying the criteria of section 1.2, a video mixer of a different type is certainly required.

A new method is thus introduced for the mixing of two remotely-situated, independently-generated, crystal-referenced video signals. This results in the design and manufacture of a new type, low-cost VMU, suitable for use with this gait analysis device. This VMU may be used for a variety of different applications and will be shown to be revolutionary in its field of use.

4.4.1 PROPOSED VIDEO MIXING TECHNIQUE

If two remote, crystal-referenced video signals are of the same frequency but different in phase, their phase may be equalised by momentarily increasing or decreasing the frequency of one video signal relative to that of the other until coincidence occurs.

In maintaining the RF measuring link for this gait analysis device, alteration to the computer's crystal frequency is required. The generation of the computer's video timing-signals take place in a dedicated IC, called the Uncommitted Logic Array (ULA), and are shown in Fig 4.14.

14 MHz		crystal frequency
÷ 4	3.5 MHz	Z80 A CLK
÷ 224	15625 Hz	video line frequency
÷ 312.5	50 Hz	video frame frequency

Fig 4.14 : Crystal-referenced video timing-signals

It is seen that the effect of altering the crystal's operating frequency is more pronounced in the line than in the frame frequency. This dictates the pulse locking order for the synch process, namely, frame before line. It thus becomes necessary to alter the crystal frequency twice - first by a large amount to obtain coincidence between the frame synch pulses, and then by a smaller amount to obtain coincidence between the line synch pulses. In obtaining this condition, the two video signals will be synchronised with each other and can then be mixed, for display on a TV monitor, by a simple resistive technique. On this basis, the new VMU is configured, and is schematically shown in Fig 4.15.

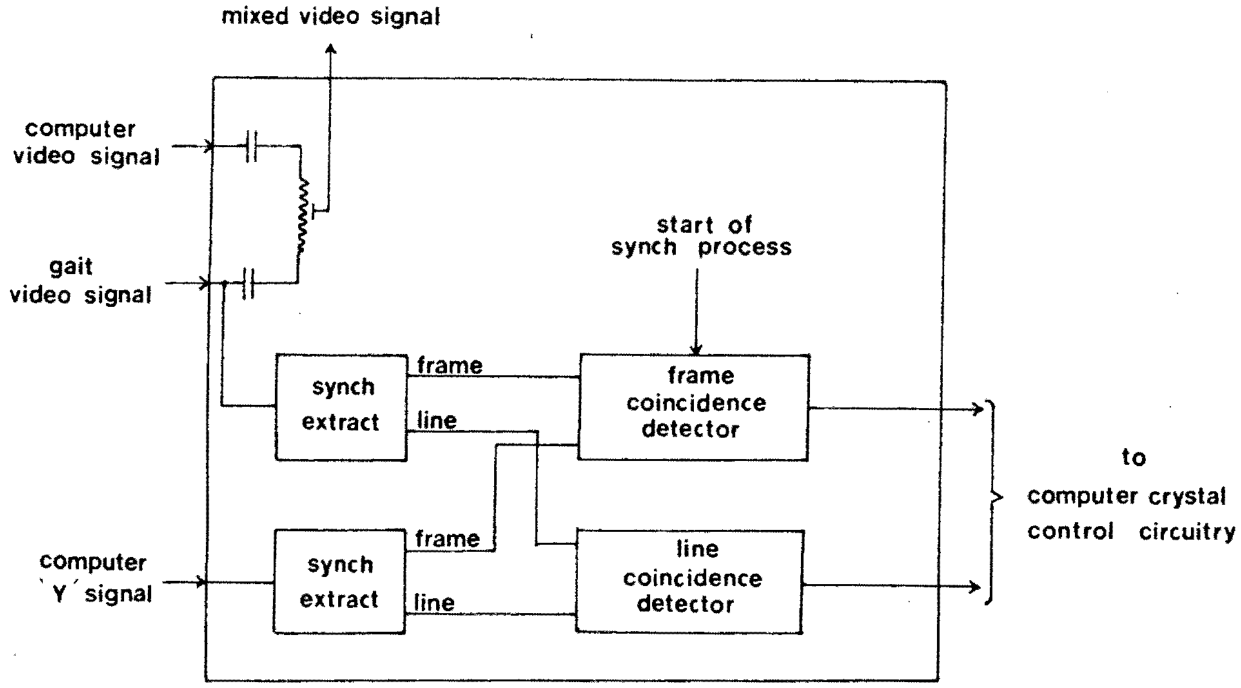


Fig 4.15 : Proposed design of a new VMU

4.4.2 METHODS USED IN CONTROLLING CRYSTAL FREQUENCY

In determining the best method for altering the computer's crystal frequency, the method used in driving this crystal must be examined (Fig 4.16).

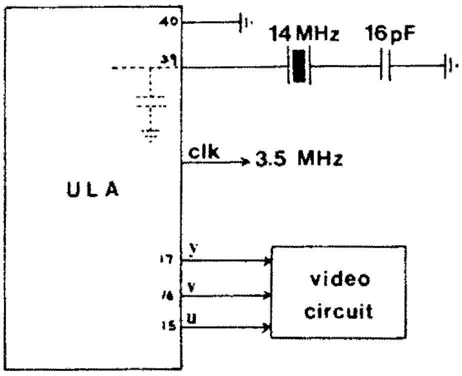


Fig 4.16 : Computer's crystal circuit[30]

The crystal is depicted as part of a π network. Insertion of additional load capacitance C_L, will increase the crystal's operating frequency by an amount Δf, given by [31]

$$\Delta f \sim f_s \frac{C_1(C_{L2} - C_{L1})}{2(C_o + C_{L1})(C_o + C_{L2})} \qquad \ldots\ldots \; 4.8$$

where : f_s = crystal's series resonant frequency (= 14 MHz)

C_o = crystal's shunt capacitance (= 5 pf)

C_1 = crystal's motional capacitance (= 20 ff)

C_{L1}, C_{L2} = minimum and maximum value of load capacitance.

The relative variation in crystal operating frequency resulting from a 20 pf variation in series load capacitance is shown in Fig 4.17 for a 10 MHz fundemental AT - crystal with C_o = 6 pf and C_1 = 20 ff.

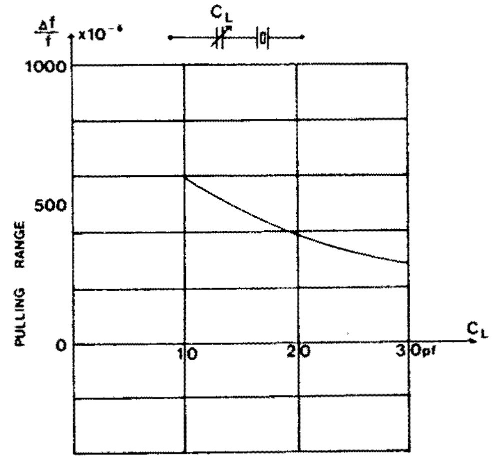


Fig 4.17 : Pulling range of a series load capacitor

A considerably larger pulling rate is achieved by the insertion of a series LC circuit instead of a pure capacitive load. The resulting frequency variation Δf, may be expressed as

32

$$\Delta f \sim f_s \frac{C_1}{2\left(C_o - \dfrac{1}{\omega_s{}^2 L_v - \dfrac{1}{C_v}}\right)} \qquad \ldots\ldots\ldots\ 4.9$$

where : f_s , C_1 , C_o as before

$\omega_s = 2\pi f_s$

L_v = series load inductance

C_v = series load capacitance.

L_v and C_v can be optimally chosen, allowing for the pulling range to be symmetrical to the crystal series resonance. By increasing L_v , the pulling range may be greatly extended towards the lower frequencies. This pulling behaviour is shown in Fig 4.18 for a 10 MHz fundamental AT - crystal with C_v = 10 - 30 pf, C_o = 6 pf and C_1 = 20 ff.

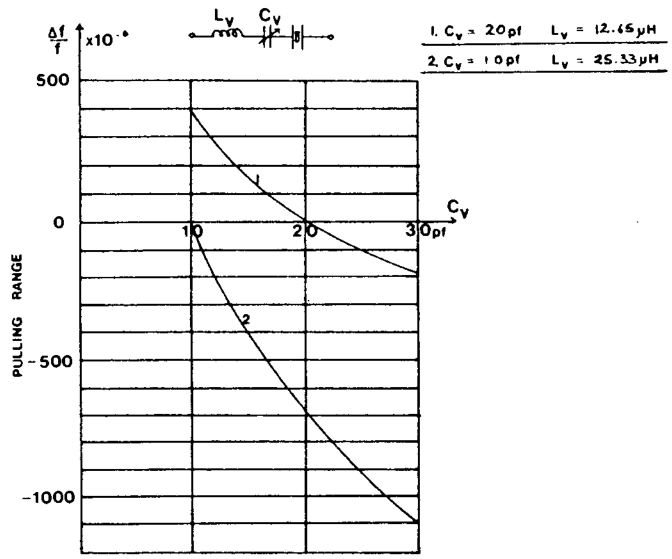


Fig 4.18 : Pulling range of a series resonant circuit

4.4.3 HARDWARE TECHNIQUES FOR THE ALTERATION OF CRYSTAL OPERATING FREQUENCY

In calculating the frequency variation resulting from the inserted crystal-control components, the stray capacitances between the components and the conductors, as well as the internal capacitance of the semiconductors, must be taken into consideration.

a) Large Frequency Variation

This method requires the controlled insertion of a series LC circuit in shunt with the ULA input (Fig 4.19). This circuit is electronically connected at the start of the synch process and disconnected upon detection of frame coincidence.

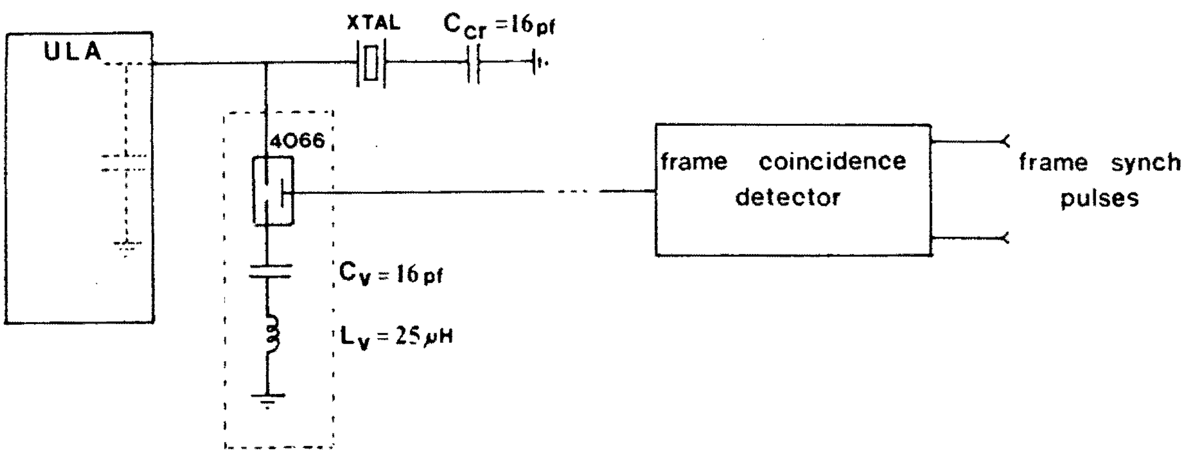


Fig 4.19 : Frame synch control circuitry

Using equation 4.9, a frequency variation of 75.5 kHz should result from the insertion of component values, C_v = 16 pf and L_v = 25μH, assuming a switch resistance (CD 4066) of 5 pf (approx).

b) Small Frequency Variation

Exclusive use of a pulling capacitor C_L, results in the smallest pulling range. To acquire line synch, only minor alterations to the crystal operating frequency are required. A controlled pulling range may be obtained by use of a varactor diode in place of C_L. Use of a type-I phase detector for altering the varactor diode bias voltage, and hence capacitance, would allow for synchronisation and locking of the two line-synch pulses. A schematic of the crystal configuration for varactor diode frequency control is shown in Fig 4.20.

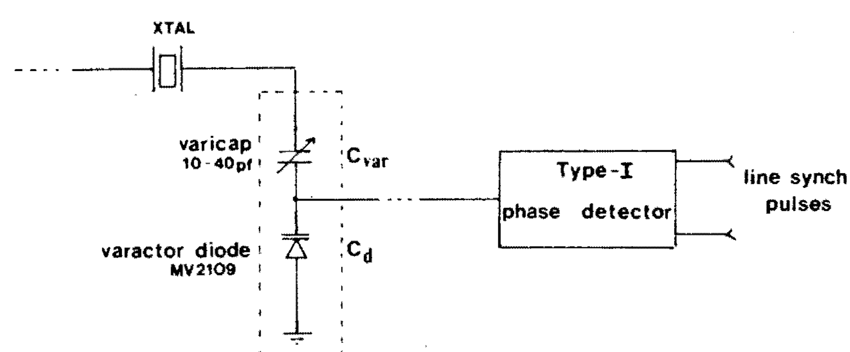


Fig 4.20 : Line-synch control circuitry

A varicap is connected in series with that of the varactor diode to decrease the pulling range even further.

The insertion of these components requires that C_{cr} = 16 pf (fixed) be disconnected. The varicap, C_{var}, is so adjusted making its combined capacitance with that of C_d at line synch equal to C_{cr}.

Since C_{var} = 36 pf and $22 \leqslant C_d \leqslant 29$ pf for $6 \leqslant V_d \leqslant 12$ volts, should result in a maximum frequency variation of 856 Hz, as calculated using equation 4.8.

4.4.4 VMU CIRCUIT IMPLEMENTATION

The frame and line synch pulses of both video signals are required for reference purposes. This is accomplished by use of synch separators at the VMU input.

The maximum frequency variations resulting from the implementation of the crystal control-circuits of Fig 4.19 and Fig 4.20 were measured as 105 kHz and 936 Hz respectively. The large discrepancy in the first measurement can be attributed to the unknown internal configuration of the ULA due to its proprietary design. The additional capacitance and inductance of this device should be taken into consideration in determining the exact anticipated frequency variation. The small discrepancy in the latter measurement can be attributed to the effect of stray and lead capacitance in connecting the relevant components.

In preventing 'picture roll' in the event of momentary RF signal loss, a monostable is used for producing the frame synch pulses. This monostable is triggered each time its integrated frame input signal reaches a predetermined amplitude.

For measurements in a clinical environment, use of a long video cable between the backpack and VMU can cause group delay in the gait video signal. This manifests itself as a visual misalignment between the pressure bars and software overlay on the TV monitor. Therefore, the inclusion of an external graticule shift control has been found necessary to overcome this phenomonen. This control introduces a variable time delay into the computer's line synch pulse.

4.5 COMPUTER

A number of modifications and circuit additions as discussed earlier, are required before this computer may be incorporated with the gait analysis device.

The limited current capability of the computer's power supply requires that the interface circuits be externally powered. An interlocking power protection circuit, making use of a low-current consumption relay, protects the unbuffered computer bus at power-up and power-down.

4.6 MEMORY MODULE INTERFACE

This interface (Fig 4.21), facilitates the downloading of stored gait information to a computer.

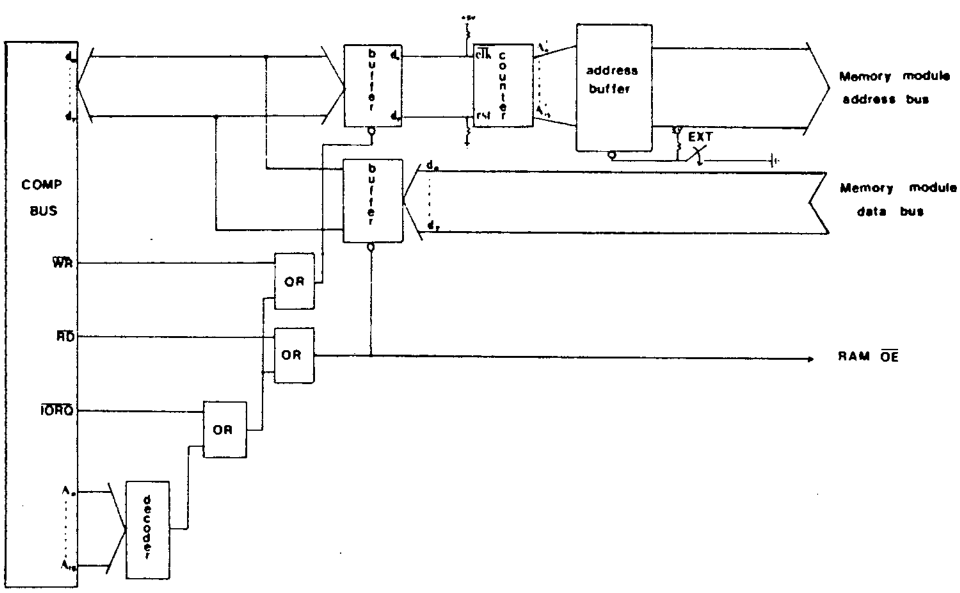


Fig 4.21 : Memory module interface

A 14 bit software-strobed counter, is connected to the memory module's address bus. Prior to start of data transfer, this counter is reset, allowing for gait data in the memory range 0000H - 3FFF to be addressed.

Gait information, appearing on the memory module's data bus, may be written to any designated location within the computer's memory. This information may be retrieved and analysed, or stored on a microdrive cassette for historical purposes.

4.7 SYSTEM CALIBRATION

For calibration purposes, the plug-in attachment of Fig 4.22 is inserted into the backpack unit in place of the memory module. This attachment makes use of a dual in-line package (DIP) switch as a channel selector, allowing for the simultaneous monitoring of any two sequential transducer channels. Digital voltmeters, DVM1 and DVM2, are used in obtaining an accurate representation of the transducer outputs.

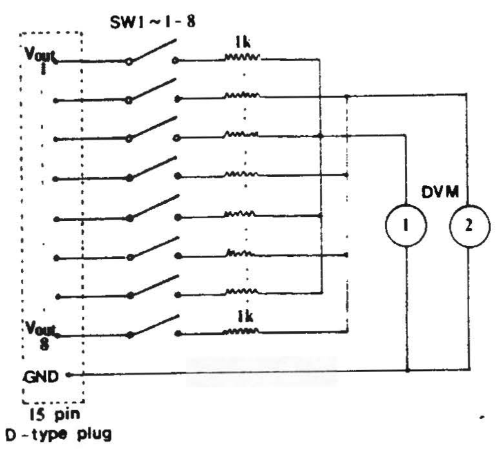


Fig 4.22 : Transducer output monitoring attachment

In calibrating the individual transducer channels, an insole is connected to the backpack. With the insole in its unloaded state, the individual transducer channels are first balanced by activating the appropriate external switch. For each channel in turn, the amplitude of V_{out} is set to 4.74 volts in response to a statically applied load of 250 N (Fig 4.23), by adjusting the relevant value of R_G (see Fig 4.8).

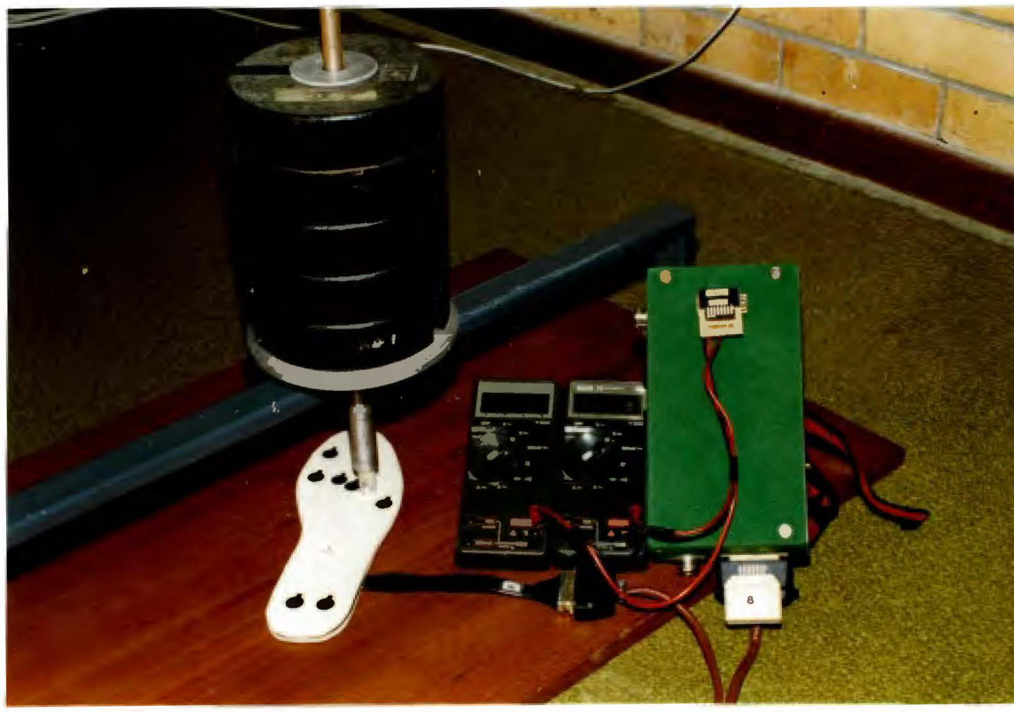


Fig 4.23 : Method of transducer load application for calibration purposes

Use of an 8 bit ADC with a 5 volt reference for the memory module allows equation 2.11 to be expressed as,

$$a = 255\ (1 - e^{-1/90}) \qquad \text{......}\ 4.10$$

where : a = binary equivalent of V_{out}, and $0 \leqslant a \leqslant 239$ for $0 \leqslant 1 \leqslant 250$ N.

This relationship between V_{out} and 'a' is used in calibrating the memory module.

4.8 SUMMARY

All the hardware techniques implemented in the design of this new gait analysis device have been presented. Exclusive use of CMOS and low-power devices minimise power consumption, thus permitting portable operation.

A novel transducer-monitoring technique was introduced in allowing variable lengths of insole-connecting cables to be used without affecting system accuracy. This relieves the patient from wearing the backpack for measurements conducted with a treadmill.

A new technique of synchronising two remotely-generated video signals was presented. This resulted in the design and manufacture of a new, low-cost VMU. Analysis of this technique shows its flexibility in steering and controlling crystal-related signals.

Stored gait information may be downloaded to any computer having a similar bus architecture to that of the memory module. Use of an IBM-PC or compatible would allow for a more extensive analysis of sampled gait information, in addition to providing sophisticated graphic pressure displays.

CHAPTER FIVE

GAIT ANALYSIS SOFTWARE

The incorporation of a computer in a gait analysis system can result in a more effective diagnosis of an abnormality of gait. This requires that all support software be extensive enough to provide the relevant gait information, and be presented in such a manner so as facilitates both intra and interpatient gait comparison.

The complexity of the mathematical operations to be performed are dependent on the requirement of the practitioner. The software written for this project was specifically aimed at a) the identification of gait abnormalities, b) the diagnosis of running-related injuries, and c) the therapeutic assessment of certain shoes, shoe inserts or insoles.

5.1 DISPLAY SOFTWARE

During ambulation, the incoming gait information is displayed in the form of a bar graph on a video monitor. To enhance this display, a software-generated pressure graticule, as well as a transducer location profile (Fig 5.1), is superimposed employing the video mixer described in section 3.1.3. Two software overlay programs are thus required, allowing for the display of either the left or right foot pressure profile.

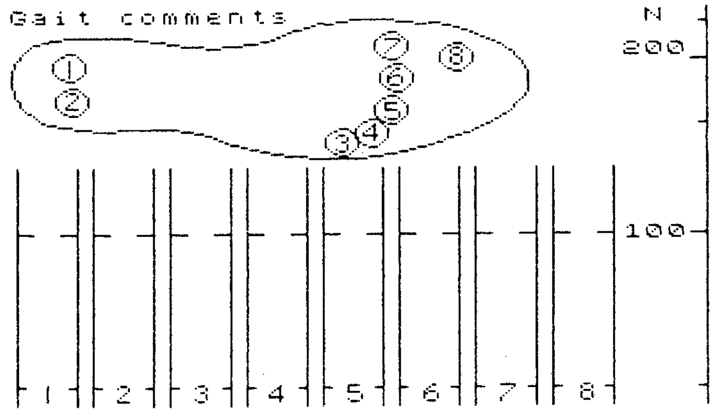


Fig 5.1 : Right foot pressure graticule overlay

A VCR can be used for recording the pressure bars and software overlay for analysis and historical purposes. Comments may be added to the software overlay prior to video recording by way of the computer's keyboard. In this manner, a patient's databank may be established.

5.2 ANALYSIS SOFTWARE

The 16K memory module allows for limited gait storage which can amount to a few footsteps at most. There are a number of ways in which this information can be meaningfully displayed. One such method is by the generation of eight, time-sequential force curves (Fig 5.2), which describes the plantar pressures as the foot progresses through the gait cycle.

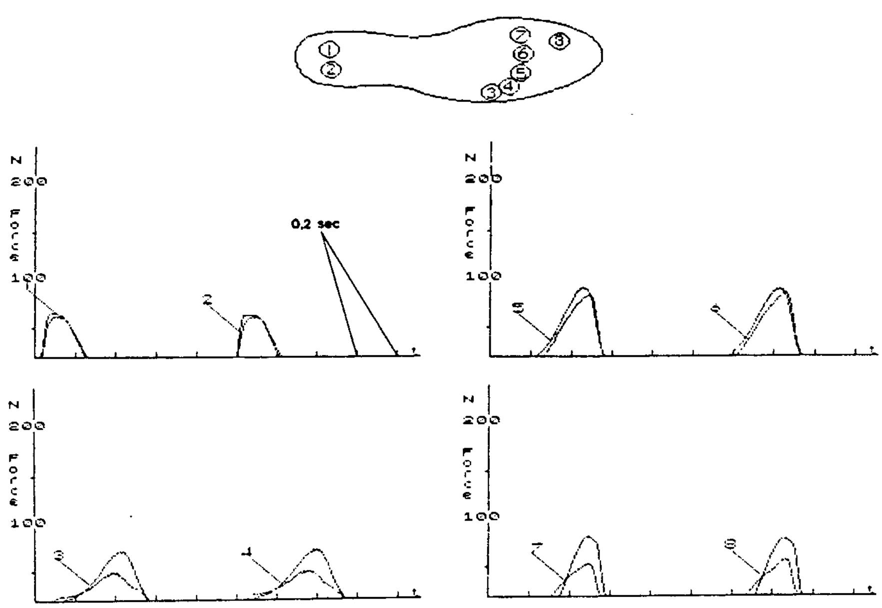


Fig 5.2 : Discrete force-time curves

The above set of curves may not be the most appropriate means of comparison in all cases.

In allowing the acquired gait measures to be compared with those of other researchers, it is necessary for the results to be standardised (Lord et al.[33]). In an effort to optimise interstudy gait comparison, a tabular format containing the most characteristic and representative gait parameters is produced (Fig 5.3). The software listing used in generating this table is shown in Appendix C.

John Rex 10/6/87 12km/h

C	Imp	t/o	dur	t/f	%t	s′o	s′f	Fmx
1	5	ref	124	124	57	1	2	63
2	1	12	24	36	11	2	1	31
3	4	20	132	152	61	3	3	37
4	6	32	144	176	67	7	4	48
5	10	32	180	212	83	6	7	73
6	11	32	184	216	65	5	5	75
7	17	32	180	212	83	4	8	135
8	12	40	176	216	81	8	6	95
T/cycle: 736				I: 66		F: 557		

Venue (wo) 6b 71 kg +00deg

C	t′Fm	Ix	F/t↑	F/t↓	Imp	Fmx	Trg
1	24	1	2625	630	6	94	36
2	12	0	2583	2583			
3	52	2	712	462	10	85	156
4	72	3	667	667			
5	88	5	830	793	21	146	180
6	92	6	815	815			
7	84	8	1607	1406	29	230	184
8	100	7	950	1250			
%N: 169 P: 165 S: 128					I′: 32		t′: 100

Fig 5.3 : Sample hardcopy printout of computed gait parameters

The gait information contained within this printout is described with the aid of the force-time curves shown in Fig 5.4.

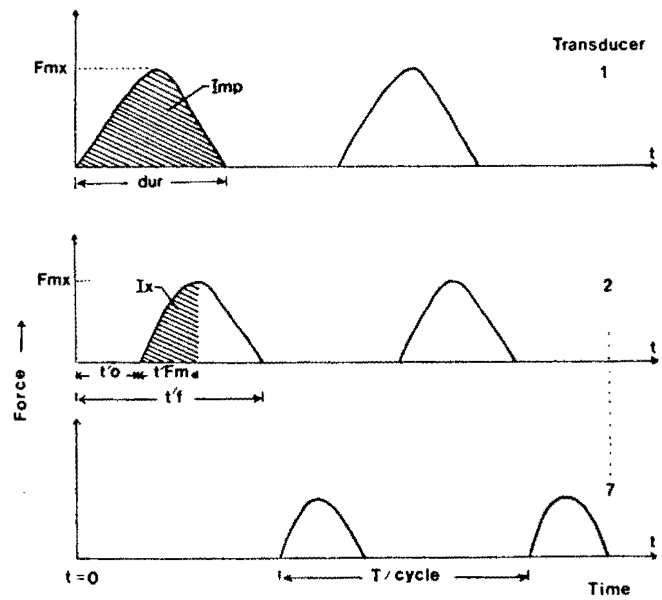


Fig 5.4 : Description of hardcopy printout abbreviations

In the majority of cases, foot contact occurs on the posterior heel and ceases after contact release of the big toe during walking or jogging. Since heel-strike is the first event occurring during a normal gait cycle, this is chosen as the time reference. The two heel transducers situated orthogonally to the foot axis allow for the detection of an inverted heel-strike. Computer analysis then determines which heel transducer was first activated, and assigns its moment of contact as the time reference (t = 0). In the event of exclusive forefoot foot-strike, another time reference would have to be used.

5.2.1 COMPUTED GAIT PARAMETERS

An explanation of the abbreviations used in Fig 5.3 is provided below. All timimg information is given in milliseconds and relates to one complete gait cycle.

C : "Channel". Ranging from 1 to 8, it identifies a particular transducer.

Imp : "Impulse". Localised impulse value, measured in newton-seconds, is evaluated by integrating the areas under the individual force-time curves relating to one gait cycle.

t/o : "Transducer on-time". Time prior to transducer registering a force. Measured relative to the preassigned reference point.

dur : "Duration" of force contact on each transducer.

t/f : "Transducer off-time". Time for a particular transducer to switch-off.

% t : "Percentage time". Ratio of transducer force duration 'dur' to total foot contact time, expressed as a percentage.

s'o : "Switch-on". Transducer switch-on sequence.

s'f : "Switch-off". Transducer switch-off sequence.

Fmx : "Force maximum". The maximum amplitude of force registered by each transducer during the stance phase of the gait cycle. Related to the impulse value.

T/cycle : "Time of gait cycle". The time between successive toe-on to toe-on moments of contact. Measured using the transducer located under the first metatarsal head.

I : The sum total of all localised impulse values.

F : The sum total of all the maximum forces.

t'Fm : "Time to force maximum". Time taken for the transducer to register its maximum force.

Ix : "Impulse to force maximum". Similar computation as that of 'Imp' except terminated on occurrence of 'Fmx'.

F/t↑ : "Force divided by time". Measured in newtons per second, it defines the slope of the ascending curve before peaking.

F/t↓ : Similar to above except defines the slope of the descending curve after peaking.

* Imp : The sum of impulses pertaining to selected pairs of transducers.

* Fmx : The sum of maximum forces pertaining to selected pairs of transducers.

Trg : "Time in region". Denotes the nett time during which selected plantar regions make contact with the force transducers.

%N : "Percentage normal". Percentage of total foot contact time that the foot, when in the normal position (transducers 5 & 6) registers a force.

---

* Found in second table of Fig 5.3.

S : "Percentage supinate". Percentage of total foot contact time that the foot, when in the supinated position (transducers 3 & 4) registers a force.

P : "Percentage pronate". Percentage of total foot contact time that the foot, when in the pronated position (tranducers 7 & 8) registers a force.

I' : Sum of all the Ix values.

t' : Sum of 't'Fm' times pertaining to transducers 2 and 5.

Small markers are included next to the regional time computations. These allow the practitioner to identify quickly which of the transducers within each pair first registered a force.

The above measurements have proved adequate in the diagnosis of many gait-related disorders. For diverse studies, additional information may be included.

Gait studies in the clinical environment often make use of variable speed treadmills. In comparative tests, it is important that speed as well as elevation be the same. The computer control of the more sophisticated treadmills allows for these parameters to be closely monitored. Thus, a facility is provided permitting this information to appear on the gait examination printout. In addition, information such as the patient's name, date of test, patient mass, insole size and shoe type are also given on the printout. This information is inputted to the computer prior to start of gait computation.

5.3 MENU-DRIVEN INSTRUCTION SET

The "auto-run" facility of this computer allóws for programs to run automatically when loaded. This facilitates the configuration of a menu-driven software package for use with the gait analysis device.

The program is started by having the operator insert the program cassette into the microdrive unit after system power-up and pressing two keys, namely, "r" and "ENTER". Thereafter, the computer takes over. Machine code programs are automatically loaded before the operating instructions of Fig 5.5 appear on the monitor display.

```
INSTRUCTIONS FOR GAIT ANALYSIS
USING A VARIABLE SPEED TREADMILL

1.Select insole and calibrate
  system before inserting into
  patient's shoe by pressing the
  switch marked 'CAL'

2.Ensure interface switch is in
  the 'off' position prior to
  gait sampling

3.At the desired treadmill speed
  press the data sample switch
  once on the Gait Analyser Unit

4.Put interface switch in the
  'on' position and press "c"
```

Fig 5.5 : Gait examination instructions

The menu screen of Fig 5.6 appears on pressing the "c" key. It provides the operator with a number of options for the processing or storing of sampled gait information.

```
** MENU - SCREEN **

DO YOU WANT?

1. TO RELOAD GAIT ANALYSER
2. IMPULSE + FORCE CALCULATIONS
3. STORE GAIT INFORMATION
4. PROCESS STORED INFORMATION
5. ERASE FILENAME
6. FILE CATALOG
7. FORMAT CARTRIDGE

Your choice ?
```

Fig 5.6 : Menu screen

Any particular menu option can be selected by pressing the corresponding numeric key. An explanation as to the function of each option follows.

1. TO RELOAD GAIT ANALYSER : Clears screen of any information and displays gait examination instructions of Fig 5.5.

2. IMPULSE + FORCE CALCULATIONS : Provides prompts for inputting patient particulars and treadmill settings before producing the hardcopy printout of Fig 5.3. The menu screen of Fig 5.6 is displayed on completion.

3. STORE GAIT INFORMATION : Provides prompt for patient's filename. Upon entering, 12K of gait information is stored in the corresponding file. On completion, the menu screen is displayed. Blank microdrive cartridges have the ability of storing seven files of gait information.

4. PROCESS STORED INFORMATION : Provides prompt for patient filename. It loads previously stored gait information into the computer for computation. On completion, option 2 is automatically selected.

5. ERASE FILENAME : Provides prompt for the filename to be deleted. On completion, the menu screen is displayed.

6. FILE CATALOG : Provides a listing of the stored files and displays the cartridge status. The pressing of any key causes the menu to reappear on the screen.

7. FORMAT CARTRIDGE : Provides prompt for global cartridge name. Used for formatting the cartridge in event of corruption. All files are deleted during this process. On completion, the menu screen is displayed.

In the event of choosing an incorrect option (3, 4, 5, and 7), return to the menu screen is achieved by pressing the ENTER key on presentation of the filename prompt.

User-friendly comments are displayed for all menu options. Besides simplifying device operation, it allows for its use by operators with no computer or programming knowledge.

5.4 SUMMARY

The software permitting computer gait analysis has been described. Although various methods exist for displaying the computed gait information, a tabular format is found to be advantageous for comparative studies. It provides a rapid quantitive evaluation for the diagnosis and treatment of gait disorders.

The program has been configured in a menu-driven format thus simplifying its operation. User-friendly comments are provided for operators with no previous computer experience. Alterations to the analysis program may be made depending on the practitioner's requirements.

CHAPTER SIX

GAIT ANALYSIS DEVICE EVALUATION

Many clinical examinations have been performed using the new gait analysis device. For these gait analyses to be valid, they should be repeatable and provide the same results on subsequent testing. Research conducted by Soames[34] has shown that the intersubject variability in the pattern and magnitude of the peak pressure distributions highlights the individuality of foot function during gait. This, together with the consistency of these patterns in both the short and long term, is of considerable importance to the clinical situation, as it provides a baseline form which change can be measured.

Three subjects, two males and one female, were chosen to illustrate some of the applications of the system. Since these subjects' data form part of a larger study, it is not possible to draw conclusions from the results obtained from these subjects alone.

All tests were conducted in a clinical environment under controlled laboratory conditions. In addition, measurements were taken from the right foot only, at predetermined treadmill speeds.

6.1 ASSESSMENT OF INTRASUBJECT VARIABILITY AND GAIT MEASUREMENT REPRODUCIBILITY

Research conducted by Miller et al.[15] has shown that no two consecutive gait cycles are the same during walking or running. The variability in measured values must therefore be investigated in determining whether examination of one discrete footstep suffices in characterising the gait cycle.

The day-to-day difference in measured gait parameters must also be determined. Knowledge of this is required when evaluating gait performance over a period of time, for example, of patients undergoing surgical treatment.

6.1.1 INTRASUBJECT GAIT VARIABILITY

In evaluating the variation in consecutive gait cycles, a subject was made to run on a treadmill at constant speed, during which time multiple gait cycles were sampled. The mean variation in computed gait parameters for twelve consecutive tests is shown in Table 6.1.

Table 6.1 : Percentage variation in measured gait parameters

Imp	t/o	dur	t/f	%t
4,5	5,5	4,2	3,6	5,4

Fmx	T/cycle	% N	% P	% S
4,2	0,6	1,4	2	2,9

Since the intercycle gait variations are not significantly different, the information of one cycle can be used for gait characterisation. These variations may be reduced by taking the mean computed gait data of a number of consecutive footsteps.

6.1.2 DAY-TO-DAY GAIT VARIABILITY

With ambulation conditions kept constant (13 km/h), the gait cycle of one of the previous subjects was recorded on five different days to determine the day-to-day variation in gait characteristics. Analysis of results showed a maximum variation of 7.2% in any measured gait parameter (Table 6.2). This probably represents a true day-to-day variation.

Table 6.2 : Percentage day-to-day variations in gait parameters

Imp	t/o	dur	t/f	% t
6,7	7,2	5	4,1	6,1

Fmx	T/cycle	% N	% P	% S
6,4	1,3	1,8	1,9	3,2

6.2 METHOD OF SYSTEM EVALUATION

It is important that the pattern and magnitude of the gait pressures be similar to those reported by other investigators. Experiments pertaining to previous gait studies were repeated, thus allowing system accuracy and versatility to be evaluated.

To allow for a direct comparison of results, required that the display of gait information be similar to that of previous studies. As a result, only a small part of the computed gait information was used. LOTUS 123™, a spreadsheet software package with graphing facilities, was used for producing bar-graphs relating values of impulse, force and percentage plantar contact times to discrete plantar locations.

During the stance phase of a normal subject, Manlay et al.[35] have shown that normal foot-loading is transferred from heel-centre to the metatarsal heads and thence to the big toe. Gait abnormalities are usually identified by the registration of high regional impulse (and associated force) values. In addition, an incorrectly sequenced pressure pattern may also be observed.

For the subsequent results to be meaningful, a clinical assessment of each subject should be available. These are presented in Appendix D.

Subjects A, B and C, as referenced in Appendix D, participated in the following experiments. The treadmill speed, once selected, was kept constant throughout the tests. In addition, the treadmill inclination was kept constant at zero.

6.2.1 GAIT EXPERIMENTS

The following experiments were performed, the results of which are presented in Appendix E.

a) THE EFFECT OF DIFFERENT SHOE APPAREL ON WALKING GAIT

It is well-known that different shoe types have different supportive and cushioning properties. Grundy et al.[36] have observed that the rigidity of the soles of a shoe can influence peak pressures and contact times. For example, increasing the rigidity leads to the metatarsal heads being in contact with the ground for a longer period, and consequently, the centre of foot pressure moves towards the forefoot move rapidly.

Subject A was used in examining the shock-absorbing and supportive (anti-pronation) properties of three different shoe types, namely, a running shoe, a canvas tackie (sneaker) and a leather casual shoe. The gait pattern in each shoe was recorded at a treadmill speed of 6 km/h. Results are presented in Appendix E (i) and E (ii), the latter showing more precise timing information. This display format, used by Miller et al.[15], allows for the plantar load-bearing sequence to be described.

Wearing a pair of good running shoes for walking purposes instead of tackies or leather casuals, decreases the impulses (Appendix E (i) a) and associated forces (Appendix E (i) b) under the plantar surface without changing their relative relationship. This makes the distribution of forces more even.

One of the criticisms levelled at the tackie is the discomfort experienced after long usage. This is not surprising since both the registered impulse and force values are relatively high.

The higher impulses and forces associated with the medial heel edge (transducer 1) are indicitive of an inverted heel-strike, since its magnitude is higher than that of the lateral edge (transducer 2). This is evident in the timing diagram of Appendix E (ii) since transducer 1 is the first to register a force.

b) COMPARISON OF SHOD AND UNSHOD FOOT FORCES

The consensus of opinion is that the effect of wearing shoes spreads the load over a wider area of the foot and for the foot contact times to be increased.[37,38,39]

To acquire the unshod gait pattern, the instrumented insole was attached to the plantar surface of subject B, and covered with a sock. Gait information was recorded at a treadmill speed of 12 km/h. Thereafter, the gait pattern pertaining to a pair of running shoes and a pair of casual shoes was acquired for comparison. The results are shown in Appendix E (iii).

These results are in agreement with Soames[34] and Pollard et al.[8]. The peak vertical forces are decreased when wearing shoes, with the majority of changes being associated with the lateral side of the foot (transducers 2, 3, 4). The contact times generally increase when wearing shoes, particularly under the medial heel (transducer 1) and mid foot (transducers 5, 6).

The inability of walking shoes to reduce the high peak forces associated with running can easily be seen (Appendix E (iii) b). Even though a soft sole material is used, 'bottoming out' is seen to occur (transducer 7) by virtue of the material being compressed too rapidly and thus not being able to absorb the shock.

c) MEASUREMENT OF RUNNING SHOE CUSHIONING PROPERTIES

Biomechanical testing has shown that a running shoe can influence the movement pattern of the lower leg, knee joint angle and rearfoot angle, which can have an influence on impact forces. This prevents the accurate assessment of the cushioning properties of a running shoe by material testing only. As shown in the previous experiment, a shoe with a very soft mid-sole may actually result in a higher impact force than one with a mid-sole made of harder material.

Three different running shoes were evaluated by subject B running on a treadmill at 12 km/h. Gait information pertaining to each shoe was recorded and is shown in Appendix E (iv).

Examination of the force and impulse graphs shows that the shoes tested differ in their shock-absorbing and pronation-control properties. The high impulse and force registered under the first metatarsal head (transducer 7) in the Trilogy results from use of the "Kinetic Wedge" design. This causes the first metatarsal to drop, and in doing so, allows for the arch to remain in a raised position, thus preventing excessive pronation. Observation of the contact times show that subject B still overpronates by virtue of the time spent in the medial forefoot region (transducers 6, 7, 8) and thus requires a correcting insert.

The Pony running shoe, being the cheapest of the three, exhibited the best cushioning property of the shoes tested. By virtue of the material used in the sole, it is envisaged that this trend will reverse after a few kilometres of running as the material becomes compressed. Better-quality cushioning material may not show compaction to the same extent.

d) <u>THE EFFECT OF INSERTS ON THE GAIT PATTERN</u>

Examination of subject B's gait patterns (Appendix E (iv)) shows excessive pronation during load-bearing as shown by the contact times of the medial forefoot region (transducers 6, 7, 8).

To correct this disorder, a half-medial arch support was suggested for use by subject B. To test its effectiveness, the subject's gait pattern was recorded for each running shoe in turn, first with the insert and then without. The results of this test are shown in Appendix E (v) for two pairs of shoes.

The computed percentage variation in pronation and supination resulting from insert use is shown for three different shoes in Table 6.3.

Table 6.3 : Computed pronation and supination variation resulting from insert use

SHOE	%P	%N	%S
Venue w-o	165	169	128
Venue w	152	163	120
Trilogy w-o	158	160	119
Trilogy w	150	143	84
Pony w-o	156	154	102
Pony w	135	147	123

w-o : without insert w : with insert

It is obvserved that the insert performs differently in each shoe with the largest variation taking place with the Pony running shoes. Closer examination of this shoe shows that it does not have a built-in arch-support, a feature that helps prevent over-pronation. Thus, the orthotic will exhibit a more pronounced effect when used in this shoe than in one with a built-in arch-support.

The insert material, being compressible, has the ability to absorb shock. Since this would influence the force and impulse values, these parameters cannot be used in assessing motion control. Plantar contact times are thus more suitable as the main criteria in assessing the effectiveness of an insert. For example, less time spent on the medial edge of the sole indicates less over-pronation.

e) EFFECT OF WORN-OUT RUNNING SHOES

The ideal running shoe material should be able to absorb shock without breaking down. Most mid-sole materials currently employed eventually lose their resilience and cushioning properties.

The average life-span of a good running shoe is approximately 2500 km. After this distance, it should be discarded due to permanent compaction of the mid-sole material. The lack of shock absorption can result in tiredness and/or injury while running.

An experiment to examine the shock-absorbing and support properties of a worn-out running shoe was conducted on subject B. Gait patterns were recorded for two pairs of the same running shoe, one old and one new, under similar conditions. The results are presented in Appendix E (vi).

Most of the impulses and associated forces are higher in the case of the worn-out shoe, with the majority of changes being associated with the medial foot region (transducers 6, 7, 8).

Use of the gait analyser allows for the long-term monitoring of forces associated with a particular shoe. By comparison, the shock absorbing efficiency of the shoe can be evaluated.

f) EFFECT OF HIGH-HEELED SHOES ON WOMEN

Many shoes currently available constrain the foot in what must be regarded as an unphysiological position. Shoes, particularly those that are fashion-oriented, have long been implicated in the development of hallux valgus. As yet, quantitative data which indicates how this condition is brought about is limited.

The distribution of forces resulting from the use of high-heeled shoes was recorded by having subject C walk at constant speed while wearing the relevant shoe apparel. For comparative purposes, the gait patterns of two other shoe types were also recorded, the results of which are shown in Appendix E (vii).

These results are in agreement with Soames et al.[40] who have shown that as heel height increases, pressures under the medial metatarsal heads increase (transducers 6, 7), while those under the lateral metatarsal heads decrease (transducers 2, 3, 4).

Running and flat shoes provide a redistribution of forces and are thus more beneficial than their high-heeled counterparts.

6.3 SUMMARY

The versatility and accuracy of the gait analysis device has been demonstrated by performing a number of experiments pertaining to foot function while walking and running. It was shown that there exists a high degree of correlation between the results obtained experimentally and those obtained by previous researchers.

The large variation in the gait pattern of "normal" individuals makes the formulation of a database difficult. It is suggested that each be used as a reference against which change can be measured.

All computed gait measures are derived from information contained in one gait cycle. Although intercycle variations in gait behaviour exist, it was found to be insignificant for diagnostic purposes.

CHAPTER SEVEN

CONCLUSIONS

There is consensus amongst researchers that the localisation and quantisation of plantar pressures are of greater importance than the measurement of global foot pressures in the study of lower extremity foot function. To date, research in this field is limited by the unavailability of suitable measuring equipment.

A brief review of the pressure measuring systems currently in use shows that restrictions in measurement result from the transducer type used and on-line method of monitoring the transducer outputs. This, and use of expensive computing and display equipment, invariably restricts gait measurements to a clinical environment by a small spectrum of users. To allow for gait measurements by the orthopaedic and sports fraternity alike, requires the availability of a low-cost, portable gait analysis device.

In configuring this new device, use of an instrumented insole was considered the most unobtrusive way of measuring the plantar pressures. The lack of suitable, robust pressure transducers necessitated the design and manufacture of these components.

The features and specifications of the new gait analysis device were devised to allow for full portability, real-time gait monitoring and computer-aided gait analysis. A RF measuring link was considered a prerequisite for measurements in an outdoor environment.

Two methods of analysing a patient's gait pattern were incorporated. One such method makes use of a VCR for recording the real-time pressure display whereas the latter makes use of digital sampling and storage.

A new state of the art video mixing technique was used in producing the visual gait display. This allowed for the easy identification and quantisation of plantar pressures.

The system was evaluated for accuracy and repeatability. A number of previously conducted gait experiments were repeated, enabling the results to be compared with those of other researchers.

Two prototype versions of the gait analysis device have been built and are in extensive use. Research currently in progress includes the biochemical measurement of running-shoe cushioning properties and the indentification of biomechanical factors contributing to common running injuries, namely, runner's knee, shinsplints and archilles tendonitis. The results of these tests are to be published shortly.

The overall conclusion is that the new gait analysis device is far more flexible and easy to operate than those in current use. It is envisaged that the use of this device in gait studies will help towards gaining a better understanding of lower extremity foot function.

Worldwide interest has been shown in this device. It is hoped that licensing agreements with manufacturing companies will be agreed upon shortly. In addition, device patents are pending.

7.1 PROPOSED GAIT STUDIES

In addition to the gait studies referred to in section 6.2, the following will be undertaken :-

a) determination of the force distribution under the hallux valgus foot before and after different surgical procedures;
b) determination of the force distribution under the claw foot before and after different types of surgery;

c) in the diabetic foot, determining the relationship between the force distribution and the site of ulcers, with a view to their prevention;
d) determination of the effect of arthritis, including stills' disease on foot function; and
e) determination of the harmful shoulder forces relating to incorrect crutch use after lower limb surgery. This study requires that transducers be mounted in both crutches and insoles, as shown in Fig 7.1.

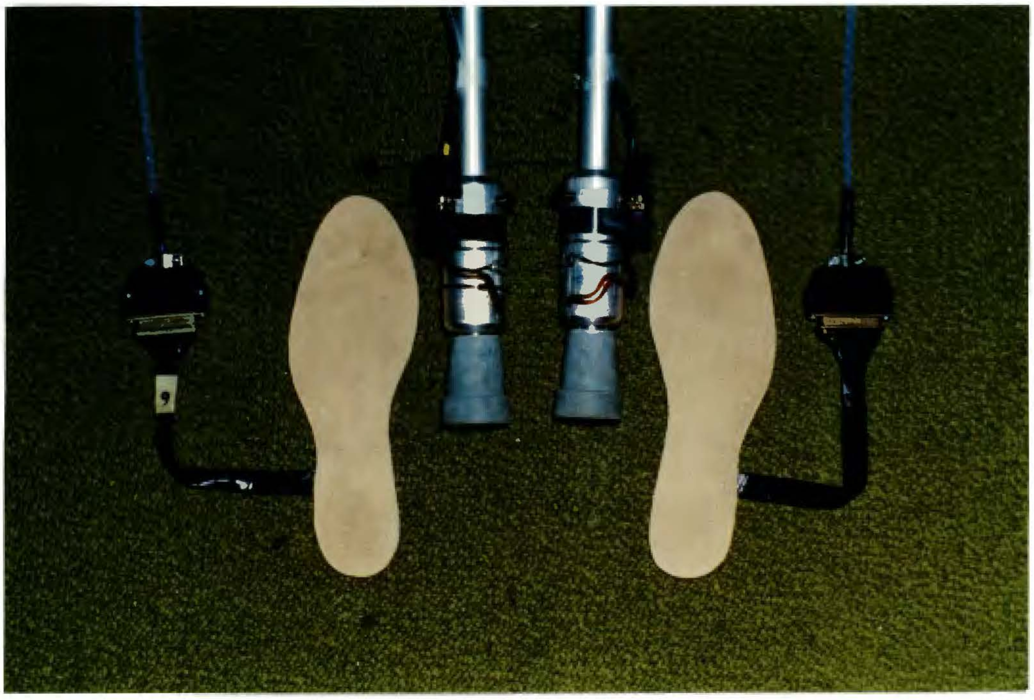


Fig 7.1 : Transducer placement for crutch study

7.2 ADDITIONAL DEVICE APPLICATIONS

Besides gait evaluation, the analysis device can be used for a variety of additional tasks, some of which will be described.

Occupational therapists may use this device in the rehabilitation process of patients with neuromotor disorders or limb deformity/loss. Events, such as the switching-on of a machine, may be controlled by having the patient load a transducer or matrix of transducers with a particular force or pattern of forces.

Many deaf people suffer from imbalance resulting from damaged circular canals in their ears. A method used in assessing the effectiveness of a rehabilitation program makes use of a see-saw arrangement. While in the neutral position, the patient is made to stand in the fulcrum region with their feet slightly parted. Upon release, the degree of patient imbalance is evaluated by monitoring the see-saw movement. Pressure transducers, when placed under their feet, allow for these balancing forces to be quantified in both amplitude and frequency, thus providing an indication of their rehabilitation progress.

Running shoes differ in their shock-absorbing and pronation control properties. The individuality and variation in the "normal" gait pattern results in a variation in shoe behaviour amongst individuals. It thus becomes imperative that a gait analysis device be used in the prescription of running shoes. Excessive pronation and/or supination, if detected, can be corrected by way of inserts.

Many wheelchair-confined para/quadraplegics with no lower limb sensation suffer from pressure sores in their lower back region. To prevent this, pressure transducers can be mounted within the supporting surface to detect areas of high pressure loading. These transducers, when used in a similar manner, can be used in the prevention of bed sores.

The gait analyser could be extensively used in the footwear manufacturing industry. Shoe designs could be evaluated for their fit and support, allowing interaction between the designer and manufacturer. The customisation of running shoes for professionals could be based on their gait patterns. This could prove highly lucrative for their sponsors.

In addition to the above applications, these transducers may be used for measuring pressures in confined areas, for example, under prosthetic devices, under the saddle and hooves of horses, in the boots or skiers, in the boots of army personnel, etc. The applications are virtually endless.

A "rugby scrummaging analyser" is presently under development for the detection of harmful and excessive pressures in the shoulder and neck region of rugby players while scrummaging. Similar transducers as are used with the gait analyser are implemented, a prototype of which is shown in Fig 7.2.

Fig 7.2 : Prototype shoulder pressure transducer

To facilitate the simultaneous display of up to twenty transducers, it is necessary for the video display to be split.

Lastly, the video mixer can be used for the synchronisation and mixing of any two crystal-referenced video signals. With full fading facilities, previously recorded events may be enhanced by the addition of text or graphics from an ordinary home computer.

7.3 PROPOSED SYSTEM IMPROVEMENTS

A modular construction technique was implemented for the gait analysis device. Simplified operation and guaranteed portability require the device to be light and compact. This means that the system should be housed in the confines of an attaché case allowing for its use in a variety of institutions and environments.

To enhance the gait pattern of flat-footed individuals, the transducer configuration within the insole should be changed. This is easily implemented and in time, a variety of insole configurations will be made available.

APPENDIX A

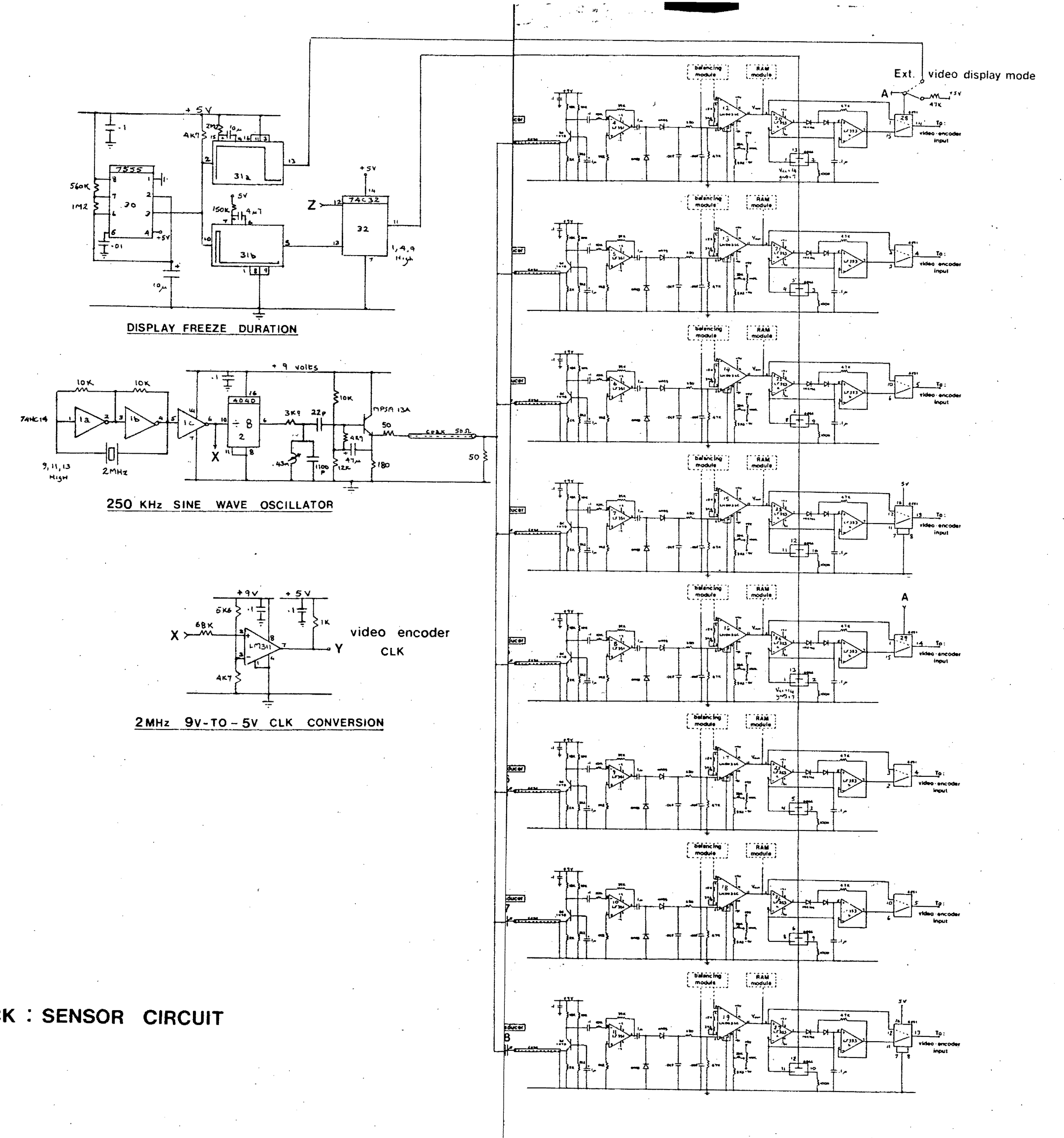
CIRCUIT DIAGRAMS OF NEW GAIT ANALYSIS DEVICE

* Sensor Module
* Channel-Balancing Module
* Video Encoding Module
* 16K RAM Module
* 1 Watt Tx Module
* Regulated Power Supplies

* Video Mixer Unit

* Computer Modifications

* Memory and Printer Interfaces

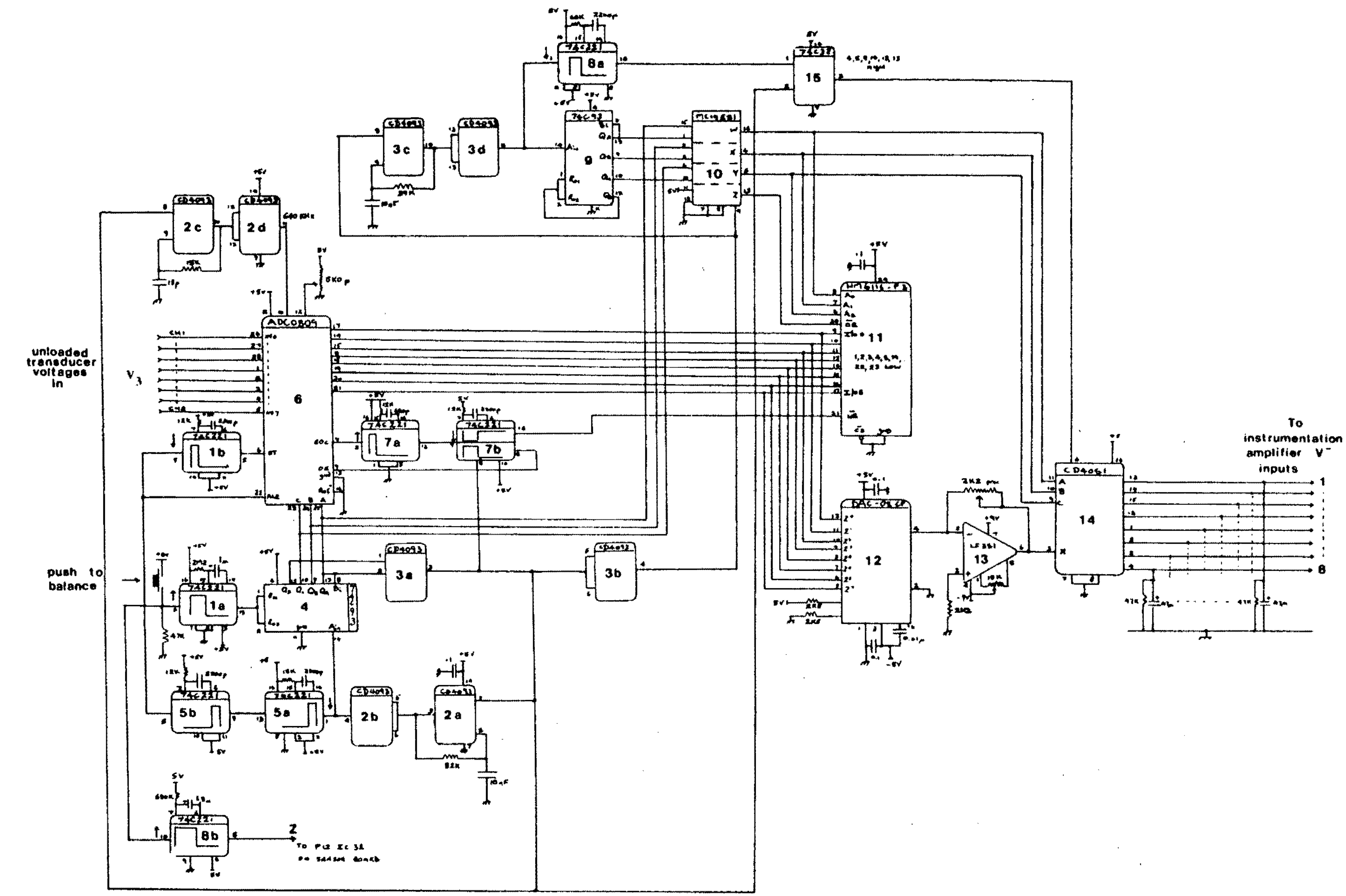


DISPLAY FREEZE DURATION

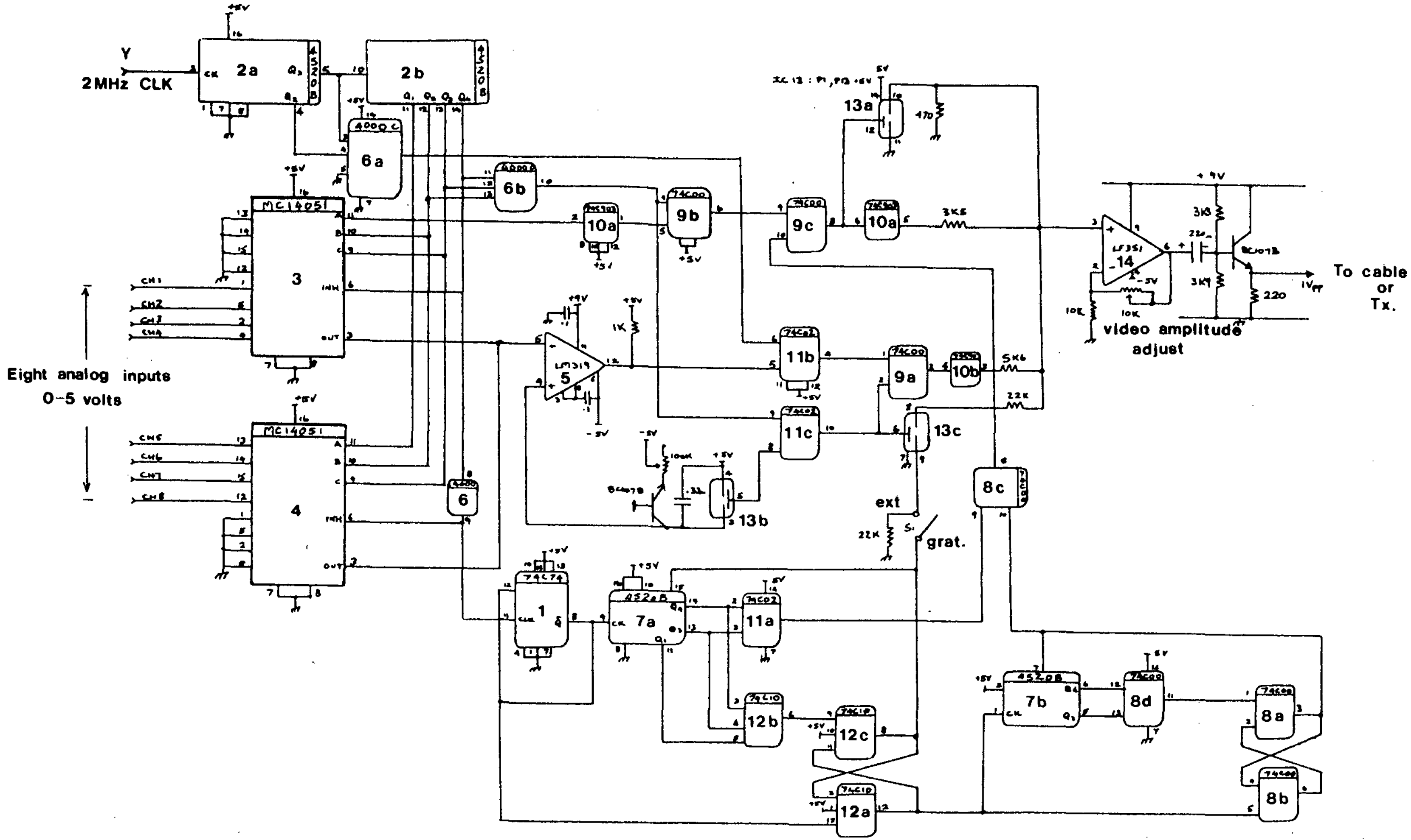
250 KHz SINE WAVE OSCILLATOR

2MHz 9V-TO-5V CLK CONVERSION

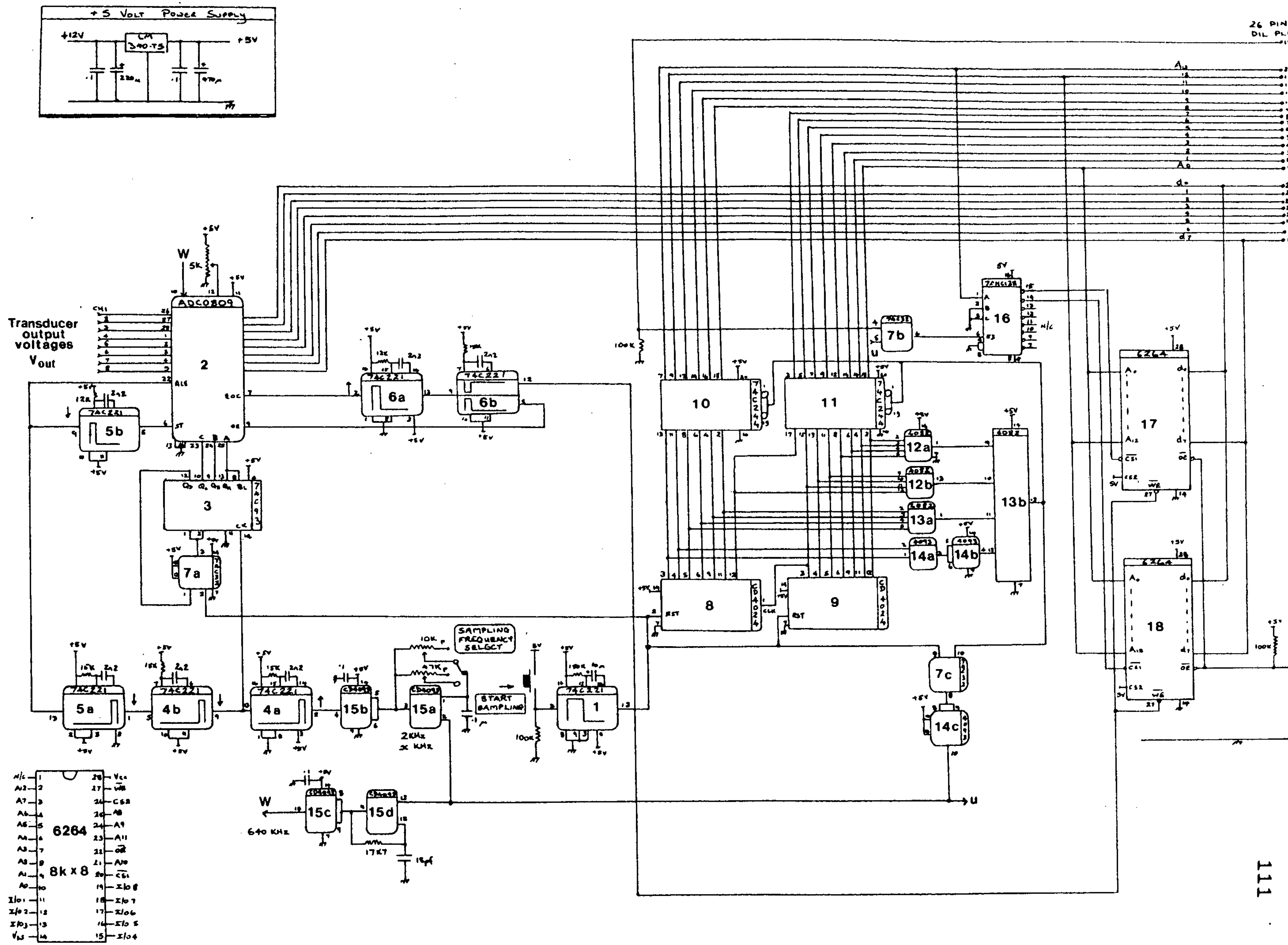
BACKPACK : SENSOR CIRCUIT



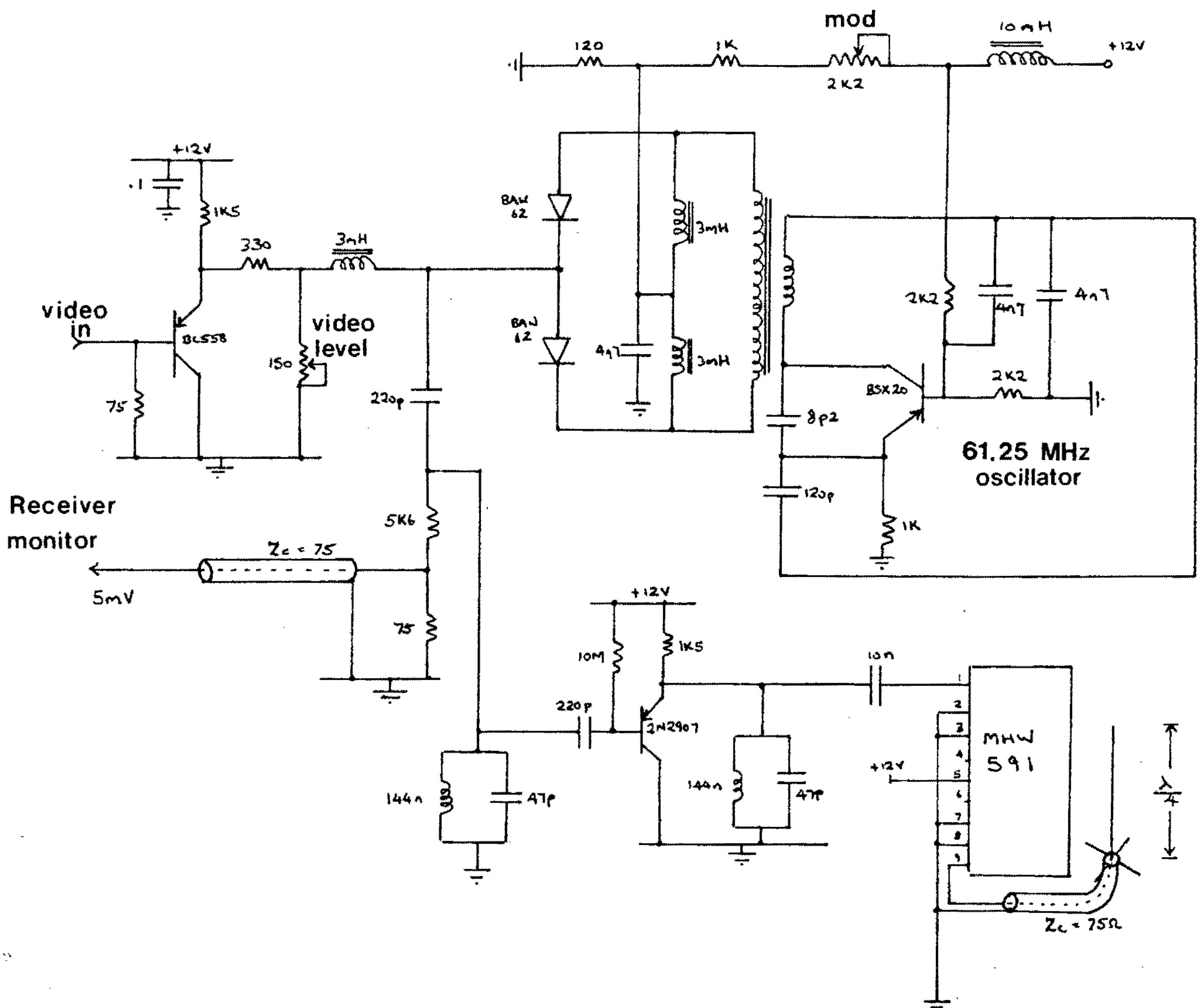
BACKPACK: CHANNEL BALANCING CIRCUIT



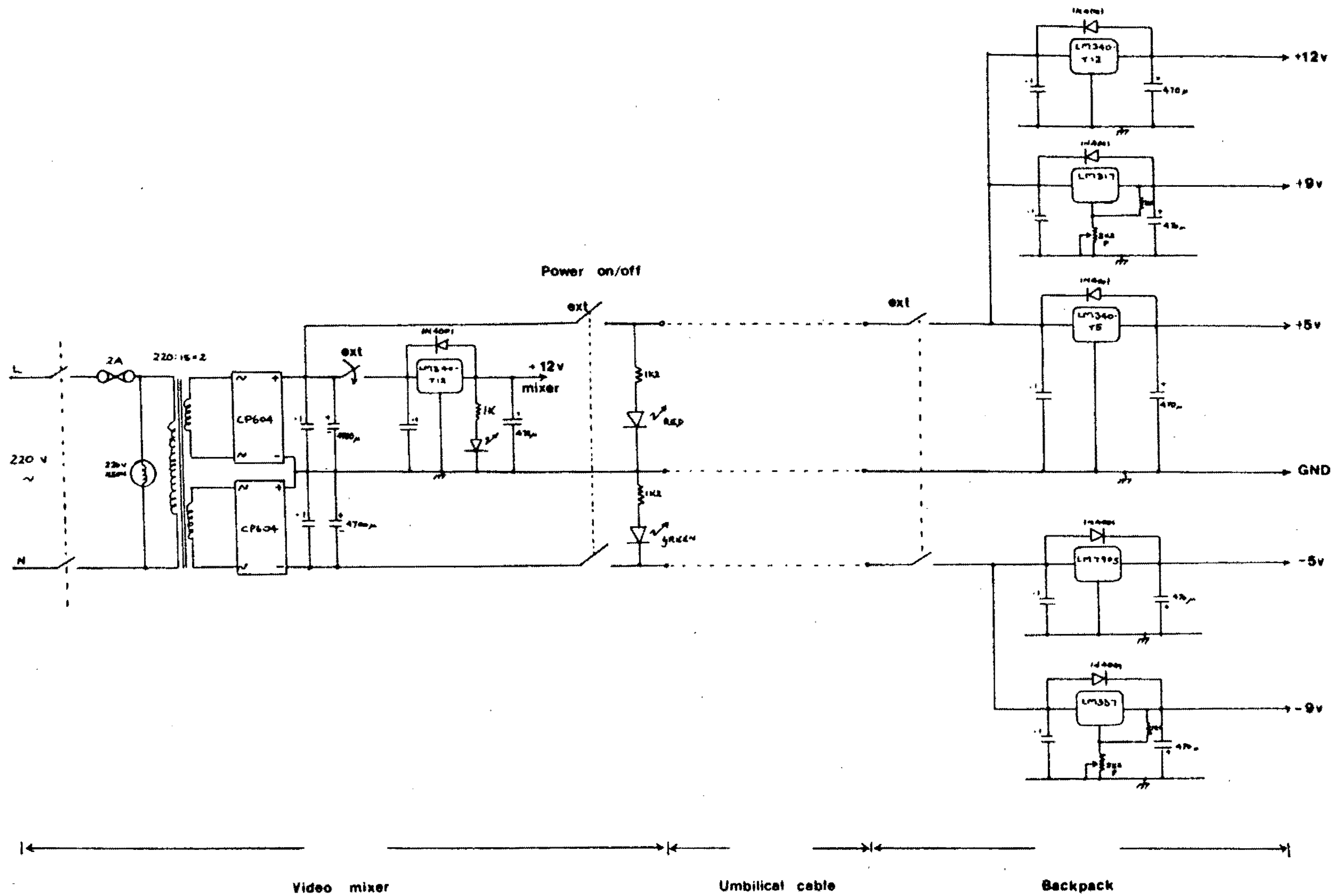
BACKPACK : VIDEO ENCODER CIRCUIT



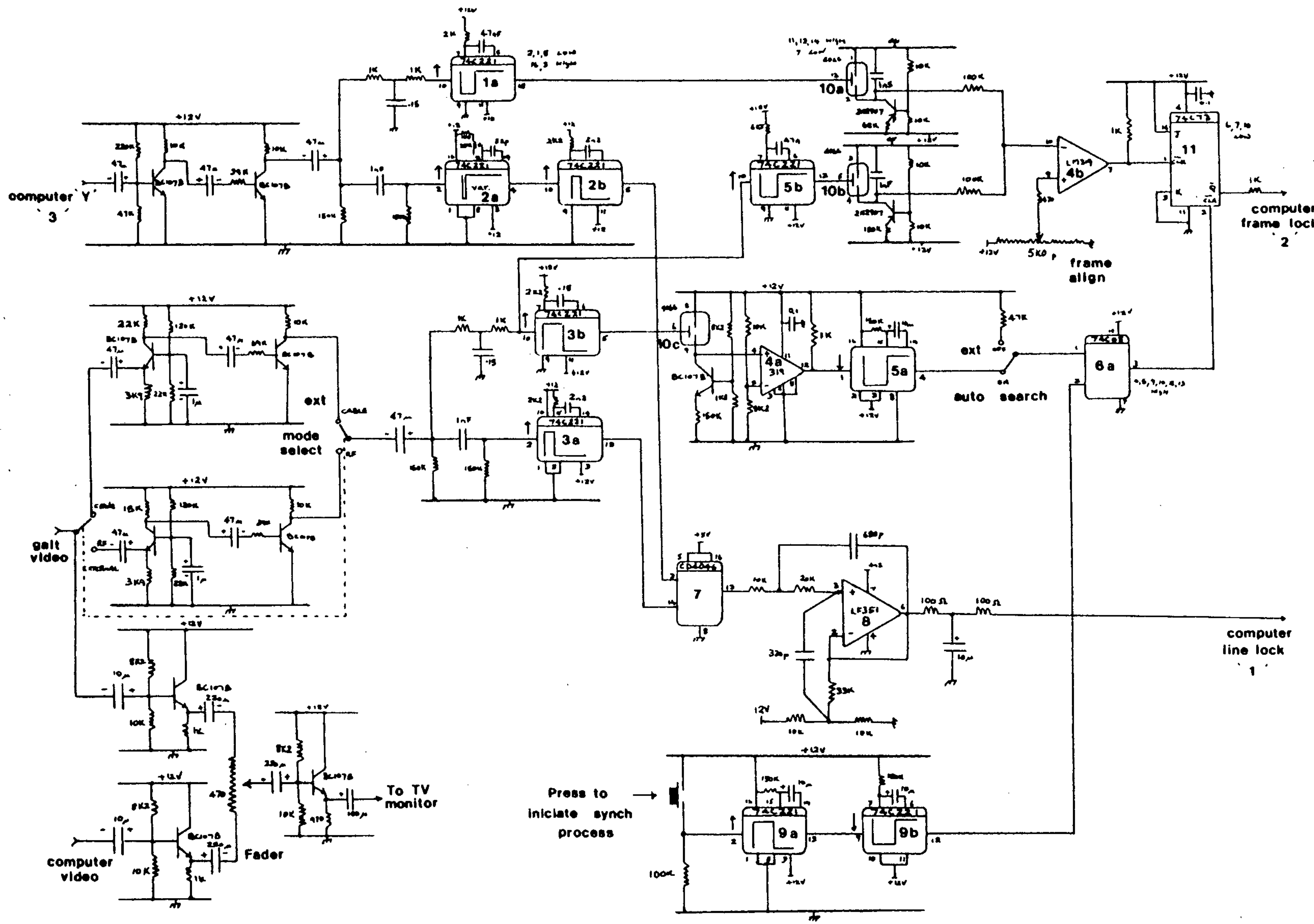
BACKPACK : 16K MEMORY MODULE



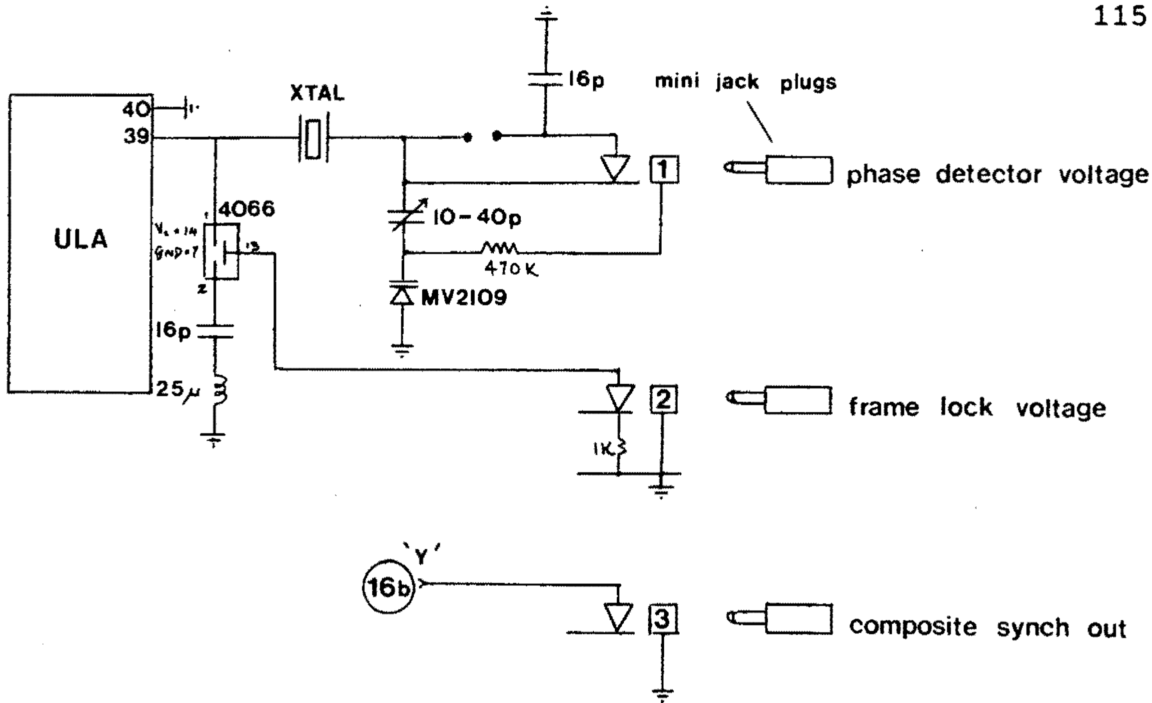
I WATT BAND I TELEVISION TRANSMITTER



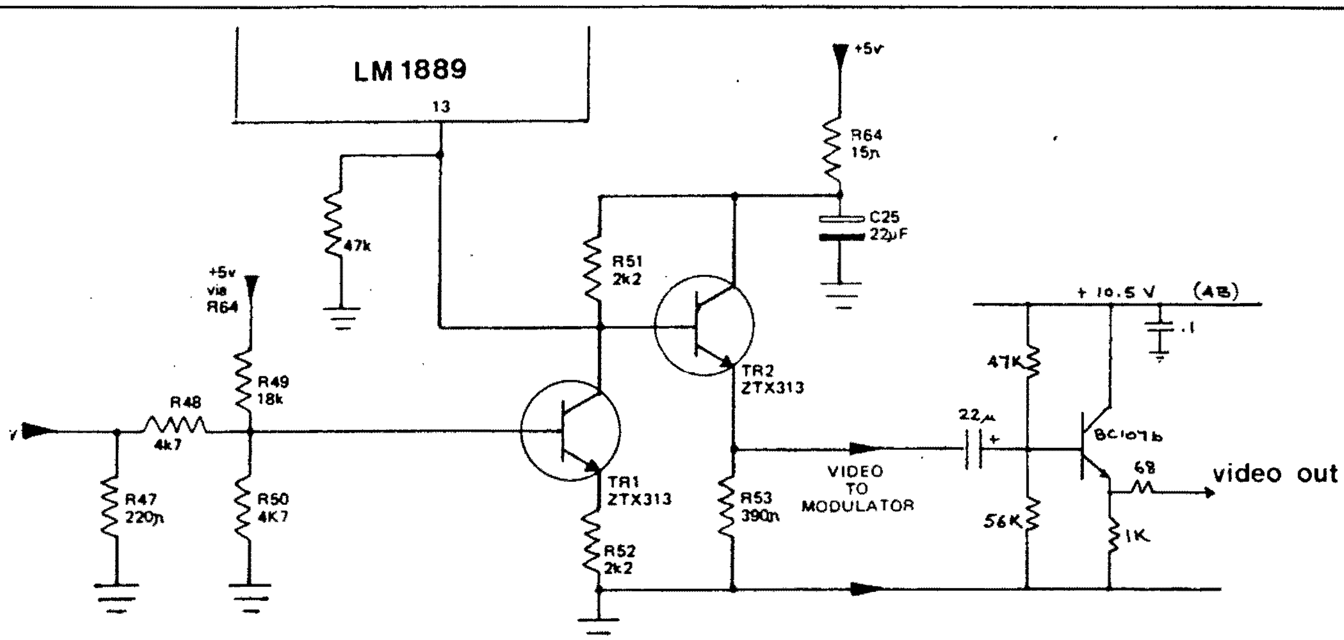
BACKPACK: POWER SUPPLIES



VIDEO MIXER UNIT

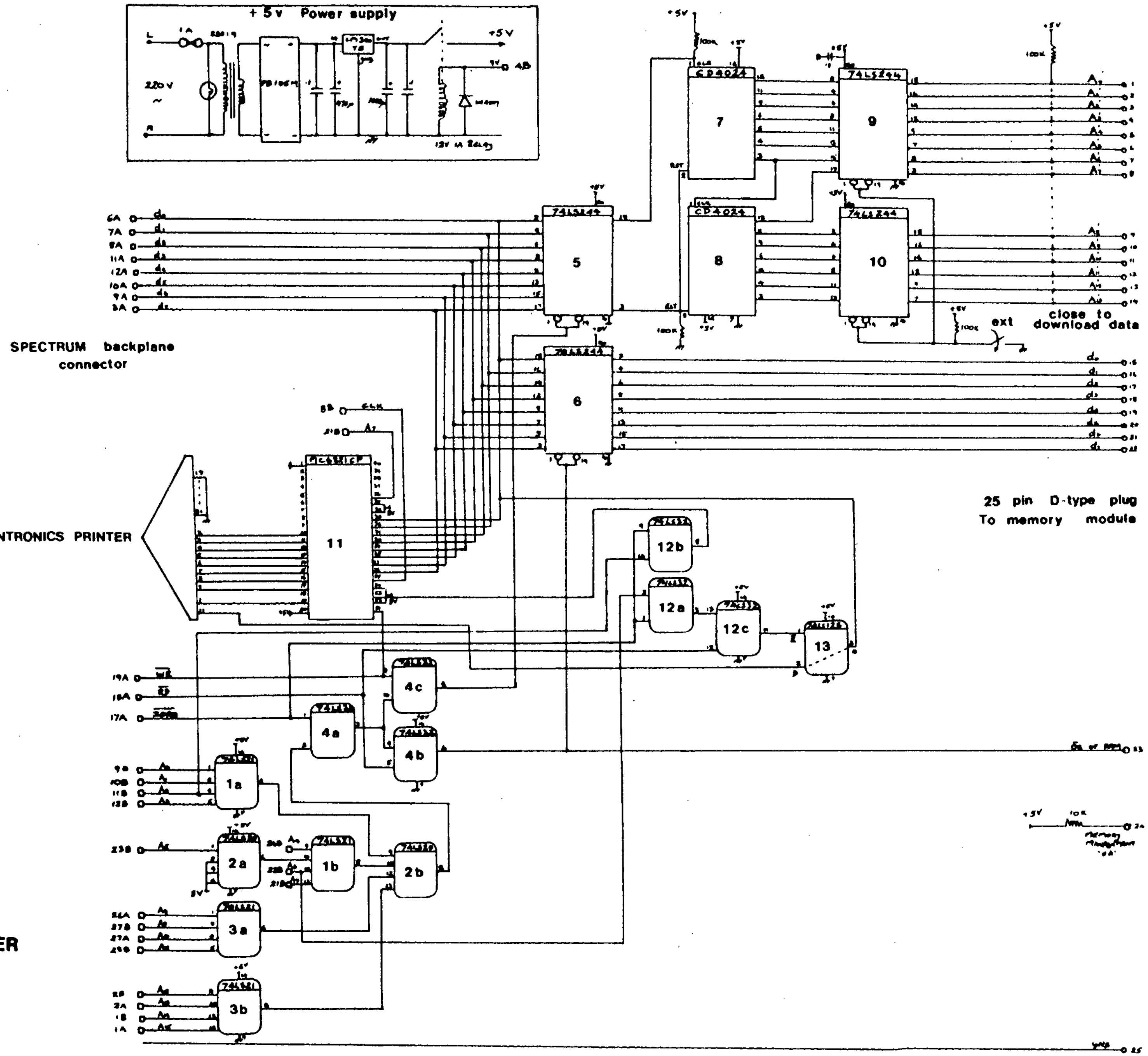


Crystal control circuitry



Computer video output stage

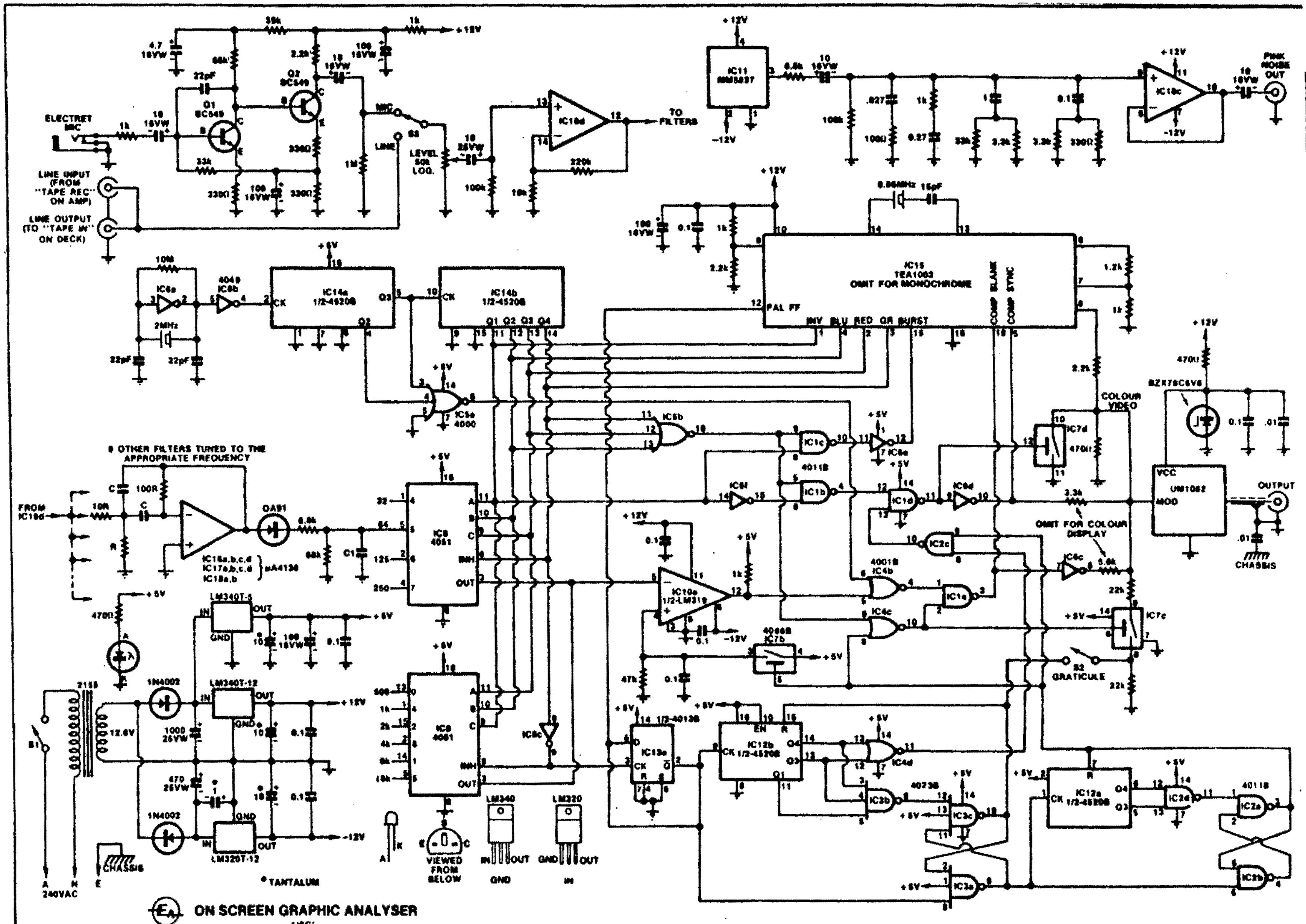
SPECTRUM COMPUTER MODIFICATION



MEMORY/ PRINTER INTERFACE

APPENDIX B

VIDEO ENCODER CIRCUIT OF R. DE JONG[29]


ELECTRET MIC
LINE INPUT (FROM "TAPE REC" ON AMP)
LINE OUTPUT (TO "TAPE IN" ON DECK)
TO FILTERS
PINK NOISE OUT
IC11 MM5837
IC15 TEA1002 OMIT FOR MONOCHROME
COLOUR VIDEO
OMIT FOR COLOUR DISPLAY
OUTPUT
CHASSIS
UM1082
8 OTHER FILTERS TUNED TO THE APPROPRIATE FREQUENCY
FROM IC18d
IC8 4051
IC9 4051
IC10a 1/2-LM319
S2 GRATICULE
LM340T-5
LM340T-12
LM320T-12
240VAC
VIEWED FROM BELOW
LM340
LM320
* TANTALUM
ON SCREEN GRAPHIC ANALYSER
1/SC/-

APPENDIX C

GAIT ANALYSIS SOFTWARE LISTING

```
  5 REM **** 21 January, 1987 ***
 20 REM ******* Program written by MARK LEVIN ... University of
Cape Town ******
 25 REM *** Input by PROFESSOR TIM NOAKES and ANDREW BOSCH ***
 30 REM This program is for use with the "MK1 Insole Gait Analys
er"
 40 REM Backpack to Spectrum memory transfer routine
 50 PRINT AT 0,0;"INSTRUCTIONS FOR GAIT ANALYSIS"''"USING A VARI
ABLE SPEED TREADMILL": PLOT 0,163: DRAW 239,0: PLOT 0,147: DRAW 2
55,0
 60 PRINT AT 4,0;"1.Select insole and calibrate"'"  system befor
e inserting into"'"  patient's shoe by pressing the"'"  switch ma
rked 'CAL'"
 70 PRINT AT 10,0;"2.Ensure interface switch is in"'"  the 'off'
 position prior to"'"  gait sampling"
 80 PRINT AT 15,0;"3.At the desired treadmill speed"'"  press th
e data sample switch"'"  once on the Gait Analyser Unit"
 90 PRINT AT 20,0;"4.Put interface switch in the"'"  'on' positi
on and press ""c"""
190 IF INKEY$="" THEN GO TO 190
200 IF INKEY$<>"c" THEN GO TO 190
210 CLS
220 REM Call up the machine code program which performs the memo
ry transfer
230 RANDOMIZE USR 48512
240 BEEP 1,30
260 PRINT AT 0,6;"** MENU - SCREEN **": PLOT 48,165: DRAW 152,0
270 PRINT AT 3,10;"DO YOU WANT?"
280 PRINT AT 6,0;"1. TO RELOAD GAIT ANALYSER"''"2. IMPULSE + FOR
CE CALCULATIONS"''"3. STORE GAIT INFORMATION"''"4. PROCESS STORED
 INFORMATION"''"5. ERASE FILENAME"''"6. FILE CATALOG"''"7. FORMAT
 CARTRIDGE"
290 PRINT AT 21,0;"Your choice ?"
300 PAUSE 0
310 LET y$=INKEY$
320 IF y$="1" THEN CLS : GO TO 40
330 IF y$="2" THEN CLS : GO TO 900
340 IF y$="3" THEN CLS : GO TO 400
350 IF y$="4" THEN CLS : GO TO 500
360 IF y$="5" THEN CLS : GO TO 580
370 IF y$="6" THEN CLS : GO TO 650
380 IF y$="7" THEN CLS : GO TO 700
390 GO TO 310
400 REM * Store gait information *
410 PRINT AT 0,7;"GAIT DATA STORAGE": PLOT 56,164: DRAW 136,0
```

```
 420 PRINT AT 4,0;"At completion of gait sampling"''"insert patie
nt cartridge into"''"microdrive unit"''''"Enter patient's filenam
e (11 ch) "''"and press 'Enter' key"
 430 INPUT "Filename :";n$
 440 IF LEN n$=0 THEN CLS : GO TO 260
 450 SAVE *"m";1;n$CODE 48630,12000
 460 PAUSE 10: CLS : GO TO 260
 500 REM * Process stored data *
 510 PRINT AT 1,2;"RETRIEVAL OF STORED GAIT DATA": PLOT 16,156: D
RAW 232,0
 520 PRINT AT 8,0;"Insert patient's cartridge into"'''"the microd
rive unit and enter"'''"the relevant filename"
 530 INPUT "Filename :";o$
 540 IF LEN o$=0 THEN CLS : GO TO 260
 550 LOAD *"m";1;o$CODE
 560 PAUSE 10: CLS : GO TO 900
 580 REM * Erase filename *
 590 PRINT AT 1,5;"ERASE PATIENT FILENAME": PLOT 40,156: DRAW 176
,0
 600 PRINT AT 8,0;"With patient cartridge inserted"'''"in the mic
rodrive unit, enter "'''"the filename then press 'Enter'"
 610 INPUT "Filename :";o$
 620 IF LEN o$=0 THEN CLS : GO TO 260
 630 ERASE "m";1;o$
 640 PAUSE 10: CLS : GO TO 260
 650 REM * File catalog *
 660 CAT 1
 670 PRINT AT 21,2; FLASH 1;"PRESS ANY KEY TO CONTINUE"
 680 PAUSE 0
 690 CLS : GO TO 260
 700 REM * Cartridge format *
 710 PRINT AT 1,8;"CARTRIDGE FORMAT": PLOT 64,156: DRAW 128,0
 720 PRINT AT 6,0;"Insert cartridge into microdrive"'''"unit and
enter global filename"''''''"WARNING : All patient files will "'"
         be erased!"
 730 INPUT "Global name: ";p$
 740 IF LEN p$=0 THEN CLS : GO TO 260
 750 FORMAT "m";1;p$
 760 CLS : GO TO 260
 900 REM ** Start of calculations **
 910 REM Input patient's code,umbulation conditions etc.
 920 INPUT "Patient's name:";a$
 930 INPUT "Patient's weight (kg):";b$
 940 INPUT "Date of test:";c$
 950 INPUT "Shoe type:";d$
 960 INPUT "Insole size:";e$
 970 INPUT "Treadmill speed (km/h):";f$
 980 INPUT "Inclination:";g$
 990 CLS
1000 REM Print table outlines with headings
1010 PRINT AT 2,1;"C": PRINT AT 2,3;"Imp": PRINT AT 2,7;"t/o": PR
INT AT 2,11;"dur": PRINT AT 2,15;"t/f": PRINT AT 2,19;"%t": PRINT
 AT 2,22;"so": PLOT 182,158: DRAW 1,1: PRINT AT 2,25;"sf": PLOT 2
06,158: DRAW 1,1: PRINT AT 2,28;"Fmx": PRINT AT 4,1;"1": PRINT AT
 6,1;"2": PRINT AT 8,1;"3": PRINT AT 10,1;"4": PRINT AT 12,1;"5":
PRINT AT 14,1;"6": PRINT AT 16,1;"7": PRINT AT 18,1;"8": PRINT AT
 20,1;"T/cycle:": PRINT AT 20,15;"I:": PRINT AT 20,22;"F:"
```

```
1020 PRINT AT 0,0;a$: PRINT AT 0,16;" █": PRINT AT 0,17;c$: PRINT
AT 0,25;" █": PRINT AT 0,26;f$: PRINT AT 0,28;"km/h"
1030 PLOT 4,3: DRAW 0,159: DRAW 246,0: DRAW 0,-159: DRAW -246,0:
PLOT 4,19: DRAW 246,0: PLOT 4,35: DRAW 246,0: PLOT 4,51: DRAW 246
,0: PLOT 4,67: DRAW 246,0: PLOT 4,83: DRAW 246,0: PLOT 4,99: DRAW
 246,0: PLOT 4,115: DRAW 246,0: PLOT 4,131: DRAW 246,0: PLOT 4,14
7: DRAW 246,0: PLOT 19,19: DRAW 0,143: PLOT 51,19: DRAW 0,143: PL
OT 83,19: DRAW 0,143: PLOT 115,3: DRAW 0,159: PLOT 147,19: DRAW 0
,143: PLOT 173,3: DRAW 0,159: PLOT 195,19: DRAW 0,143: PLOT 219,1
9: DRAW 0,143
1040 LET m1=0: LET m2=0: LET m3=0: LET m4=0: LET m5=0: LET m6=0:
LET m7=0: LET m8=0: LET f1=0: LET f2=0: LET f3=0: LET f4=0: LET f
5=0: LET f6=0: LET f7=0: LET f8=0
1050 LET o1=0: LET o2=0: LET o3=0: LET o4=0: LET o5=0: LET o6=0:
LET o7=0: LET o8=0: LET t1=0: LET t2=0: LET t3=0: LET t4=0: LET t
5=0: LET t6=0: LET t7=0: LET t8=0: LET c1=0: LET c2=0
1060 LET i1=0: LET i2=0: LET i3=0: LET i4=0: LET i5=0: LET i6=0:
LET i7=0: LET i8=0: LET h1=0: LET h2=0: LET r1=0
1070 LET t9=0: LET t10=0: LET t11=0: LET t12=0: LET t13=0: LET t1
4=0: LET t15=0: LET t16=0: LET t17=0: LET w1=0: LET w2=0: LET w3=
0: LET w4=0: LET w5=0: LET w6=0: LET w7=0: LET w8=0: LET x=0.004
1100 REM Locate instant when heel transducers unloaded (a1=a2=0)
1110 LET u=8: LET n1=48632: LET n2=48633
1120 LET a1=PEEK n1: LET a2=PEEK n2
1130 IF a1<3 AND a2<3 THEN LET p1=n1: LET p2=n2: GO TO 1370
1140 LET n1=n1+u: LET n2=n2+u
1150 IF n1>53000 THEN CLS : BEEP 2,20: BEEP 3,10: BEEP 1,3: FOR w
=1 TO 100: PRINT AT 10,0; FLASH 1;"*** TEST ABORT !  TRY AGAIN **
*": NEXT w: CLS : GO TO 260
1160 GO TO 1120
1360 REM Assume heelstrike occurs....find which heel transducer s
trikes first
1370 LET b1=0: LET v1=0: LET b2=0: LET v2=0
1380 LET a1=PEEK p1: LET a2=PEEK p2
1390 IF b1=0 AND v1=0 AND a1<3 THEN LET b1=1
1400 IF b2=0 AND v2=0 AND a2<3 THEN LET b2=1
1410 IF b1=1 AND v1=0 AND a1>2 THEN LET v1=1: LET j=p1: LET h1=1
1420 IF h1=1 THEN LET q=j: PRINT AT 4,7;"ref": GO TO 1500
1430 IF b2=1 AND v2=0 AND a2>2 THEN LET v2=1: LET j=p2: LET h2=1
1440 IF h2=1 THEN LET q=j-1: PRINT AT 6,7;"ref": GO TO 1800
1450 LET p1=p1+u: LET p2=p2+u
1460 IF p1>n1+5000 THEN CLS : BEEP 2,20: BEEP 3,10: BEEP 1,3: FOR
 w=1 TO 100: PRINT AT 9,0; FLASH 1;"**** NO HEELSTRIKE DETECTED *
***"'''"*** TEST ABORT !   TRY AGAIN ***" : NEXT w: CLS : GO TO 2
60
1470 GO TO 1380
1500 REM ** Impact first occurs at No.1 (h1=b1=v1=1) **
1510 REM e= accumulated impulse value; z=time before transducer r
egisters pressure; g= count of impulse on-time; f= instantaneous
impulse value; x= 8*(1/2000) seconds; i1-i8= individual impulse v
alues; t1-t8= individual on-time values; t9= max value of t'f; t1
0 - t17= time to reach max pressures; k= count for t10 - t17; o1-
o8= individual off-times
1520 REM No.1 (h1=1)
1540 LET e=0: LET f=0: LET g=0: LET l=0: LET k=0: LET w=0: LET p=
q
1545 IF INKEY$="q" THEN CLS : GO TO 260
```

```
1550 LET a=PEEK p
1560 IF a>239 THEN LET a=239
1570 IF a<3 THEN LET i1=INT (e+0.5): LET t1=g*4: PRINT AT 4,3;i1:
PRINT AT 4,11;t1: LET f1=o1+t1: PRINT AT 4,15;f1: LET m1=INT ((-9
0*LN (1-1/255))+0.5): PRINT AT 4,28;m1: LET t10=k*4: LET w1=INT (
w+0.5): GO TO 1630
1580 LET f=-90*LN (1-a/255)*x: LET g=g+1
1585 IF a>=1 THEN LET l=a: LET k=k+1: LET w=w+f
1590 LET e=e+f
1600 LET p=p+u
1610 GO TO 1550
1620 IF INKEY$="q" THEN CLS : GO TO 260
1630 REM No.2
1640 REM z= counter for transducer off-time
1650 LET e=0: LET f=0: LET g=0: LET z=0: LET l=0: LET k=0: LET w=
0: LET p=q+1
1655 IF INKEY$="q" THEN CLS : GO TO 260
1660 LET a=PEEK p
1670 IF a>239 THEN LET a=239
1680 IF b2=0 AND v2=0 AND a<3 THEN LET b2=1
1690 IF b2=1 AND v2=0 AND a>2 THEN LET v2=1: LET o2=z*4: PRINT AT
 6,7;o2
1700 IF NOT (b2 AND v2) THEN LET z=z+1
1710 IF b2=1 AND v2=1 AND a<3 THEN LET i2=INT (e+0.5): LET t2=g*4
: PRINT AT 6,3;i2: PRINT AT 6,11;t2: LET f2=o2+t2: PRINT AT 6,15;
f2: LET m2=INT ((-90*LN (1-1/255))+0.5): PRINT AT 6,28;m2: LET t1
1=k*4: LET w2=INT (w+0.5): GO TO 2100
1720 IF b2=1 AND v2=1 THEN LET f=-90*LN (1-a/255)*x: LET g=g+1: I
F a>=1 THEN LET l=a: LET k=k+1: LET w=w+f
1730 LET e=e+f
1740 LET p=p+u
1750 IF p>q+800 THEN PRINT AT 6,3;"Undetected value": GO TO 2100
1760 GO TO 1660
1800 REM ** Impact first occurs at No.2 (h2=b2=v2=1) **
1810 REM No.1
1820 LET e=0: LET f=0: LET g=0: LET z=0: LET l=0: LET k=0: LET p=
q: LET w=0
1825 IF INKEY$="q" THEN CLS : GO TO 260
1830 LET a=PEEK p
1840 IF a>239 THEN LET a=239
1850 IF b1=0 AND v1=0 AND a<3 THEN LET b1=1
1860 IF b1=1 AND v1=0 AND a>2 THEN LET v1=1: LET o1=z*4: PRINT AT
 4,7;o1
1870 IF NOT (b1 AND v1) THEN LET z=z+1
1880 IF b1=1 AND v1=1 AND a<3 THEN LET i1=INT (e+0.5): LET t1=g*4
: PRINT AT 4,3;i1: PRINT AT 4,11;t1: LET f1=o1+t1: PRINT AT 4,15;
f1: LET m1=INT ((-90*LN (1-1/255))+0.5): PRINT AT 4,28;m1: LET t1
0=k*4: LET w1=INT (w+0.5): GO TO 1940
1890 IF b1=1 AND v1=1 THEN LET f=-90*LN (1-a/255)*x: LET g=g+1: I
F a>=1 THEN LET l=a: LET k=k+1: LET w=w+f
1900 LET e=e+f
1910 LET p=p+u
1920 IF p>q+800 THEN PRINT AT 4,3;"Undetected value": GO TO 1940
1930 GO TO 1830
1940 REM No.2
1960 LET e=0: LET f=0: LET g=0: LET l=0: LET k=0: LET w=0: LET p=
q+1
```

```
1965 IF INKEY$="q" THEN CLS : GO TO 260
1970 LET a=PEEK p
1980 IF a>239 THEN LET a=239
1990 IF a<3 THEN LET i2=INT (e+0.5): LET t2=g*4: PRINT AT 6,3;i2:
PRINT AT 6,11;t2: LET f2=o2+t2: PRINT AT 6,15;f2: LET m2=INT ((-9
0*LN (1-1/255))+0.5): PRINT AT 6,28;m2: LET t11=k*4: LET w2=INT (
w+0.5): GO TO 2100
2000 LET f=-90*LN (1-a/255)*x: LET g=g+1: LET w=w+f
2010 IF a>=1 THEN LET l=a: LET k=k+1
2020 LET e=e+f
2030 LET p=p+u
2040 GO TO 1970
2100 REM No.3
2110 IF INKEY$="q" THEN CLS : GO TO 260
2120 GO SUB 4850
2130 LET p=q+2
2140 LET a=PEEK p
2150 IF a>239 THEN LET a=239
2160 IF b=0 AND v=0 AND a<3 THEN LET b=1
2170 IF b=1 AND v=0 AND a>2 THEN LET v=1: LET o3=z*4: PRINT AT 8,
7;o3
2180 IF NOT (b AND v) THEN LET z=z+1
2190 IF v=1 AND b=1 AND a<3 THEN LET i3=INT (e+0.5): LET t3=g*4:
PRINT AT 8,3;i3: PRINT AT 8,11;t3: LET f3=o3+t3: PRINT AT 8,15;f3
: LET m3=INT ((-90*LN (1-1/255))+0.5): PRINT AT 8,28;m3: LET t12=
k*4: LET w3=INT (w+0.5): GO TO 2240
2200 IF v=1 AND b=1 THEN LET f=-90*LN (1-a/255)*x: LET g=g+1: IF
a>=1 THEN LET l=a: LET k=k+1: LET w=w+f
2210 LET e=e+f: LET p=p+u
2220 IF p>q+2400 THEN PRINT AT 8,3;"Undetected value": GO TO 2240
2230 GO TO 2140
2240 REM No.4
2250 IF INKEY$="q" THEN CLS : GO TO 260
2260 GO SUB 4850
2270 LET p=q+3
2280 LET a=PEEK p
2290 IF a>239 THEN LET a=239
2300 IF b=0 AND v=0 AND a<3 THEN LET b=1
2310 IF b=1 AND v=0 AND a>2 THEN LET v=1: LET o4=z*4: PRINT AT 10
,7;o4
2320 IF NOT (b AND v) THEN LET z=z+1
2330 IF v=1 AND b=1 AND a<3 THEN LET i4=INT (e+0.5): LET t4=g*4:
PRINT AT 10,3;i4: PRINT AT 10,11;t4: LET f4=o4+t4: PRINT AT 10,15
;f4: LET m4=INT ((-90*LN (1-1/255))+0.5): PRINT AT 10,28;m4: LET
t13=k*4: LET w4=INT (w+0.5): GO TO 2380
2340 IF v=1 AND b=1 THEN LET f=-90*LN (1-a/255)*x: LET g=g+1: IF
a>=1 THEN LET l=a: LET k=k+1: LET w=w+f
2350 LET e=e+f: LET p=p+u
2360 IF p>q+2400 THEN PRINT AT 10,3;"Undetected value": GO TO 238
0
2370 GO TO 2280
2380 REM No.5
2390 IF INKEY$="q" THEN CLS : GO TO 260
2400 GO SUB 4850
2410 LET p=q+4
2420 LET a=PEEK p
2430 IF a>239 THEN LET a=239
```

```
2440 IF b=0 AND v=0 AND a<3 THEN LET b=1
2450 IF b=1 AND v=0 AND a>2 THEN LET v=1: LET o5=z*4: PRINT AT 12
,7;o5
2460 IF NOT (b AND v) THEN LET z=z+1
2470 IF v=1 AND b=1 AND a<3 THEN LET i5=INT (e+0.5): LET t5=g*4:
PRINT AT 12,3;i5: PRINT AT 12,11;t5: LET f5=o5+t5: PRINT AT 12,15
;f5: LET m5=INT ((-90*LN (1-1/255))+0.5): PRINT AT 12,28;m5: LET
t14=k*4: LET w5=INT (w+0.5): GO TO 2520
2480 IF v=1 AND b=1 THEN LET f=-90*LN (1-a/255)*x: LET g=g+1: IF
a>=1 THEN LET l=a: LET k=k+1: LET w=w+f
2490 LET e=e+f: LET p=p+u
2500 IF p>q+2400 THEN PRINT AT 12,3;"Undetected value": GO TO 252
0
2510 GO TO 2420
2520 REM No.6
2530 IF INKEY$="q" THEN CLS : GO TO 260
2540 GO SUB 4850
2550 LET p=q+5
2560 LET a=PEEK p
2570 IF a>239 THEN LET a=239
2580 IF b=0 AND v=0 AND a<3 THEN LET b=1
2590 IF b=1 AND v=0 AND a>2 THEN LET v=1: LET o6=z*4: PRINT AT 14
,7;o6
2600 IF NOT (b AND v) THEN LET z=z+1
2610 IF v=1 AND b=1 AND a<3 THEN LET i6=INT (e+0.5): LET t6=g*4:
PRINT AT 14,3;i6: PRINT AT 14,11;t6: LET f6=o6+t6: PRINT AT 14,15
;f6: LET m6=INT ((-90*LN (1-1/255))+0.5): PRINT AT 14,28;m6: LET
t15=k*4: LET w6=INT (w+0.5): GO TO 2660
2620 IF v=1 AND b=1 THEN LET f=-90*LN (1-a/255)*x: LET g=g+1: IF
a>=1 THEN LET l=a: LET k=k+1: LET w=w+f
2630 LET e=e+f: LET p=p+u
2640 IF p>q+2400 THEN PRINT AT 14,3;"Undetected value": GO TO 266
0
2650 GO TO 2560
2660 REM No.7 (Cycle time calculated using this transducer)
2670 IF INKEY$="q" THEN CLS : GO TO 260
2675 GO SUB 4850
2680 LET z1=0: LET d=0: LET p=q+6
2690 LET a=PEEK p
2700 IF a>239 THEN LET a=239
2710 IF b=0 AND v=0 AND a<3 THEN LET b=1
2720 IF b=1 AND v=0 AND a>2 THEN LET v=1: LET o7=z*4: PRINT AT 16
,7;o7
2730 IF NOT (b AND v) THEN LET z=z+1
2740 IF b=1 AND v=1 AND d=0 AND a<3 THEN LET d=1: LET i7=INT (e+0
.5): LET t7=g*4: PRINT AT 16,3;i7: PRINT AT 16,11;t7: LET f7=o7+t
7: PRINT AT 16,15;f7: LET m7=INT ((-90*LN (1-1/255))+0.5): PRINT
AT 16,28;m7: LET t16=k*4: LET w7=INT (w+0.5)
2750 IF b=1 AND v=1 AND d=0 THEN LET f=-90*LN (1-a/255)*x: LET g=
g+1: IF a>=1 THEN LET l=a: LET k=k+1: LET w=w+f
2760 IF d=1 AND a<3 THEN LET g=g+1
2770 IF d=1 AND a>2 THEN LET z1=g*4: PRINT AT 20,10;z1: GO TO 281
0
2780 LET e=e+f: LET p=p+u
2790 IF p>q+2400 THEN PRINT AT 16,3;"Undetected value": GO TO 281
0
2800 GO TO 2690
```

```
2810 REM No.8
2820 IF INKEY$="q" THEN CLS : GO TO 260
2830 GO SUB 4850
2840 LET p=q+7
2850 LET a=PEEK p
2860 IF a>239 THEN LET a=239
2870 IF b=0 AND v=0 AND a<3 THEN LET b=1
2880 IF b=1 AND v=0 AND a>2 THEN LET v=1: LET o8=z*4: PRINT AT 18
,7;o8
2890 IF NOT (b AND v) THEN LET z=z+1
2900 IF v=1 AND b=1 AND a<3 THEN LET i8=INT (e+0.5): LET t8=g*4:
PRINT AT 18,3;i8: PRINT AT 18,11;t8: LET f8=o8+t8: PRINT AT 18,15
;f8: LET m8=INT ((-90*LN (1-1/255))+0.5): PRINT AT 18,28;m8: LET
t17=k*4: LET w8=INT (w+0.5): GO TO 3100
2910 IF v=1 AND b=1 THEN LET f=-90*LN (1-a/255)*x: LET g=g+1: IF
a>=1 THEN LET l=a: LET k=k+1: LET w=w+f
2920 LET e=e+f: LET p=p+u
2930 IF p>q+2400 THEN PRINT AT 18,3;"Undetected value": GO TO 310
0
2940 GO TO 2850
3100 REM ** Calculation of % times **
3110 REM Find max value of "t/f"
3120 LET n=0
3130 IF f1>=n THEN LET n=f1
3140 IF f2>=n THEN LET n=f2
3150 IF f3>=n THEN LET n=f3
3160 IF f4>=n THEN LET n=f4
3170 IF f5>=n THEN LET n=f5
3180 IF f6>=n THEN LET n=f6
3190 IF f7>=n THEN LET n=f7
3200 IF f8>=n THEN LET n=f8
3210 LET t9=n
3220 PRINT AT 4,19;INT (((t1/t9)*100)+0.5)
3230 PRINT AT 6,19;INT (((t2/t9)*100)+0.5)
3240 PRINT AT 8,19;INT (((t3/t9)*100)+0.5)
3250 PRINT AT 10,19;INT (((t4/t9)*100)+0.5)
3260 PRINT AT 12,19;INT (((t5/t9)*100)+0.5)
3270 PRINT AT 14,19;INT (((t6/t9)*100)+0.5)
3280 PRINT AT 16,19;INT (((t7/t9)*100)+0.5)
3290 PRINT AT 18,19;INT (((t8/t9)*100)+0.5)
3300 REM ** Accumulation of impulses **
3310 PRINT AT 20,18;i1+i2+i3+i4+i5+i6+i7+i8
3320 REM  Accumulation of max forces
3330 PRINT AT 20,25;m1+m2+m3+m4+m5+m6+m7+m8
4470 REM * Pressure sequence of transducers *
4480 LET c=4
4490 DIM x(8,2): DIM s(2)
4500 FOR i=1 TO 8
4510 LET x(i,1)=i
4520 NEXT i
4530 LET x(1,2)=o1: LET x(2,2)=o2: LET x(3,2)=o3: LET x(4,2)=o4:
LET x(5,2)=o5: LET x(6,2)=o6: LET x(7,2)=o7: LET x(8,2)=o8
4540 FOR i=7 TO 1 STEP -1
4550 FOR j=1 TO i
4560 IF x(j,2)<x(j+1,2) THEN LET s(1)=x(j,1): LET s(2)=x(j,2): LE
T x(j,1)=x(j+1,1): LET x(j,2)=x(j+1,2): LET x(j+1,1)=s(1): LET x(
j+1,2)=s(2)
```

```
4570 NEXT j
4580 PRINT AT c,23;x(i+1,1): LET c=c+2
4590 NEXT i
4600 PRINT AT 18,23;x(1,1)
4610 REM *** Bubble-sort of "t'f";from minimum to maximum ***
4620 LET c=4
4630 DIM y(8,2): DIM r(2)
4640 FOR i=1 TO 8
4650 LET y(i,1)=i
4660 NEXT i
4670 LET y(1,2)=f1: LET y(2,2)=f2: LET y(3,2)=f3: LET y(4,2)=f4:
LET y(5,2)=f5: LET y(6,2)=f6: LET y(7,2)=f7: LET y(8,2)=f8
4680 FOR i=7 TO 1 STEP -1
4690 FOR j=1 TO i
4700 IF y(j,2)<y(j+1,2) THEN LET r(1)=y(j,1): LET r(2)=y(j,2): LE
T y(j,1)=y(j+1,1): LET y(j,2)=y(j+1,2): LET y(j+1,1)=r(1): LET y(
j+1,2)=r(2)
4710 NEXT j
4720 PRINT AT c,26;y(i+1,1): LET c=c+2
4730 NEXT i
4740 PRINT AT 18,26;y(1,1)
4750 BEEP 1,40
4760 RANDOMIZE USR 23296
4770 CLS
4780 REM Advance paper between screens
4790 LPRINT : LPRINT : LPRINT
4800 GO TO 5550
4850 LET e=0: LET f=0: LET g=0: LET z=0: LET b=0: LET l=0: LET v=
0: LET k=0: LET w=0
4860 RETURN
5550 REM Generation of new screen yielding manual calculations
5555 PRINT AT 2,1;"C tFm Ix F/t  F/t  Imp Fmx Trg": PRINT AT 20,1
;"%N:    F:    S:    I:     t:": PRINT AT 4,1;"1"""" 2"""" 3"""" 4
"""" 5"""" 6"""" 7"""" 8"
5560 PLOT 4,3: DRAW 0,159: DRAW 247,0: DRAW 0,-159: DRAW -247,0:
PLOT 20,19: DRAW 0,143: PLOT 48,19: DRAW 0,143: PLOT 76,19: DRAW
0,143: PLOT 116,19: DRAW 0,143: PLOT 156,3: DRAW 0,159: PLOT 188,
19: DRAW 0,143: PLOT 204,3: DRAW 0,16: PLOT 220,19: DRAW 0,143: P
LOT 4,19: DRAW 247,0: PLOT 4,35: DRAW 151,0: PLOT 4,51: DRAW 247,
0: PLOT 4,67: DRAW 151,0: PLOT 4,83: DRAW 247,0: PLOT 4,99: DRAW
151,0: PLOT 4,115: DRAW 247,0: PLOT 4,131: DRAW 151,0: PLOT 4,147
: DRAW 247,0
5565 PRINT AT 0,15;" █    █   kg █   deg": PLOT 31,160: DRAW -1,-1:
PLOT 169,16: DRAW -1,-1: PLOT 215,16: DRAW -1,-1: PLOT 108,153: D
RAW 0,6: DRAW -2,-2: DRAW 2,2: DRAW 2,-2: PLOT 148,158: DRAW 0,-6
: DRAW -2,2: DRAW 2,-2: DRAW 2,2
5590 PRINT AT 0,0;d$: PRINT AT 0,16;e$: PRINT AT 0,20;b$: PRINT A
T 0,26;g$
5600 PRINT AT 20,28;t11+t14: PRINT AT 4,3;t10: PRINT AT 6,3;t11:
PRINT AT 8,3;t12: PRINT AT 10,3;t13: PRINT AT 12,3;t14: PRINT AT
14,3;t15: PRINT AT 16,3;t16: PRINT AT 18,3;t17
5610 REM * Impulse till max force *
5620 PRINT AT 4,7;w1: PRINT AT 6,7;w2
5630 PRINT AT 8,7;w3: PRINT AT 10,7;w4
5640 PRINT AT 12,7;w5: PRINT AT 14,7;w6
5650 PRINT AT 16,7;w7: PRINT AT 18,7;w8
5660 PRINT AT 20,22;w1+w2+w3+w4+w5+w6+w7+w8
```

```
5680>REM * Up and down slope calculation *
5690 IF t1>1 AND t10>1 THEN LET c1=INT ((m1/t10)*1000+0.5): LET c
2=INT ((m1/(t1-t10))*1000+0.5): IF LEN STR$ c1<5 AND LEN STR$ c2<
5 THEN PRINT AT 4,10;c1: PRINT AT 4,15;c2
5700 IF t2>1 AND t11>1 THEN LET c1=INT ((m2/t11)*1000+0.5): LET c
2=INT ((m2/(t2-t11))*1000+0.5): IF LEN STR$ c1<5 AND LEN STR$ c2<
5 THEN PRINT AT 6,10;c1: PRINT AT 6,15;c2
5710 IF t3>1 AND t12>1 THEN LET c1=INT ((m3/t12)*1000+0.5): LET c
2=INT ((m3/(t3-t12))*1000+0.5): IF LEN STR$ c1<5 AND LEN STR$ c2<
5 THEN PRINT AT 8,10;c1: PRINT AT 8,15;c2
5720 IF t4>1 AND t13>1 THEN LET c1=INT ((m4/t13)*1000+0.5): LET c
2=INT ((m4/(t4-t13))*1000+0.5): IF LEN STR$ c1<5 AND LEN STR$ c2<
5 THEN PRINT AT 10,10;c1: PRINT AT 10,15;c2
5730 IF t5>1 AND t14>1 THEN LET c1=INT ((m5/t14)*1000+0.5): LET c
2=INT ((m5/(t5-t14))*1000+0.5): IF LEN STR$ c1<5 AND LEN STR$ c2<
5 THEN PRINT AT 12,10;c1: PRINT AT 12,15;c2
5740 IF t6>1 AND t15>1 THEN LET c1=INT ((m6/t15)*1000+0.5): LET c
2=INT ((m6/(t6-t15))*1000+0.5): IF LEN STR$ c1<5 AND LEN STR$ c2<
5 THEN PRINT AT 14,10;c1: PRINT AT 14,15;c2
5750 IF t7>1 AND t16>1 THEN LET c1=INT ((m7/t16)*1000+0.5): LET c
2=INT ((m7/(t7-t16))*1000+0.5): IF LEN STR$ c1<5 AND LEN STR$ c2<
5 THEN PRINT AT 16,10;c1: PRINT AT 16,15;c2
5760 IF t8>1 AND t17>1 THEN LET c1=INT ((m8/t17)*1000+0.5): LET c
2=INT ((m8/(t8-t17))*1000+0.5): IF LEN STR$ c1<5 AND LEN STR$ c2<
5 THEN PRINT AT 18,10;c1: PRINT AT 18,15;c2
5950 PRINT AT 5,20;i1+i2: PRINT AT 9,20;i3+i4: PRINT AT 13,20;i5+
i6: PRINT AT 17,20;i7+i8
5960 PRINT AT 5,24;m1+m2: PRINT AT 9,24;m3+m4: PRINT AT 13,24;m5+
m6: PRINT AT 17,24;m7+m8
5965 REM ** Specific regional contact times **
5970 IF o2>=o1 THEN CIRCLE 246,141,1: LET r1=o2-o1+t2: IF LEN STR
$ r1<4 THEN PRINT AT 5,28;r1
5980 IF o1>=o2 THEN CIRCLE 246,122,1: LET r1=o1-o2+t1: IF LEN STR
$ r1<4 THEN PRINT AT 5,28;r1
5990 IF o4>=o3 THEN CIRCLE 246,109,1: LET r1=o4-o3+t4: IF LEN STR
$ r1<4 THEN PRINT AT 9,28;r1
6000 IF o3>=o4 THEN CIRCLE 246,90,1: LET r1=o3-o4+t3: IF LEN STR$
 r1<4 THEN PRINT AT 9,28;r1
6010 IF o6>=o5 THEN CIRCLE 246,77,1: LET r1=o6-o5+t6: IF LEN STR$
 r1<4 THEN PRINT AT 13,28;r1
6020 IF o5>=o6 THEN CIRCLE 246,58,1: LET r1=o5-o6+t5: IF LEN STR$
 r1<4 THEN PRINT AT 13,28;r1
6030 IF o8>=o7 THEN CIRCLE 246,45,1: LET r1=o8-o7+t8: IF LEN STR$
 r1<4 THEN PRINT AT 17,28;r1
6040 IF o7>=o8 THEN CIRCLE 246,26,1: LET r1=o7-o8+t7: IF LEN STR$
 r1<4 THEN PRINT AT 17,28;r1
6070 PRINT AT 20,4;INT ((((t5+t6)/t9)*100)+0.5)
6080 PRINT AT 20,10;INT ((((t7+t8)/t9)*100)+0.5)
6090 PRINT AT 20,16;INT ((((t3+t4)/t9)*100)+0.5)
6100 BEEP 1,40
6110 RANDOMIZE USR 23296
6120 CLS
6130 FOR w=1 TO 56
6140 LPRINT
6150 NEXT w
6160 GO TO 260
```

APPENDIX D

CLINICAL ASSESSMENT OF SUBJECTS

Subject A

Subject B

Subject C

CLINICAL INFORMATION FOR USE WITH THE GAIT ANALYSER

Patient's name : Subject A

Age : 29 Weight : 73 kg Height : 1.76 m

Examination Date : 14 May 1987 Foot : Right

Foot type ;
Off-bearing weight.
Right foot :

_ Low arch
x Medium arch
_ High arch

Left foot :

_ Low arch
x Medium arch
_ High arch

Foot type ;
On-bearing weight.
Right foot :

_ Low arch
x Medium arch
_ High arch

Left foot :

_ Low arch
x Medium arch
_ High arch

Rearfoot position.
Right foot :

_ Pronation
x Neutral
_ Supination

Left foot :

_ Pronation
x Neutral
_ Supination

Frontal plane knee infleunces.
Right leg :

_ Genu varum
x Straight
_ Genu valgum

Left leg :

_ Genu varum
x Straight
_ Genu valgum

Comments :

Good biomecanical alignment

CLINICAL INFORMATION FOR USE WITH THE GAIT ANALYSER

Patient's name : Subject B

Age : 40 Weight : 71 kg Height : 1.75 m

Examination Date : 10 June 1987 Foot : Right

Foot type ;
Off-bearing weight.
Right foot :

_ Low arch
_ Medium arch
x High arch

Left foot :

_ Low arch
_ Medium arch
x High arch

Foot type ;
On-bearing weight.
Right foot :

_ Low arch
x Medium arch
_ High arch

Left foot :

_ Low arch
x Medium arch
_ High arch

Rearfoot position.
Right foot :

_ Pronation
x Neutral
_ Supination

Left foot :

_ Pronation
x Neutral
_ Supination

Frontal plane knee infleunces.
Right leg :

_ Genu varum
x Straight
_ Genu valgum

Left leg :

_ Genu varum
x Straight
_ Genu valgum

Comments :

Good biomechanical alignment.
Callouse under 1st metatarsal head.

CLINICAL INFORMATION FOR USE WITH THE GAIT ANALYSER

Patient's name : Subject C

Age : 24 Weight : 61 kg Height : 1.65 m

Examination Date : 30 May 1987 Foot : Right

Foot type ;
Off-bearing weight.
Right foot :

x Low arch
_ Medium arch
_ High arch

Left foot :

x Low arch
_ Medium arch
_ High arch

Foot type ;
On-bearing weight.
Right foot :

x Low arch
_ Medium arch
_ High arch

Left foot :

x Low arch
_ Medium arch
_ High arch

Rearfoot position.
Right foot :

x Pronation
_ Neutral
_ Supination

Left foot :

x Pronation
_ Neutral
_ Supination

Frontal plane knee infleunces.
Right leg :

_ Genu varum
x Straight
_ Genu valgum

Left leg :

_ Genu varum
x Straight
_ Genu valgum

Comments :

Normal standing position is pronated. Might be prone to injuries caused by over-pronation.
Callouse under 2nd metatarsal head.

APPENDIX E

EXPERIMENTAL GAIT ANALYSIS RESULTS

i) The Effect of Different Shoe Apparel on Walking Gait.

ii) Plantar Pressure Timing Information for (i) above.

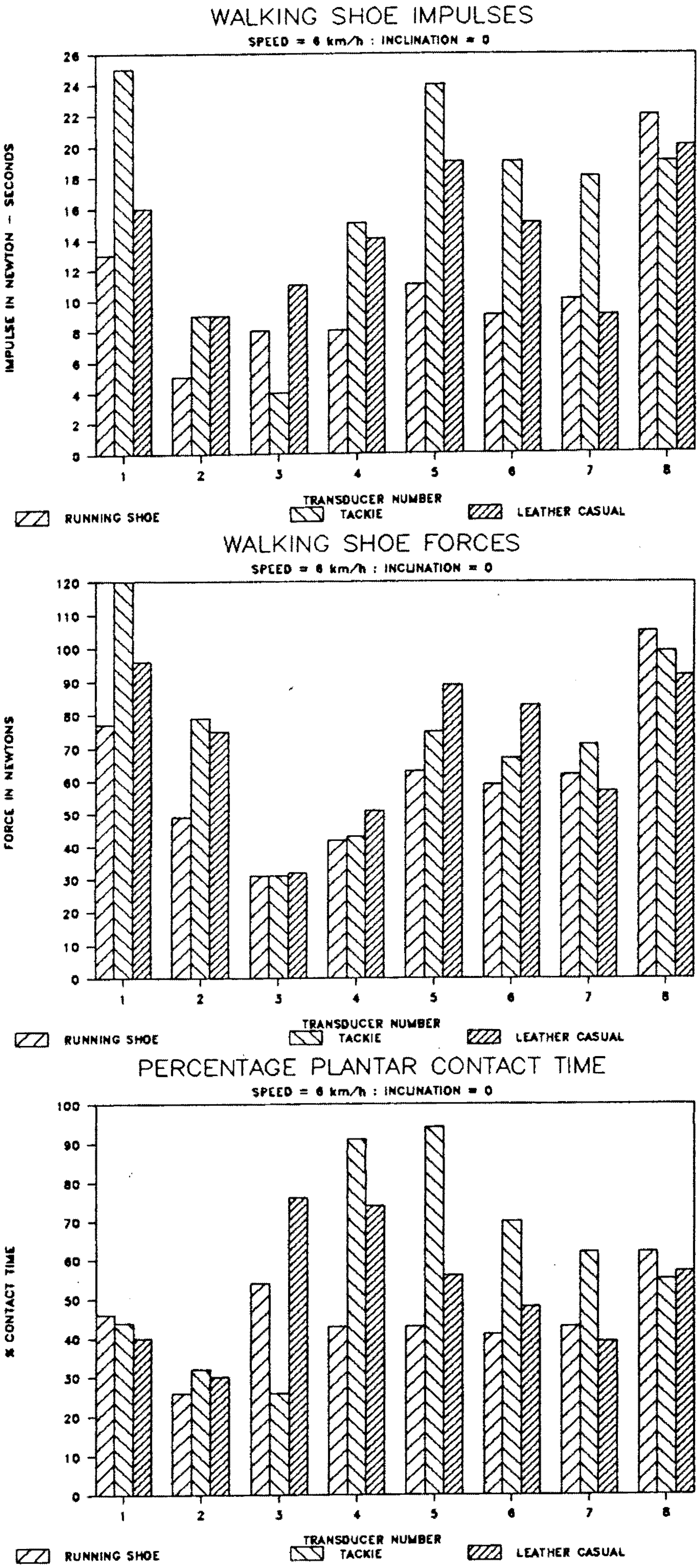
iii) Comparison of Shod and Unshod Foot Forces.

iv) Measurement of Running Shoe Cushioning Properties.

v) The Effect of Inserts on the Gait Pattern.

vi) Effect of Worn-out Running Shoes.

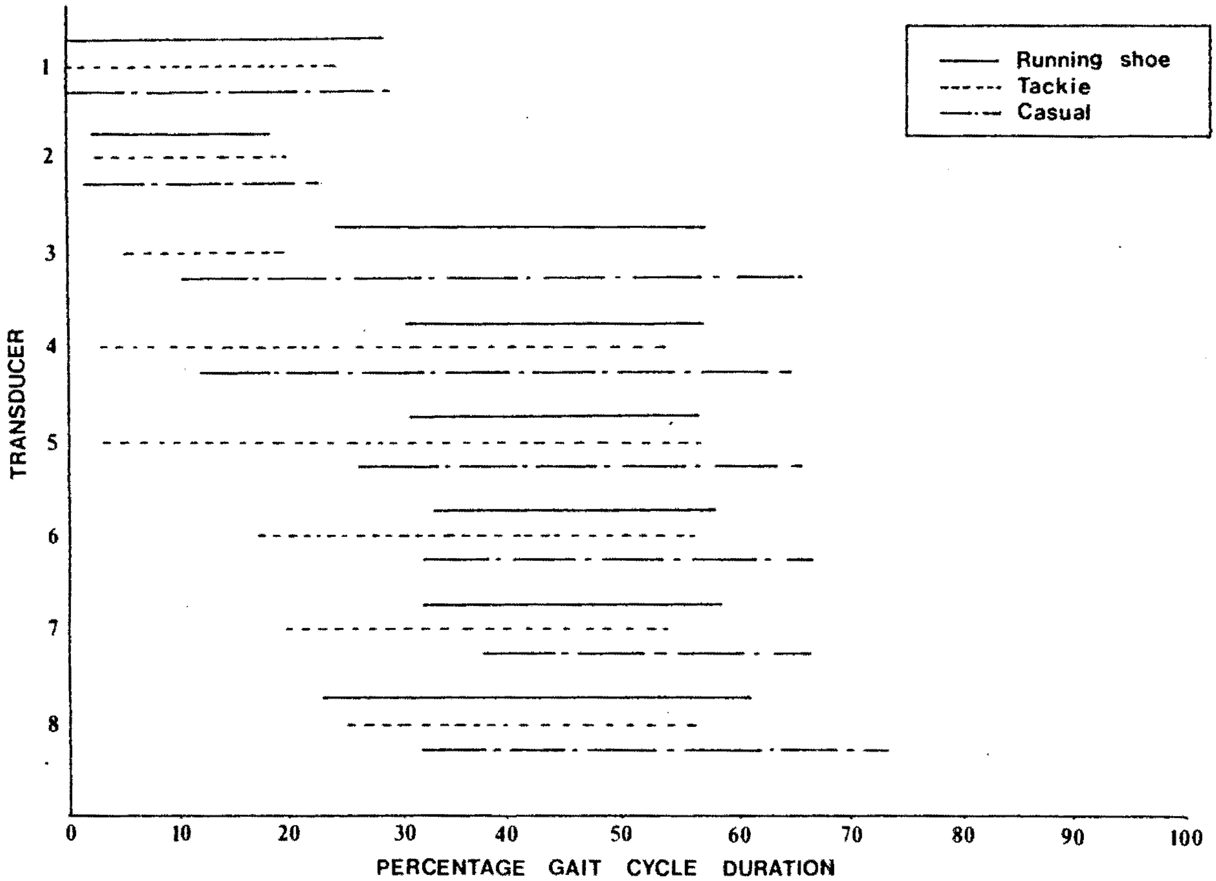
vii) Effect of High-Heeled Shoes on Women.


WALKING SHOE IMPULSES
SPEED = 6 km/h : INCLINATION = 0
IMPULSE IN NEWTON – SECONDS
0
2
4
6
8
10
12
14
16
18
20
22
24
26
1
2
3
4
5
6
7
8
TRANSDUCER NUMBER
RUNNING SHOE
TACKIE
LEATHER CASUAL
WALKING SHOE FORCES
SPEED = 6 km/h : INCLINATION = 0
FORCE IN NEWTONS
0
10
20
30
40
50
60
70
80
90
100
110
120
TRANSDUCER NUMBER
RUNNING SHOE
TACKIE
LEATHER CASUAL
PERCENTAGE PLANTAR CONTACT TIME
SPEED = 6 km/h : INCLINATION = 0
% CONTACT TIME
0
10
20
30
40
50
60
70
80
90
100
TRANSDUCER NUMBER
RUNNING SHOE
TACKIE
LEATHER CASUAL

a

b

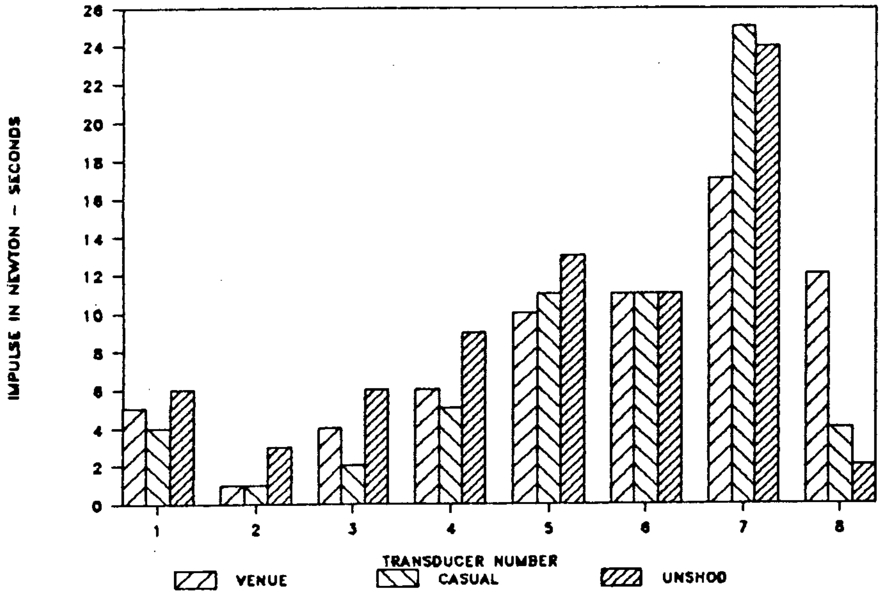
c



Running shoe
Tackie
Casual
TRANSDUCER
1
2
3
4
5
6
7
8
0
10
20
30
40
50
60
70
80
90
100
PERCENTAGE GAIT CYCLE DURATION

SHOD / UNSHOD IMPULSES

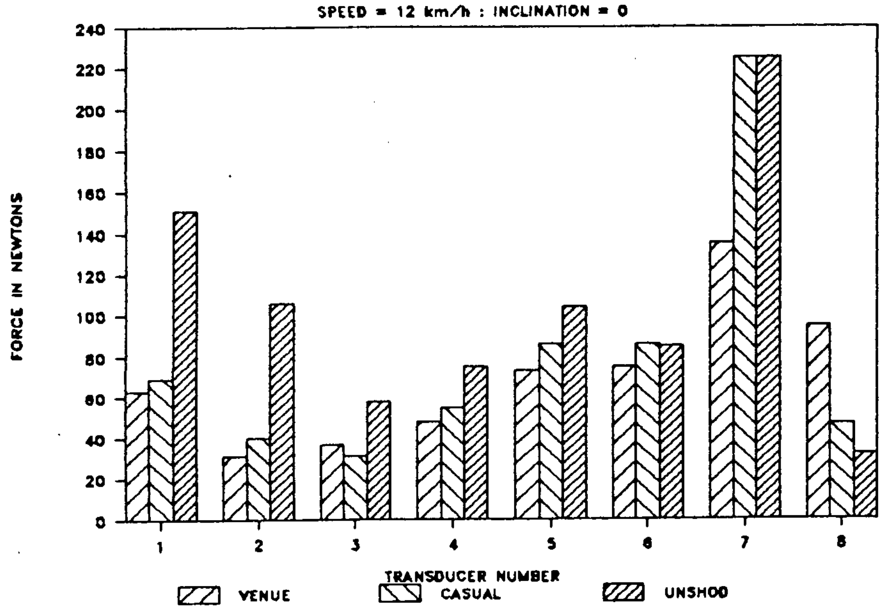
SPEED = 12 km/h : INCLINATION = 0



a

SHOD / UNSHOD FORCES

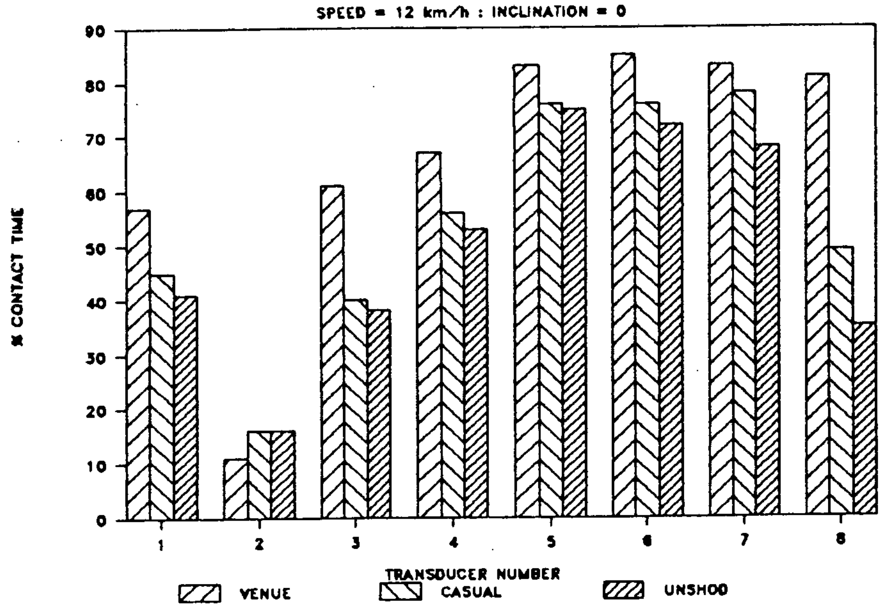
SPEED = 12 km/h : INCLINATION = 0



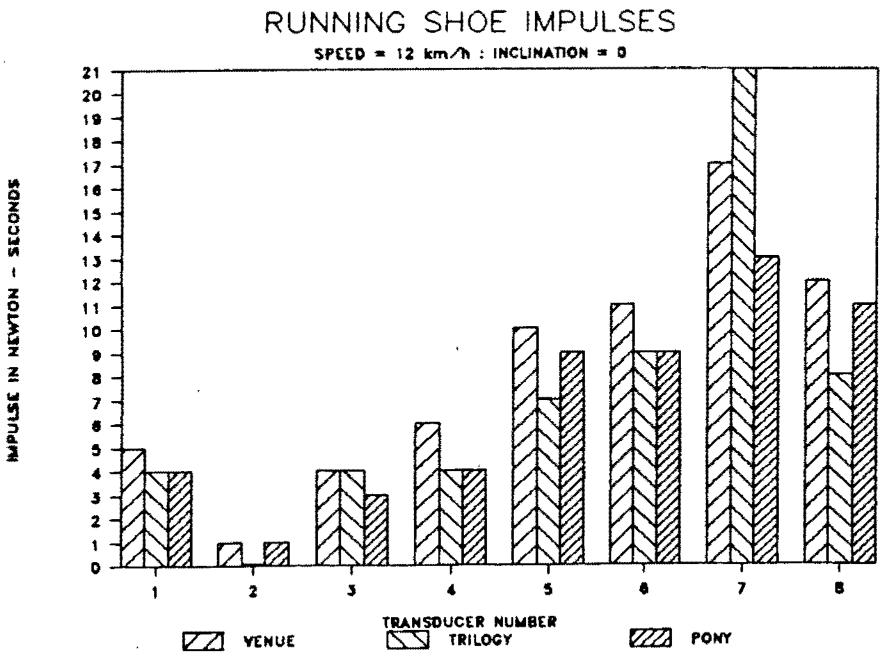
b

PERCENTAGE PLANTAR CONTACT TIME

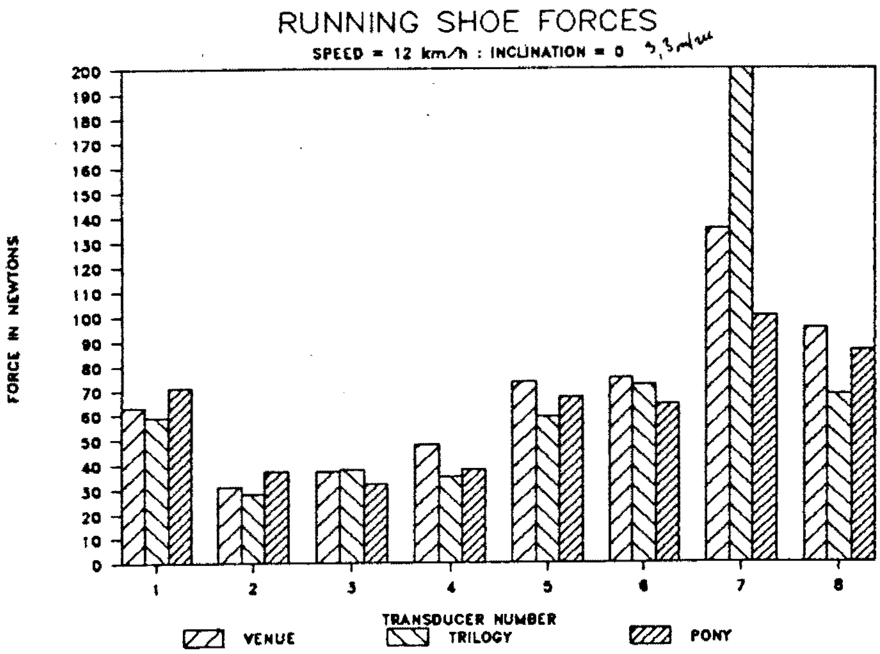
SPEED = 12 km/h : INCLINATION = 0



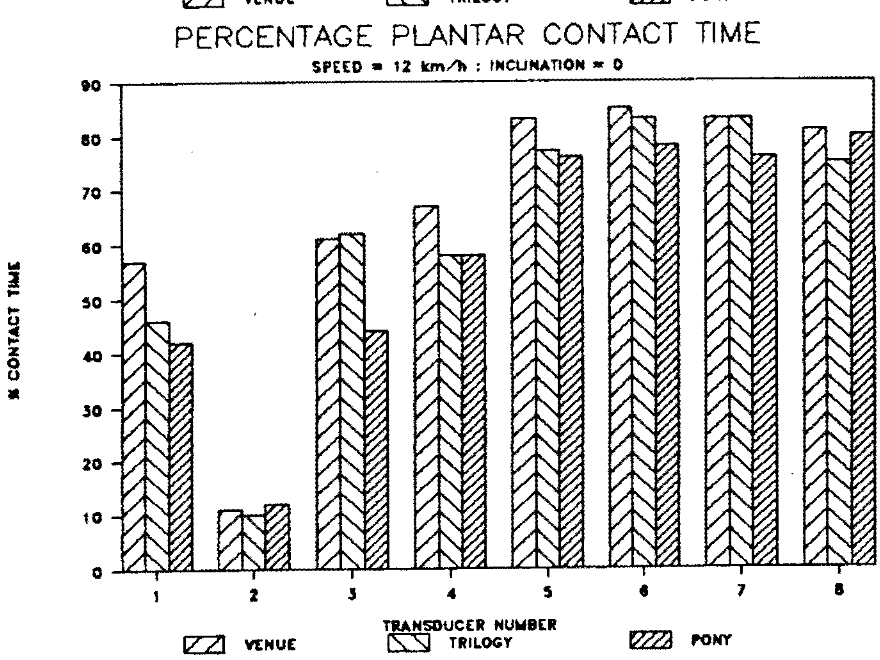
c


RUNNING SHOE IMPULSES
SPEED = 12 km/h : INCLINATION = 0
IMPULSE IN NEWTON – SECONDS
TRANSDUCER NUMBER
VENUE
TRILOGY
PONY

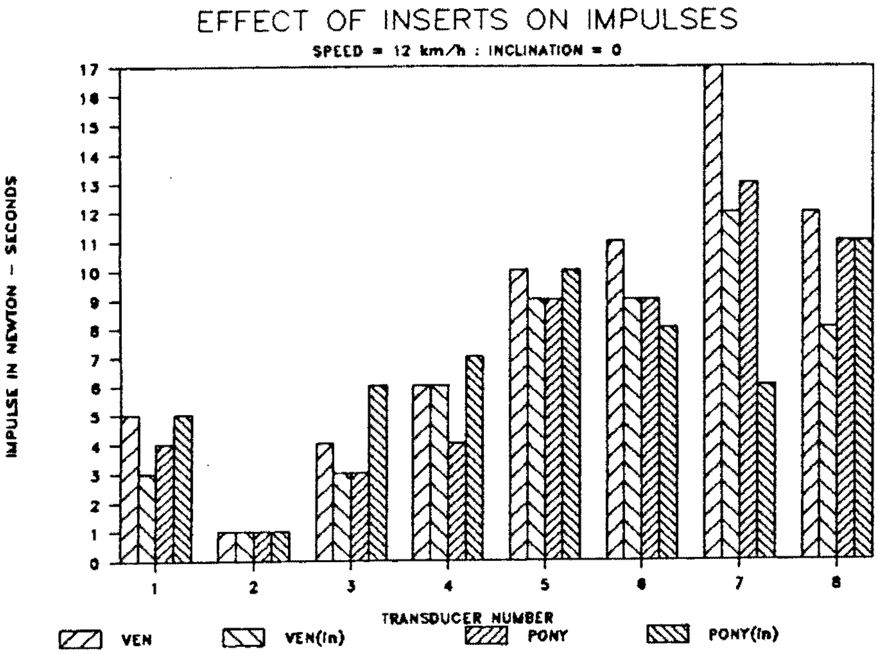
a


RUNNING SHOE FORCES
SPEED = 12 km/h : INCLINATION = 0
3,3 m/sec
FORCE IN NEWTONS
TRANSDUCER NUMBER
VENUE
TRILOGY
PONY

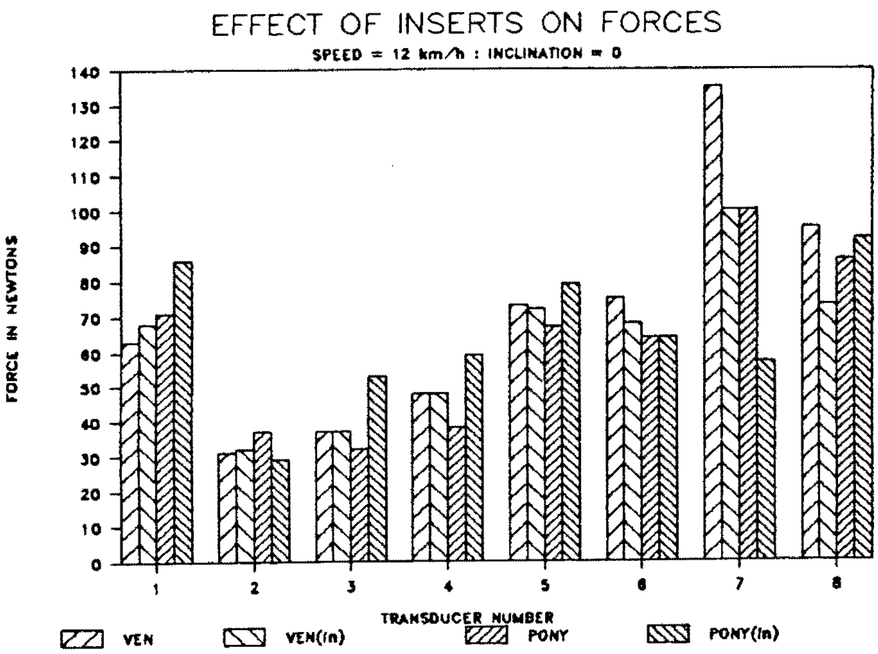
b


PERCENTAGE PLANTAR CONTACT TIME
SPEED = 12 km/h : INCLINATION = 0
% CONTACT TIME
TRANSDUCER NUMBER
VENUE
TRILOGY
PONY

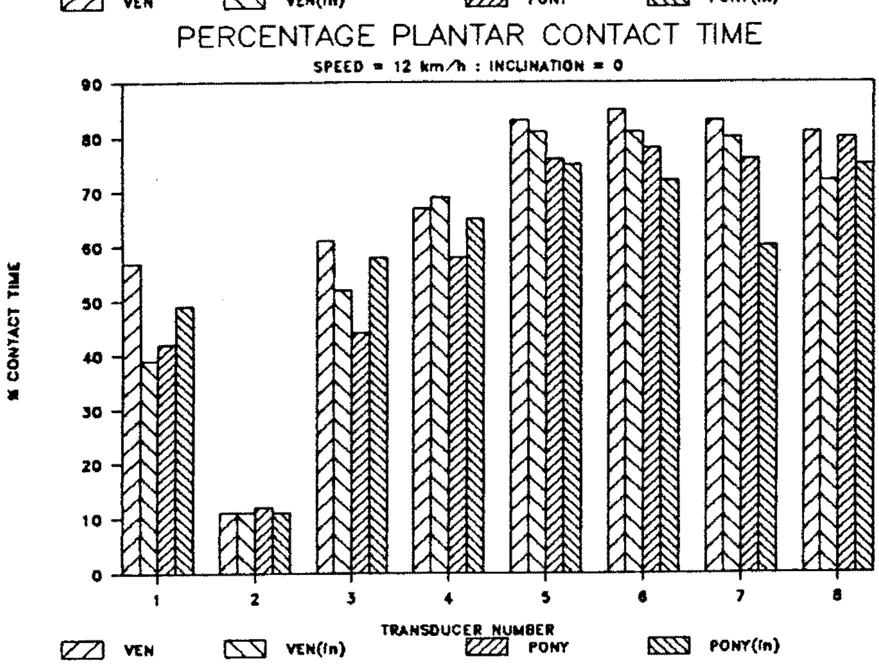
c


EFFECT OF INSERTS ON IMPULSES
SPEED = 12 km/h : INCLINATION = 0
IMPULSE IN NEWTON - SECONDS
0
1
2
3
4
5
6
7
8
9
10
11
12
13
14
15
16
17
1
2
3
4
5
6
7
8
TRANSDUCER NUMBER
VEN
VEN(In)
PONY
PONY(In)

a


EFFECT OF INSERTS ON FORCES
SPEED = 12 km/h : INCLINATION = 0
FORCE IN NEWTONS
0
10
20
30
40
50
60
70
80
90
100
110
120
130
140
1
2
3
4
5
6
7
8
TRANSDUCER NUMBER
VEN
VEN(In)
PONY
PONY(In)

b


PERCENTAGE PLANTAR CONTACT TIME
SPEED = 12 km/h : INCLINATION = 0
% CONTACT TIME
0
10
20
30
40
50
60
70
80
90
1
2
3
4
5
6
7
8
TRANSDUCER NUMBER
VEN
VEN(In)
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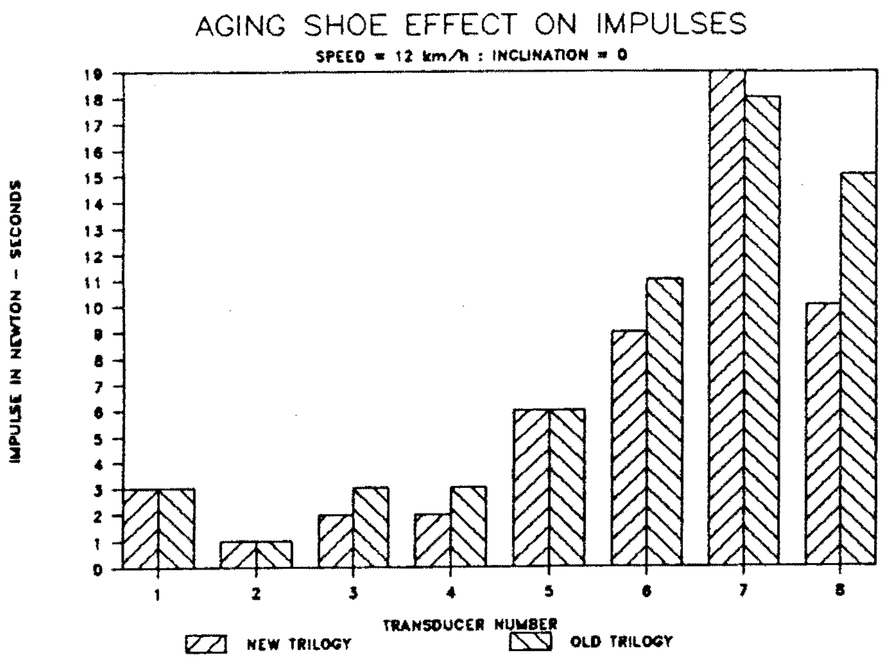
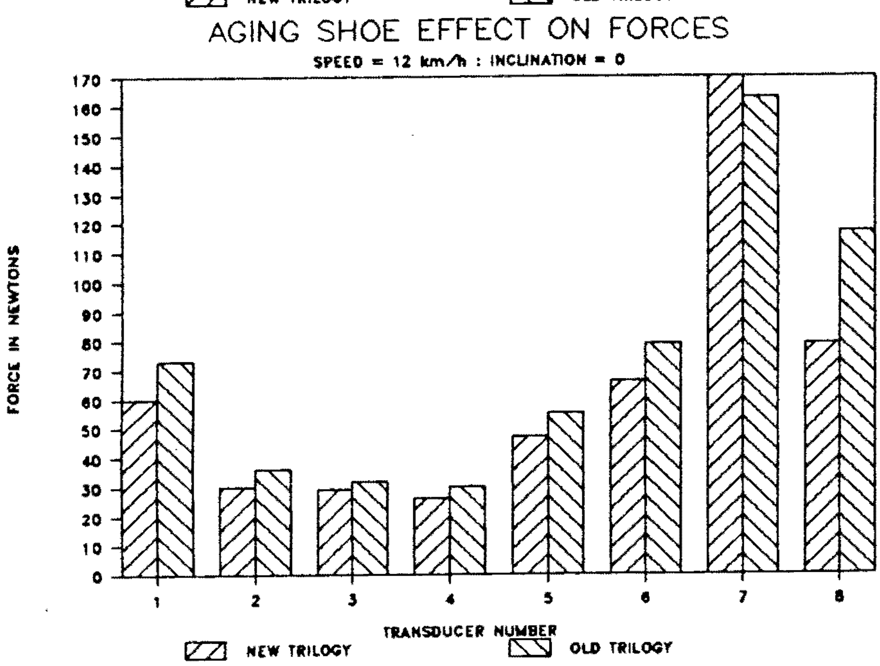
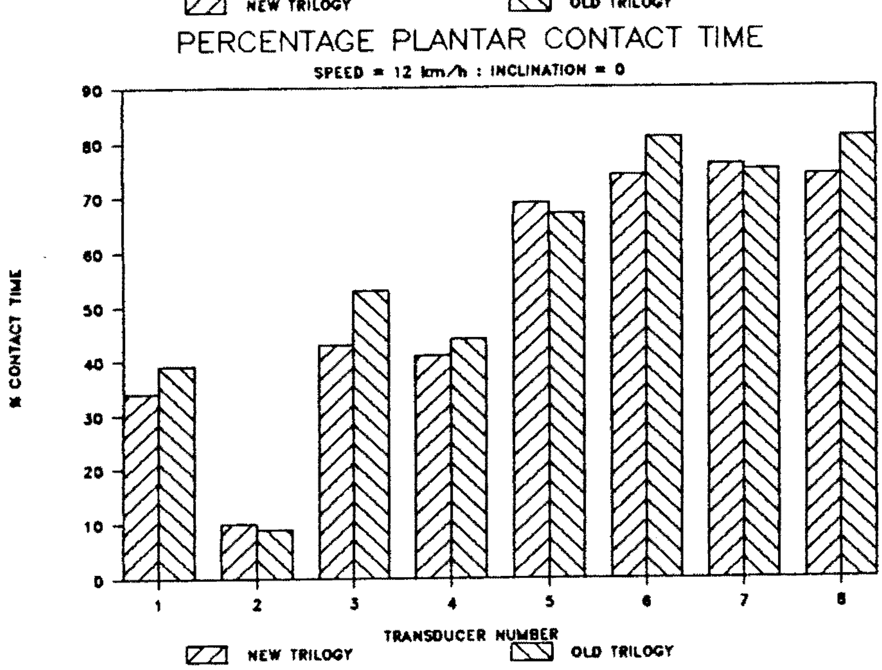

AGING SHOE EFFECT ON IMPULSES
SPEED = 12 km/h : INCLINATION = 0
IMPULSE IN NEWTON – SECONDS
TRANSDUCER NUMBER
NEW TRILOGY
OLD TRILOGY

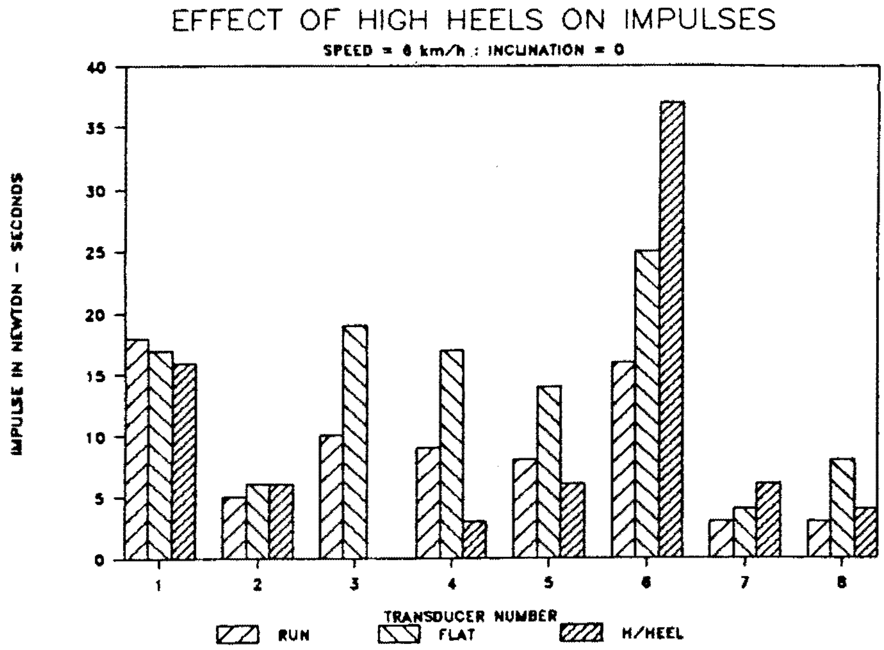
a


AGING SHOE EFFECT ON FORCES
SPEED = 12 km/h : INCLINATION = 0
FORCE IN NEWTONS
TRANSDUCER NUMBER
NEW TRILOGY
OLD TRILOGY

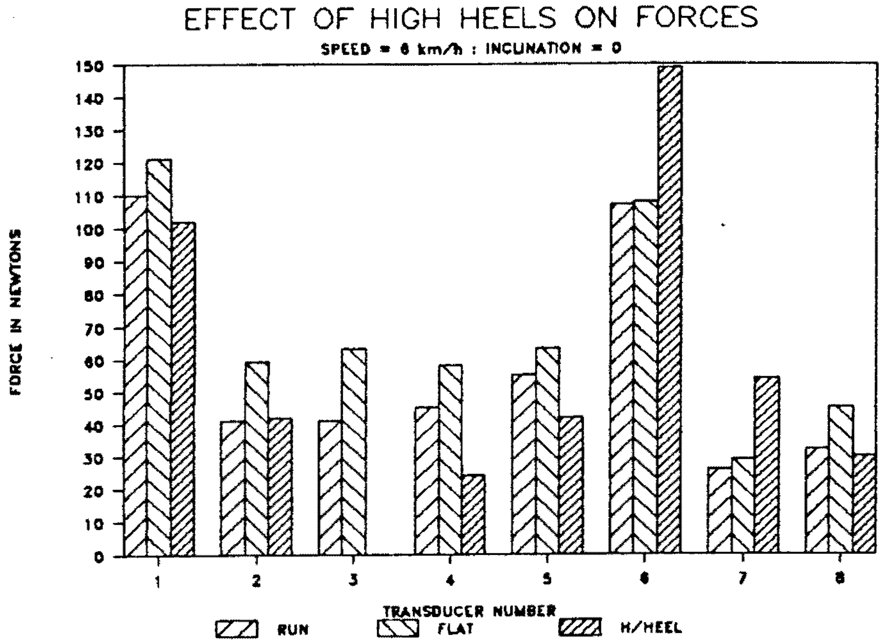
b


PERCENTAGE PLANTAR CONTACT TIME
SPEED = 12 km/h : INCLINATION = 0
% CONTACT TIME
TRANSDUCER NUMBER
NEW TRILOGY
OLD TRILOGY

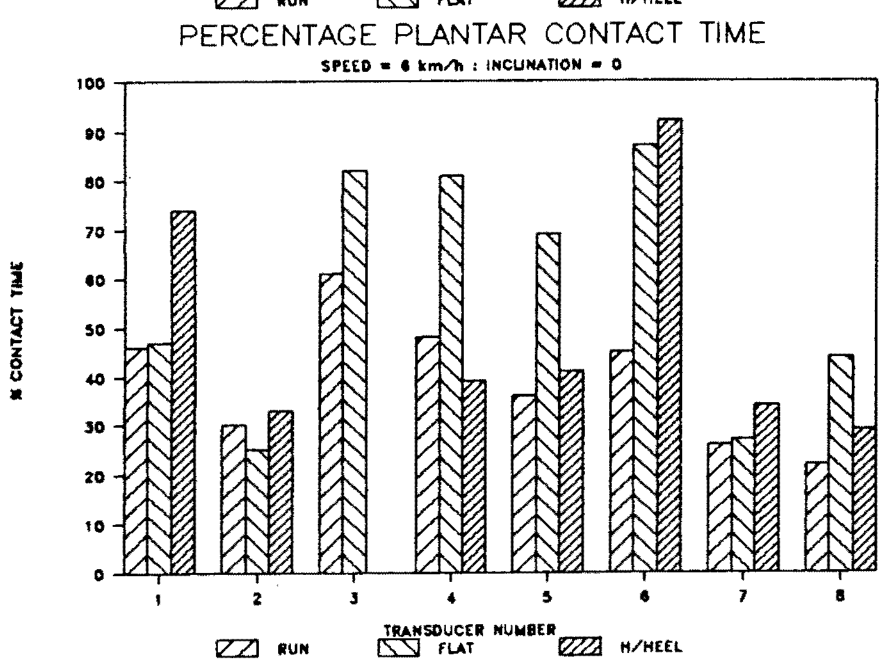
c


EFFECT OF HIGH HEELS ON IMPULSES
SPEED = 6 km/h : INCLINATION = 0
IMPULSE IN NEWTON - SECONDS
TRANSDUCER NUMBER
RUN
FLAT
H/HEEL

a


EFFECT OF HIGH HEELS ON FORCES
SPEED = 6 km/h : INCLINATION = 0
FORCE IN NEWTONS
TRANSDUCER NUMBER
RUN
FLAT
H/HEEL

b


PERCENTAGE PLANTAR CONTACT TIME
SPEED = 6 km/h : INCLINATION = 0
% CONTACT TIME
TRANSDUCER NUMBER
RUN
FLAT
H/HEEL

c

APPENDIX F

PUBLICATION

The following brief article was published in the "Proceedings of the Ninth Annual Conference of the American Society of Biomechanics", held in Ann Arbor, Michigan, 2 - 4 October 1985.

A PORTABLE IN-SHOE PRESSURE MONITOR

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The assessment of lower extremity function can be greatly enhanced by the measurement of the mechanical environment at the foot/shoe interface. This was recognized a century ago by Marey (1886), and more recently has been the focus of two groups in the U.S.A., Cavanagh et al (1983) and Wernick (1983). The purpose of our study was to develop an in-shoe pressure monitor to overcome some of the problems associated with previous systems.

A block diagram of our system is illustrated in Figure 1. A thin insole, having eight pressure sensing transducers, is placed inside the shoe. With a mass of 170 g and thickness of 3 mm, the insoles come in different sizes with various tranducer configurations and can be used for either foot. The insole is connected to the processing circuitry by means of a thin cable running up the outside of the patient's leg. The backpack (mass of 1 kg) carried by the patient performs two functions : it stores the pressure data in memory (4 seconds at 250 Hz per channel); the video generator produces bar graphs which are

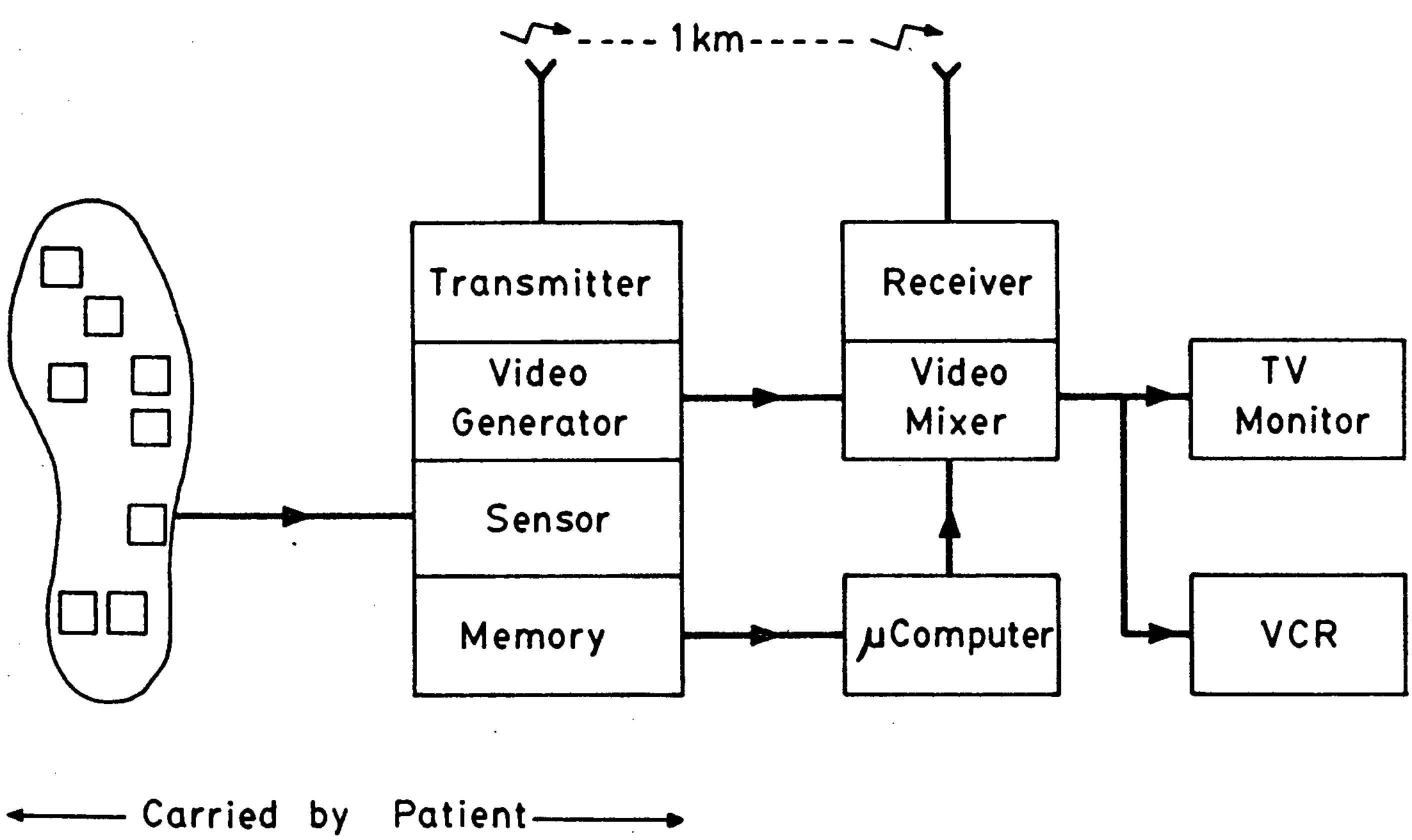

FIGURE 1

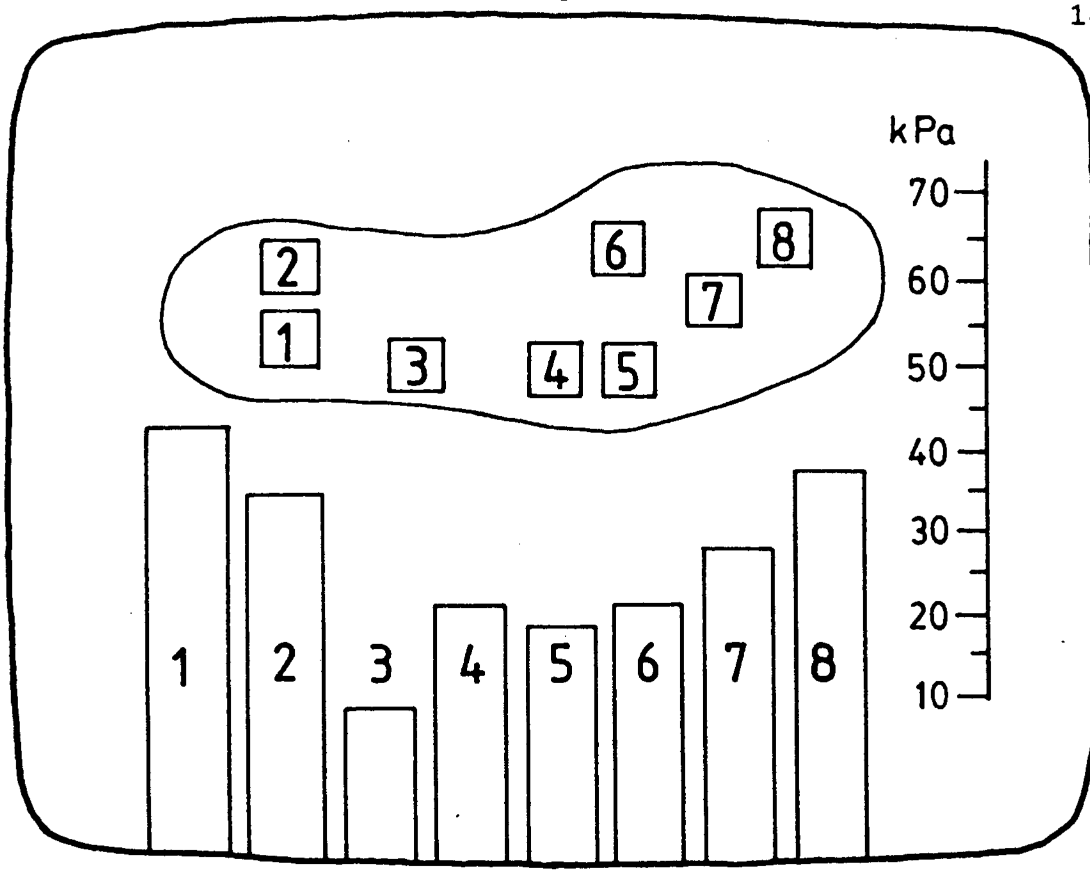

FIGURE 2

transmitted either by an umbilical cable or as an RF signal to a video mixer. The mixer combines a software generated mask from a ZX Spectrum microcomputer with the bar graphs to produce a display on the TV monitor as shown in Figure 2. The display, which functions in real-time, is dynamic for 15 seconds, and static for 5 seconds when it highlights the maximum value at each transducer location. All the information, together with any patient particulars typed on the screen, may be stored on a domestic video cassette recorder (VCR) with freeze frame facility. The digital data stored in memory may be downloaded via an interface to the microcomputer for further processing and display.

We believe that our in-shoe pressure monitor has a number of features -- low cost, instant data display, portability, simple operation, RF transmission range of 1 km, thin insole -- that enhance its potential as a clinical assessment tool. At present it is being used to study normal subjects and patients with running injuries.

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