MECHANICAL AND ELECTRICAL
STUDIES ON THE HORSE

A thesis presented in partial fulfilment
of the requirements for the degree of
Doctor of Philosophy at Massey University

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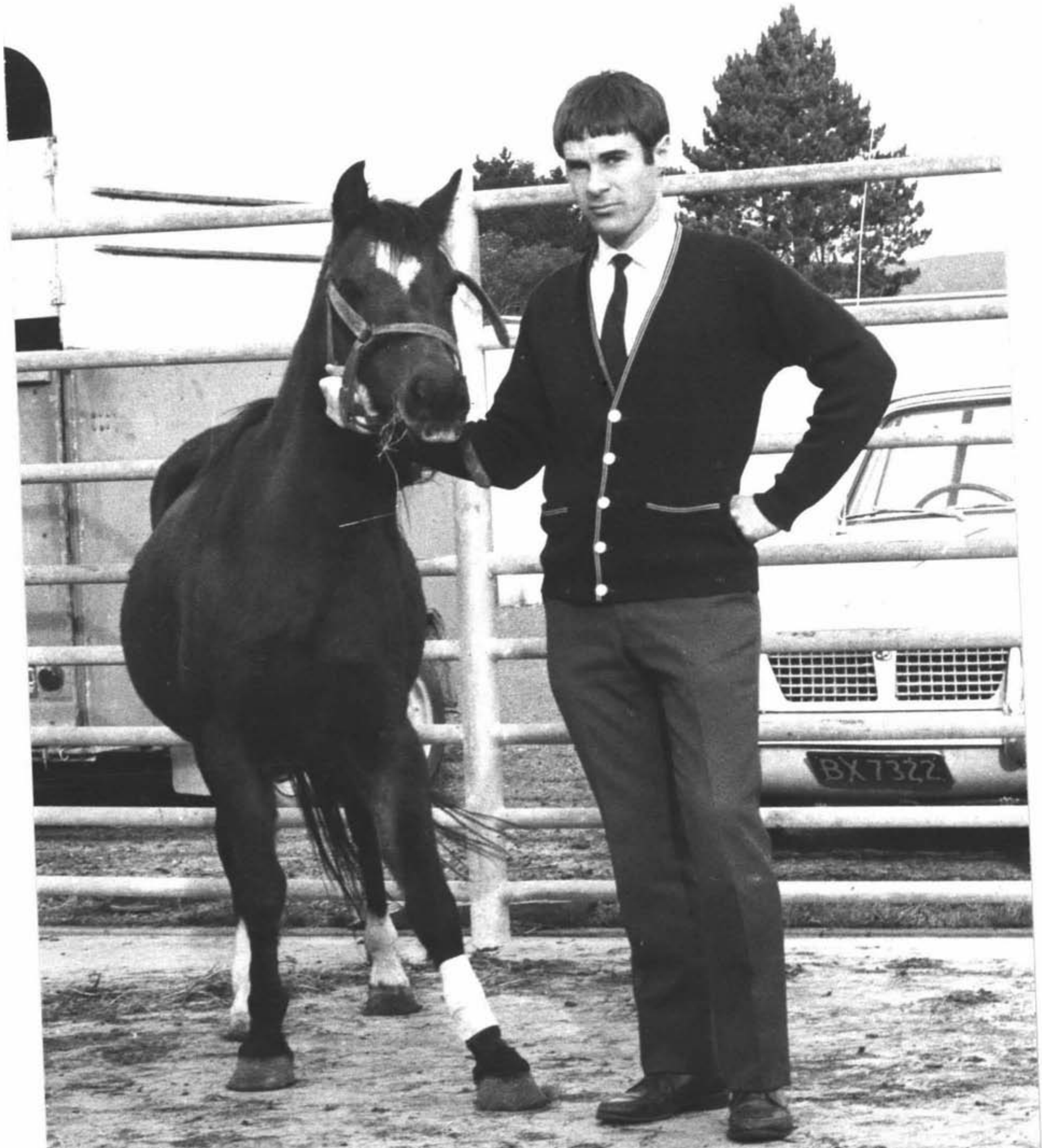
FRONTISPIECE

Great Expectations - the author and horse 001.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ABSTRACT

Muscle forces may have profound influences on stress and strain within the muscle - skeletal system of an animal. These forces have previously been measured by indirect methods often with severe restrictions on the type of movement (if any) that the skeleton was allowed to undergo. A direct method of muscle force (or tendon tension) measurement in the conscious animal was sought which did not necessitate restriction of skeletal movement.

This project reports the manufacture and successful *in vivo* functioning of a tendon tension 'buckle' transducer. The use of the device was demonstrated by correlating tension in the common digital extensor tendon with strain on the lateral and medial aspects of the bone of the walking horse. Strain was monitored by bonding foil electrical resistance strain gauges to the bone surface using a contact cement.

The swing and support phases of the forelegs were monitored by switches attached to each forefoot and occasionally monitored by cinematography. Thus it was possible to make a detailed analysis of the lateral bone strain recording in terms of body weight bearing and of muscle action via the monitored tendon. Also considered were the influences of tension in other tendons of the limb and the action of the head and neck during walking.

A further application of the tendon tension transducer was its use in monitoring tendon tension during recordings of the electrical activity generated by the corresponding muscle. This activity, which was recorded from intramuscular wire electrodes, was later rectified, integrated and compared with the directly measured force.

Experience in electromyography was gained through extensive studies of the equine larynx during which the instrumentation was improved.

The early attempts at bonding strain gauges and flanges to living equine bone failed because the epoxy resin flanges which held the lead wires became detached from the bone. The epoxy resin mouldings were consequently screwed in place but this caused slight lameness immediately following surgery.

Experiments with a tendon tension transducer proved fruitless until the 'buckle' configuration was adopted. This device, when suitably constructed, did not significantly affect the gait of the animal.

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GLOSSARY

Generally, a term is defined only in the context in which it is used in the script.

abduct:	to draw away from a centre or median line.
adduct:	to draw toward a centre or median line.
aspect:	that part of a surface viewed from a particular direction.
autoclave:	a self-locking apparatus for the sterilization of materials by steam under pressure.
bolus:	a rounded mass, e.g., a quantity of food entering the oesophagus at one swallow.
caudal:	the tail end of the body.
contralateral:	pertaining to or situated on the opposite side.
cranial:	the head end of the body.
distal:	away from the long axis of the body.
dorsal surface:	that directed away from the plane of support (the ground), OR (of the manus) opposite to palmar surface.
dowel:	a peg or pin for fastening.
electrogoniometer:	an electrical instrument for measuring angles.
electromyography:	the recording of the changes in electric potential of a muscle.
EMG:	electromyography or electromyogram.
endotracheal:	within the trachea.
fascia:	a band or sheet of tissue enclosing and connecting muscles.
fire:	discharge.
gauge factor:	ratio of the fractional change in resistance and the strain sensed by an electric resistance strain gauge.
insertion:	attachement, as the site of attachement of a skeletal muscle on the bone that is moved when the muscle contracts.

interosseus:	between two bones.
intratracheal:	endotracheal.
in vitro:	in glass; referring to studies performed on tissues removed from the living organism under artefical conditions in the laboratory.
in vivo:	in life; referring to studies performed on tissues not removed from the living organism.
ischaemia:	local diminution in the blood supply.
isometric:	maintaining the same length (e.g. of muscle).
isotonic:	of the same strength.
kineplasty:	utilization of the stump of an amputated extremity for producing motion in a prosthesis.
kinesiology:	scientific study of movement of body parts.
laparotomy:	incision of the abdominal wall.
lateral:	pertaining to or situated at the side.
medial:	pertaining to or situated toward the midline.
motor unit:	the motor fibre of a cranial nerve, together with the striated muscle fibres innervated by its terminal branches.
motor unit threshold:	muscle force beyond which motor unit is activated.
movement artefact:	the signal resulting from a disturbance of the steady state potential of an EMG electrode.
MPa:	Megapascal.
muscle head:	a segment of a muscle.
myofibril:	a muscle fibril running parallel with the long axis of the fibre and representing the contractile elements.
neutral surface:	the locus of points in an object at which there is no longitudinal extension.
origin:	the end of a skeletal muscle that remains relatively fixed when the muscle contracts.
palmar:	pertaining to the palm.
percutaneous:	performed through the skin.

periosteum:	a specialized connective tissue covering all bones and having bone - forming potentialities.
phalanges:	pl. of phalanx.
phalanx:	any bone of a finger or toe.
phrenic:	pertaining to the diaphragm.
piezoelectric effect:	electric polarization produced by compression or extension of a material.
plasticity:	the retention of deformation after removal of stress.
p.p.m.	parts per million.
prosthesis:	an artificial substitute for a missing body part.
proximal:	near the long axis of the body.
recovery (of strain):	tendency of deformed body to resume original shape after removal of applied forces.
recumbent:	lying down.
reflection:	a turning or bending back.
relaxation (of stress):	decreasing stress required to maintain a given deformation.
resection:	surgical removal of a considerable portion of an organ or body part.
rheology:	the study of deformation and flow of materials.
sarcomere:	the unit of length of a myofibril.
settling time:	the time taken by the step response of a filter to remain within 5% of its final value.
Steinmann bone pin:	a surgical nail inserted in distal portions of bones for skeletal tractions.
subcutaneous:	beneath the layers of the skin.
suture:	a stitch,series of stitches, or the application of such.
systolic:	pertaining to the contraction, or period of contraction, of the heart, especially of the ventricles.
tensile strength:	the tensile stress at which a body will fracture, or will continue to deform with decreasing load.

tetanus:	sustained contraction of a muscle.
time constant:	the time taken for current, voltage or charge to reach $(1 - 1/e)$ of its final value.
tonic:	characterized by continuous tension.
trace:	graph with time as abscissa.
ultimate strength:	tensile strength.
ventral surface:	that directed toward the plane of support (the ground).
viscoelastic substance:	a solid or liquid which can both store and dissipate energy.

PREFACE

Many horses during races or while training have injured themselves beyond recovery by the overstrain of a bone or tendon in a leg. It seemed appropriate to begin a study of this biophysical topic, since at Massey University the experimenter (a physicist) could count on the advice and help of people from the Veterinary Faculty. Experiments began in 1972.

The money available was small, and we feel much indebted to the owner of the local knackery, Mr J.C. Clement, for his willingness to make horses available for limited periods, between the time when he received them and the time when he needed to destroy them and prepare the meat for the big-cat cage at Wellington Zoo. The University fed and stabled the horses. The period varied between seven and thirty days. On one occasion a horse seemed to have dislocated its shoulder while recovering from the anaesthetic. Mr Clement's advice by telephone was for us to destroy the horse, and at once to bleed the carcass, gut it, and chill it in a particular way. With good help this was achieved and we did not incur the cost of a spoiled carcass.

The biophysical study involved the progressive development and evaluation of a comprehensive monitoring system for recording biomechanical and bioelectrical information from the horse foreleg. We wished to demonstrate that it was possible to monitor various phenomena simultaneously and then to interpret these measurements in terms of the walking gait of the animal. We developed systems for monitoring bone strain, then developed techniques for monitoring those gait and tendon tension phenomena thought likely to influence bone strain. Lastly we extended the study to include electromyography of horse limb muscles.

The first operations on legs were attempts to fix resistance strain - gauges to a bone of the lower forelimb. Work on hind legs was avoided as needlessly risky.

The next operations demonstrated (for the first time) that one could fit transducers to indicate tension in an uncut tendon. Recordings were made as the animal stood, walked or trotted.

In the third stage, intramuscular wire electrodes were inserted to detect electrical signals activating the muscles. This proved easy, and it was done in the hope that a suitably smoothed EMG signal might give an imitation of tendon tension good enough for some purposes. This sequence of tests involved nine operations, using eight horses in all. The horse's gait was originally recorded on cine film, but it proved simpler to record the contact of foot switches to show whether each forefoot was on the ground or off it.

The complete system, making simultaneous records of bone strain, tendon tension, myoelectric activity and foot position, was then fitted in succession to two horses. Chapters 3 and 4 show some records and attempt some comparisons.

The use of knacker's horses had the advantage of low cost. It also allowed the forelimb to be retrieved from the knackery, so that the transducers and their sites could be inspected for damage.

The arrangement with the knackery, however, had two disadvantages which are reflected in the work we were able to do. Firstly we had little control over the availability of horses or over the length of time that we could work with any one particular horse. This meant that in general we were restricted to only one operation per horse. It was not practicable to perform a whole series of separate operations on one particular horse even if the horse could have withstood the consequent limb damage.

The second disadvantage of our horse supply was that we had little or no control over the type of horse supplied, many were old, some were hacks, one was a thoroughbred, some were used to being handled, some were semi-wild and bit or fought. There was no possibility of training these animals to walk in a preferred gait in the available time so the variability in recordings taken from different horses was expected to be large. Also, recordings were taken over a period of four years as the transducers were developed. Consequently no effort was expended to calculate mean values and standard errors for the monitored parameters as data collected under such different conditions could not be meaningfully combined.

The data was often recorded at night to avoid interference with and interference from the more usual activities of the large animal hospital of the University Veterinary Faculty, however, the late hour caused drowsiness and hence variability of horse gait. This, coupled to the fact that the horses were not trained in a particular walking gait meant that even for one particular horse the variation in gait from stride to stride caused decreased repeatability in the monitored parameters. Consequently a statistical analysis of results would reflect the variability of stride rather than the uncertainty of the measurements.

Extensive myoelectric laryngeal studies were made on anaesthetised horses, partly as a means of developing expertise in electromyography for studies on the equine limb. Fourteen normal and more than forty abnormal larynges have been electromyographically probed in a programme to elucidate the function of laryngeal muscles in both 'normal' and 'roaring' horses, but it was thought best not to attempt an analysis of these results in this thesis.

CHAPTER 1

INTRODUCTION

1.1 RESEARCH GOALS

The object of this study was the *in vivo* monitoring of parameters associated with a bone, a tendon and a muscle in an animal's leg during normal locomotion. More specifically, the correlation of features in bone strain and tendon tension recordings were studied. In addition, tendon tension (or muscle force) was related to the electrical activity in the muscle. It was considered that the simultaneous monitoring of these parameters would allow a unified approach to the investigation of the mechanics of the animal limb.

The emphasis in the project was more on the experimental techniques involved in the *in vivo* monitoring of the parameters than with attempts to determine the "typical record" or the spread in results recorded from a number of animals.

Initially we had hopes of signal averaging the mechanical data but its inconsistency, even from one animal, indicated that little useful purpose would have been served. Analysis of the raw data seemed more useful. Also spectral analysis of electromyograms (Barnes, Pinder and Goulden, 1974), although interesting, has not (yet) proved of great value, so again the raw data was preferred.

Kear and Smith (1975) stated that recordings of *in vivo* tendon strain were very similar between animals but it is also evident from their results that there can be variations in mean tendon strain of up to 300% between animals. The style of gait of the animal and the degree of lameness (if any) are likely to influence these variations.

1.2 CHOICE OF SPECIES

The animal chosen was the horse. This species is of interest to many people in New Zealand and throughout the rest of the world. Many millions of dollars are spent annually in attempting to predict the performance of horses and their locomotor systems. These systems, although highly developed, are prone to damage especially during a race.

One common problem in the thoroughbred racehorse is the longitudinal fracture of the third metacarpal bone (Figure 2.2-1) (Cheney et al., 1973). This bone is known variously as the foreleg cannon bone and the metacarpus (Turner et al., 1975). In the present study, the third metacarpal bone is sometimes abbreviated as Mc III.

Fortunately, New Zealand is well stocked with horses so they are not prohibitively expensive to acquire for experimental purposes. Massey University houses the only veterinary school in the country and so is ideally suited to large animal research.

Another factor favouring the horse as an animal suitable for this sort of experiment is its large size. This makes the construction and attachment of transducers less difficult. However because of its size and strength the horse demands a degree of respect not required when handling the smaller domestic animals.

1.3 SURVEY OF BONE STRAIN STUDIES

Research into the mechanical properties of bone may be classified in a variety of ways, for example the studies may be *in vitro* or *in vivo*, static or dynamic, on bone segments or whole bones, on wet or dry, on fresh or embalmed specimens, etc. The present study involves *in vivo* measurements of surface strain on (whole) bones.

1.3.1 *In Vivo* Studies

It seems that all of the *in vivo* bone strain studies have detected only surface strain by the bonding of electric resistance strain gauges to the surface of living bone. Perry and Lissner (1962) mentioned an early experiment in which collodion was used to bond a wire strain gauge to a chronically exposed area of canine leg bone (tibia) to record surface bone strain while the animal was walking. Evans (1953) described a similar experiment. The installation was viable for up to only 36 hours. The flesh over the bonding site was permanently removed to expose the bone. Such a procedure is not desirable if an animal is to be kept alive for long periods.

A significant improvement in experimental technique was heralded by the papers of Lanyon and Smith (1969, 1970). These authors employed a contact cement, isobutyl 2 - cyanoacrylate monomer to bond a semiconductor strain gauge to a tibia in each of several trained sheep which walked and trotted during recordings. These papers were the first to describe the long term (several weeks) successful functioning of strain gauges for measuring bone strain during normal voluntary activity. The semiconductor gauges used had a very high gauge factor (sensitivity) but covered only a small area of bone. The adhesive used was highly suited to <u>in vivo</u> bonding in contrast to methyl 2 - cyanoacrylate monomer (Cochran, 1972) which had been widely used for <u>in vitro</u> bonding. The latter adhesive was employed by Lissner and Roberts (1966) to bond strain gauges to vertebrae of living anaesthetized dogs. The animals were secured in a box and were subjected to large accelerations while still under anaesthesia. The tests were repeated after sacrificing and embalming the animals. These tests were usually completed within 100 hr of sacrificing the animal, although some took place a week later. The paper also discussed similar experiments on human cadavers.

Lanyon (1971a, 1972) described the measurement of strain in sheep vertebrae during quiet respiration, walking and trotting. He noted that it was impossible to find the blend of factors causing the observed strain pattern without more instrumentation and suggested that the single (semiconductor) gauges used should be replaced by groups of strain gauges. Such replacement would alleviate the problem of recorded strain being critically dependent on the alignment of a single gauge.

In vivo bone strain research was further advanced by the use of foil electrical resistance strain gauge rosettes (Lanyon, 1973; Lanyon, 1974) which enabled the changing directions and magnitudes of the principal surface strains to be calculated. One of the significant features of the techniques pioneered by Lanyon is that they enabled direct measurements to be made of surface bone strain in a nearly normal mechanical and biological environment.

Cochran (1972) reported the simultaneous recording of strain from two regions of the tibia as well as an electromyogram signal from the gastrocnemius muscle of a walking dog. That author used foil strain gauges in his independently developed methods for monitoring bone strain. He made a case for using this type of strain gauge instead of the small semiconductor gauges used by Lanyon and Smith. Foil gauges are, it was suggested, more rugged and convenient to encapsulate, and more readily available in sizes covering an adequate bone area, than semiconductor strain gauges.

Another step forward was reported by Lanyon et al. (1974) who recorded surface strain from the tibial shaft of one of the authors. That was apparently the first report of *in vivo* bone strain measurements recorded from a human.

The first report of *in vivo* bone strain measurements from a large animal was that of Barnes and Pinder (1974) who recorded longitudinal surface strains from the lateral and medial aspects of the equine Mc III bone. Subsequently there has appeared a more thorough account of strains recorded from the lateral, palmer, medial and dorsal aspects of the equine tibia, metatarsus, radius and Mc III bone (Turner et al., 1975). In both of these reports, foil electrical resistance strain gauges were bonded in pairs as a check on the performance of the installations.

A possible alternative to the electrical resistance strain gauge for the detection of bone deformation is the use of electrodes to sense the piezoelectrical effect in bone, as discussed by Bassett, (1966). He speculated on the origin of the effect and its relation to bone architecture.

Cochran (1974) used Ag : Ag Cl electrodes and an electrical resistance strain gauge to simultaneously record electromechanical and surface strain data from the radius of a walking dog. A good degree of correlation existed, but the record of the electromechanical data was marred by superimposed electromyographic (EMG) signals. The piezoelectric signals picked up in that experiment were not just from the bone surface. Holes 2 mm deep were drilled in the bone cortex to accommodate the Ag : Ag Cl pellets which constituted the sensitive region of the electrode assemblies.

1.3.2 *In Vitro* Studies

Many *in vitro* studies involve small samples cut at specified orientations from a whole bone. Cheney et al., (1973) performed single and cyclic loading on uncut equine Mc III bones as well as elasticity tests on specimens cut from similar bones. The authors also suggested that bone marrow fluid might play a role in the fracture of this bone of the racehorse.

Small blocks cut from bone have been used to determine their piezoelectric constants (Fukada and Yusada, 1957). Boiling a specimen did not cause it to loose its piezoelectric property, thus it was not of biological origin. All specimens were thoroughly dry for testing.

Machined specimens of bone have also been employed in studies of the elastic, viscoelastic and plastic properties of this material. A variety of techniques have been used to determine the strain (or deformation) experienced by the specimen which was usually mounted in a mechanical testing machine. McElhaney et al. (1970) measured deformation with very complient strain gauged cantilever contact arms. This system was adequate for the quasi - static loading (0.01 in./min/) used.

Black et al., (1973) detected grip to grip strain with two linear variable displacement transformers in their studies of the dynamic mechanical properties of human bone. Elastic moduli were found to vary within the frequency range 35 to 350 Hz used.

Bonfield and Datta (1974) used both a capacitance gauge and a Tuckerman optical gauge in their studies on the Young's modulus of compact bone. The authors found good agreement (generally within 5%) between results from the two types.

The plastic properties of bone were investigated by Burstein et al., (1972) who used the well known strain gauge extensometer to measure displacement (see also Simkin et al., 1973 and Reilly et al., 1974).

A technique for studying stress and strain in whole bones involves coating the structures with a brittle lacquer which cracks in response to tensile deformation in the underlying material (Evans, 1953).

The mechanical properties of bone and the appropriate tests for determining them were surveyed by Evans (1973). Yamada (1970) gave a broader coverage which included tendon, muscle and ligament.

Ultrasonic methods fordeterminingthe elastic modulus have also been applied to bone (Lang, 1970; Abendschein et al., 1970). These studies incurred only small bone strains.

The viscoelastic nature of bone has been investigated by several authors. McElhaney (1966) studied the dynamic response of bone and muscle tissue over a wide range of applied strain rates. Displacement was measured with a specially designed capacitance transducer having a frequency response to almost 30 kHz. The dependence of elastic properties on strain rate was clearly evident.

Smith et al., (1965) concluded that both the elastic and viscous moduli for bone were independent of frequency within the range 500 Hz to 3 500 Hz. However their bone specimens were kept in 50% alcohol which, as they pointed out, would have a dehydrating action. Also, they admitted the bone modulus was beyond the range of their equipment design, casting doubt especially on their assessment of the viscous modulus.

Laird et al., (1973) modelled a bone specimen as a massless spring in an experiment which indicated that bovine bone is linearly viscoelastic but that the three element viscoelastic model is not entirely adequate in the 1 to 16 kHz frequency range.

However Lakes et al., (1974) in a theoretical study based on the experimental results of other workers, concluded that for some frequencies, the linear viscoelastic theory does not apply to bone in compression.

1.3.2

It was initially envisaged that the present study would be expanded to include determinations of the visco-elastic properties of the equine Mc III bone. The method to be used was that of Robinson and Edgar (1970) with a preliminary determination of the elastic modulus to density ratio using resonant free vibrations of a cylindrical bone specimen (Robinson, 1974).

1.4 GAIT SENSING

Many different types of gait sensor have been proposed in the literature. Some were designed to do more than merely detect foot - ground contact, others require additional information for this contact to be identified.

Attenburrow et al., (1974), citing Marey (1874), described a very early form of accelerometer attached to each foot of a horse. The rider carried a "portable registering instrument". This was a remarkable piece of research for those times. The authors then gave results they obtained using a photoresistor attached to the sole of each foot. Such a sensor is very attractive for it has no moving parts, is very light in weight and should give reliable results if operated in well illuminated, clean areas.

A modern application of an accelerometer used in gait monitoring was given by Lanyon (1971b). Accelerometers are reliable in as much as they do not need adjustments, are independent of ambient lighting and are little affected by cleanliness of the foot. They could be used with animals on almost any type of ground. Accelerometers do however require additional information (e.g. cinematography) for unambiguous interpretations of their output signals.

The photographic method has also been known in gait studies for about a century (Muybridge, 1957). This method is effective for giving an overview of the gait but requires synchronization with other data records. An interesting extension of the photographic approach was reported by Frederickson et al. (1971) wherein high-speed cinematography was used to observe the orientation of glass-fibre moulds attached to the shoes of a trotting horse. Computer-aided analysis of the film indicated the support and swing phases of the limbs as well as the pitch, yaw and roll of the hooves. This technique is very sophisticated and was adapted from a monitoring system developed in the aviation industry.

Finley et al., (1969) constructed a conductive copper walkway for swing and support phase identification while simultaneously monitoring joint position with an assembly of electrogoniometers attached to human subjects.

Methods intended to measure the force between foot and ground would also indicate support and swing phases. In an extensive study on the draught force in horses, Bjorck (1958) had elaborate measuring shoes constructed which were fitted with strain gauges and resistors to measure both vertical and horizontal forces. Those shoes provided a wealth of information but were far more complex than our application required.

Frederick et al., (1970) mentioned strain gauges bonded to the hoof, but rejected this scheme because of difficulties expected in the calibration of the system. However it should be adequate in sensing the swing and support phases of the leg, provided that good bonding of the strain gauge could be maintained. The above authors favoured a specially constructed shoe containing load washers as force sensing elements. These washers were commercially available rings fitted with strain gauges positioned so as to respond to compression of the washers. Their output signals were easily interpreted but specially constructed and milled horse shoes were required. Similar shoes were used by Cheney et al., (1973) in their studies of track hardness in relation to fracture of the Mc III bone.

Another force measuring system involved force platforms planted in the ground, but to record a sequence of steps a considerable number of such platforms were required. Frederick et al., (1970) noted that horses endeavoured to avoid these platforms.

Human locomotion studies were undertaken by Leavitt et al., (1972) and Canzoneri et al., (1973) who attached binary foot switches to the heel, midfoot and toe of each shoe. The switches consisted of copper strips. Using a similar arrangement, Tokuriki (1973) banded pairs of metal plates to the soles of a dog during electromyographic gait studies. Switches such as these would not have been sufficiently rugged for attachment under a horse's hoofs.

Winter et al., (1972) inserted five custom made microswitches into each shoe and interconnected them to give various voltage levels in a manner similar to that of Canzoneri et al., (1973).

The type of gait indicator used in most of the present investigation was a commercially available microswitch attached to the lateral aspect of each forefoot. This type was chosen partly because the Devices chart recorder used in the initial stages of the research was equipped with two event markers which could be operated externally. Also it was felt that the most reliable electrical contact would be had from a commercially available switch. This type of switch worked well on the hard surfaces used during the present research, but might have been unreliable had the horse walked on soft or well grassed ground. The faster gaits could also have caused unreliable switch function because of inertia in the operating mechanism (see Figures 3.6-2 and -3). However for the purposes for which they were intended, the switches used proved to be quite adequate.

1.4

1.5 MUSCLE FORCE, TENDON TENSION AND STRAIN

Very little work has been published of the direct measurement of tendon tension or strain during life. *In vivo* muscle force or tendon tension have been calculated from assumed instantaneous anatomical geometries and an assessment of at least one external force. Frequently however, no attempt was made to calculate the actual muscle force, many authors being content to merely monitor the externally measured force (Lippold, 1952; Bigland and Lippold, 1954a,b; Rose et al., 1967; Holt et al., 1969; Lloyd, 1971; Currier, 1972; Fusfeld, 1972; Rosentswieg et al., 1972; Cain et al., 1973). However Bigland and Lippold did state the approximate ratio of the measured force to tendon tension.

There is a rapidly developing interest in the elastic and viscoelastic properties of tendon measured *in vitro*, and attempts have been made to relate these properties to conditions experienced during life. This topic is discussed further in Section 4.7.5.

1.5.1 Direct *In Vivo* Force Measurement

An early measurement of *in vivo* contractile force was made on a section of the right ventricle (Walton et al., 1947). The authors used calibrated springs attached to myocardiograph levers to measure isometric systolic tension of the myocardium. The technique involved levers pulling at two points on the canine heart surface during open - chest surgery.

A well known way of measuring muscle force in the physiology laboratory is to severe the tendon of insertion and attach a force measuring instrument to the free end. Teig (1972a, 1972b) used a similar technique to measure force and contraction velocity of the middle ear muscles of cats and rabbits. The animals were placed under general anaesthesia and subjected to major surgery, during which supramaximal stimulations were applied to nerves supplying the muscles. The resulting force and/or length changes of the muscles were monitored by suitable transducers.

Inman et al., (1952) used the semi-detached muscles of human kineplastic amputees in studies of the relationship between muscular tension and electrical activity.

Techniques such as these are not suited to the monitoring of voluntary muscle force in the normal conscious animal.

The tension of normal, voluntarily contracting human muscle was measured directly by Close et al., (1960) but in order to do so the tibia was absolutely immobilized by a Steinmann bone pin. A force gauge was rigidly attached to the skeleton by another Steinmann pin. That study was restricted to isometric and isotonic contractions. The authors pointed out that "unless the muscle end is free, or bone-pin techniques are used, muscle tension cannot be measured without elaborate leverage-system calculations". This statement has been superseded by the following developments.

The first report of direct <u>in vivo</u> tendon tension measurement in an unrestrained limb appears to be that of Shaw (1968) who resected a length of canine tendon and replaced it with a strain gauge sutured to the tendon stumps. Changes in tension on the strain gauge were recorded when the muscle was stimulated (300 g), when the dog stood (600 g) and when it walked (1000 g). No recordings of tension against time were published.

The first publication reporting the <u>in vivo</u> pattern of tendon tension in a mobile animal appears to be that of Barnes and Pinder (1974) (see Appendix II). The technique used did not require the severing of the tendon but adopted a modified form of the 'buckle transducer' first suggested by Salmons (1969) to directly monitor tendon tension. That author has independently suggested variations of his basic design (Salmons, 1972; Salmons, 1975), and the types have been discussed by Barnes, Pinder and Goulden (1975).

1.5.2 Indirect *In Vivo* Force Measurements

Indirect measurements of tendon tension during life are typified by the work of Alexander (1974) who used a force platform to obtain records of the forces exerted on the ground by a jumping dog during take-off. These forces, together with cinematographic and x-radiographic information, were used to calculate the instantaneous forces exerted by the limb muscles during take-off. Examples of the calculations required for isometric and anisometric contractions have been given by Bouisset (1973). Wani et al., (1974) gave detailed measurements and calculations for tensions developed in the biceps, brachialis and triceps muscles in man. Account was taken of forearm inertia, muscle friction and length - tension relationships. These are a complicated procedures which could be obviated by the direct monitoring of tendon tension.

Qualitative comparisons of human muscle force variation were made by Carlsoo (1966) who used "power-plates" set into the floor as well as measuring plates fitted to one shoe. Variations in the three dimensional force between shoe and floor during the initiation of walking could thus be made. Intramuscular electromyograms were simultaneously recorded from wire electrodes placed in certain leg muscles.

In order to monitor the floor contact forces during subsequent strides, more force platforms or plates would have been required.

1.5.3 Tendon Strain Measurement

An extensive *in vitro* study of the strain variation in the components of the extensor apparatus of the finger has been made by Sarrafian et al., (1970). They used cyanoacrylate cement to bond strain gauges to the surface of the tendons which were then tensed to control the movement of the mounted finger. The strain curves obtained were taken as assessments of the relative changes in tendon tension necessary to manipulate the finger.

Kear and Smith (1975) reported the first *in vivo* tendon strain measurements. They bonded a type of strain gauge arch to a tendon in each of several sheep and recorded while the animals walked and trotted. Section 4.7.7 discusses tendon strain and tendon tension transducers.

1.5.4 In Vitro Tendon Measurement

Arnold (1972, 1974a, 1974b) has published many papers on the mechanical and rheological properties of biological tissues. The cited papers refer to human tendon. The papers delt with force - length relations, ultimate tensile strength, damping, and rheological relaxation and recovery curves. Minns et al., (1973) related stress - strain and relaxation characteristics of human tendon to the protein fibres and amorphous matrix which constitute the composite material of tendon. The stress - strain curve for collagenous fibres has been discussed in terms of cross-linking and intermolecular forces present in polymers, (LaBan, 1962).

Simultaneous mechanical and light microscopic studies of collagen fibres have been made by Viidik (1972), and similar papers are discussed in Section 4.7.5. The relationship between cyclic extensions and biological aging of collagen has been investigated by Rigby (1964) using rat tail tendon.

These papers are but a few of the more than one hundred which have appeared reporting the artificial stessing of tendons to determine their physical properties (Kear and Smith, 1975).

1.6 ELECTROMYOGRAPHY IN BIOMECHANICAL STUDIES

According to Long (1974), the science of electromyographic kinesiology was launched by the work of Inman et al., who, in 1944 and subsequently (e.g., 1952) reported major attempts at correlating electromyographic (EMG) findings in moving muscles with the motion of the moved body segments. Although the term kinesiology implies movement, a considerable amount of research has been focussed on muscle in the isometric state.

EMG kinesiology aims to elucidate the function and co-ordination of muscles in different movements and postures (Jonsson, 1973). The approach to the subject used in the present investigation was to correlate the EMG from the humeral head of the equine common digital extensor muscle with tension in the corresponding tendon. Furthermore the EMG was suitably processed to yield (where applicable) a good indication of this tension. Many papers have appeared concerning correlation of the EMG signal with posture but the following survey will be restricted to papers dealing not only with this correlation but also with processing of the EMG signal to predict muscle activity.

A linear relationship between voluntary isometric tension and the (mechanically) integrated electromyogram was found by Lippold (1952). Surface electrodes were placed over the human gastrocnemius - soleus muscle group of the right leg and muscle tension was indirectly monitored with the aid of a rotating foot plate.

Since this pioneering work, integration has usually been accomplished electronically. Bigland et al., (1954a) found that both methods of integration gave similar results.

These authors also found a linear relation between muscular electrical activity and tension for constant velocity of both muscular shortening and lengthening. The integrated electrical activity was less dependent on tension during lengthening than during shortening. This dependence on tension increased with velocity of muscle shortening. However, the integrated electrical activity during muscle lengthening appeared to be independent of velocity.

By electrically stimulating the muscle, force - velocity curves were obtained which appeared to satisfy Hill's force - velocity equation (Hill, 1938).

The authors also compared surface and intramuscular EMGs, finding that both gave similar correlations of integrated electrical activity with force provided that the needle electrodes had large bared tips. These would approximate the wire electrodes (Section 5.3) used in the present investigation.

In their second 1954 paper, Bigland et al. found that muscle tension varied in a distinctly non-linear S-shaped fashion with motor unit firing rate. Those authors also deduced that motor unit recruitment was more important than increasing firing rate in controlling muscle force. These findings are at variance with those of Milner-Brown et al., (1972, 1973b, 1973c and 1975) whose results are discussed below.

Inman et al., (1952) used an electronic integrator having charge and discharge times of 10 and 45 ms respectively, (somewhat shorter than the 66 ms averaging period used in the present study). The authors simultaneously recorded voluntary muscle force in amputees and electromyograms from the corresponding limb muscles, as well as the integrated rectified electromyograms. All of the surface, needle and wire electrodes used provided EMGs which, when integrated, mirrored the isometric muscle force recordings.

When the muscle length was allowed to vary, no such similarity in the records was found. Also, the tension peak followed the processed EMG peak by about 80 ms.

A nonlinear relationship between isometric muscle tension and averaged rectified EMG was found by Zuniga et al., (1969). They full-wave rectified the surface EMG signal and used an R-C filter with a 200 ms time constant. This corresponds to an averaging period (95% filter settling time) of 600 ms. A time constant of 200 ms was preferred to the 500 ms, 100 ms and 50 ms values also tried as it provided the desired amount of smoothing and retained reasonably detailed monitoring of EMG activity. The isometric contractions were about 4 s in duration.

The nonlinear relationship was such that the EMG activity increased more rapidly than the corresponding tension. Similar relationships were found by Vredenbregt et al., (1966). Such relationships are in contrast to the linear one found by Bigland et al., (1954a) but they apparently tested their data for only linear regression, not a nonlinear fit. Also, at tensions well below that corresponding to maximum effort, the nonlinear relationship found by Zuniga et al., does appear linear. These authors suggested that the curvilinear nature of the relationship at higher tension levels could be explained by synchronization of motor unit activity.

Another interesting feature of the results of Zuniga et al., is that a rapid initial increase in tension was not accompanied by a corresponding increase in processed EMG. Similar results were found in the present study and are discussed in Section 5.7.3.

Three types of low pass filter were tested by Kreifeldt (1971) for suitability in the smoothing of rectified EMG signals. The third order averaging filter consistently produced higher signal to noise ratios than either the simple RC filter (first-order Butterworth) or the third-order Butterworth filter. However the superiority of the averaging filter was diminished for smaller filter-settling times (250 ms compared with 1 s).

These three types of filter produce a continuous output roughly averaged over a time equal to their settling times. A less satisfactory arrangement is the resetting integrator whose output is discontinuous, being updated at the end of each averaging period.

The resetting integrator (intermittent averager) has been contrasted with the ideal integrator (moving averager) (Rubow et al., 1971) in studies on electromyographic learning in which the subject was asked to use his processed EMG signal to control a simulated prosthesis. The moving averager allowed better control of the device.

An unusual display of integrator output was used by Bouisset and Maton (1972) who monitored EMG activity in anisometric anisotonic muscle contractions performed against inertia at a non - constant speed. The integrated EMG was in the form of pulses, the pulse rate being proportional to the area under the electromyogram. This display form is difficult to interpret readily. Also, the authors did not (explicitly) state the averaging period of the integrators used.

Gottlieb and Agarwal (1970) point out the shortcomings of the first order filters commonly used to average the rectified EMG. These filters severely obscure the dynamics of the EMG signal so that the time of activation and to a much greater extent that of deactivation are difficult to deduce from the integrated record. The authors point out the advantages of the ideal linear filter and give the circuit of a third order filter approximation to the ideal having a 10 ms averaging period.

Jaw muscles were monitored to determine the relationship between chewing force and the EMG, both direct and integrated

(Ahlgren and Owall, 1970). During biting without food, the integrated EMG could be used as a reliable index to the biting force. Unfortunately, details of the integrator were not stated. Provided the bolus was homogeneous, (e.g. chewing gum) a quantitative relationship was also found during chewing. A homogeneous bolus might be expected to encourage uniform velocities of muscle contraction which in turn is conducive to a quantitative EMG - force relationship described by Bigland and Lippold (1954a). Chewing peanuts did not give a reliable relationship. However this unreliability could have been due to the stated near impossibility of reliably recording the peanut chewing force.

Ahlgren and Owall also noted a 41 ms delay between peak integrated EMG and maximal chewing force.

In studies of effort and fatigue, Cain et al., (1973) full-wave rectified and integrated EMG signals detected on the forearm surface. The integrator had a 0.2 s time constant and its output varied linearly with isometric tension for brief (3 s) muscular contractions. The brevity ensurred that the arm did not fatigue, however the tetanic force was not specified. These authors also presented traces (graphs with time as abscissa) of both force and integrated EMG for constant-effort contractions.

Angel (1974) made a study of the so-called two-burst pattern. This is the occurrence of two distinct bursts of electromyographic activity in the agonist muscle during the ballistic movement of an extremity. Position and velocity of the human hand was monitored as well as the surface EMG from the pectoralis major muscle. This EMG was also amplified, passed through a high pass filter and a precision full-wave rectifier. The rectifier output was processed by a three-pole, state-variable averaging filter (Garland et al., 1972) having an averaging period of 66 ms.

1.6

The EMG processing, including the averaging period, was virtually identical to that employed in the present study (Section 5.4). It is interesting that the same averaging period was chosen independently by Angel and the writer. This can be partly explained by the use of standard capacitors in the averaging filter (integrator). The reason the writer chose 66 ms is given in Section 5.4 but Angel (1974) did not state his reasons for so choosing.

Eldridge (1975) compared digital and analogue averaging of half-wave rectified phrenic nerve discharges and intercostal muscle EMG signals. For both sources of electrical activity an R-C time constant of 0.1 s (0.3 s settling time) yielded good agreement with the digital method of counting the pulses from a voltage to frequency converter for 0.1 s episodes. A 50 ms R-C time constant was less satisfactory because of increased "noise".

There was good correlation between integrated rectified phrenic activity and airway pressure changes generated by respiratory muscle contractions in the anaesthetized cat.

Comprehensive studies of the human motor unit have been made using the first dorsal interosseus muscle of the hand (Milner-Brown et al., 1972, 1973a, 1973b, 1973c and 1975). The results of these studies help to explain the good correlation observed in the present work between tendon tension and averaged rectified EMG. The first paper was a preliminary report. The 1973a paper described the recording of surface and intramuscular electromyograms together with the force generated by the muscle. This force was measured by a transducer held between the thumb and first finger.

The motor unit impulses recorded by the bipolar needle electrode were used to trigger a signal averager which monitored the tension to yield the twitch tension profile associated with the impulses. Also, the unrectified and rectified surface EMGs associated with these impulses were recorded.

The twitch tension profiles generated by a motor unit reached a peak 45 ms after the beginning of the motor unit impulses and declined to half the peak value 65 ms after the impulses.

The 1973b paper of Milner-Brown et al. reported that the twitch tensions of motor units spanned a two decade range and varied nearly linearly with the voluntary force level at which the units were recruited. This relationship held over the entire force range studied. The highest force level in the range was less than half the maximum voluntary force.

The authors also noted a rapid, nearly exponential decline in the number of additional motor units recruited as threshold force or twitch tension increased, and deduced that recruitment accounted for a diminishing fraction of force increase at high force levels.

In that study, force levels were maintained at constant levels while readings were taken.

Linearly changing force levels were used in motor unit firing rate studies reported by Milner-Brown et al., (1973c). They concluded that increases in motor unit firing rate are more important than motor unit recruitment in producing muscle force increases at intermediate to large forces. Recruitment dominates a low force levels. Such findings were unexpected in the light of previous papers.

The firing rate of all motor units began at about 8.4 impulses/sec regardless of their size or recruitment force.

The relation between muscle force and motor unit firing rate was linear for decreasing force regardless of its rate of decrease. A linear relation also held for intermediate rates of force increase, but non-linearities were observed both for slow and rapid rates of muscle force increase.

The 1975 paper of Milner-Brown et al., examined the relation between the surface electromyogram and muscular force. Intramuscular needle electrodes were used to trigger a signal averager which monitored the surface EMG. They found that large and small motor units were uniformly distributed throughout the particular muscle and that a motor unit's muscle fibers may be widely dispersed.

The peak to peak duration of the surface EMG wave form was independent of the motor unit threshold force but the peak to peak amplitude of this wave form increased approximately as the square root of the threshold force.

Cnockaert et al., (1975) furnished an interesting example of the use of electromyography in muscle force studies. The authors were able to deduce the relative contribution of individual muscles to the isometric contraction of a muscle group by assuming that the external torque developed by a muscle is linearly related to the integrated rectified surface EMG.

Ralston et al., (1976) noted an 80 ms delay between averaged rectified EMG and muscle force in several limb muscles and also found that tension outlasts the EMG by about 200 ms. The authors suggested that the role of muscles should be re-evaluated in terms of tension rather than just the EMG.

Perhaps the buckle transducer will prove to be suited to this task.

CHAPTER 2

ANATOMY AND SURGERY

2.1 INTRODUCTION

This study grew from an initial ambition to measure bone strain in the equine limb. It was considered desirable that the bone to be studied should be of interest to the racing fraternity, and have easy surgical access. Additionally, a simple shape and ample size for strain gauge bonding were important.

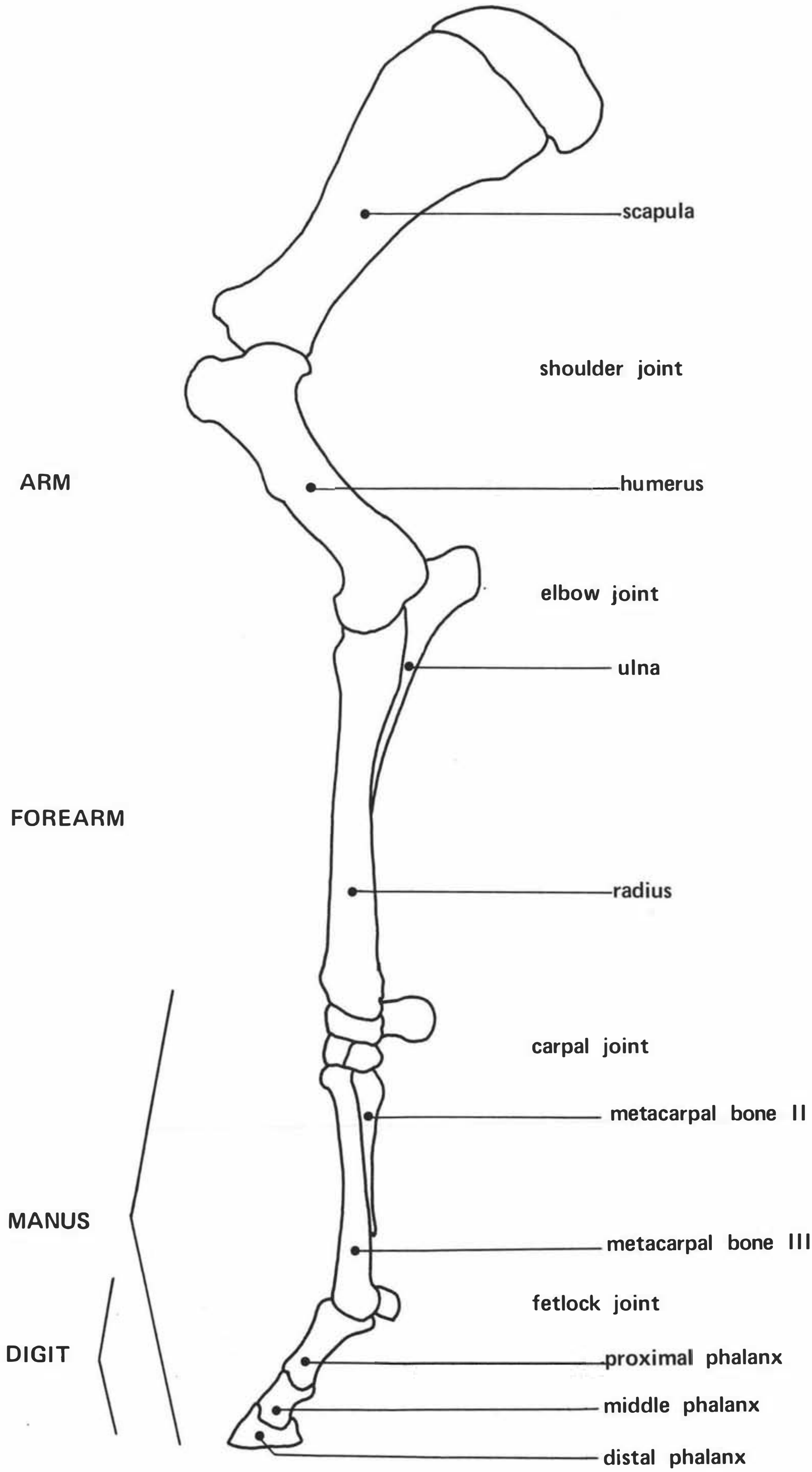
The bone chosen was the third metacarpal bone (Mc III bone, or cannon bone). This presents a number of sites suitable for the attachment of strain gauges and is believed to suffer compression stress fractures causing the common problem of 'bucked shin' (Rooney, 1969, 148-150).

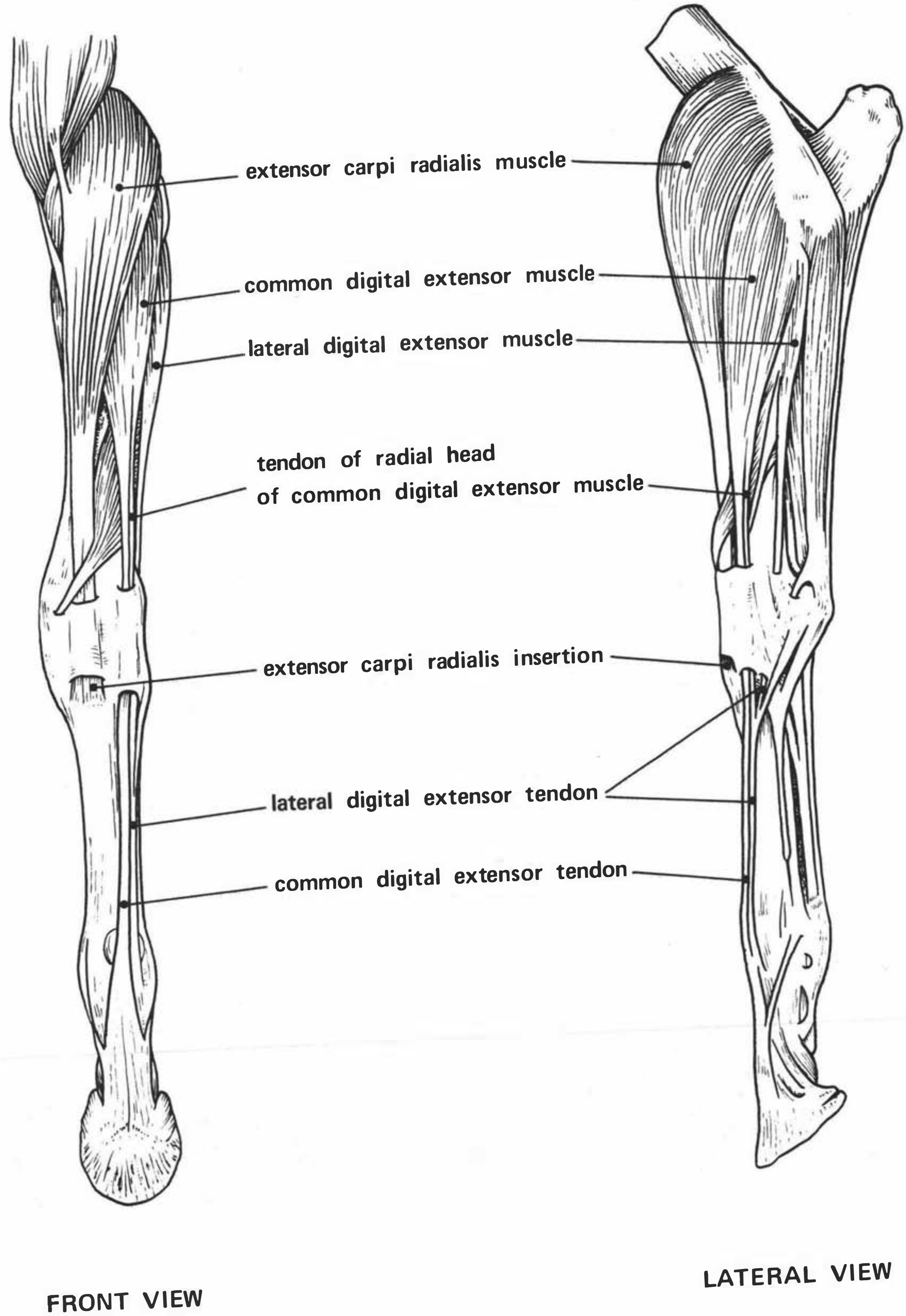
Bone strain is affected by a number of factors, one of which is muscular in origin (Lanyon and Smith, 1970). These authors mentioned that unco-ordinated muscular action may culminate in bone fracture. Muscular tension is transmitted to corresponding tendons.

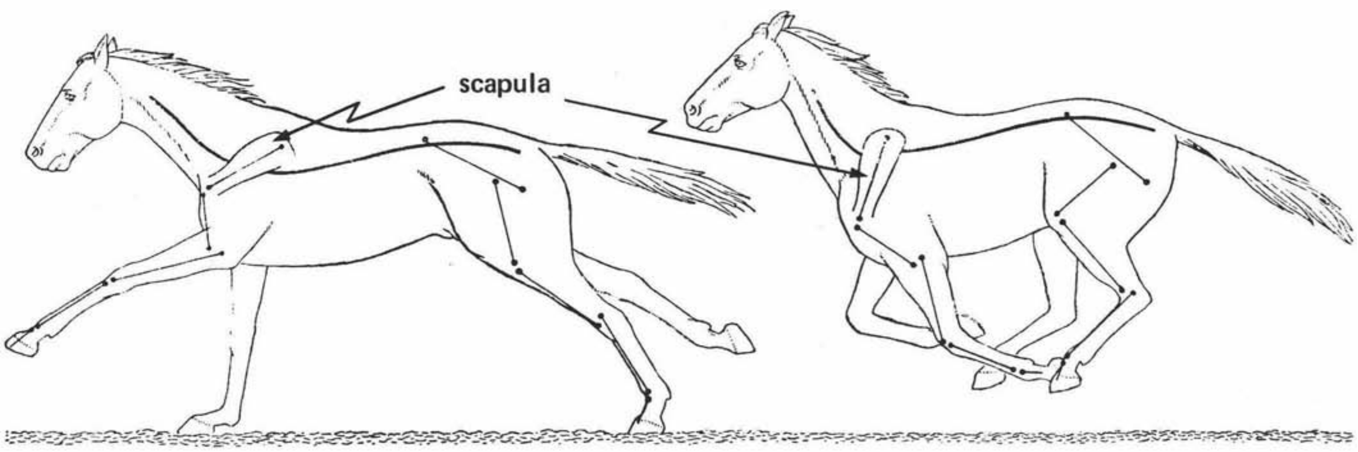
The study was therefore expanded to include the measurement of tension in a tendon thought likely to influence the strain recorded from the monitored bone. Because the flexor tendons were bulky and in close contact with one another, the two extensor tendons were considered more suitable. These were well separated in the region of the Mc III bone so that a transducer could be fitted to one tendon without unduly interfering with the other. The common digital extensor tendon was chosen as it was further from the strain gauge bonding site on the lateral aspect of the Mc III bone.

Tendon tension is largely the result of electrical activity in the corresponding muscle. It was therefore of interest to record the EMG signal from the muscle of the monitored tendon.

Three heads of the common digital extensor muscle were available, but because of its ease of access and large size, the


scapula
shoulder joint
ARM
humerus
elbow joint
ulna
FOREARM
radius
carpal joint
metacarpal bone II
MANUS
metacarpal bone III
fetlock joint
DIGIT
proximal phalanx
middle phalanx
distal phalanx


extensor carpi radialis muscle
common digital extensor muscle
lateral digital extensor muscle
tendon of radial head
of common digital extensor muscle
extensor carpi radialis insertion
lateral digital extensor tendon
common digital extensor tendon
FRONT VIEW
LATERAL VIEW


scapula

humeral head was chosen for recording the EMG signals from.

In order to gain experience in electromyography, a subsidiary project was undertaken. This involved a study of the function of the equine larynx as revealed by electromyography. For this reason, Chapter 2 includes a short description of the anatomy of the equine larynx.

2.2 ANATOMY OF THE THORACIC LIMB OF THE HORSE

Measurements were made on the third metacarpal bone, the common digital extensor tendon and the common digital extensor muscle. The function of these is influenced by their own structure and that of other entities in the leg. It is therefore useful to summarize the gross anatomy of the limb, especially the extensor apparatus.

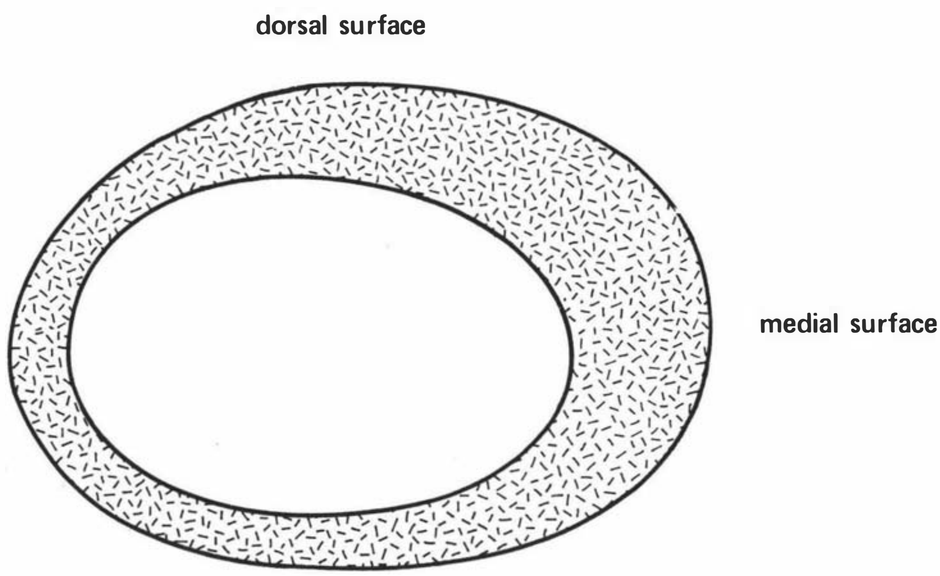
There are two systems to consider, one involving the bones and articulations of the thoracic limb, (Figure 2.2-1), the other the tendons and muscles of this leg (Figure 2.2-2).

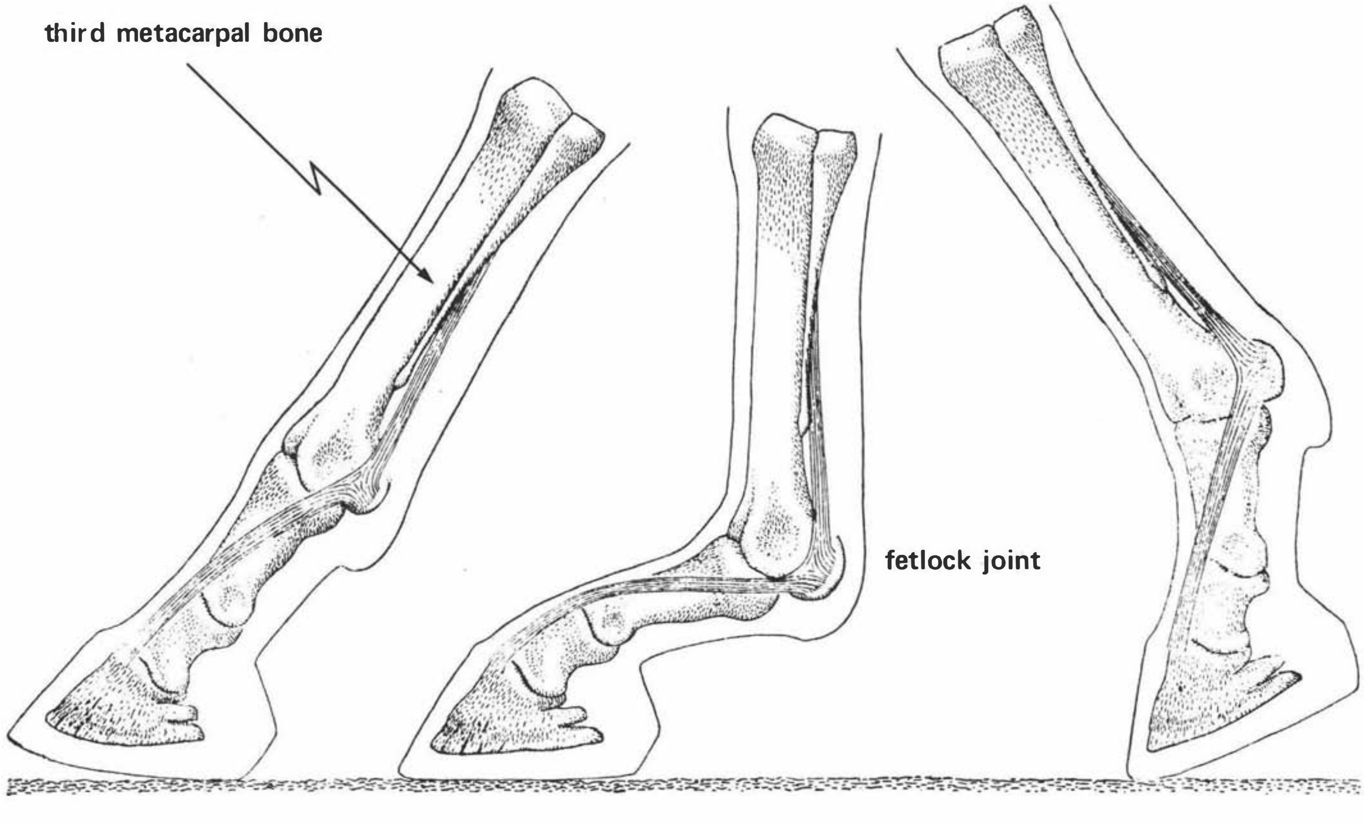
2.2.1 The Bones

The limbs of the horse have many functions, of which locomotion and the bearing of body weight are the most important. The hind (pelvic) limbs are of prime importance in locomotion. Figure 2.2-1 shows the bones of the thoracic limb in the standing position. We are interested in the shape of the bones and the sites of tendon attachment.

The scapula or shoulder blade is a flat, triangular bone which is able to rotate around a transverse axis (Sisson, 1975, p 354) as depicted in Figure 2.2-3. This action adds to the length of stride (Hildebrand, 1960). The large cranial prominence serves as attachment point for the strong biceps brachii muscle.

The humerus is a long bone having an irregularly cylindrical shaft, with a twisted appearance. The extensor carpi radialis muscle and the humeral head of the common digital extensor muscle


dorsal surface
medial surface


third metacarpal bone
fetlock joint

are attached to the lateral aspect of the distal extremity of this bone.

The radius is much the larger of the two 'forearm' bones, and is gently curved. The dorsal surface of the proximal extremity is endowed with three tuberosities. On the medial side is the radial tuberosity which gives insertion to the biceps brachii tendon. The large lateral tuberosity furnishes attachment to the radial head of the common digital extensor muscle and to part of the lateral extensor muscle.

In the adult horse the radius is fused with the ulna. This consists of a large proximal extremity (olecranon) and a shaft which tapers distally to a point. The olecranon is a lever arm for the muscles which extend the elbow.

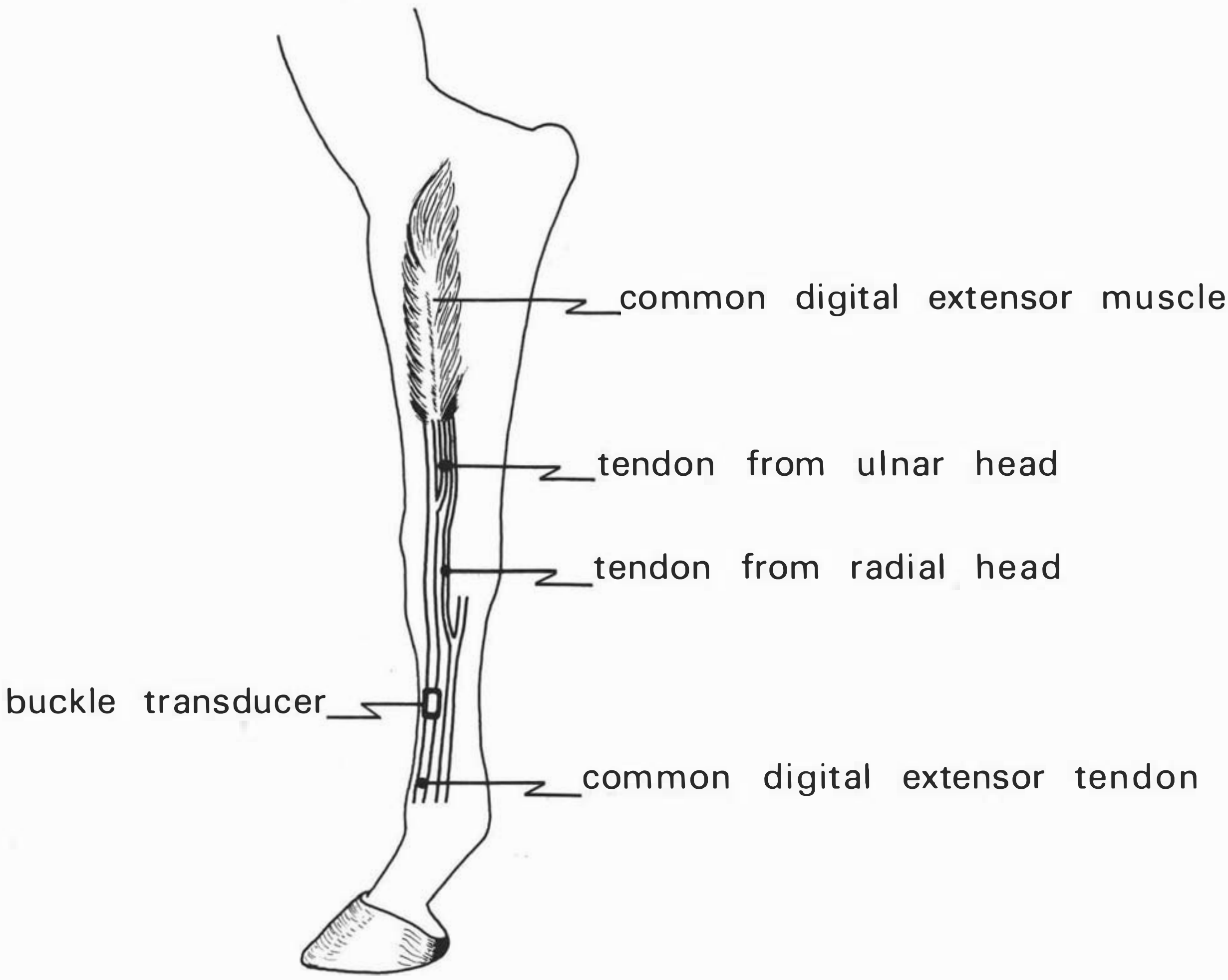
There are three metacarpal bones, coded Mc II, Mc III and Mc IV. Mc II and Mc IV are small tapering bones either side of the palmar surface of Mc III. This third metacarpal bone is the one from which bone strain was recorded, it is long, straight, and strong.

The shaft is elliptically cylindrical with a corrugated palmar surface.

The third metacarpal bone is one of the strongest bones in the skeleton. The compact substance is especially thick dorsally and medially (Figure 2.2-4). The medullary cavity extends further toward the ends than in most of the long bones of the horse and there is little spongy substance (Sisson, 1975 p. 290).

The tendon of the extensor carpi radialis muscle is inserted on the metacarpal tuberosity situated medially on the dorsal surface of the proximal extremity of Mc III. The palmar surface provides attachment for the suspensory ligament (Figure 2.2-5).

The digit of the manus is composed primarily of the proximal, middle and distal phalanges. Insertion of the lateral digital extensor tendon is on the lateral side of the proximal extremity of


common digital extensor muscle
tendon from ulnar head
tendon from radial head
buckle transducer
common digital extensor tendon

the proximal phalanx. The common digital extensor tendon is inserted on to the dorsal surface of the proximal extremities of all three phalanges.

2.2.2 The Common Digital Extensor System

The common digital extensor muscle (Figure 2.2-2) is a composite one, comprising three heads. The division is always rather artificial (Sisson, 1975) although each head is provided with a tendon (Figure 2.2-6).

The humeral head is much the largest, it is fusiform and terminates opposite the distal third of the radius. The common digital extensor tendon appears in a pennate fashion mid-belly on the surface of the muscle. This tendon passes through the lateral of two grooves in the distal extremity of the radius and over the carpal joint to incline medially down the dorsal aspect of the Mc III bone. Although at the proximal extremity of this bone the tendon is placed decidedly lateral to the middle line of the limb, this line is reached near the fetlock joint. The branches of the interosseus tendon blend with the common digital extensor tendon just distal to the middle of the first phalanx.

The action of this muscle and tendon is to extend the carpal articulation and the joints of the digit and to flex the elbow joint.

The small radial head of the common extensor muscle has a flat belly, followed by a delicate tendon which does not fuse with the common extensor tendon but may join the lateral extensor tendon (Figure 2.2-2).

The still smaller and deeper ulna head has a delicate tendon which may join the principal one.

2.2.3 Transducer Sites

The mid section of the lateral aspect of the McIII bone was the usual site for strain gauge bonding. There is a clear region of bone between the lateral digital extensor tendon (Figure 2.2-2) and the second metacarpal bone (Figure 2.2-1). It was easy to operate on this region when the horse was in lateral recumbency.

In one horse, simultaneous measurements were made from both the lateral and medial aspects of the mid section of the Mc III bone. The medial aspect is clear of tendons but the horse must be rolled over to lie on the monitored leg during the bonding of the medial gauges.

The buckle transducer was attached to the common digital extensor tendon either slightly distal or proximal to the bone strain recording site.

Electromyograms were recorded from the humeral head of the common digital extensor muscle. The recording site was near the origin of the muscle, since more distal sites would involve greater movement of the wire electrodes and hence more movement artefacts whilst recording.

2.3 SURGERY FOR BONE STRAIN MEASUREMENT

2.3.1 Preparation

The horses presented for surgery were weighed, their height was measured and their approximate ages were determined from dental examination by a veterinarian.

The animals were prepared for anaesthesia by the intramuscular administration of 20 mg of acetyl promazine. Anaesthesia was induced by intravenous injection of thiopentone sodium (1 g per 100 kg body weight). An endotracheal tube was placed in position and attached to a large animal Water's canister and rebreathing bag. The anaesthesia was maintained by use of a halothane-oxygen mixture delivered through a Fluotec Vaporizer.

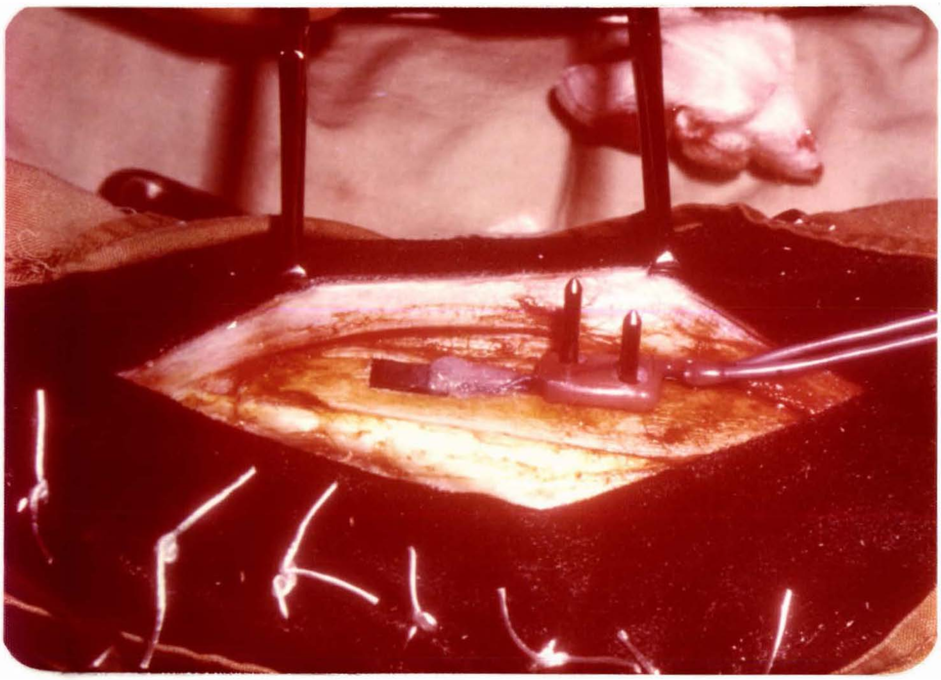
The strain gauge mounting unit (Figures 3.3-1, 3.3-2) and lead out wires were disinfected in 70% alcohol solution for at

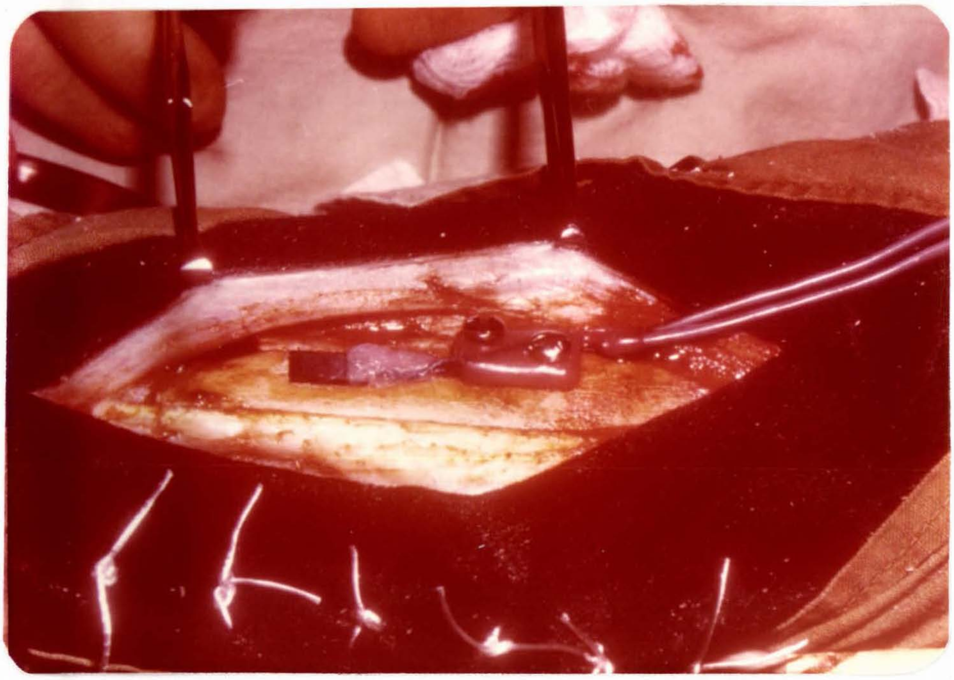
FIGURE 2.3-1

Photograph showing the strain gauges and flange bonded to the third metacarpal bone. The flange was slid down the guide pins which were placed in holes drilled in the bone.

FIGURE 2.3-2

Photograph showing the flange screwed to the bone to secure the installation.





least 5 minutes. After soaking, the strain gauges and flange were suspended by the unsoaked plugs on the lead out wires and lowered through previously sterilized 'Portex' autoclavable 2" nylon tubes. The surgeon assisted in feeding the gauges through the nylon and then secured the nylon end to the lead out cables with sterile autoclavable adhesive tape about 10 cm from the flange. The unsterile plugs were thus shielded by the nylon from the disinfected gauges and sterile drapes. Also immersed in 70% alcohol were the template and guide pins (Figure 2.3-1), used to assist in the screwing of the flange to the bone, (Figure 2.3-2).

Autoclaving had to be avoided since the temperature reached in this process is usually more than 120°C, which is greater than the 80°C maximum temperature specified for some of the strain gauges used. In addition the plastic coatings on the lead out cables probably would not have withstood the temperature of the autoclave.

2.3.2 The Operation

For an operation on the left foreleg, the horse was placed in right lateral recumbency and prepared for surgery.

An Esmarch bandage was applied to the leg to prevent blood from invading the operative site. A longitudinal incision about 7 cm long was made in the skin on the lateral aspect of the third metacarpal bone. The skin was retracted and the exposed periosteum was incised and reflected from the underlying metacarpal bone.

A template constructed from perspex was used to guide the hand drill in forming two holes in the lateral surface of the bone. The holes in the template matched those in the flange of the strain gauge assembly.

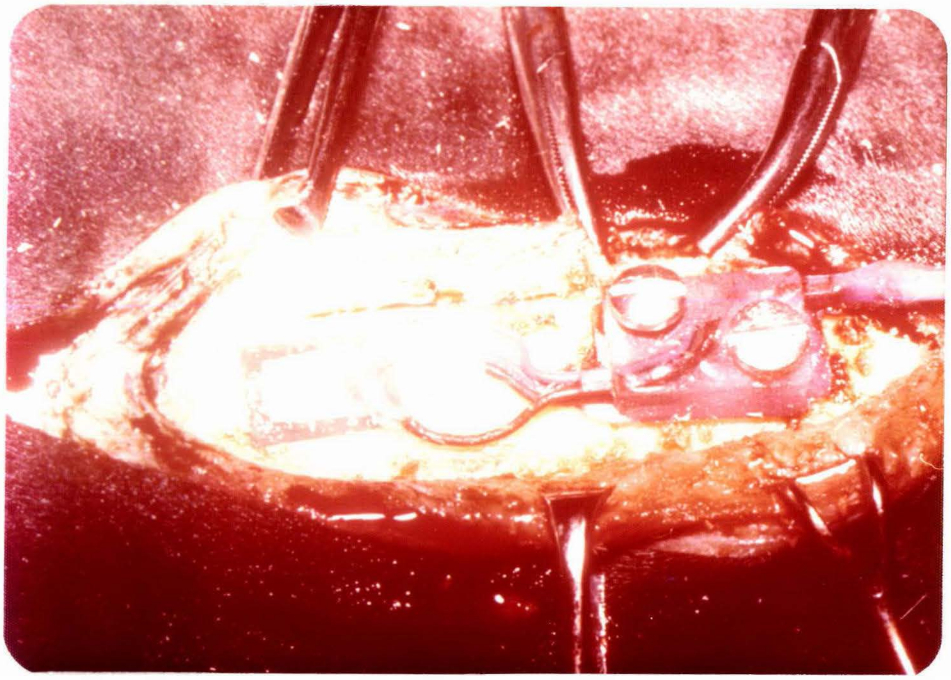
The bonding site was wiped and then swabbed with acetone followed by swabbing with a 1 : 1 mixture of ether and alcohol (Perry and Lissner, 1962). The flange was then screwed to the bone,

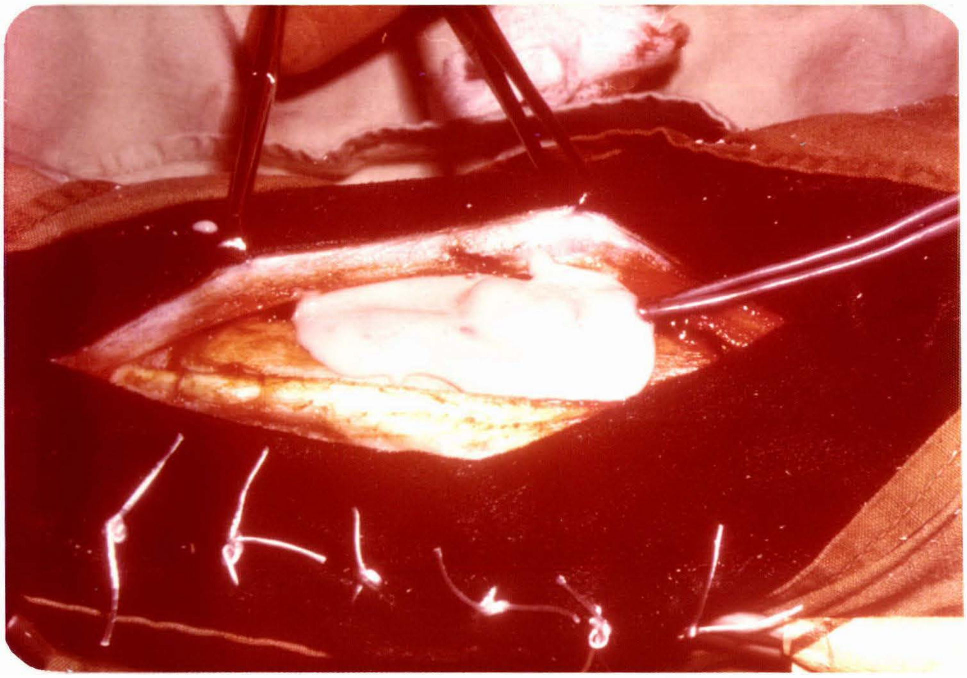
FIGURE 2.3-3

Close-up photograph of a strain gauge unit secured to a bone. In this early strain gauge unit the lead out wires were soldered directly to the gauges.

FIGURE 2.3-4

Photograph of a secured strain gauge unit coated with a protective layer of rapid curing silicone rubber.





the strain gauge wires being slightly bent so that the gauges were tilted away from the bone.

The bone beneath the strain gauges was swabbed again with acetone and with the 1 : 1 ether and alcohol mixture. As soon as this had evapourated, the bone and gauges were sprayed with surface activator (I.S. Activator A.C. Intercontinental Chemical Company Ltd, Dublin). When this dried the bond site was flooded with surgical cement (isobutyl 2 - cyanoacrylate monomer) and the gauges pressed swiftly to the bone. The gauges were held thus for two minutes while the contact cement set.

With the flange and gauges secured to the bone (Figures 2.3-2, 2.3-3), cold cure silicone rubber (Walker - Chemie GMBH, Muchen) was spread over the installation (Figure 2.3-4). The amount of catalyst used was such that the rubber set within 4 minutes.

To decrease the chance of infection, aqueous penicillin solution was instilled into the wound. The subcutaneous tissue was sutured, thus enveloping the installation in a thin flexible protective coating. The skin wound was then sutured, leaving an 'S' bend in each lead out cable under the skin. The lead cables were directed up the leg so that the plugs were positioned just below the carpal joint.

In order to protect the installation, especially during the horse's recovery from anaesthesia, a pressure bandage including a generous layer of cotton wool was applied, extending from immediately below the carpus to the coronet.

2.4 SURGERY FOR TENDON TENSION TRANSDUCER

2.4.1 Preparation

The frame, bridge and lead wires of the 'buckle transducer' were disinfected by immersion in 70% alcohol solution. Autoclavable nylon sheathing was positioned around the lead wires as discussed in Section 2.3.1.

2.4.2 The Operation

An incision approximately 10 cm long was made on the dorsal aspect of the left foreleg superficial to the common digital extensor tendon. The incision continued over the third metacarpal bone in such a manner that the buckle transducer and the bone strain installation could be positioned well apart.

With the lead cable end away from the bone strain site, the 'buckle transducer' was placed on the common digital extensor tendon. This was then drawn up and the bridge (Figure 4.2-1) slid across the frame under the tendon.

The bridge was located in position with two dowel pins, one each side of the transducer (Figure 4.2-1). The pins were formed from gauge 20, 0.89 mm diameter stainless steel suture wire. This was threaded through both bridge and frame, then bent over top and bottom to form a 'C' shape.

As soon as the 'buckle transducer' had been fitted to the tendon the device was calibrated as described in Section 4.5.

The implant was then washed with aqueous penicillin solution. A loop in each lead cable was left under the skin, the subcutaneous tissue sutured and the wound closed.

2.5 SURGERY FOR EMG ELECTRODES

Initially the wire electrodes were inserted while the horse was standing under the influence of xylazine, an hallucinogenic drug which, when administered intravenously at a dosage of 0.45 mg per kg, makes the animal drowsy and easy to manage for a period of about 20 minutes.

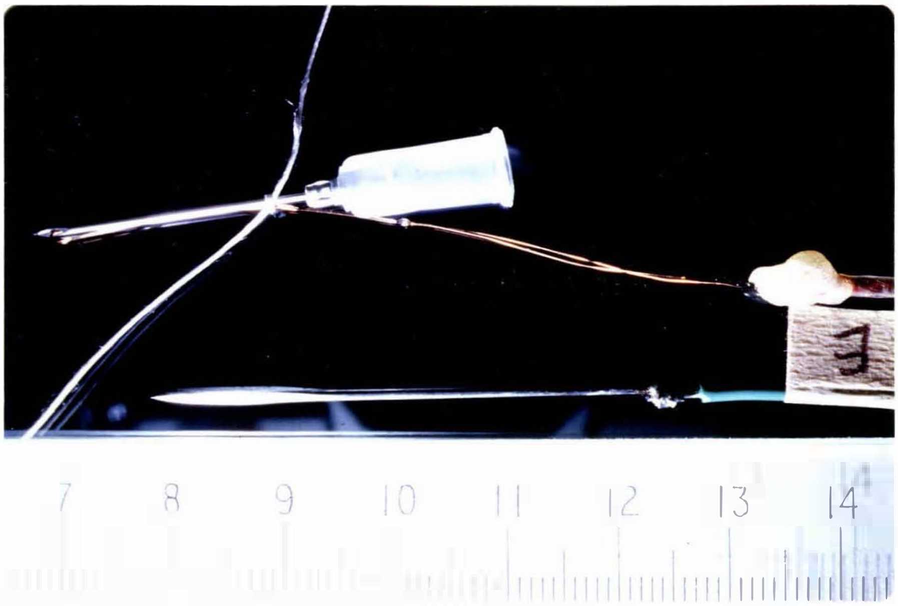
A local anaesthetic drug was injected subcutaneously in the operative area and a small incision made cranial and distal to the lateral tuberosity of the radius (Section 2.2). A small incision was then made in the fascia of the common digital extensor muscle where the fine wires were to be inserted.

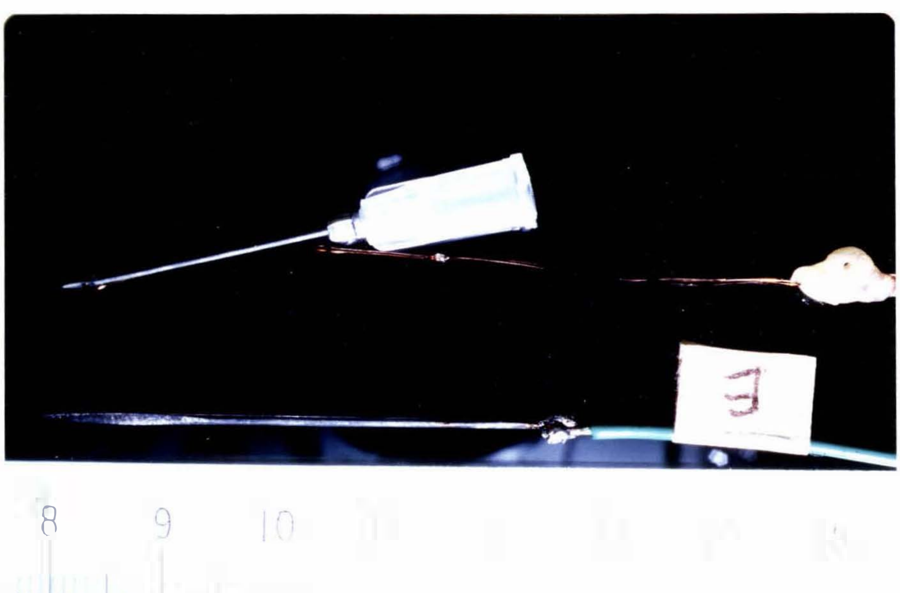
FIGURE 2.5-1

Photograph of fine wires and earth electrode (suture needle) ready for the surgeon to handle. A cotton thread temporarily holds the fine wires to the hypodermic needle. The twin core shielded cable and drilled epoxy resin moulding are shown at right. Numbers indicate centimeters.

FIGURE 2.5-2

Photograph of the earth electrode and fine wires ready for implantation after slipping the cotton thread off the hypodermic needle. Note the bead of epoxy resin binding the fine wires together about half way along their length. Numbers indicate centimeters.


7 8 9 10 11 12 13 14


8 9 10

The epoxy resin encasing the fine wire-cable joint was sutured subcutaneously to the proximal end of the skin incision. The cotton knot tying the fine wire to the hypodermic needle (Figure 2.5-1) was then removed and the needle with the fine wire hooked over its tapered end (Figure 2.5-2) inserted into the humeral head of the common digital extensor muscle. The hypodermic needle was then withdrawn while the exposed fine wires were lightly held, thus leaving the electrodes securely in the muscle. A few centimetres of cable were left coiled in a subcutaneous position and the skin wound sutured, except for a small opening at the distal part of the wound through which the cable emerged.

Chemical restraint of the animal with xylazine proved to be unsatisfactory in some horses because of twitching of the extensor muscle during the surgical procedures. Consequently this technique was abandoned in favour of placing the electrodes in the muscle while the horse was under general anaesthesia.

For the installation of muscle, bone and tendon transducers and calibration of the buckle transducer in the one operative session, two surgeons were used. Under these circumstances the operative period was effectively reduced and the muscular problems associated with prolonged recumbency in the horse were not encountered.

2.6 POST OPERATIVE RECOVERY

Some of the horses used in the initial studies experienced difficulty in getting to their feet as a result of post-anaesthetic forelimb lameness. This condition dissuaded each of these animals from putting weight on the foreleg on which it had been lying during the operation. In general this was not the leg on which surgery had been performed!

The exact cause of the lameness was not clear. Pearson and Gibbs (1970) reported "post-operative radial paralysis" as

FIGURE 2.7-1

Photograph of an equine larynx removed from the animal. The glottis, g, is the diamond shaped opening formed by the arytenoid cartilages (upper sides) and the two vocal folds (lower sides). The epiglottis is the leaf-shaped cartilage at the bottom of the photograph and forms the lower part of the laryngeal entrance.

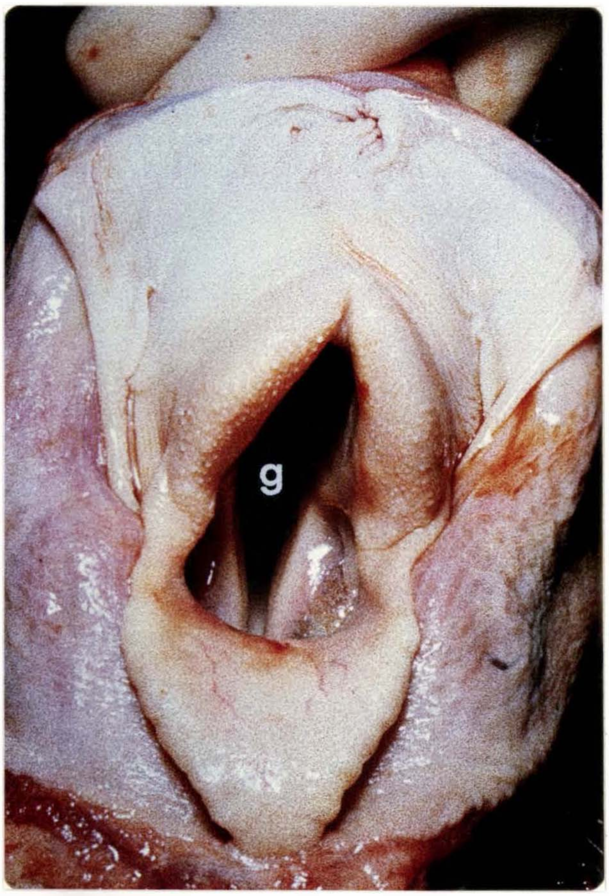
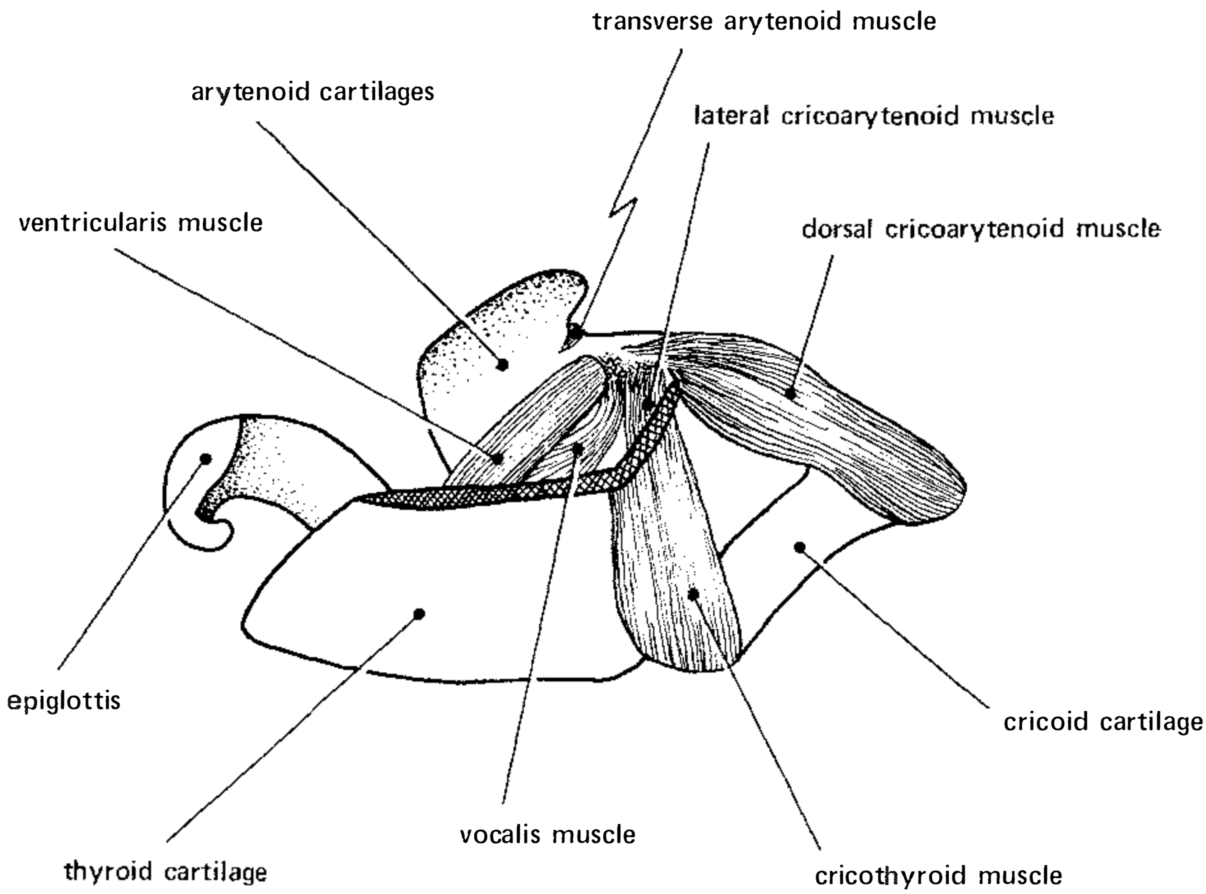

g

FIGURE 2.7-2

Diagram showing a lateral view of the cartilages and intrinsic musculature of the equine larynx. The upper part of the thyroid cartilage has been removed to reveal the ventricularis, vocalis and lateral cricoarytenoid muscles. The signet ring shaped cricoid cartilage is partially obscurred by the thyroid cartilage.


transverse arytenoid muscle
arytenoid cartilages
lateral cricoarytenoid muscle
ventricularis muscle
dorsal cricoarytenoid muscle
epiglottis
cricoid cartilage
thyroid cartilage
vocalis muscle
cricothyroid muscle

a hazard following laparotomy in lateral recumbency. Trim and Mason (1973) suggested that post-anaesthetic forelimb lameness in horses anaesthetized in lateral recumbency resulted from muscular ischaemia during the recumbent phase.

After the operation the horse was laid in the recovery stall on the side opposite to that laid on during surgery and subsequently lameness in the unmonitored forelimb was not encountered.

A consequence of the recovery stall recumbency of the horse was that the horse was lying on the operated leg and as the horse emerged from the anaesthetic, paddling of the forelegs subjected the installations and lead out cables to unknown stresses.

The implantation of the bone strain unit generally caused more lameness than that caused by the attachment of the tendon tension transducers of later design. Two days post operative recovery were usually required following bone surgery for the walking gait of the animal to become virtually normal. One day was required if only a tendon transducer was fitted.

The presence of the EMG electrodes did not appear to affect the gait of, or distress any horse. However a few horses would sometimes twitch the foreleg muscles.

2.7 LARYNGEAL ANATOMY

The larynx is a short, complex, valvular apparatus connecting the pharynx and trachea. This organ produces voice, prevents aspiration of foreign material and regulates air flow in respiration. The study of the equine larynx was concerned with the activity of the intrinsic laryngeal muscles and the corresponding movement of the laryngeal cartilages in relation to respiration.

The glottis (Figure 2.7-1) is the laryngeal sphincter formed by the vocal folds and the paired arytenoid cartilages (Figure 2.7-2). Rotation of these cartilages outwards abducts the vocal folds. The airway is restricted by the glottis during expiration and phonation.

The epiglottis (Figures 2.7-1 and 2.7-2) is shaped like a pointed ovate leaf and forms the lower part of the laryngeal entrance.

The signet ring shaped cricoid cartilage (Figure 2.7-2) articulates with the arytenoid cartilages.

The thyroid cartilage consists of two lateral laminae joined ventrally under the epiglottis. The laminae form a large part of the lateral walls of the larynx.

The intrinsic laryngeal muscles rotate the arytenoid cartilages at the cricoarytenoid joints. The dorsal cricoarytenoid muscle (Figure 2.7-2) rotates the arytenoid cartilages outward, causing abduction of the vocal folds and glottic dilation.

The opposite effect is achieved by the transverse arytenoid, lateral cricoarytenoid, ventricularis and vocalis muscles which cause glotticconstriction by rotating the arytenoid cartilages inward. The vocal folds are consequently adducted.

2.8 DISCUSSION

The animal chosen for this project was one of the largest of the domestic species. The size of the horse allowed the transducers to be relatively small, yet large enough to facilitate manufacture and handling.

The width of the buckle transducer was in part determined by the size of the strain gauges (Figure 4.2-1A). A large leg and tendon would therefore reduce the relative size of the transducer and hence the degree of interference the animal might experience. Similar arguments apply to the bone strain and EMG transducers.

In the horse the common digital extensor tendon is large and easily accessible as is the McIII bone. Furthermore, their simple shapes make for ease of transducer design and positioning.

The sites chosen for the attachment of the transducers were relatively distant from sources of possible interference. Had more difficult sites been chosen, it is likely that different transducers and implantation techniques would have had to be employed.

Isobutyl 2-cyanoacrylate monomer, the contact cement used to bond the strain gauges to bone, was first brought to our notice in a paper by Lanyon and Smith (1969). Their suppliers of the adhesive were unable to supply the substance as it was still under investigation in Britain and was not for sale. A similar ban on the sale of the adhesive was in force in the U.S.A. Supplies of the adhesive, marketed under the name of Coapt Surgical Adhesive, were ultimately obtained from Ethicon G.M.B.H., Germany through Ethnor Pty Ltd, Australia.

Similar materials had been known for a decade but were histotoxic (Woodward et al.,1965). We are not aware of the results of investigations into isobutyl 2 - cyanoacrylate monomer, but decided to use this adhesive following Lanyon and Smith (1969).

In Section 2.3.2 it was stated that the epoxy resin flange of the strain gauge mounting unit was screwed to the bone prior to bonding the gauges to the bone. This is the reverse of the order described by Barnes and Pinder (1974) - appendix II. It was subsequently felt that a safer procedure was to screw the flange to the bone first.

The silicone rubber coating was applied to the strain gauge mounting unit to help protect the strain gauges from damage should the horse strike its leg against some object. However, since this silicone rubber does not adhere very well to the bone, a force on the silicone rubber, tangential to the bone, might be readily transmitted to the strain gauges. This situation could arise during the recovery of the horse from anaesthesia, when the horse may 'paddle' with its forelegs while in lateral recumbency. The horse was usually placed in the recovery stall on the same side as that of the leg which had undergone surgery. When paddling occurs, the bandages and hence the skin are subjected to a torque which could cause a tangential force to be transmitted to the strain gauges. The repeated application of such a force might lead to failure of the insulation of the gauges. The advisability of using the quick setting silicone rubber during surgery is therefore questionable.

In the early tendon tension transducers, the ends of the dowel pins were twisted together (Figures 4.3-2, 4.3-4) to ensure their security. Lameness associated with these early transducers was possibly caused by the dowel pin ends catching some subcutaneous tissue. Lameness connected with the implantation of the bone strain implantation was probably caused by soreness associated with scraping of the periosteum from the bone surface.

CHAPTER 3

BONE STRAIN

3.1 INTRODUCTION

The object of the equine bone strain study was the correlation of strain in the Mc III bone with tendon tension and with phases of the stride while the horse was walking. This gait was chosen because it presented fewer instrumentation difficulties than the faster gaits.

The particular bone chosen is of interest since it is known to experience longitudinal fractures. These extend into the articulation at its distal end (Adams, 1966). This bone is subjected to great forces especially during a full gallop. Unevenness of terrain can produce uneven pressures in the joint which may cause a portion of the Mc III bone to split away.

3.2 BONE STRAIN TRANSDUCER AND CEMENT

The electric resistance strain gauge is a widely used transducer of strain, it consists of a thin conductor which is strained through being bonded directly or indirectly to the surface under investigation. The stretching or compressing of the thin conductor alters its length and cross-sectional area and hence its resistance which is detected with appropriate circuitry (Section 3.4).

Longitudinal strains on the lateral aspect of the Mc III bone and occasionally on the medial aspect were monitored. Gauges were bonded to the bone in pairs, both for comparison purposes, and in case one gauge failed. Each pair consisted of both a large and a small strain gauge. The small gauges had a gauge length of 1 mm and gauge widths of 2 mm (Fb-1, Showa Measuring Instruments Co., Ltd, Tokyo, Japan), or 1 mm (KFC-1-C1, Kyowa Electronic Instruments Co., Ltd, Tokyo, Japan). The large strain gauges had a gauge length of 5 mm and a gauge width of 4.5 mm (KF-5-C1, Kyowa, Japan).

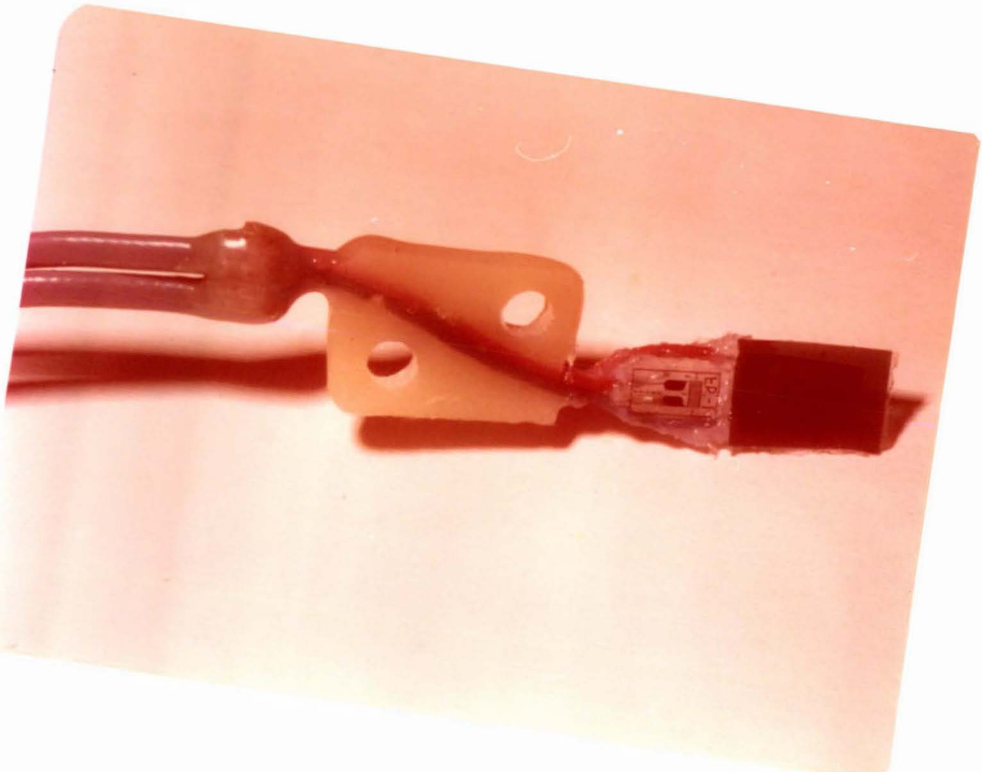
The temperature coefficient of the gauges was not deemed to be of major consequence, since the gauges were used inside the body of the animal, and so were in an approximately constant

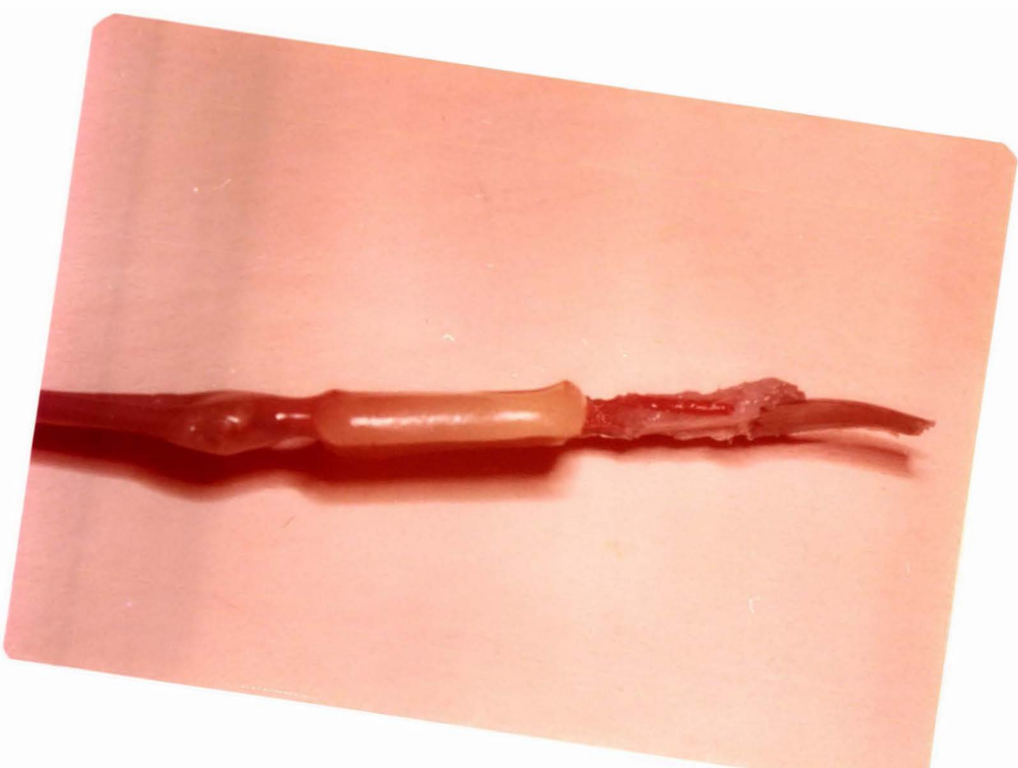
FIGURE 3.3-1

Photograph of the underneath of a strain gauge mounting unit. Note the beads of solder on the wires embedded in the epoxy resin flange. Silicone rubber covers the upper surface of each gauge and the soldered joints. The green gauge backing is 11 mm in length.

FIGURE 3.3-2

Photograph showing a side view of the strain gauge unit pictured in Figure 3.3-1 above.





temperature environment. Furthermore, any slight variation in temperature would be very slow compared with the variation in bone strain during several strides. Gauges designed for steel and aluminium having approximate temperature coefficients of 11 and 23 p.p.m. / ^{o}C respectively were used.

The cement used to bond the strain gauges to bone was a contact cement, isobutyl 2 - cyanoacrylate monomer, after Lanyon and Smith (1970). The cement was marketed under the name of 'Coapt Surgical Adhesive' (Ethicon G.M.B.H., Hamburg).

Application of pressure to a thin film of the adhesive was sufficient to harden the cement. Polymerization was accelerated by spraying the surfaces, with a catalyst, 'I.S. activator A C' (Intercontinental Chemical Co., Ltd, Dublin).

3.3 STRAIN GAUGE MOUNTING UNIT

A bone strain transducer unit consisted of both a large and a small foil electric strain gauge, each soldered to separate twin core shielded cables, (Figures 3.3-1, 3.3-2). The cores of the cables passed through an epoxy resin ('Araldite', Ciba Ltd, Switzerland) moulding designed to be screwed to the bone.

Kyowa foil strain gauges, types KFC-1-C1-11, gauge length 1 mm, and KF-5-C1-23, gauge length 5 mm were coated on their upper surface with one part silicone rubber (Silastic 732 RTV, Dow Corning Corp., U.S.A.). This adhesive/sealant protected and insulated the foil from contact with moisture once the unit was implanted in the animal.

A strip of adhesive tape was temporarily placed over the silicone rubber while it set, holding each gauge down and forming a mould for the rubber. The lead wires of each gauge were bent up and over the tape in a 'U' so that the two cores of the cable could be soldered to the strain gauge wires forming two joints above each gauge.

The shielding and two cores of the cable were embedded in an epoxy resin moulding to avoid tension in the lead out cable being transmitted to the strain gauges. Prior to this embedding, a small section of the insulation of each core was removed and the exposed wire enveloped with a whisp of solder. Once embedded, these whisps prevented the lead out wires from sliding through their insulation and disturbing the strain gauge.

The mould was formed by fashioning a rectangular hole in a small sheet of perspex that had been heated and bent to conform with the surface of a short plastic pipe of suitable curvature. Grooves were cut in the concave surface of the perspex sheet to accommodate the cable cores during the moulding operation.

A small 'S' bend was made in the two centimetres of each core which were allowed to extend from the mould for soldering to the strain gauge wires. Once good soldering joints had been confirmed, they were coated with silicone rubber adhesive sealant. When this had set, this insulation was checked by dipping the joints in a beaker of water and checking resistances between the lead out cable and an electrode in the water.

3.4 MEASURING AND RECORDING APPARATUS

Each strain gauge bonded to the bone could be individually connected to a three element strain gauge rosette, so forming a Wheatstone bridge configuration.

Two arrangements were used to allow balancing of the bridge. The first used a plug-in module housing a 500 k ohm trim potentiometer which was split into two 250 k ohm sections. The wiper of the potentiometer could place a resistance of up to 250 k ohm in parallel with either the active strain gauge or an inactive one. The module plugged in to the preamplifier box which also housed the inactive arms of the Wheatstone bridge. This arrangement worked well until the wiper ceased to make reliable contact with the conductive wafer.

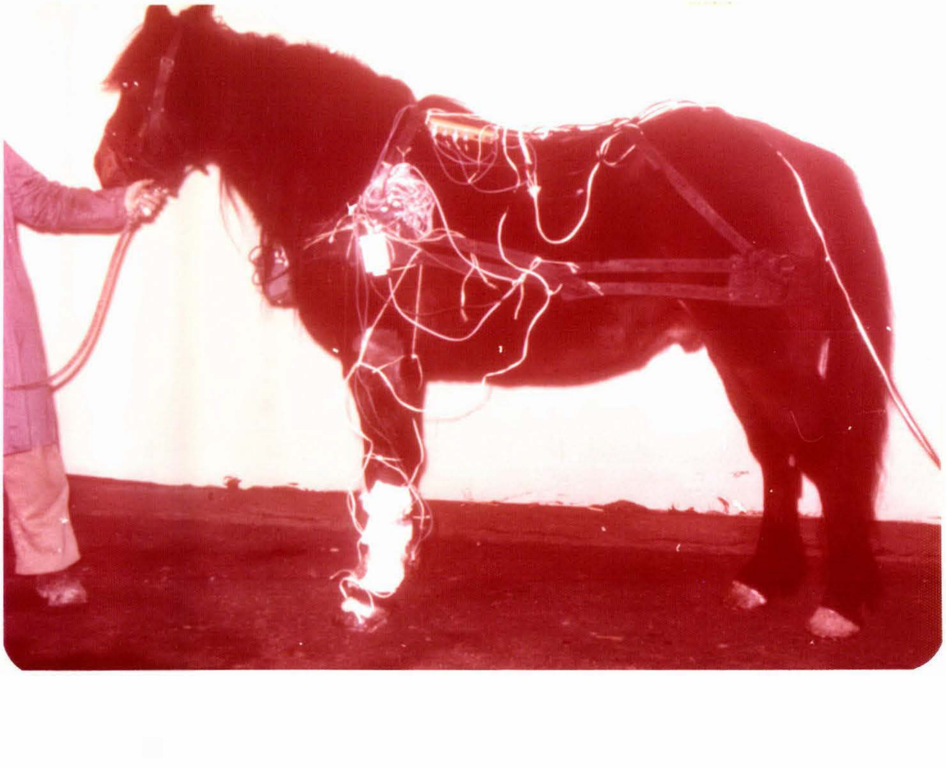
A more rugged, sealed, 50 k ohm potentiometer was used in the subsequent arrangement. Since the conductive wafer was inaccessible, the potentiometer was placed across two adjacent arms.

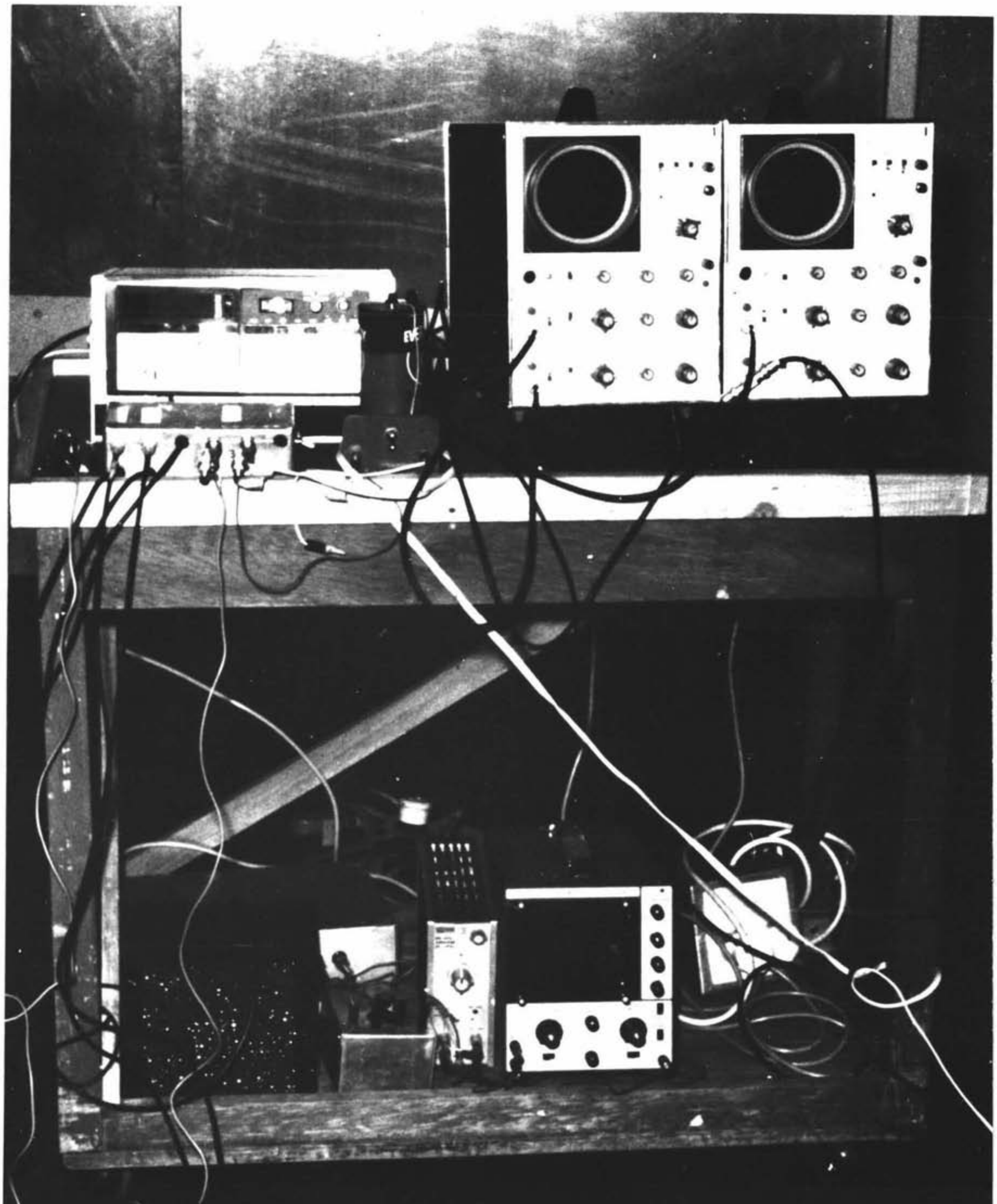
FIGURE 3.4-1

Close-up photograph of equine foreleg with foot-switch and pre-amplifiers attached.

FIGURE 3.4-2

Photograph of a harnessed and instrumented experimental horse.





Their junction was connected to the wiper of the potentiometer which was housed within the preamplifier box.

The d.c. power supply for the bridge was provided by a 2.8 V Hg battery.

The single stage preamplifier consisted of an operational amplifier having a gain of 400.

Because the preamplifier was to be positioned on the limb near the active strain gauge (Figure 3.4-1), size and weight of the preamplifier were important considerations. A heavy preamplifier assembly might have affected the gait of the horse and would more easily work loose. Thus the Hg batteries for the Wheatstone bridge and preamplifier power supplies were housed in a combined power pack and junction box on the horse's back (Figure 3.4-2). Five core shielded cable linked this box and the match-box sized aluminium preamplifier box.

Fifty yards of coaxial cable linked the junction box and the amplifier unit (Figure 3.4-3) placed near the frequency modulated tape recorder (Philips analog cassette recorder, Mini-log 4). The unit had a gain of 18.3 to make its output well within the input voltage requirements of the tape recorder.

A vocal commentary was superimposed on the gait data because the speech channel for the tape recorder was unavailable at that time.

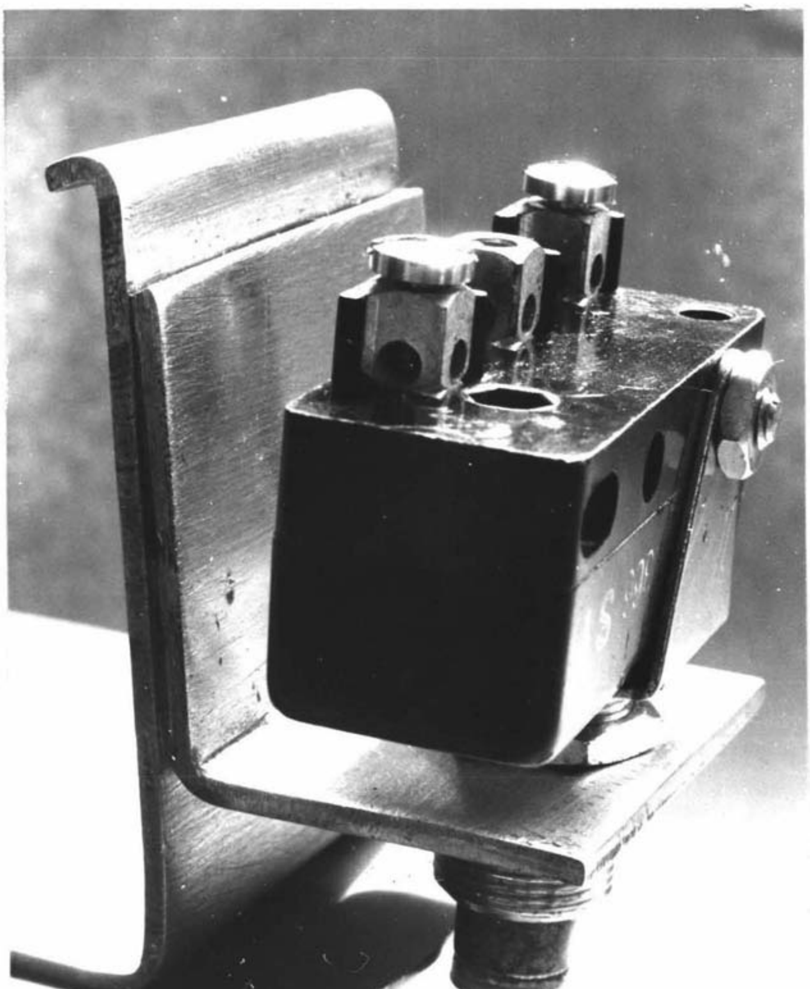
Initial recordings were made directly on heat sensitive paper by a multi-channel chart recorder, (type M.8, Devices, Gt. Britain). A modified stereo tape deck recorded either bone strain or tendon tension data in a frequency modulated form (Appendix I).

3.5 CALIBRATION

Two methods were used to calibrate the bone strain recording system. For a given fractional change in resistance of the active gauge, the output voltage of the system

FIGURE 3.6-1

Photograph of the type of foot switch used initially. The base plate and bracket were bound to the hoof and the industrial microswitch was then screwed to the bracket. The plunger is shown overhanging the curved edged of a surface representing ground level. The bracket protruded 26 mm from the base plate.



was calculated knowing the Wheatstone bridge supply voltage and the amplifier gain. The known gauge factor was then used to relate this output voltage to strain.

A more direct method was subsequently employed. A laboratory resistance box was set to 120 ohm (the nominal active gauge resistance) and connected to the recording system in place of the active gauge. Another (megohm) resistance box was periodically placed in parallel with the first. The fractional resistance change on making the parallel connection caused an electrical output which was recorded. This recorded signal corresponded to some strain, related to the fractional resistance change through the gauge factor.

3.6 GAIT SENSOR

Two methods of sensing gait were used in this study, cinematography and the use of a foot switch. The former was useful for monitoring the animal's four legs simultaneously, as well as for observing the movements of other body parts, especially the head and the neck. Since the main purpose of the cinematographic record was to aid the interpretation of other biophysical data, some way of identifying corresponding parts of each was necessary. The method adopted was to mount a light emitting diode on the camera just within its field of view. A foot switch, described below, operated both this diode and an event marker on the device recording bone strain or tendon tension. Extra marks could be generated by a manual marker button to assist in matching the records.

A foot switch attached to the lateral aspect of the hoof indicated the swing and support phases of the corresponding leg.

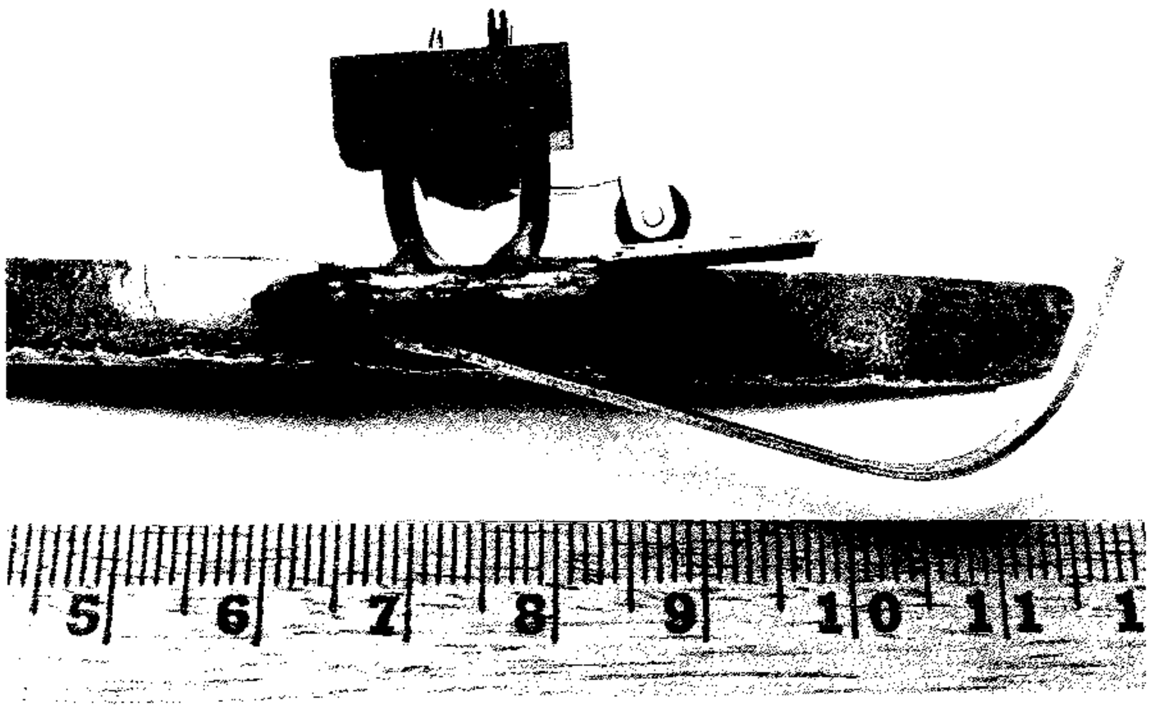
The initial design employed a heavy duty industrial microswitch as sensor (Figure 3.6-1). This had a sturdy plunger needing minimal movement to operate the switch, but also had a movement absorbing capability of 0.5 cm. Thus the switch operated immediately the plunger contacted the ground.

FIGURE 3.6-2

Photograph of a foot switch in the 'swing phase' position as would occur when the shoe was off the ground. The lever hangs below the shoe. Note the water-proofed flexible cap under the microswitch to protect it from dirt and water.

FIGURE 3.6-3

Photograph showing a foot switch in the 'support phase' position as would occur when the foot was on the ground. The lever has been rotated upwards to operate the arm of the microswitch.



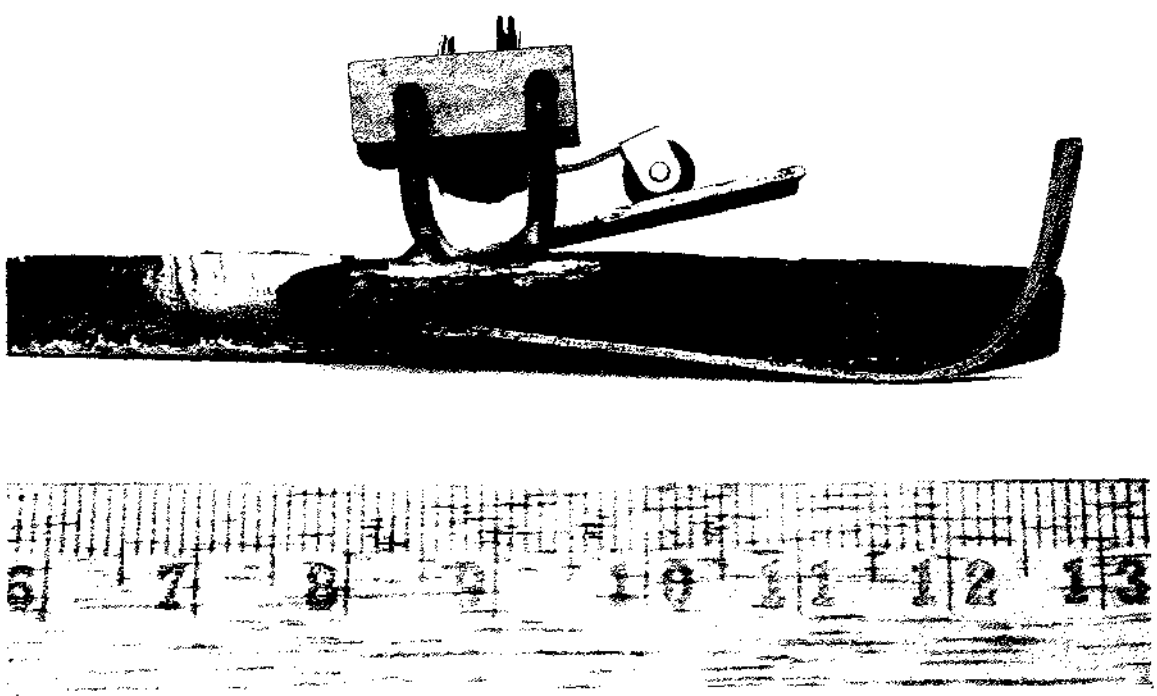
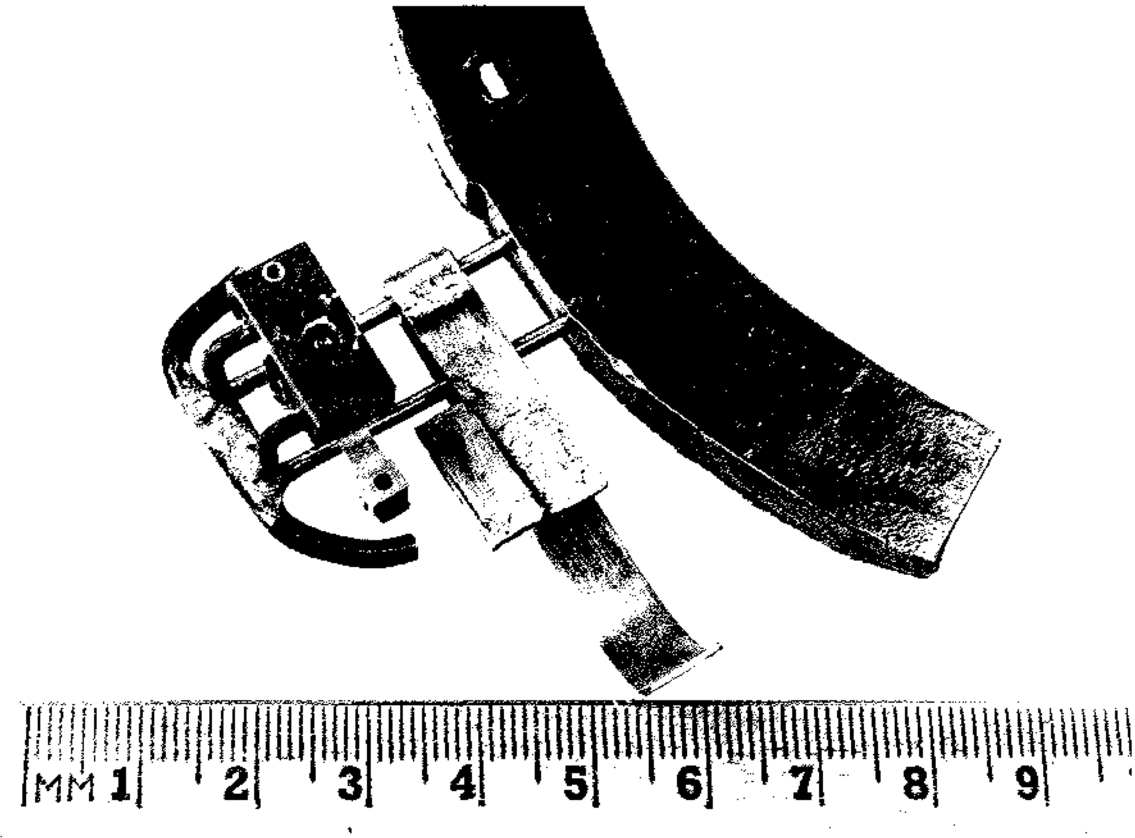


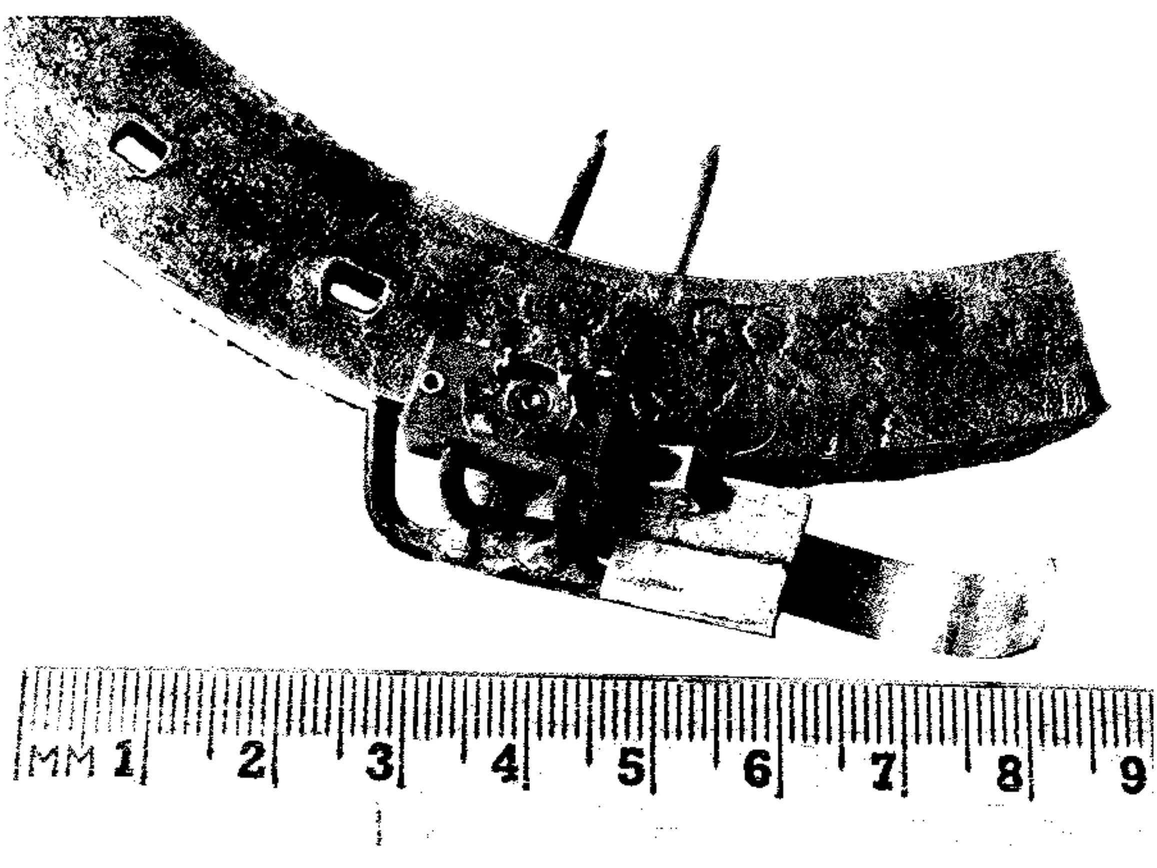
FIGURE 3.6-4

Photograph showing the components of the foot switch type finally adopted. To the right is a drilled horse shoe for the left hoof.

FIGURE 3.6-5

Photograph of an assembled foot switch. In practice the microswitch was wired before mounting.


MM 1 2 3 4 5 6 7 8 9


MM 1 2 3 4 5 6 7 8 9

Crushing of the switch was obviated by the play in the plunger. Vertical adjustment of the switch was necessary to ensure that the plunger contacted the ground just before the hoof. There also had to be movement of the plunger following its ground contact sufficient to ensure reliable operation of the switch even when the animal was walking on a slightly uneven surface.

The microswitch was mounted on a bracket attached to the lateral aspect of the hoof (Figure 3.6-1). The base plate of the bracket extended 2 cm under the hoof and was bound to it with twine and adhesive bandage. The metal under the hoof measured 2 cm x 4 cm x 1.6 mm. However the gait of the horse was not observed to be altered by the strip. Unfortunately, walking the horse tended to loosen the binding which bound bracket to hoof.

The design subsequently adopted did not allow the small microswitch to touch the ground. A lever (Figure 3.6-2), held down during the swing phase by the arm of the microswitch, was rotated 'upwards' on touching the ground (Figure 3.6-3). The upper part of the lever pushed up on the roller of the microswitch arm, causing the plunger to be retracted and the switch operated. The delicate microswitch and arm were thus protected from the rigours of ground contact.

The microswitch and lever were held in position by a frame (Figure 3.6-4) having two prongs which friction fitted into a pair of holes drilled in the horse shoe (Figure 3.6-5). The first horse shoes so drilled also included a hard rubber strip attached to the inner side of each shoe. This strip provided extra friction which was found to be unnecessary. Very adequate friction fit was also achieved by simply inserting the two prongs between the shoe and the hoof, thus dispensing with the need to drill holes in the shoe.

The second foot switch was considerably more reliable than the first version, no problems of the installation working loose being encountered.

FIGURE 3.7-1

Bone strain and foot switch data recorded from the right foreleg of a 590 kg horse standing 1.7 m .

(a) Trace from foot switch on the monitored foreleg, where the durations of the support phase and swing phase are indicated by Td and Tu respectively.

(b) Strain on the lateral aspect of the third metacarpal bone. The interval AA is redrawn in Figure 3.7-2.

(c) Seconds marks.

(d) Strain on the medial aspect of the third metacarpal bone.

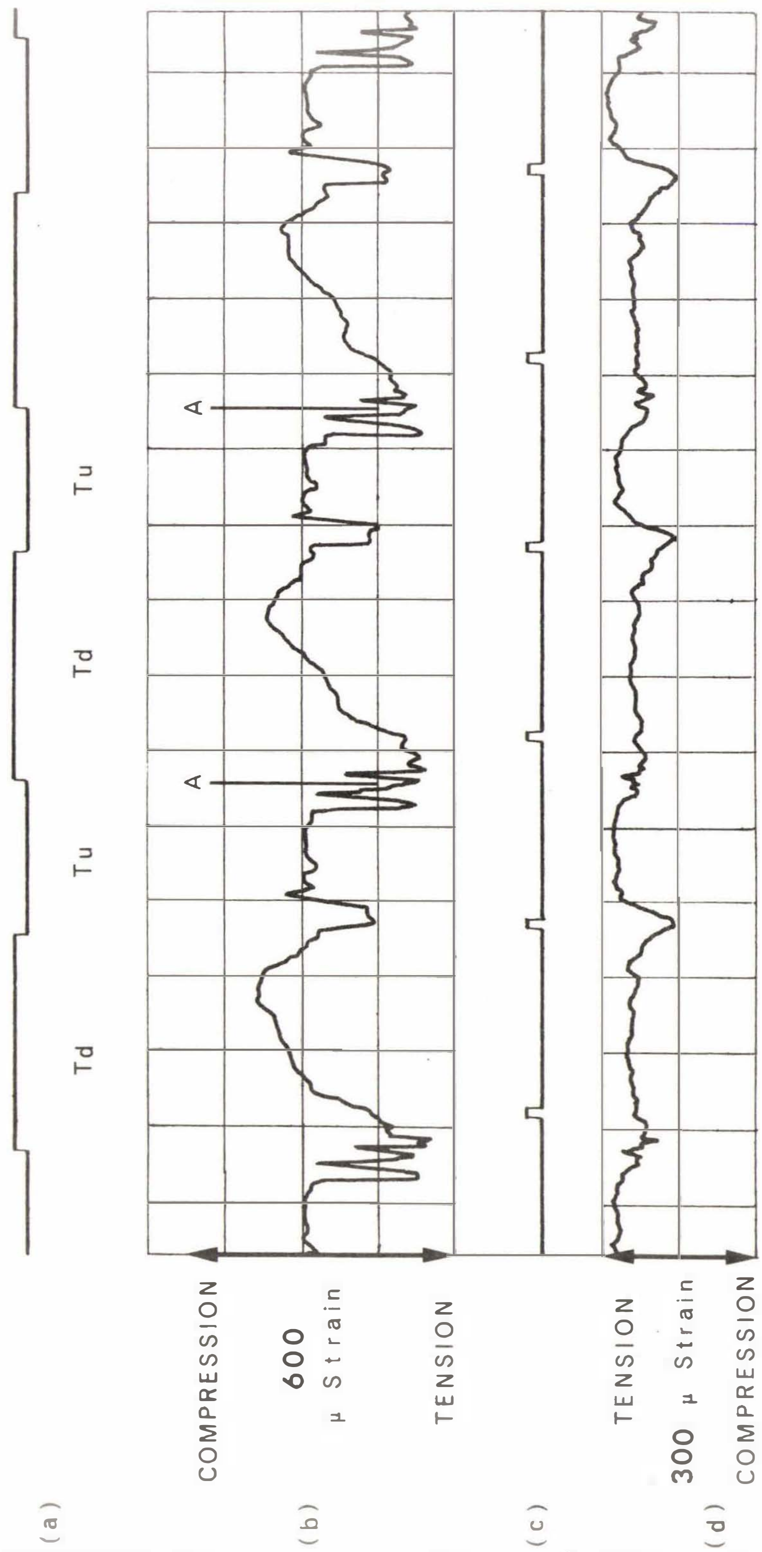

(a)
Td
Tu
Td
Tu
(b)
COMPRESSION
600
μ Strain
TENSION
A
A
(c)
(d)
TENSION
300 μ Strain
COMPRESSION

FIGURE 3.7-2

Section AA of Figure 3.7-1 labelled and freely drawn.

A Indicates hoof - ground contact, that is the end of the swing phase, Tu, of the monitored foreleg and the start of its support phase Td.

B Marks the lateral compression spike shortly after hoof - ground contact.

C Is an estimate of the start of the swing phase of the contralateral foreleg.

D Is a region of increasing compression.

E Corresponds to the maximum of compression on the lateral aspect of the Mc III bone.

F Is an estimate of the end of the swing phase of the contralateral foreleg.

G Indicates the end of the support phase of the monitored foreleg and the beginning of its swing phase.

H Indicates the rapid increase in lateral bone tension in the early moments of the swing phase.

I, J and K: Indicate a triplet of compression peaks superimposed on varying backgrounds.

L Corresponds to another rapid increase in lateral tension.

M Marks the lateral compression spike shortly before hoof - ground contact.

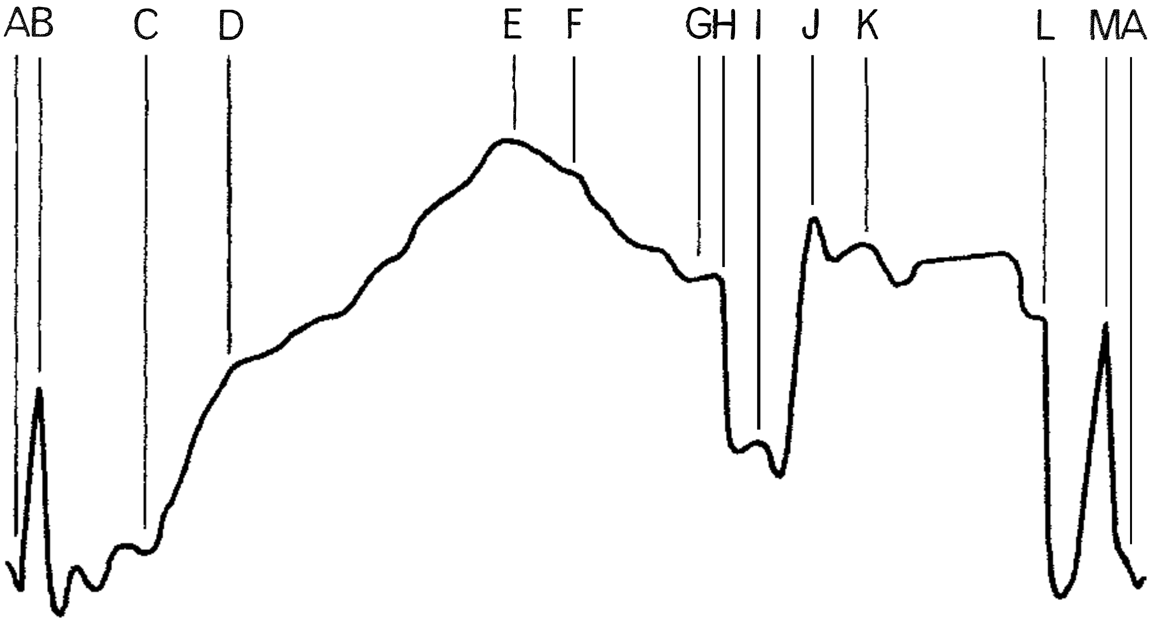

AB
C
D
E
F
GH
I
J
K
L
MA

FIGURE 3.7-3

Data recorded simultaneously from the left foreleg of a 390 kg mare standing 1.57 m.

A: Bone strain on the lateral surface of the third metacarpal bone. Increasing compression is in the positive y direction.

B: Tension in the common digital extensor tendon. Increasing tension is in the negative y direction.

C: Seconds marks generated by the chart recorder.

D: Trace from foot switch on each foreleg. L and R indicate the swing phases of the monitored (left) foreleg and right foreleg respectively.

E: Averaged rectified EMG signal. The electromyogram was recorded from the humeral head of the common digital extensor muscle.

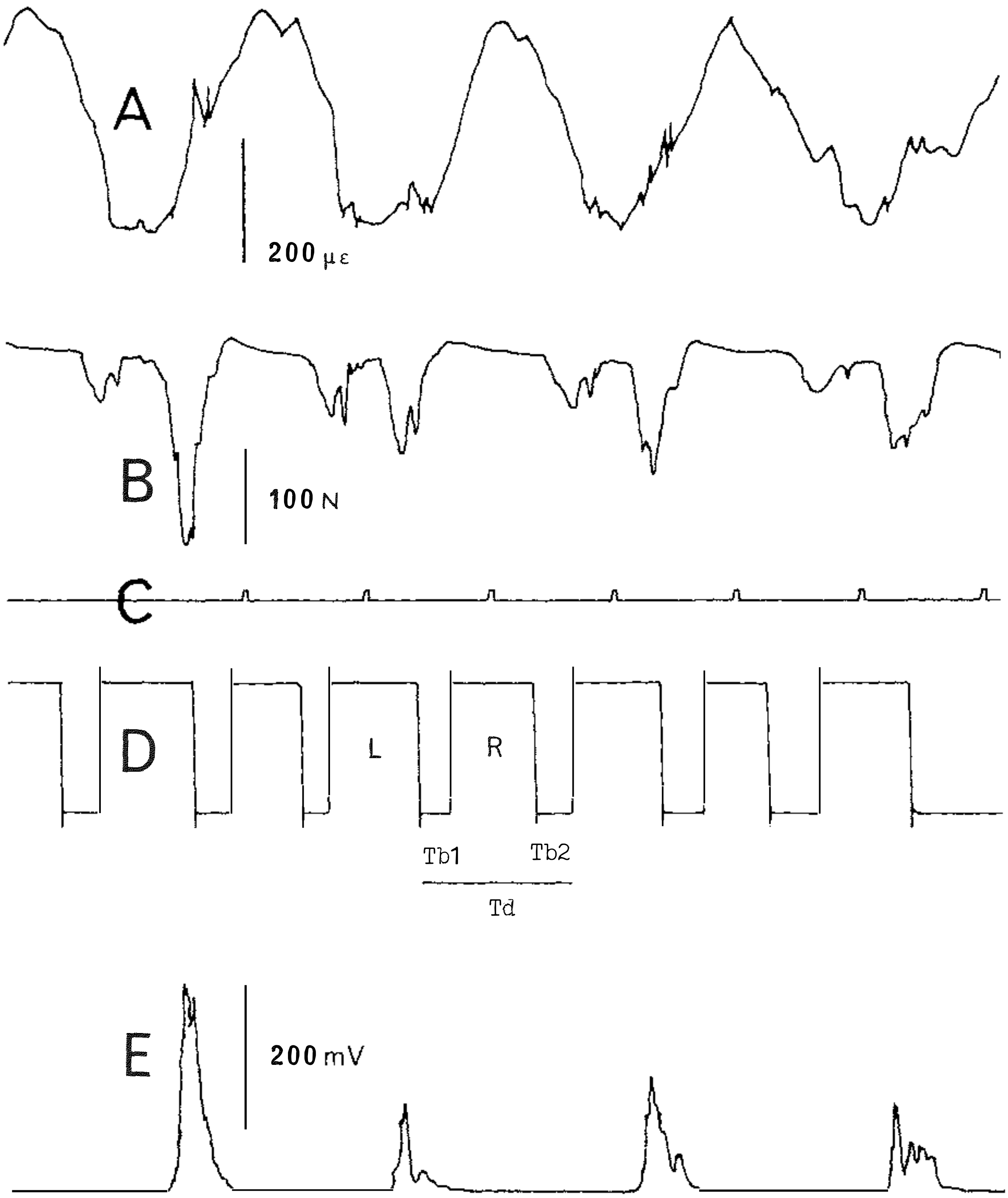

A
200 με
B
100 N
C
D
L
R
Tb1
Tb2
Td
E
200 mV

FIGURE 3.7-4

Original recording of data presented in Figure 3.7-3 but with different scales. Traces A, B, C and D of Figure 3.7-4 correspond respectively to traces D, A, C and B of Figure 3.7-3.

The frequency response of the chart recorder was zero to 120 Hz. Note that Figure 3.7-4A includes superimposed speech signals which were omitted from Figure 3.7-3D.

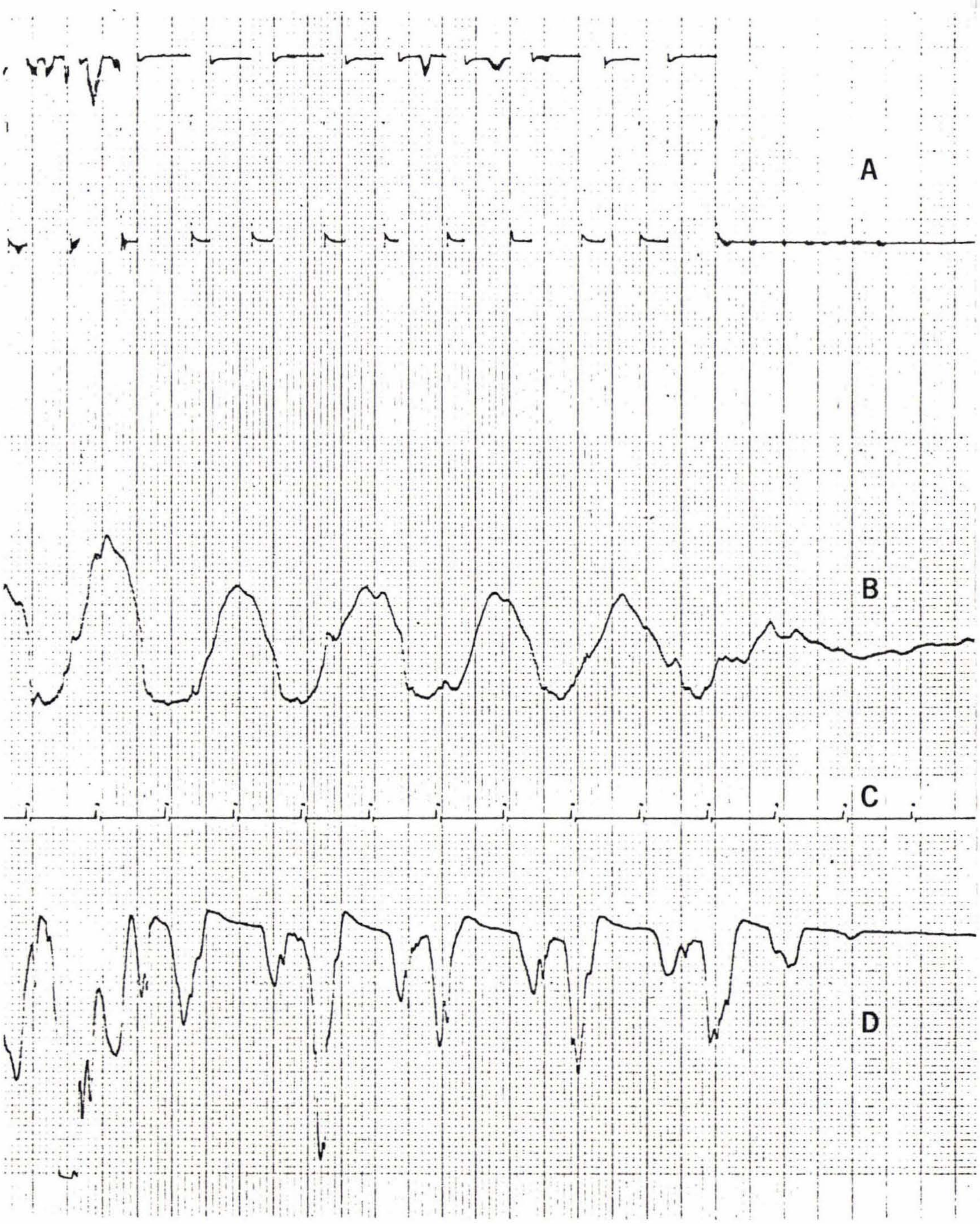
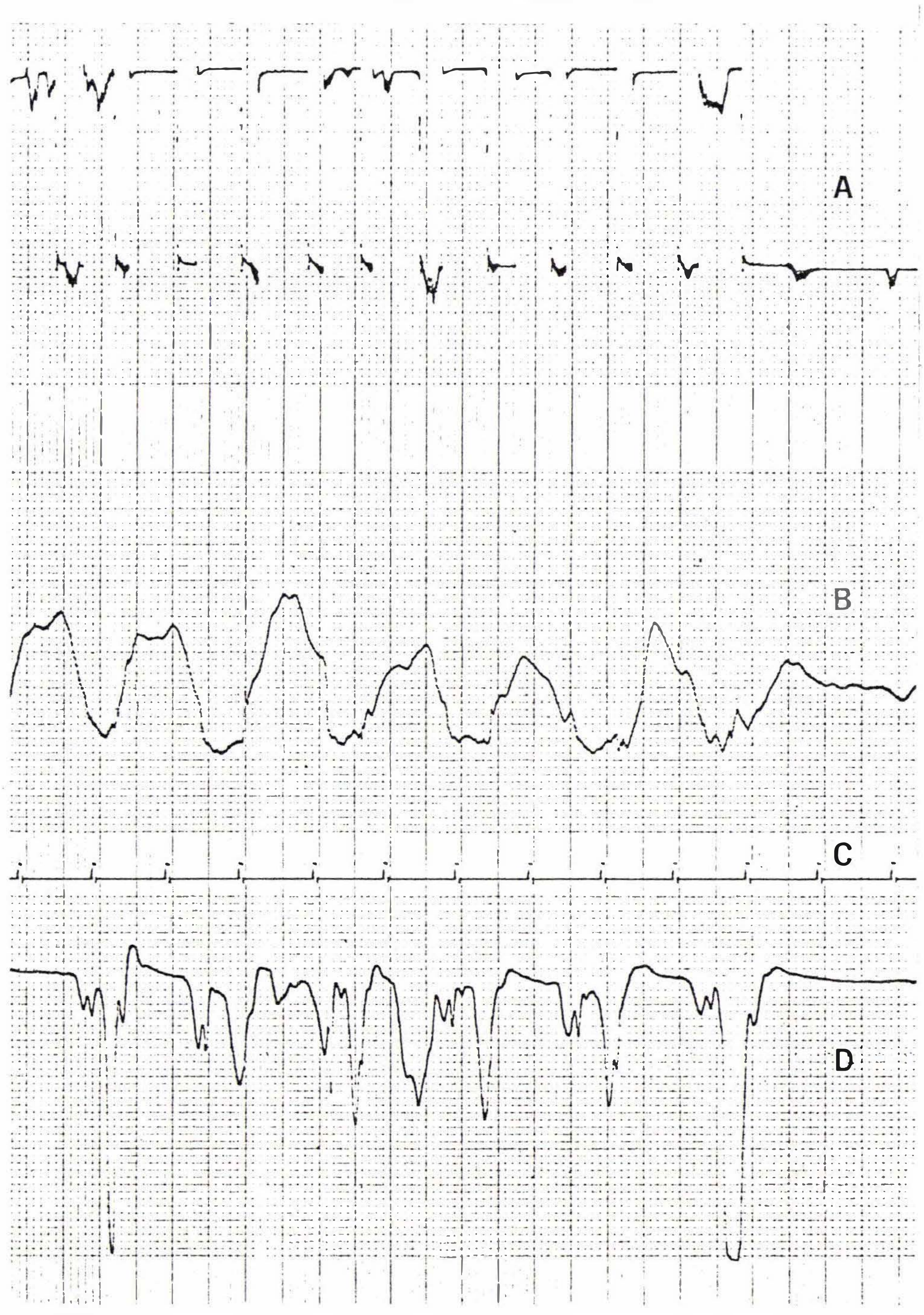

A
B
C
D

FIGURE 3.7-5

Original data recorded from the same animal and in the same way as Figure 3.7-4.


A
B
C
D

3.7 BONE STRAIN RESULTS

Figures 3.7-1 and II-5 show redrawn bone strain recordings from a 590 kg horse which stood 1.7 m high. Trace (b) of each figure was recorded from the lateral aspect of the Mc III bone in the right foreleg. Trace (c) of Figure II-5 and trace (d) of Figure 3.7-1 were from the medial aspect of this bone. In all cases, the recorded strain was in a direction parallel to the long axis of the bone.

The first trace of each figure was from a foot switch on the right foreleg and indicates the support phase Td and swing phase Tu of this leg. Time marks in seconds are displayed between the bone strain traces.

To aid later discussion, section A A of Figure 3.7-1 has been freely redrawn and labelled as Figure 3.7-2.

Strain in the left Mc III bone, lateral aspect, is shown in Figure 3.7-3A. As in Figure 3.7-1(b), compression is in the positive 'y' direction and tension in the negative 'y' direction. The phases of the bone strain recording may be determined by correlation with the foot switch trace, Figure 3.7-3D. Here L denotes the swing phase of the left foreleg and R denotes the swing phase of the right foreleg.

Correlation between bone strain and tendon tension is a major consideration and is discussed in Section 4.7.2 with special reference to Figures 3.7-3A and 3.7-3B.

Figure 3.7-4 is a photocopy of original data, part of which is redrawn as Figure 3.7-3. Figures 3.7-3A, B, C and D correspond with Figures 3.7-4 B, D, C and A respectively, but the scales are not the same.

Figures 3.7-4 and 3.7-5 have identical scales and represent similar data from the same animal.

3.8 DISCUSSION

Transducers were implanted in the left foreleg of most of the experimental horses, but some had their right foreleg operated on instead. It is therefore convenient during the discussion to refer to the monitored foreleg and the unmonitored foreleg, the former being host to the implanted transducer. The contralateral foreleg is synonymous with the unmonitored foreleg. It should be noted that although the term unmonitored is used, the leg thus described had, in general, its swing and support phases determined by a foot switch.

Bone strain in the walking mammal is engendered by a complex set of causes. These involve the weight of the animal, its style of gait, the nature of the ground, the action of muscle - tendon and ligament systems, and the shape and viscoelastic properties of the bone itself. All these factors influence the recorded bone strain, so it is difficult to unambiguously ascribe features of the bone strain recordings to specific causes.

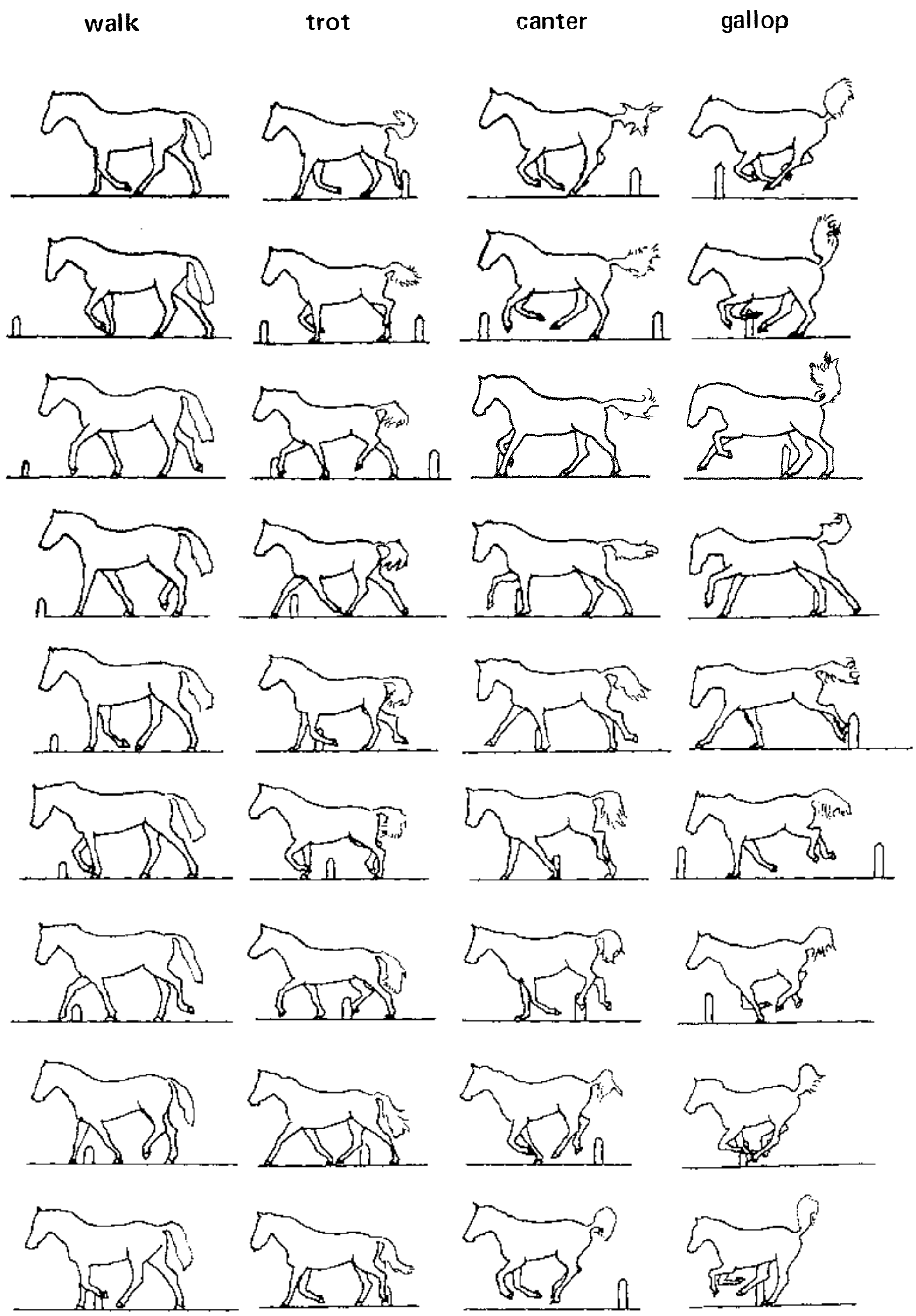
The approach used in this study was to monitor three of the factors likely to produce a detectable effect on the bone strain recording. These factors were the swing and support phases of each foreleg, and common digital extensor tendon tension of the monitored leg. The most obvious of the three factors was foot - ground contact of the monitored foreleg.

Although legs are primarily for propulsion, they must nevertheless, for most mammals, also support body weight. This is certainly so for the horse, whose forelegs carry about 60% of the body weight (Bjorck, 1958). During the support phase of the monitored leg, the contralateral leg is lifted from the ground, thereby placing about 60% of body weight on the monitored foreleg alone. The transfer of this extra weight is expected to be reflected in the bone strain recording. Thus the swing and support phases of the 'unmonitored leg' were recorded.

This transfer of weight need not be abrupt and could occur gradually during the time Tb that both forefeet are on the ground. However this period (Tb in Figure 4.6-1) can be relatively brief, being about 0.4s for a slow walk (Figure 4.6-1, first stride)

FIGURE 3.8-1

Drawings of four types of equine gait. The duration of each sequence is : walk, 1 s; trot, 0.7 s; canter, 0.5 s; and gallop, 0.4 s.


walk
trot
canter
gallop

and about 0.2s for a faster walk (Figure 4.6-1, last stride). The Tb period of Figure 3.7-3D gradually increased from 0.2s to 0.4s (and then became indefinite) as the animal slowed to a halt. By comparison, a galloping horse has both its forefeet touching the ground for about 0.05 sec in each cycle of this gait. (Figure 3.8-1).

In each period Td of the support phase of the monitored foreleg, there are two periods in which both forefeet are on the ground. The first (Tb1) occurs at the beginning of Td and the second (Tb2) at the end (Figure 3.7-3).

Peak compressive strain (labelled E, Figure 3.7-2) occurs on the lateral aspect of the Mc III bone during Td (Figures 3.7-1 and 3.7-3). Furthermore, this strain occurs during the swing phase of the unmonitored foreleg (labelled R in Figure 3.7-3). This is to be expected, especially if the bone bends in compression so that the lateral surface tends to become concave.

There is a steady decrease in lateral bone compression during the time interval Tb2 (interval F G, Figure 3.7-2) when weight is being transferred from the monitored foreleg to the unmonitored one.

A sharp decrease in compressive strain (or increase in tension) (H, Figure 3.7-2) is evident immediately following the lifting of the monitored leg (G, Figure 3.7-2). This sharp decrease ends abruptly, often with rapid variations in bone strain (I, J, K, Figure 3.7-2) presumably due to tendon tension.

In contrast to the steady decline in compression during the time Tb2, there is usually a great deal of fluctuation in bone strain during the time interval Tb1 (interval A C, Figure 3.7-2). These fluctuations start just before Tb1 begins and are taken to be caused by tendon tension. This matter is persued further in Chapter 4.

Although the chief aim of the bone strain measurements was their correlation with tendon tension, an attempt was made to determine whether the bone bends or is axially compressed. These attempts involved the bonding of two strain gauges to the bone, one on its lateral aspect and one on its medial aspect. The variations in strain experienced by these gauges could be recorded individually as indicated in Figure 3.7-1 or could be added or subtracted by connecting both gauges as the appropriate two arms of the same Wheatstone bridge circuit. This dual processing of the data yielded results that were less repeatable than having only one active gauge in each bridge, and the results were correspondingly more difficult to interpret. This technique was subsequently abandoned.

It is interesting to note that since this work was done, a recent paper by Turner et al. (1975) presents recordings of bone strain on the lateral, medial, palmar and dorsal surfaces of the Mc III bone while the horse was walking and while trotting. These authors concluded that the Mc III bone is loaded almost axially during the stride.

A feature of the results reported by Turner et al. is that the recordings of bone strain from the lateral side of the Mc III bone are much less regular than from the other surfaces. It is possible that the common and lateral digital extensor muscles are partly responsible for this since their tendons lie near or on the lateral surface of the bone. If this is the case, the common and lateral digital extensor muscles might be regarded as providing the fine adjustment controls in correcting the course of the foot.

Additionally, a transverse assymetry exists when a horse walks in that its head and massive neck are able to turn toward an object that attracts the attention of the animal. This turning of the head and neck is more likely to influence the lateral and medial sides of the bone than the dorsal and palmar ones. The lateral side is likely to be affected more than the medial since the bone on the former side is thinner (Figure 2.2-4).

The possible irregular nature of recordings from the lateral aspect of the Mc III bone makes their correlation with other recordings more tenuous.

Temperature compensation of the strain gauges or of the lead wires was not employed in this study. The chief aim of the study was the correlation of dynamic measurements. Although temperature compensation for the leads would not have been difficult, it would have entailed an extra wire and solder joint for each gauge within the animal, and a consequent increased chance of a failure of the recording system.

Special temperature compensated strain gauges could have been used, but since the body temperature of the animal was not expected to vary widely, use of such gauges was not deemed necessary. Any body temperature variations would have been of much longer period than that of a stride.

The sensing element of strain gauges may be fine wire, foil, or semiconductor which has a sensitivity factor (gauge factor or k - factor) fifty or sixty times that of foil. However the width of the semiconductor strain elements is about 0.2 mm which would cover only a few osteons, the building units of bone. The surface irregularities of bone could cause a small semiconductor strain gauge to yield non-representative results (Cochran, 1972). Foil strain gauges have a grid width exceeding 1 mm and should give satisfactory results. Additionally, semiconductor strain gauges are significantly more expensive than foil ones, so foil electric resistance strain gauges were selected for the *in vivo* experiments. The extra sensitivity that would have been afforded by semiconductor gauges proved not to be required.

3.8

CHAPTER 4

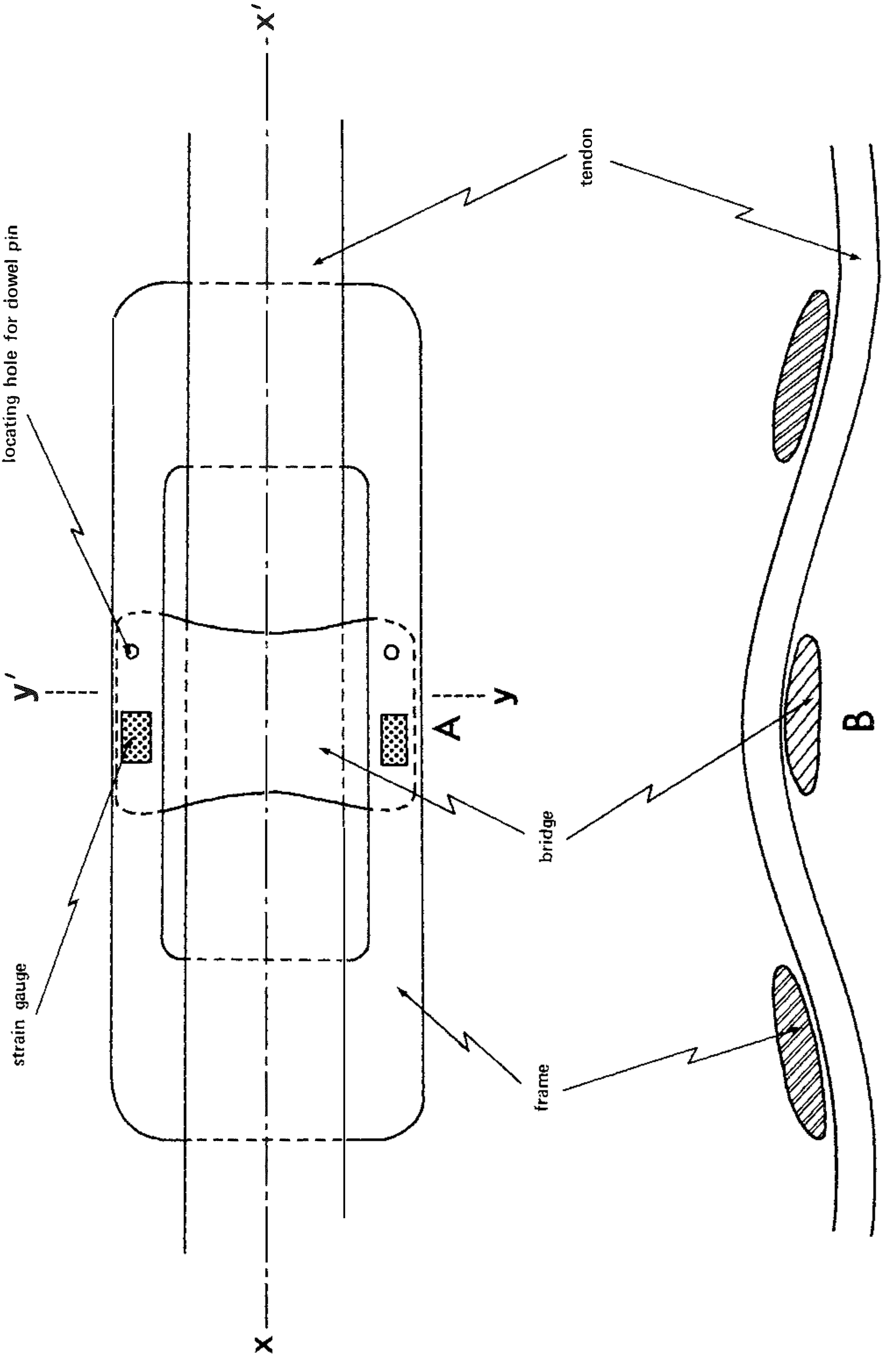
TENDON TENSION

FIGURE 4.2-1A

Schematic diagram of a buckle transducer fitted to a tendon as viewed from the bone. $x\ x'$ and $y\ y'$ indicate planes normal to the page.

FIGURE 4.2-1B

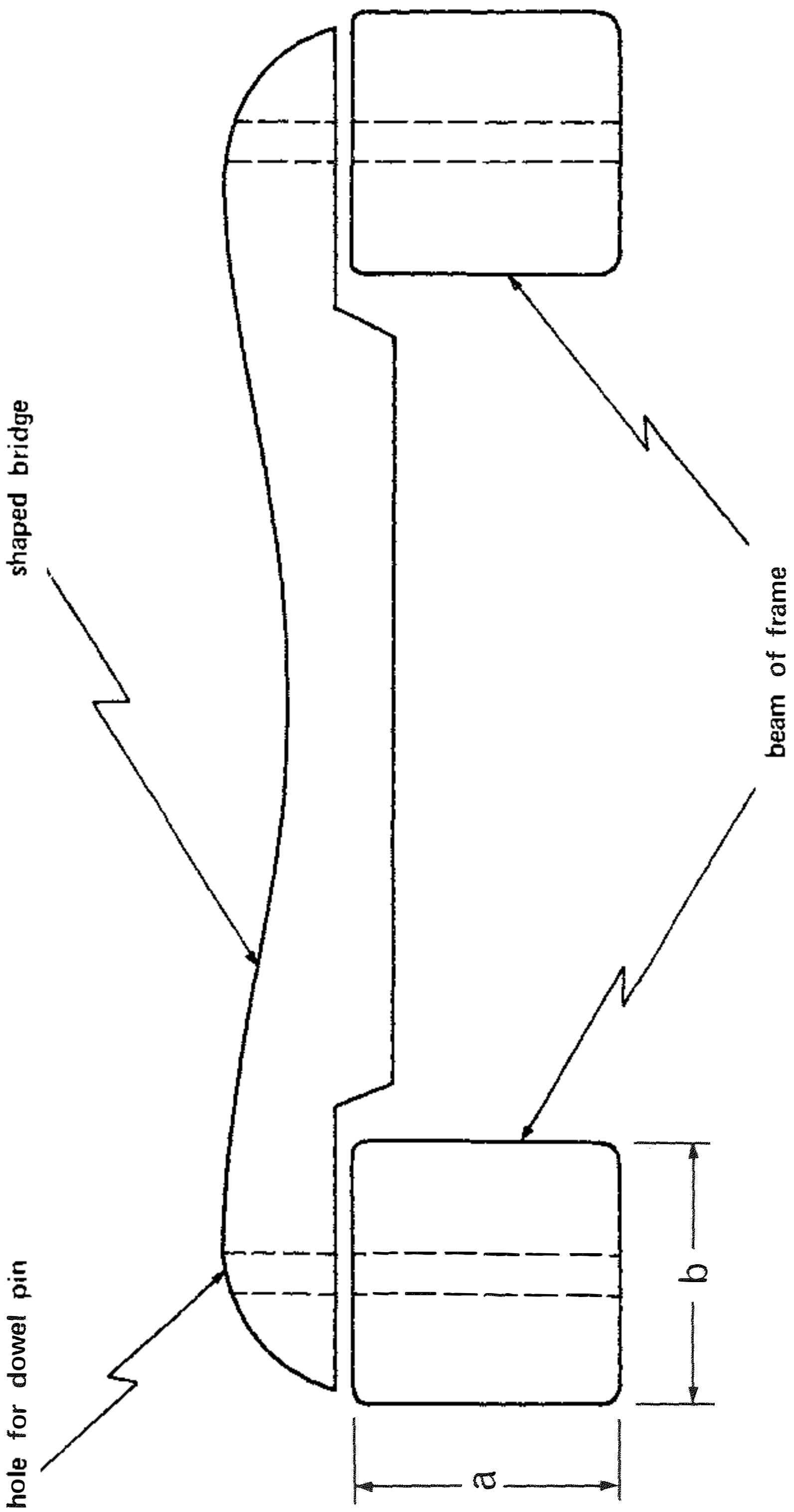
Schematic diagram showing intersection of the plane $x\ x'$ with the transducer and tendon of A above.



locating hole for dowel pin
tendon
x
x'
y
y'
A
B
bridge
frame
strain gauge

FIGURE 4.2-1C

Schematic diagram showing intersection of the plane y y' with the transducer of Figure 4.2-1A. a and b indicate respectively the thickness and breadth of the beams of the frame.


shaped bridge
beam of frame
hole for dowel pin
a
b

4.1 INTRODUCTION

The notion to correlate bone strain and tendon tension emanated from an observation in a paper by Lanyon and Smith (1970). These authors related bone strain to weight-bearing alone, but pointed out that the greatest potential source of bone strain is muscular. This brought to mind a brief report by Salmons (1969) describing the principle of operation of the 'buckle transducer' designed to measure skeletal muscle tension _in vivo_. It seemed appropriate, therefore, to combine the technique of Salmons with that of Lanyon and Smith to look for correlations between bone strain and tendon tension.

4.2 TENDON TENSION TRANSDUCER CONCEPTS

To measure tendon tension in a conscious animal, both the length of the tendon and its normal function should be maintained as much as possible. An ideal situation is one in which the object being investigated is also its own transducer as in a piezoelectric specimen, for then its effective length need not be modified.

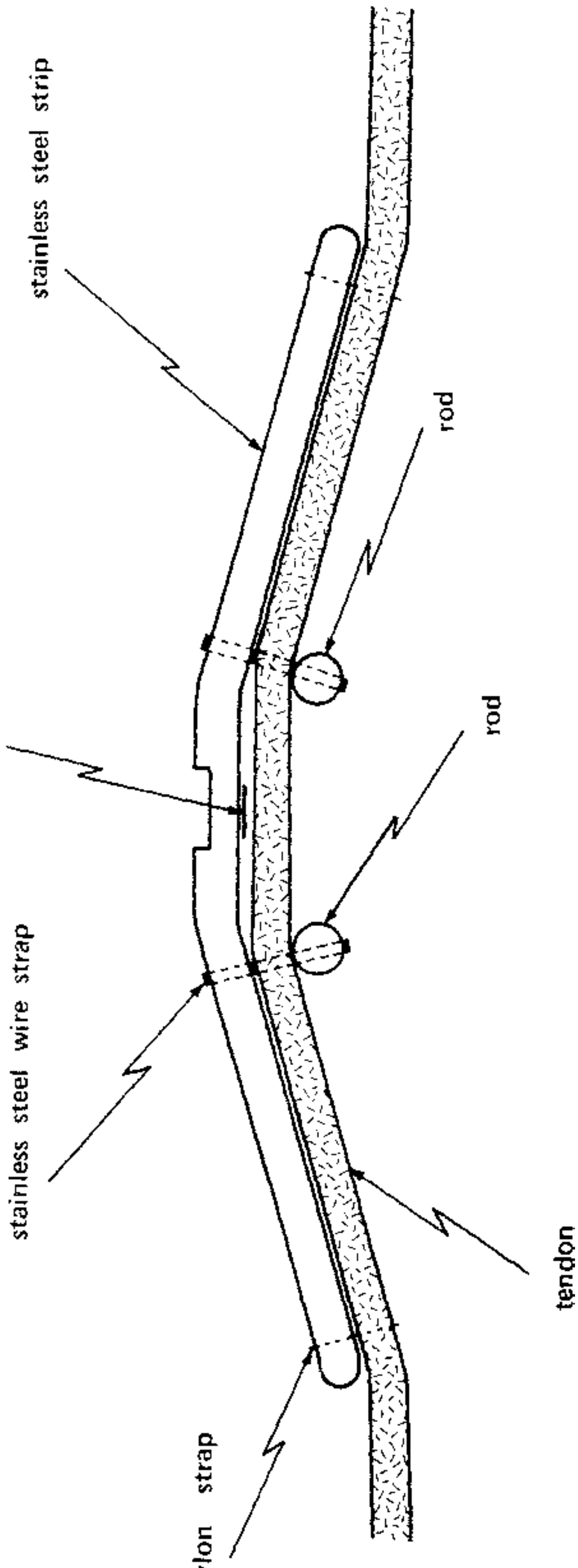
A viable method of monitoring tendon tension _in vivo_ is to introduce a slight kink in the tendon and use the tendency to straighten when stretched. Such a technique was suggested by Salmons (1969), wherein a loop of tendon was drawn up through a rectangular frame and a cross-piece slid under the tendon to form a kind of 'buckle transducer'.

A strain gauge bonded to the cross-piece (or bridge) monitored its bending and hence the tendon tension. Development of this basic concept led to the devices ultimately employed in this project.

A modification made was the sensing of strain in the sides of the frame rather than in the bridge (Figure 4.2-1).

FIGURE 4.2-2

Schematic diagram showing the side view of a tendon tension transducer with a tendon strapped to it. As the tendon tension is increased, the reduced mid-section of the stainless steel strip bends slightly resulting in stretching of the strain gauge. The nylon straps prevent the tendon from slipping from the steel strip. The stainless steel straps hold the tendon to the steel strip.


stainless steel strip
rod
strain gauge
rod
stainless steel wire strap
tendon
nylon strap

Salmons (1972) had independently suggested such a version. He favoured computation of tendon tension from the geometry of a buckle transducer having sharp corners (Salmons 1975 Figure 1.). We favour rounded corners, chamferred edges and calibrating the transducer directly (Barnes et al., 1975). This topic is discussed further in Section 4.5.

Other configurations employing a kinked tendon are possible. One transducer built and tested *in vivo* involved a strip of stainless steel bent slightly to form a shallow "V" (Figure 4.2-2). An electric resistance strain gauge bonded to the reduced mid-section of the strip indicated expanding (straightening) of the "V" as the tendon, strapped to the strip, was increasingly tensed.

The inner straps used consisted of 20 gauge (0.035 in. diameter) stainless steel suture wire threaded through each end of a smoothed stainless steel rod. This drew the tendon to the stainless steel strip when the suture wire, passing through small holes in the "V" strip, was pulled tight and the ends twisted to hold the rod secure. Two such rods were used, one each side of the strain gauge.

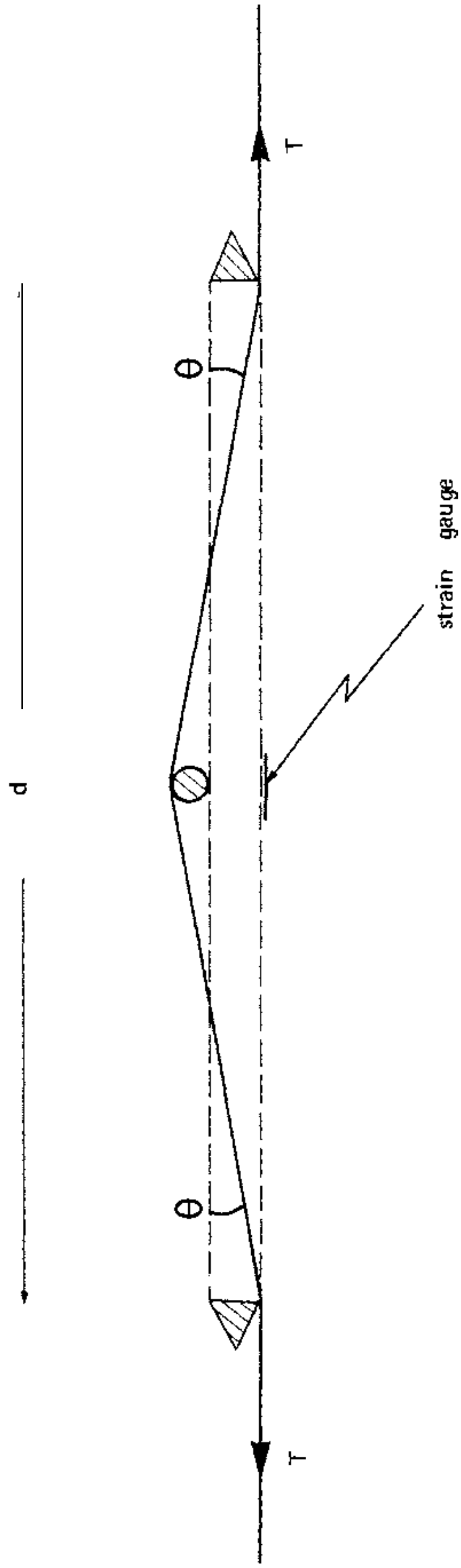
Two unfortunate features of this design led to its abandonment. Firstly, the rods used were of insufficient radius and tended to erode the tendon. This problem could have been overcome by the use of larger rods, but not so large as to cause undue bulkiness of the device. The second problem involved the two inner straps which were designed to hold the tendon securely to the stainless steel strip.

Any slackening of these inner straps caused by untwisting or strethcing of the stainless steel wire would reduce the bending moments applied to the stainless steel strip and yield a reduced output from the device.

4.2

FIGURE 4.3-1

Diagram showing a longitudinal section of a stylized buckle transducer fitted to a stylized tendon subjected to a tension T. The shaded portions of this Figure correspond to those of Figure 4.2-1B.


T
θ
d
strain gauge
θ
T

Both of these undesirable features were obviated by the use of the 'buckle' configuration (Figure 4.2-1).

Other designs were considered but not tested. One method which would incur minimal kinking of the tendon would be to insert a hollow cylindrical or spherical probe into the body of the tendon. Strain gauges internally bonded to the probe would signal its compression and relaxation as the tendon tensed and relaxed. Such an arrangement would necessitate splitting the tendon longitudinally. Although this may seem to be mutilation of the tendon, Johnson and Bartels (1972) recommend longitudinal splitting of the tendon fibres as a treatment for chronically damaged tendons.

Since tendon splitting would definitely wound the tendon and the 'buckle transducer' might not, the former was not attempted.

4.3 'BUCKLE TRANSDUCER' DESIGN AND MANUFACTURE

4.3.1 Principle of Operation

The principle of operation of the 'buckle transducer' may be discussed by considering a thread in position on a stylized angular frame and cylindrical bridge (Figure 4.3-1).

The thread is deviated an amount θ by the cylinder and subjected to a tension T. The total force downward at the centre of the frame side is $2\ T \sin \theta$ ($T \sin \theta$ on each beam) provided that the bridge is maintained in a central position. There will then be a vertical force acting at either end of the frame of magnitude $T \sin \theta$. Thus each frame side may be considered a beam supported at its ends and loaded at the centre where surface strain is detected.

In order to have a stable bridge which would not slide on the frame, the cylindrical bridge was replaced by a broad bridge through which holes were drilled for a dowel pin (Figure 4.2-1). As Salmons (1975) rightly pointed out, such location of the bridge is essential when strain is sensed at the centre of the frame sides, since then the transducer output is critically dependent upon bridge position.

The sensitivity of the 'buckle transducer' depends, among other things, on the extent to which the tendon is deviated. This deviation is influenced by the thickness of the tendon (Figure 4.2-1). Thus the output signal from the transducer is dependent not only on the tension in the tendon, but also on the thickness of the tendon when in position in the 'buckle transducer'.

Minimization of tendon damage was imperative, so all surfaces of the transducer which pressed on the tendon were shaped to avoid abrupt deviations of the tendon. Additionally, these surfaces were slightly grooved to conform approximately to the natural shape of a tendon cross-section.

The form of the shaped surfaces influences the cross-sectional shape of the tendon and hence its thickness. In the course of time the tendon may be compressed or eroded by the shaped surfaces which must minimize these effects. Compression or erosion of the tendon would reduce both the thickness of the tendon and the output of the transducer. Therefore a compromise had to be struck between the need for surface areas large enough to reduce contact pressure to an acceptable level, and small enough to avoid bulkiness.

4.3.2 Mechanical Characteristics

The material used for construction of the frame and bridge had to have the appropriate elastic and mechanical properties, and also had to be non-toxic and immune from corrosion by body fluids. Stainless steel was the most readily available suitable material and was therefore used. Titanium alloy (Salmons 1972, 1975) has a density advantage over stainless steel but this advantage was of little importance in view of the weight of the leg.

All tendon tension transducers were fashioned from a bar of stainless steel of cross-section 25 mm x 5 mm. Five 'buckle transducers' were manufactured, each differing in dimensions or method of construction. The width of the transducer was determined by the width of the tendon, taken to be 12 mm, and the width of the two side beams. The beam dimensions had to be such that the maximum expected tendon tension would not cause those beams to exceed the proportional limit of stainless steel.

Case and Chilver (1971) noted that for a high strength steel, the strain at the limit of proportionality is about 0.003. This is also the approximate strain limit quoted for Kyowa foil strain gauges.

The bending moment M at the central cross-section of one beam of the 'buckle transducer' frame may be deduced to be

$$M = \frac{d}{2} \times \frac{T}{2} \sin \theta = \frac{E\,I}{R}$$

where E is Young's modulus for stainless steel, I is the second moment of area about the neutral surface and R is the radius of curvature of the central part of the beam, see Figure 4.3-1.

For a rectangular cross-section of thickness 'a' and breadth 'b' (Figure 4.2-1C)

$$I = \frac{b\,a^3}{12}$$

Where the strain gauges are bonded, the longitudinal surface strain is

$$e = \frac{a}{2\,R}$$

so $\quad M = \dfrac{E\,b\,a^3}{R\;12} = \dfrac{E\,2\,e\,b\,a^3}{a\;\;12}$

or $\quad M = \dfrac{E\,e\,b\,a^2}{6}$

Since the overall width of the Kyowa KFC-03-C1-11 foil strain gauge is 2.5 mm, 'b' was chosen to be 4 mm, allowing ample space on the under surface of each beam for strain gauge bonding.

From Figure 4.3-1, if $d \doteqdot 4$ cm and $\theta = 14.5^{\circ}$, then $\sin\theta \doteqdot 0.25$. These values approximate the design parameters that were subsequently used. The Young's modulus for stainless steel was taken as 193 GPa, this value being obtained from a local engineering company. For a tendon tension of 400 N (see Section 4.7.5) and a strain at the strain gauge site of 0.001, the required thickness 'a' of each beam may be calculated. Since the strain 'e' is dependent on the second power of 'a' and on only the first power of 'b', the magnitude of the former is especially important to the sensitivity of the buckle transducer.

The above formulae may be manipulated to yield

$$a^2 = \frac{1.5\ T\ d\ \sin\theta}{E\ b\ e} = 1.56 \times 10^{-5}\ m^2$$

on substituting the above values. Hence $a = 3.95$ mm. We took $a = 4$ mm, giving a square cross-section for each beam.

In deriving this result, the width of the bridge, as depicted in Figure 4.2-1, was neglected. For a given angle θ, this width of contact with each beam would serve to decrease M, the bending moment, and so enable the transducer to accommodate tendon tensions even greater than 400 N.

A dowel pin hold would introduce a new strain pattern in its immediate vicinity. For the case of a circular hole in a thin plate, Nadeau (1964) shows that the axial stress is unaffected by the presence of the hole, and, as Timoshenko (1958) points out, any stress concentration is of a very localized character. We assumed that this holds approximately for the case of a hole in a beam. Provided that there is no

FIGURE 4.3-2

Photograph of the first and smallest buckle transducer fitted to the common digital extensor tendon. The bridge was secured to the beams by three of the planned four stainless steel sutures.

The length of the frame was 36 mm.

FIGURE 4.3-3

Reverse side of the buckle transducer shown in Figure 4.3-2. Note the epoxy resin protuberances to protect the silicone rubber covering the strain gauge on each beam.

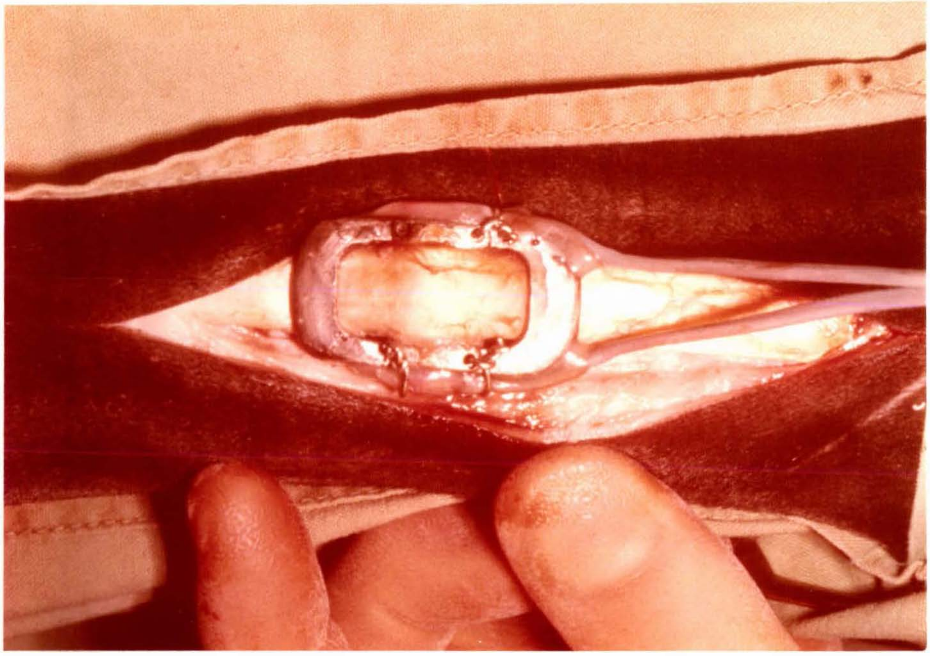
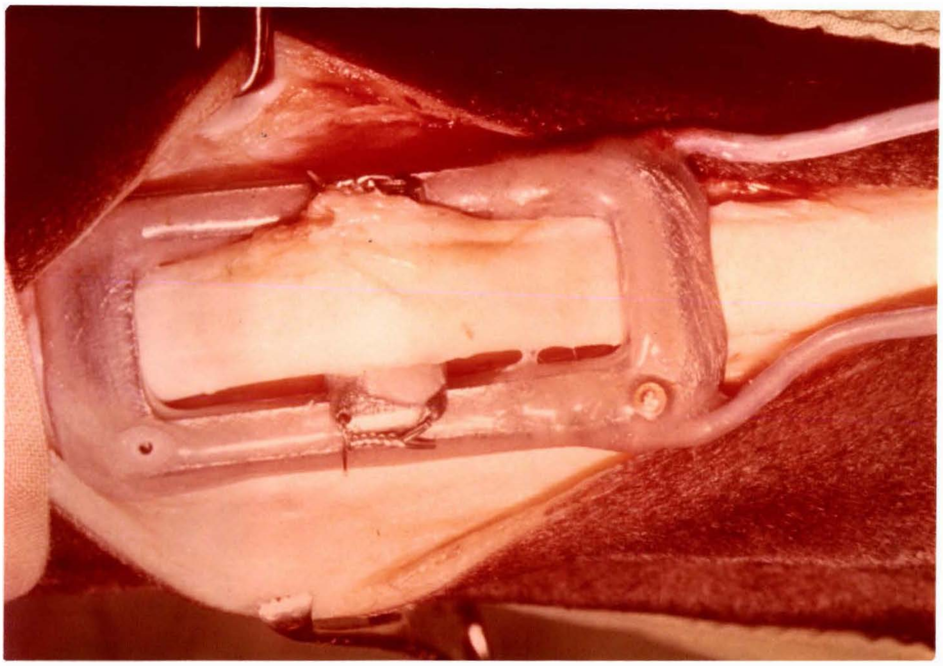


FIGURE 4.3-4

Photograph of the second buckle transducer fitted to a common digital extensor tendon. The bridge was sutured to the frame at only two places, thus allowing the lead out cable to lie under the frame. A metal strip was screwed and bolted to the bottom of one beam of the frame. This strip was included to protect the strain gauge installation on that beam.

FIGURE 4.3-5

Photograph of the reverse side of the buckle transducer shown in Figure 4.3-4. Note the protective metal strip.



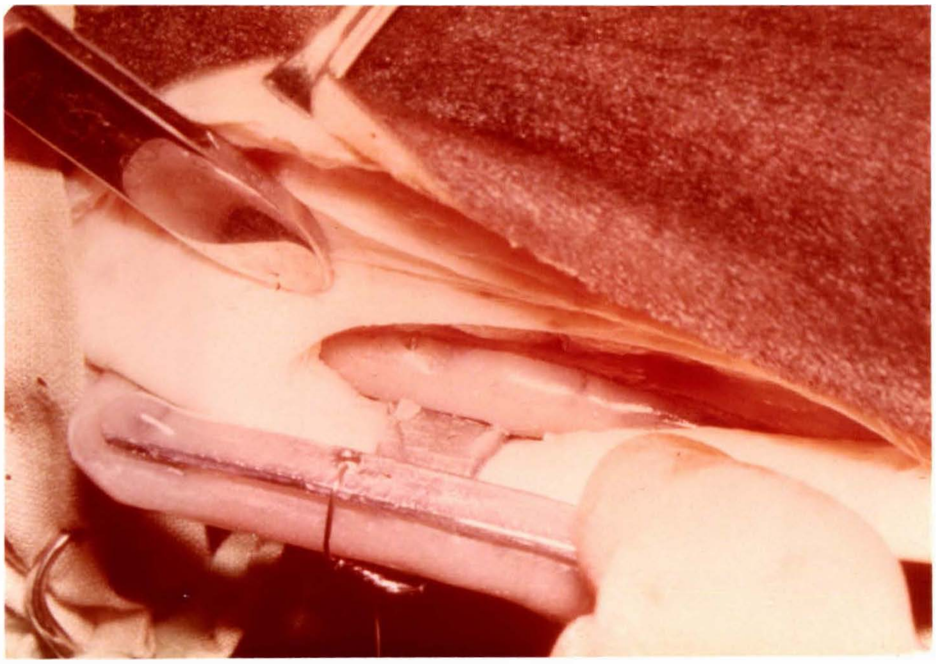


FIGURE 4.3-6

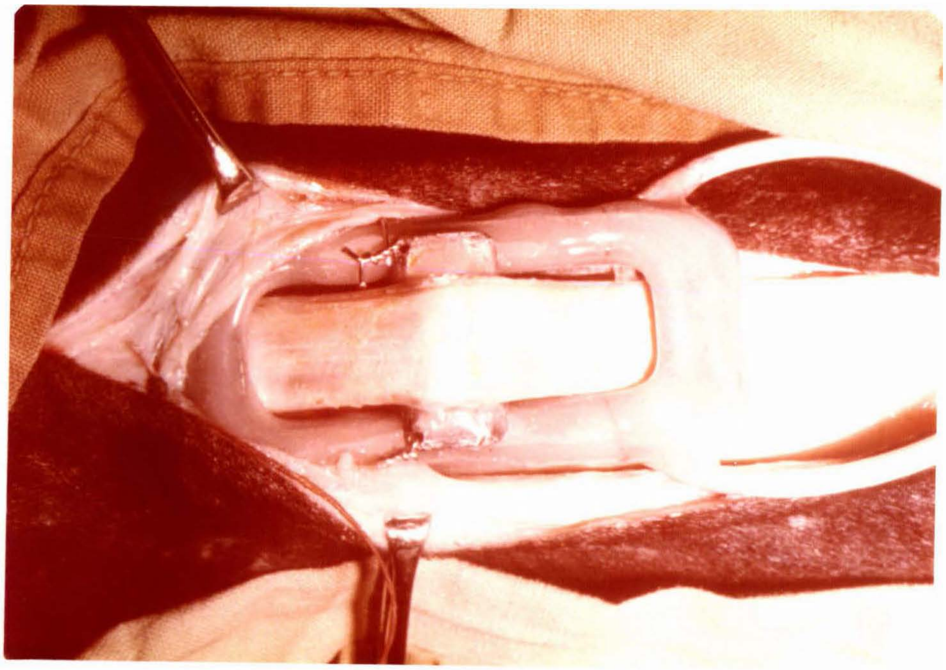
Photograph of a tendon fitted with the third buckle transducer. The length of the frame was 57 mm.

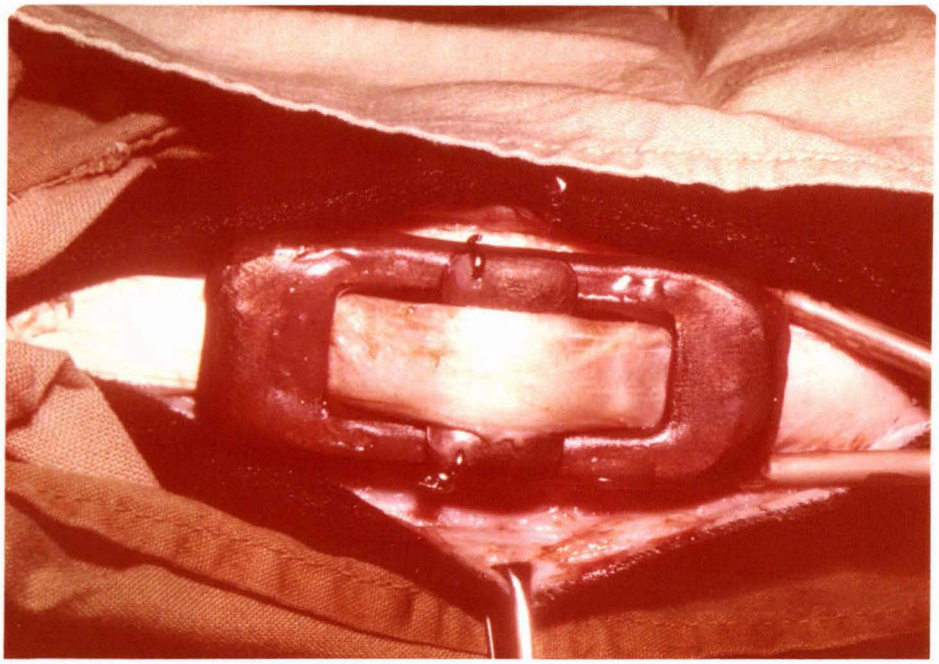
FIGURE 4.3-7

Photograph of the fourth and largest buckle transducer. This model featured large surface areas to contact the tendon which was subjected to only small curvature.

The upper dowel pin consisted of stainless steel suture wire bent over at the ends but not twisted. This type of dowel pin did not snag subcutaneous tissue.

The length of the frame was 65 mm.





yielding of the beam in this region subjected to stress concentration, there should be little effect on the strain sensed by the strain gauge. Yielding has not been observed at the dowel pin holes. We concluded that the drilling of these holes had little, if any, detrimental effect on the operation of the tension transducer, especially since we were not trying to calibrate the device by calculation.

Figures 4.3-2 and 4.3-3 show the first buckle transducer manufactured, being designed as the smallest 'buckle transducer' which would fit on the tendon and monitor its tension. The length of the frame was 36 mm, the beams being 3.5 mm thick and 3 mm in breadth. For θ about 23°, and d = 24 mm, the simple theory above indicates an upper limit of about 480 N for the tendon tension able to be withstood by the device. This is more than adequate.

The rationale behind the construction of a device of this size was that having a minimum of foreign material under the skin might cause least interference with the normal functioning of the limb. While this approach may be correct, nevertheless the device must be sufficiently large to avoid unduly high pressures between the tendon and the surfaces of the transducer contacted.

Post mortem examination of the tendon to which this first transducer was fitted revealed tendon erosion. The transducer had been implanted for two weeks. Erosion took many days to develop, and would probably not have been apparent had the transducer been removed within a few days of being implanted. Nevertheless, subsequent transducers (Figures 4.3-4 through to 4.3-7) were made longer in an effort to decrease the curvature forced on the tendon and to increase the area of the transducer surfaces contacting the tendon.

4.3.3 Manufacture

The dimensions of the beams (frame sides) and the approximate length of the frame ends were first decided. In general, the beam thickness 'a' and breadth 'b' were both 4 mm and the frame ends ranged from 8 mm to 13 mm in length for different transducers. The length 'd' (Figure 4.3-1) of the frame beams was 4 cm.

The crude frame was formed by drilling, sawing and rasping the inner area of the stainless steel bar. The beams were then filed to the required dimensions, and a crude bridge cut from the stainless steel bar. There followed the major task of filing the frame ends and bridge which determined the contours of the tendon passing through the 'buckle transducer'. The shaping of the bridge and frame ends were interdependent. The surfaces were smoothed with a fine file.

One small dowel pin hole (58 drill) was made in each beam and each end of the bridge to locate the bridge centrally. The dowel pins, consisting of stainless steel suture wire, were inserted during surgery once the tendon and bridge were in position.

Although a high degree of smoothness could have been achieved by the use of fine emery paper and polishing, such treatment was avoided in favour of separately coating the frame and bridge with epoxy resin (Araldite, CIBA Ltd, Switzerland) to yield a smooth surface for contacting the tendon. This coating also served to bind the lead out wires to the frame and protect the strain gauge installations.

In order to ensure maximum bond strength, the stainless steel was chemically pretreated (Selleys Technical Note A2 : 2/1.7.69). The surfaces were degreased and then etched by immersion for 10 minutes at 85° - 90° C in a mixture of oxalic acid (2.8 g), concentrated sulphuric acid (1.34 cc) and water (14 cc). The stainless steel was then removed from the solution and, under running water, the black deposit was removed with a stiff brush.

Following pretreatment, the strain gauge site on each beam of the frame was sprayed with surface activator (I.S. Activator A.C. Intercontinental Chemical Company Ltd) and allowed to dry. A KF - D3 - C1 - 11 foil strain gauge (Kyowa Electronic Instruments Co., Ltd, Japan), previously coated with silicone rubber (Silastic 732 RTV, Dow Corning Corp. U.S.A.) for insulation and protection, was bonded to the mid portion of each beam of the frame using contact cement (Isobutyl, 2 - cyanoacrylate monomer, Ethicon Ltd).

Double sided adhesive tape was placed on the beam under the lead wires of the strain gauge, serving to insulate them from the frame. An S bend was formed in each lead wire and held in place by the adhesive tape. These bends inhibited strain in the lead cable being transmitted to the strain gauge.

Fine nylon thread bound the twin core shielded lead cable to the underneath of each beam, the cable and strain gauge lead wires were soldered, adhesive tape was placed lightly over the S bends so that they could be effective, and the entire frame and strain gauge installations were coated with epoxy resin. Stainless steel wire placed through the locating holes in the beams prevented the epoxy resin entering there and served to support the frame while the epoxy resin set.

Extra epoxy resin was applied at the end of each beam to form pads for contacting the Mc III bone adjacent to the tendon. There was thus some clearance between the bone and the epoxy resin covering the strain gauge area allowing space for the stainless steel suture wire used to secure the bridge to the frame. Protection was also afforded the strain gauge region.

FIGURE 4.5-1

Diagram showing the forces and angles used to calibrate the buckle transducer. T and T' are the tensions in the tendon. The dashed line perpendicular to the known applied force W is used to define angles ϕ and ϕ'.

$$T = \frac{W}{\sin \phi + \cos \phi \tan \phi'}$$

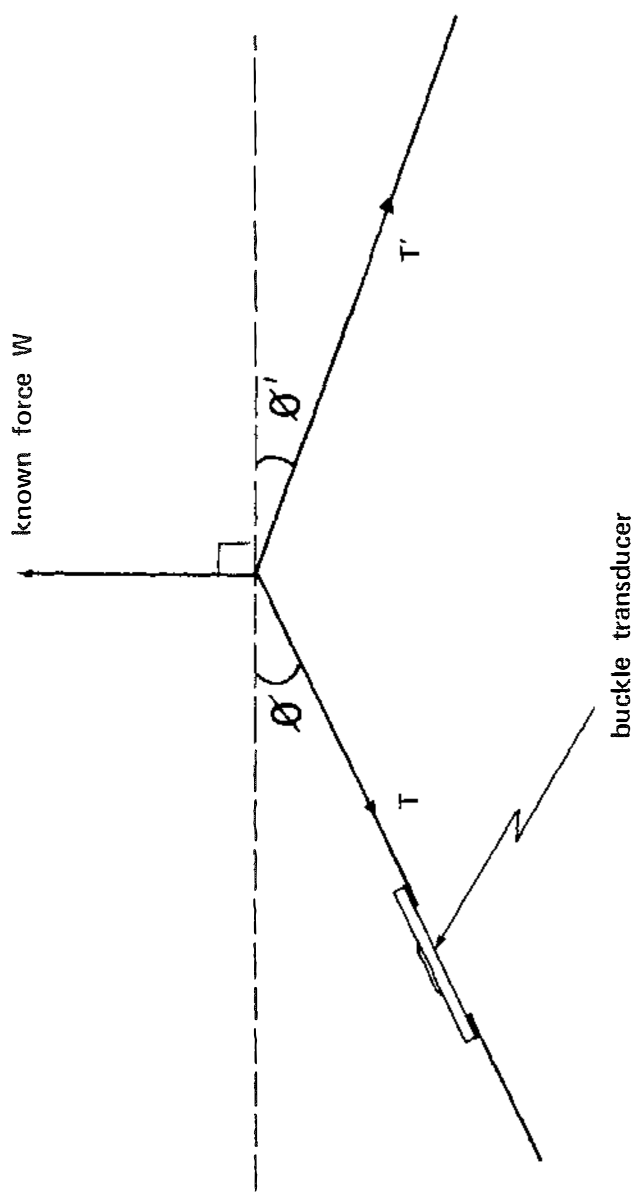
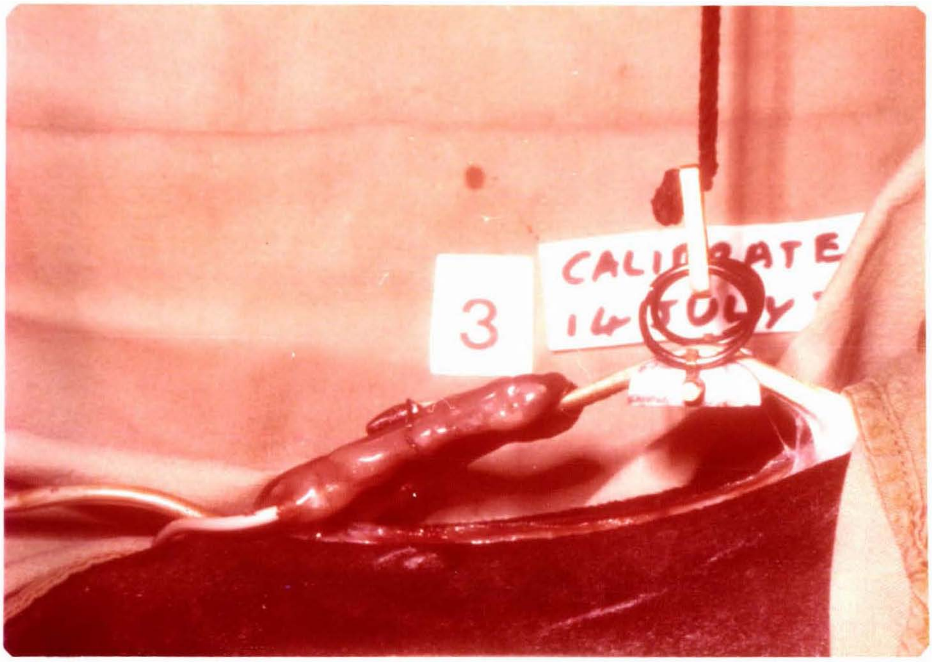

known force W
Ø'
T'
Ø
T
buckle transducer

FIGURE 4.5-2

Photograph of a buckle transducer during the calibration procedure. The carefully shaped cradle was slid under the tendon. Then the cord which was to transmit the known force was hooked onto the rings each side of the cradle.


3

4.4 MEASURING AND RECORDING APPARATUS

The strain gauges bonded to each beam of the 'buckle transducer' could be independently connected as the active arm of a Wheatstone bridge circuit. The circuitry and recording apparatus used for tendon tension measurements were the same as for bone strain recording described in Section 3.4.

The cross sectional area of the tendon was determined by the use of two slide calipers held together at right angles to each other. One set measured the tendon thickness, the other measured its width. This method was considered adequate since the common extensor is a flat tendon.

4.5 CALIBRATION

The most obvious procedure for calibrating the 'buckle transducer' was to record its output when a known tension existed in the tendon. Two methods of determining this tension were used. If the animal was to be euthanised shortly after the experiment, then the leg could be retrieved and known weights hung from the severed tendon. This procedure constituted the most direct method of calibration and was used initially.

However, the tendon could be tensed by the application of a known force to the exposed tendon while the animal was under general anaesthesia. Knowledge of the magnitude and direction of this applied force, together with measurement of the angles involved allowed the tendon tension to be calculated (Figures 4.5-1, 4.5-2). This method was subsequently adopted.

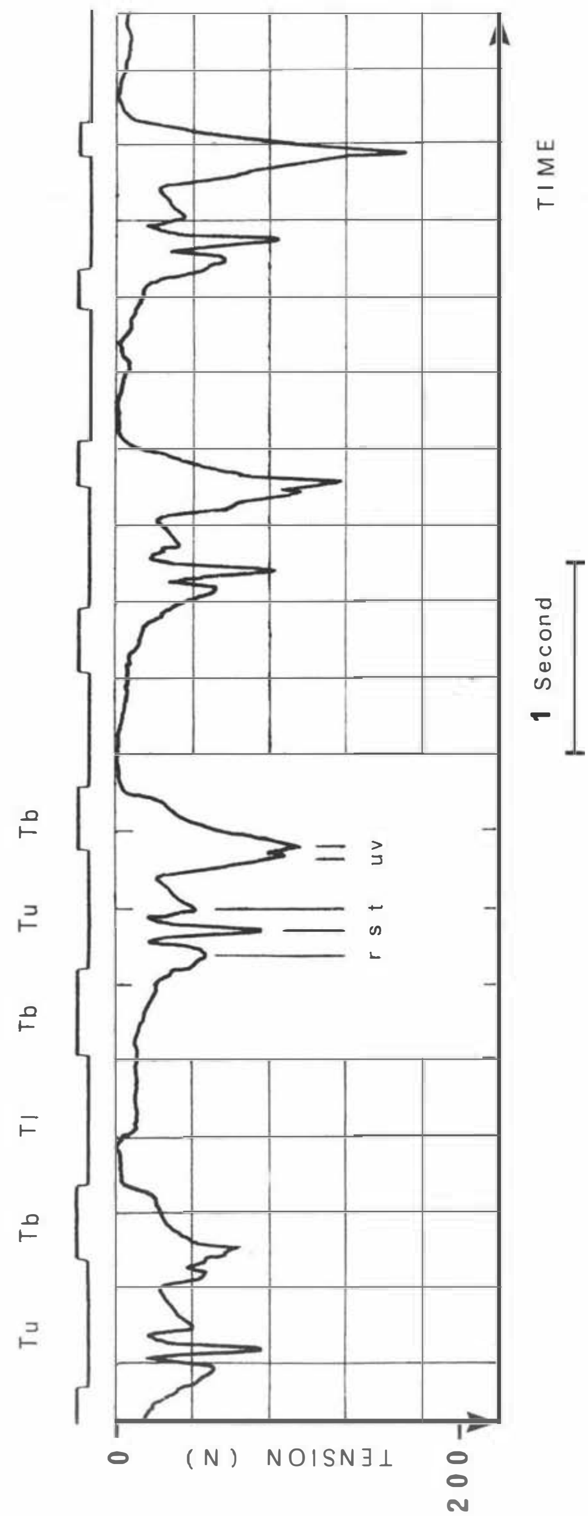
Known weights were hung from a cord which passed over a pulley and was hooked to the cradle holding the tendon. This drew the tendon from the bone. The weight W was noted, the electrical signal from the 'buckle transducer' was recorded and a photographic slide taken. (Figure 4.5-2). The weight was then changed and the process repeated. The slides were

FIGURE 4.6-1

Foot switch and tendon tension results from a 430 kg thoroughbred mare standing 1.5 m high.

The upper curve represents the output from the switches on both forefeet. Tb indicates when both forefeet were on the ground. Tu signifies the swing phase of the monitored (right) foreleg when it was not in ground contact, and Tl denotes the swing phase of the unmonitored (left) foreleg.

The lower curve is a redrawn recording of tension in the common digital extensor tendon of the right foreleg. The zero of tendon tension is not necessarily that indicated. Labels r, s, t, u and v mark five prominent tension peaks.


Tu
Tb
Tl
Tb
Tu
Tb
r s t
u v
1 Second
TIME
0
TENSION (N)
200

subsequently projected on to a translucent screen and the angles ϕ and ϕ' measured. By resolving tensions T and T' in directions parallel and perpendicular to force W, it may be shown that the tendon tension T monitored by the 'buckle transducer' is given by

$$T = \frac{W}{(\sin\phi + \cos\phi \tan\phi')}$$

4.6 TENDON TENSION RESULTS

Figures 4.6-1 and II-4 show a redrawn recording of tension in the common digital extensor tendon of the right foreleg. These results were from a 430 kg thoroughbred mare standing 1.5 m high. The zero of tendon tension was not necessarily that indicated and all tensions were possibly a little greater than shown in spite of the consistency of minimum tension in each cycle. This is discussed below.

It is evident that much tendon tension activity occurs during and just after the swing phase Tu of the monitored foreleg. There appears to be a cycle of at least five prominent peaks labelled r, s, t, u and v with some sustained tension after v. This sustained activity appears to decrease in duration as the gait quickens.

Figure 3.7-3B represents the tension versus time recording from the common digital extensor tendon in the left foreleg of a 390 kg mare. Tension again increases in the negative 'y' direction and time is indicated by the seconds marks in Figure 3.7-3C.

The foot switch trace (Figure 3.7-3D) indicates the swing phases of the monitored leg (L) and the unmonitored leg (R). Again tendon tension is associated with the swing phase of the monitored foreleg.

Figure 3.7-4 is a photocopy of original data, part of which was redrawn for Figure 3.7-3. From top to bottom, the traces in Figure 3.7-4 correspond to Figures 3.7-3D, A, C and B respectively, but with different magnifications.

The foot switch trace in Figure 3.7-4 shows the effect of having vocal information recorded on the same channel. Note that Figure 3.7-4 is of a horse moving off from a standing position, taking several strides and then coming to a halt.

Figures 3.7-4 and 3.7-5 represent data recorded in the same manner and from the same animal. The latter figure reflects a gait less regular than that suggested by Figures 3.7-3 and 3.7-4.

Figure 2 of Appendix I illustrates a regular pattern of tendon tension.

The tendon tension data presented in Figure 3.7-3 is shown again in Figure 5.6-1 together with the electromyogram recorded simultaneously from the humeral head of the common digital extensor muscle.

The cross-sectional area of the tendon whose tension is shown in Figure 3.7-3 was $(2.5 \pm 0.2 \times 10^{-5})\ m^2$ so that a tension of 100 N implied a stress of 4 MPa.

4.7 DISCUSSION

4.7.1 The Pattern of Tendon Tension

The prominent features of the tendon tension recordings are the two groups of tension peaks. These are especially evident in Figure I-2. In this recording peaks r and s (refer Figure 4.6-1) are of about equal magnitude, but peak t is only just evident. Peaks u and v are clearly resolved, the latter being the dominant peak in both Figure I-2 and Figure 4.6-1.

Figures 3.7-3B and 3.7-4D are similar to Figure I-2 in that peaks r, s, u and v are clearly evident, but peak t is much less so.

Peak r begins its rise during the overlap Tb (Figure 4.6-1) of the foreleg support phases. This peak is characterized by having a relatively long rise time (about 0.2 seconds), a steeper descent, and is not a sharp peak.

Closely following peak r is the sharp peak s (Figure 4.6-1) which is arrowed in Figure 5.6-1. Peaks r and s are probably caused by different heads of the common extensor muscle. This matter is persued further in Section 5.7 where electromyographical and anatomical considerations are discussed.

Peak t (Figure 4.6-1) is the smallest of the five coded peaks. It appears to be absent during some strides, for example the first and last 'L' swing phases of Figure 3.7-3B.

It is suggested in Appendix II that peak t assists with leg extension. But there are muscles other than the common digital extensor muscle having the function of extending the limb. These include the extensor carpi radialis and the lateral digital extensor muscles (Figure 2.2-2), which could assist leg extension during the period of the peak t. The presence of peak t probably indicates a variation in style of walking rather than a fundamental force in the walking gait.

Peaks u and v are of special interest since they correlate well with the electromyogram recorded from the humeral head of the common digital extensor muscle. This correlation is illustrated in Figure 5.6-1 and discussed in Section 5.7.

The large peaks labelled u and v (Figure 4.6-1) are sometimes the two dominent peaks in a cluster. This is evident in three of the four strides shown in Figure 3.7-3B. Peaks u and v occur at the end of the swing phase of the monitored foreleg. At this point in the stride, the leg muscles are used not for movement (Rooney, 1969) but to brace the leg for weight bearing. It is noteworthy that the last bracing represented in Figure 3.7-3 shows a lot of structure.

4.7.1

This was the last stride before the horse halted.

It is expected that the structure of the 'u v cluster' is influenced by the nature and condition of the surface on which the horse walked. At the end of the swing phase of the stride the bracing of the leg would be influenced by what the animal visually senses to be the state and nature of the ground surface. Once footfall has occurred, the animal may alter the tension in its leg muscles to counter any disturbing effect of the ground felt by the foot.

The peaks r, s and t, occurring early in the swing phase are likely to be less influenced by environmental factors than the 'u v cluster'. It is therefore likely that the peaks r, s and t are more consistent that the 'u v cluster'. Inspection of the tendon tension recordings does lend some support to this hypothesis.

Variation in the pattern of the r, s, t triad could be linked with counter measures associated with ground surface defects detected by the unmonitored forefoot which had just made ground contact. An extreme case would be if the unmonitored forefoot stumbled, necessitating rapid extension of the monitored foreleg. This would be reflected by larger than 'normal' r, s or t peaks.

4.7.2 Tendon Tension and Bone Strain Correlation

Bone strain is influenced by the properties of the bone, the weight and gait of the animal and the tensions developed in the related muscles and tendons. The correlation between longitudinal strain in the Mc III bone and tension in the common digital extensor tendon cannot be expected to be very high because of the influences of the numerous other factors. However it is possible that some of the muscles function in groups for at least part of the stride so that the above correlation would be better than if such groupings did not occur. With reference to the human finger Landsmeer (1949) affirms that it is misleading to speak of isolated, individual and specific actions of certain muscles, since they function as a co-ordinated group. A similar co-ordination could occur in some of the equine extensor and flexor muscle groups.

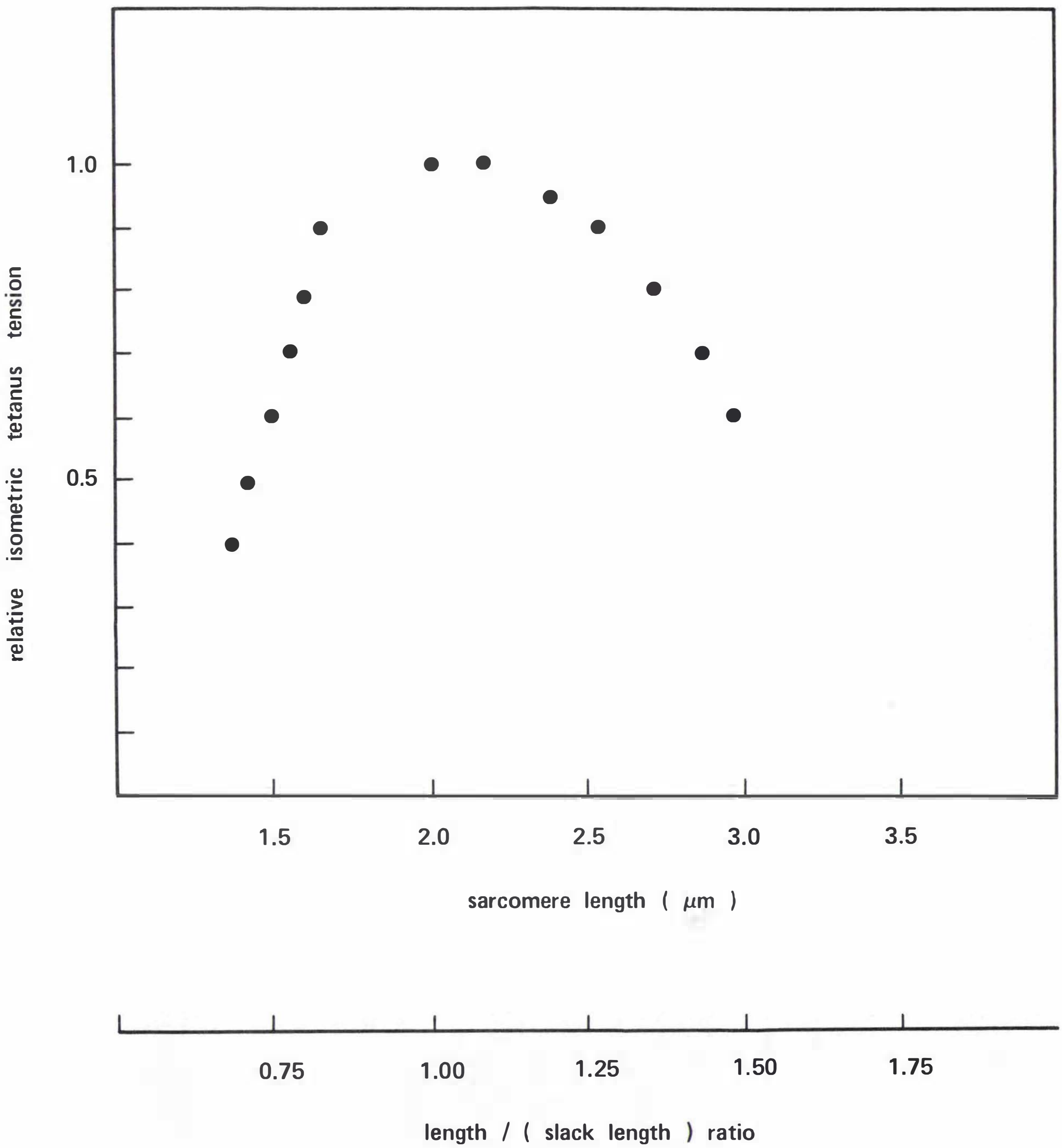
Tension in the common extensor tendon is expected to cause compressive strain on the lateral aspect of the Mc III bone because the tendon lies to that side of the bone (Figure 2.2-2). Evidence of correlation between bone strain and tendon tension in Figure 3.7-3 would be that positive going peaks in trace A should correspond with negative going peaks in trace B. Without exception, during periods of rapid variations in tendon tension there are corresponding periods of rapid variation in bone strain.

There is not always a one to one correspondence between the peaks in each trace, which is to be expected. The tension developed in the tendon is only indirectly transferred to bone compression since the force must be transmitted through a number of joints. The instantaneous orientation of the leg at a particular time in the stride will affect the way in which these joints transmit the tendon tension to the bone in question. Thus the 'transfer functions' of the joints will be time dependent and will vary as the gait varies. It is evident from Figures 3.7-3B and D that the gait was variable so that the joint 'transfer functions' were not precisely repeatable. Nevertheless large tendon tension peaks appear to correspond with large bone compression spikes.

It should be noted, however, that there is a delay of approximately 0.1 s between a tendon tension peak and its corresponding bone strain compression peak. A delay is to be expected since the muscle - tendon - joint - bone system is not perfectly elastic. Because of the viscoelastic nature of these materials (McElhaney, 1966; Abrahams, 1967), strain in any of them lags behind the stress causing it.

FIGURE 4.7-1

Diagram showing the effect of sarcomere length on the tension developed by single muscle fibres. This Figure was redrawn from Bendall (1969) Figure 6.7(a).


1.0
0.5
relative isometric tetanus tension
1.5
2.0
2.5
3.0
3.5
sarcomere length (μm)
0.75
1.00
1.25
1.50
1.75
length / (slack length) ratio

Barnes and Pinder (1974) implied that there might be a one to one correspondence between tendon tension peaks and bone strain compression peaks. Subsequent work has shown that although this is not always true, periods of tension variation in the common extensor tendon do correlate well with periods of bone strain compression on the lateral aspect of the Mc III bone.

4.7.3 Tendon Shortening

The effect of the 'buckle transducer' on the length of the tendon deserves consideration. From Figure 4.3-1 we deduce that the tendon is effectively shortened by about $d\ (\sec\theta - 1)$. For $\theta = 25^{\circ}$, $\sec\theta = 1.1$, so the tendon is effectively shortened by d/10 or 4 mm. This is less than a 1% change in tendon length, a negligible amount as far as the tendon is concerned.

Tendon shortening implies increasing the resting length of the corresponding muscle (Salmons 1975). An increase in length of 4 mm amounts to about 3% change in resting length of the muscle.

Since, in this study, we were concerned with gait, tendon tension and electromyography, the influence of tendon shortening on these deserves attention.

A 1% shortening of the common digital extensor tendon will be compensated for by a slight extension of the common digital extensor muscle. The situation is complicated by the division of this muscle into three heads. It is possible that one head will account for more readjustment than any other. This is likely to have more effect on the dynamics of the individual heads of the muscle than on the resultant gait of the animal. Since the tension in the tendon largely determines its effect on the limb, this tension is likely to be little affected by the slight tendon shortening (or muscle lengthening). Figure 6.7 of Bendall (1969) (our Figure 4.7-1) shows the effect on isometric tetanus tension of variation in sarcomere length for single muscle fibres. An increase in sarcomere length by more than 10% resulted in negligible change in the tension.

The tension - muscle length curve flattens as muscle length increases slightly beyond the resting length (Close et al., 1960; Bendall, 1969). Thus for a walking animal, where great muscle exertion and length changes are not expected, a small increase in the effective rest length of the muscle should have minimal effect on the tension developed in the tendon. If the tendon shortening results in a significant stretching of muscle fibres for one particular muscle head, then the passive tension in that head may become significant, although the developed (active) tension may decline (Close et al., 1960). The resultant effect would be an increase in force exerted on the tendon by this particular muscle head, requiring less effort by the other heads to effect a desired limb movement.

Such a situation could lead to reduced action potential count in all heads. Close et al., (1960) found that for an isometric stretch of 3 cm or more in the human soleus muscle, the action potential count was less than for the rest length, this effect becoming very pronounced for the longer muscle lengths. Thus a muscle head stretched significantly would result in a lower action potential count in the EMG and a greater (passive) tension exerted on the tendon.

For a given length of muscle, the action potential count is linearly related to tension. For a head not significantly stretched and needing to exert less force by virtue of another head exerting more, then the action potential count should be less.

Significant stretching of any head is most unlikely in view of the small tendon shortening that the buckle transducer causes. We conclude therefore that this tendon shortening

4.7.3

should have minimal effect on gait, tendon tension and EMG. To resolve these questions beyond all doubt would require detailed anatomical, biomechanical and electromyographical study of the individual heads of the common extensor muscle, and the interplay between the heads.

Finally, we note that a redistribution of tensions among the heads of the common extensor muscle would result in a different set of forces at the origins of these heads. However, there is likely to be little, if any, noticeable effect on the movements of the upper limb and shoulder as a result of such a redistribution.

4.7.4 Calibration Techniques

If several days were to elapse between the recording of tendon tension data and calibration of the 'buckle transducer', this calibration could incur a systematic uncertainty. This is because the tendon may have been eroded or have deteriorated in the intervening period so that the dimensions and mechanical properties of the tendon were different for the 'buckle transducer' calibration and for the data recording.

If only one calibrating session was to be had, it should follow soon after the *in vivo* measurements in preference to during the initial operation. Then both measurements and calibration are made after the tension transducer and tendon have been subjected to the rigours involved in the horse getting up following surgery.

If it is assumed that the condition of the tendon gradually deteriorates with time, there is some justification in also calibrating the transducer immediately it is attached to the tendon. Linear or possibly non linear interpolation of the two calibration runs, before and after measurements, might then yeild a more correct calibration corresponding to the time that the measurements were made.

After having the 'buckle transducer' attached to it *in vivo* for four days, the tendon appeared to be undamaged but its fine sheath had worn through where it pressed against the bridge and the frame ends. This effective decrease in tendon thickness was insignificant. Hence calibration of the transducer should present no special problems provided it is completed within four days of the transducer being implanted.

A difficulty may arise if calibration was accomplished neither during implantation nor soon after measurements were made. At some later time when calibration is intended, the tendon may be found to be damaged to an extent that makes it useless for calibration purposes. An alternative means of calibrating the tension transducer must be found. Since the common extensor tendons in both forelegs are normally of very similar dimensions, the undamaged contralateral tendon approximates the original dimensions of the damaged one.

Since the cross-section of the tendon is not constant throughout its length, care must be exercised to ensure that the transducer is attached to the contralateral tendon in a position corresponding to that used for tension measurements in the investigated tendon.

Measurements on both common extensor tendons from the forelegs of each of three horses indicated that each pair gave, within 5 per cent, the same calibration. However, the location of the transducer on both tendons had to correspond within 2 cm.

Calibrations were made either hours after the tension measurements were made or in some cases, on the contralateral tendon.

Another method of tension transducer calibration is that suggested by Salmons (1975) where the strain in the side beams of the frame is computed from the dimensions of transducer and tendon. He used a 'buckle transducer' having minimal rounding of edges. While this undoubtedly eases computation of strain, it would also lead to excessive wear of those parts of the tendon unfortunate enough to come into contact with the angular parts of the frame and cross-piece.

Salmons did not indicate what, if any, protective coating should cover the strain gauge installations. We chose to coat the frame and bridge with epoxy resin which protected the installations, bonded the lead out cable to the frame and provided a smooth surface for contacting the tendon. This coating, however, influenced the elastic characteristics of the buckle transducer in an unknown manner. However since the elastic modulus of this coating is at least twenty times smaller than that of the stainless steel, this influence is presumably of small proportions. Also, our shaping of the frame ends and of the bridge mould complicate the theoretical analysis of the device.

The computational approach to calibration requires that measurements of tendon thickness be made. This thickness will not be constant across a cross-section and will be influenced by the shaping of the frame ends and the bridge. The difficulties inherent in the computational method persuaded us to reject it in favour of the direct methods of calibration described previously.

4.7.5 The Working Range of Tendon Stress

A search was made of the literature to determine the range of stress which a tendon might be expected to experience during normal activity. It appears that there are several related criteria which may be used.

Abrahams (1967) examined horse extensor tendon in vitro. A sample of cross-sectional area 0.032 in ^{2} (2×10^{-5} m^2) was cyclically stretched at an average strain rate of 45% per minute to a nominal stress of 1 600 p.s.i. (11 MPa) and achieved complete strain recovery. After stressing to 3 250 p.s.i. (22.4 MPa), 0.5% residual strain was apparent. Since we would not expect residual strain to develop in the tendons of the walking horse, we may assume that stresses of about 22.4 MPa would not normally be reached. A stress of 11 MPa seems a reasonable upper limit to place on the expected tendon stress. This corresponds to a tension of 220 N . We took a tendon tension of 400 N to correspond to the safe design limit of the transducer.

The tendon tension results shown in Figure 3.7-3B indicate a peak tension slightly in excess of 200 N or a stress just above 8 MPa which compares favourably with the 11 MPa mentioned above. In a galloping horse, however, this stress level could well be exceeded.

Walker et al., (1964) suggested that in normal maximum use the living, human, plantaris tendon is stressed to about one-fourth its rupture level of roughly 14 000 p.s.i. The suggested normal maximum stress was 4 000 p.s.i. (27.6 MPa). Thus, given a value for the ultimate strength of a tendon, the working range of tendon stress may be estimated as one quarter of this given value. In agreement with this criterion, Elliott (1965) presented the results of Gratz (1931) for human fascia lata which broke at a stress of about 8 000 p.s.i. (55 MPa). The maximum safe stress was shown as about 2 100 p.s.i. (15 MPa) since greater stretches lead to irreversible changes.

From the stress-strain curves for tendon given by a number of authors (Abrahams, 1967; Evans et al., 1975), it is apparent that the toe regions of these curves extend up to a stress value approximately one quarter of that at which rupture occurs. Evans et al., (1975) showed the stress-strain curve for human palmaris longus tendon in tension. The toe region extended up to a stress of about 20 MPa, with rupture at approximately 90 MPa.

Abrahams (1967) presented the stress-strain curve for the Achilles tendon from a 54 year old woman. The peak stress before rupture was about 1 600 p.s.i. (11 MPa), and the toe region extended up to 40 p.s.i. (2.7 MPa).

In both of these cases, the toe region extends up to about one quarter of the tensile strength of the tendon. This suggests that the normal working stress range for tendon, as defined by the one-quarter tensile strength criterion, corresponds with the toe region of the stress-strain curve. In this region, the collagen fibres become oriented in the direction of the applied tensile load, resulting in disappearance of the wave pattern (Elliott, 1965; Abrahams, 1967; Diamant et al., 1972). This disappearance might be used as another criterion.

Since tendon is visco-elastic (Frisen et al., 1969; Cohen, 1974; Van Brocklin and Ellis, 1965), the rate of strain should also be considered since the stress at any stage of loading is influenced to some extent by both the strain and the strain rate (Haut and Little, 1972). Furthermore, the previous history of loading is also a factor in determining the current stress.

Elliott (1965) gave no strain rate information for the 2 100 p.s.i. (15 MPa) maximum safe stress. Likewise, Walker et al., (1964) quoted no rate of strain associated with the one-fourth rupture criterion. Evans et al., (1975) used a strain rate of 10^{-3} s^{-1} for the human palmaris longus stress-strain data. For the Achilles tendon stress-strain to failure data, Abrahams (1967) used a strain rate of 3×10^{-3} s^{-1} to the 4% strain level, well into the linear stress-strain region, and 5.7×10^{-3} s^{-1} thereafter, thus confusing the issue to some extent.

Abrahams (1967) has presented stress-strain data to the 2% strain level for Achilles tendon subjected to strain rates of 1.67×10^{-4} s^{-1}, 3.3×10^{-4} s^{-1}, 1.67×10^{-3} s^{-1}, 4×10^{-3} s^{-1} and 8.3×10^{-3} s^{-1}. All the data lies within the toe of the stress-strain curve. Abrahams seemed to imply a relationship between nominal stress σ and strain rate $\dot{e}$ of the form:

4.7.5

$$\sigma = c \; (\dot{e})^{m}$$

where c increased and m decreased with strain.
Both c and m are positive. The dot above the strain symbol e denotes the time derivative (of strain).
Abrahams (1967) noted that a specimen of horse extensor tendon did not show the type of relationship thus described. This casts doubt on the use of data obtained from other than equine common extensor tendon to deduce its visco-elastic behaviour. Unfortunately, Abrahams did not say in what manner the equine data differed from that for the Achilles tendon.

Strain rate is a significant factor in the stress-strain characteristics of equine common extensor tendon, as suggested by the hysteresis displayed in the data presented by Abrahams. We are, however, left in doubt as to the relationships among strain, strain rate and stress for this tendon.

A plausible criterion to estimate a safe working range for tendon is that when permanent damage has been done to the tendon, the working range has been exceeded. One form of permanent damage might be residual strain.

The maximum safe stress noted in Elliott's 1965 paper corresponded to a certain percentage elongation beyond which "irreversible changes" occurred. The associated diagram showed residual strain for the hysteresis curves which exceeded the allotted maximum safe stress. We might therefore presume that at least one of the "irreversible changes" was residual strain. The maximum safe stress corresponded to a percentage elongation of 3.5% for human fascia lata.

Abrahams (1967) observed no residual strain in equine common extensor tendon after cycling to 2% strain, but observed an increasing residual strain when cycling the specimen to the 3% strain level. The working range of equine common extensor tendon is, therefore, likely to have an upper limit corresponding to a strain of between 2% and 3%, for a strain rate of 0.75 s^{-1}.

4.7.5

This range covers the toe region of the stress-strain curve.

There appear then to be four ways of estimating the working stress range of a tendon. These are:

(i) A maximum safe stress of one quarter of the tensile strength.

(ii) The 'toe region' of the stress-strain curve.

(iii) The disappearance of the wave pattern.

(iv) Avoidance of permanent damage or residual strain.

Little is presently known of the effects of strain rate on these criteria.

4.7.6 The Zero of Tendon Tension

The technique used in this study to monitor tendon tension did not allow an absolute determination to be made. The tendon tension minima (least tension), although usually consistent, did not necessarily indicate zero tendon tension. The tension minima in Figures 3.7-3B, 3.7-4D and 4.6-1 are all repeatable, however, Figure 3.7-5D shows a tension minimum distinctly below all of the others in the figure.

In order to make an absolute measurement of tendon tension a technique similar to that described by Turner et al. (1975) for assessing zero bone strain would have to be used. This method involves recording from the horse under general anaesthesia, using a temperature compensated Wheatstone bridge. Additionally, either temperature compensated strain gauges would have had to be used on the buckle transducer, or the temperature of the transducer would have had to be maintained during calibration at that of the equine body.

4.7.7 Tendon Transducers

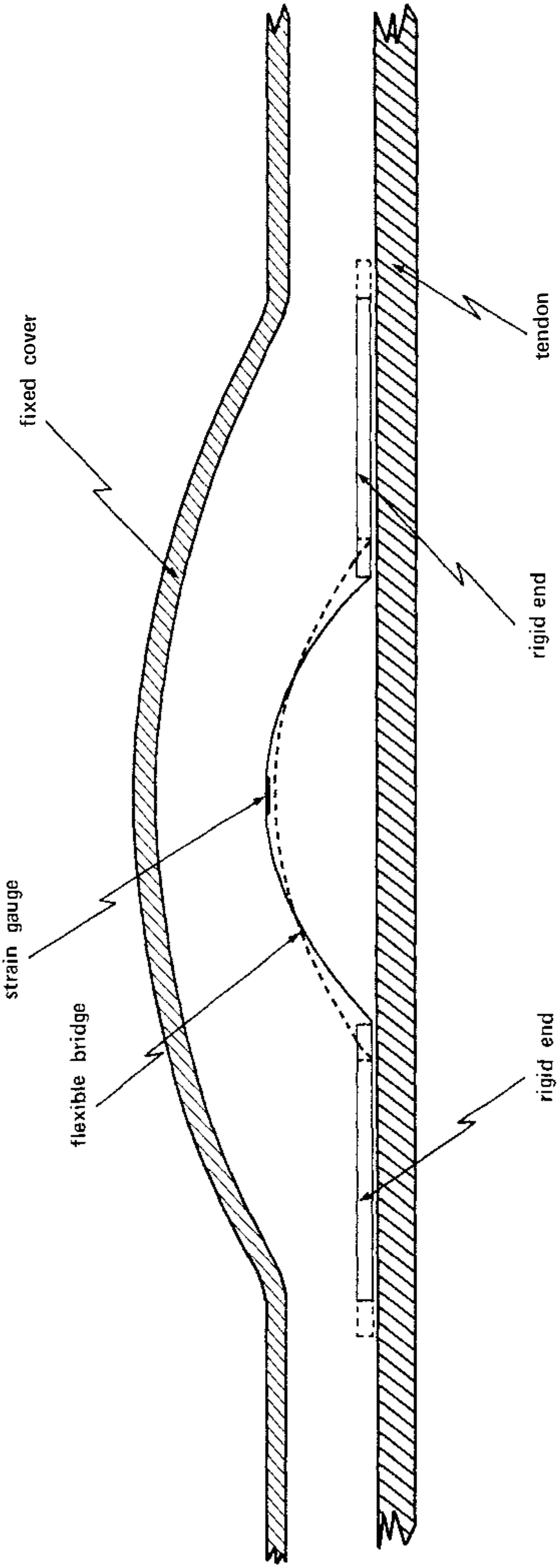
There is a growing interest in the experimental investigation of the *in vivo* mechanical function and behaviour of tendons, so it is pertinent for consideration to be given to the advantages and performance of each kind of tendon transducer. Barnes and Pinder (1974) have reported the design and _in vivo_ operation of a tendon tension buckle transducer similar to that proposed by Salmons (1969) while Kear and Smith (1975) have presented results obtained with a tendon strain transducer similar to those used for many years in other applications (Perry and Lissner, 1962; Aronson, 1971). In each case a long tendon of an animal limb was investigated. The elastic behaviour of a tendon under tension may not be described by a single elastic constant so the results obtained with a tendon strain transducer are not necessarily related to those obtained with a tendon tension transducer by a proportionality constant. However, the tendon is unlikely to be subjected to excessively large forces during normal body movement, and the experimental uncertainties associated with _in vivo_ measurements of this kind are sufficiently large that the elastic behaviour of the tendon as represented by the measurements may be described by a single elastic constant for slow gaits. Also the calibration of these tendon transducers engenders systematic errors, the causes of which may be readily identified but whose magnitude is difficult to assess.

The tendon tension buckle transducer, see Figure 4.2-1 measures the tendon tension by monitoring the force transmitted by the whole cross-section of the tendon, so there is little doubt that the force exerted by the muscle is measured. However, the transducer must be designed to present gently curved loading surfaces to the tendon to obviate tendon damage. This in turn implies that the device has to be calibrated. There are at least two methods for accomplishing this calibration.

The application of the tendon tension transducer also introduces a systematic error, for the implantation of the device reduces the effective length of the tendon. Consequently the device measures the tension in the tendon when the associated

FIGURE 4.7-2

Schematic diagram of the strain gauge arch (or clip gauge) for measuring large strains (or displacements). The rigid ends are bonded and/or sutured to the tendon. When the tendon is stretched the rigid ends and flexible bridge could occupy the positions indicated by the dashed lines. The strain gauge would then be stretched. The cover protects the clip gauge from surrounding flesh and may be secured to bone.


fixed cover
tendon
rigid end
strain gauge
flexible bridge
rigid end

muscle is abnormally extended and the relationship between the observed values of tendon tension and those existing under normal conditions is not known. The linearity of the device is not in question but the appropriate zero tension reference level is not known. A typical tendon tension transducer record is shown in Figure 4.6-1 the precise location of the base line from which the tendon tension should be measured is not known. Consequently the tendon tension transducer can be considered to monitor rather than measure the tendon tension in a normal walking animal.

The mode of action of the tendon strain transducer is illustrated in Figure 4.7-2, this transducer also suffers from systematic errors. Firstly homogeneous longitudinal strain caused by tendon tension cannot be distinguished from bending strain caused by tendon flexing, consequently the transducer must be protected by a rigid box from external lateral forces which could cause the tendon to flex and so engender spurious tendon strain measurements. Secondly the whole active site of the transducer must firmly adhere to the tendon throughout the whole period of implantation, such a requirement can prove difficult to meet. Thirdly the transducer samples strain in only one side of the tendon and this could be limited to merely a narrow surface region in the immediate vicinity of the transducer. Moreover the presence of the transducer firmly adhering to the tendon locally strengthens the tendon causing the measured strain to be less than that pertaining to normal conditions. The tendon strain transducer can be calibrated by applying the techniques used to calibrate the tendon tension transducer.

Measurements obtained from either the tendon tension or the tendon strain transducer suffer from systematic uncertainties which can render the magnitude of the measurements uncertain. Consequently these tendon transducers can be considered to monitor rather than measure tendon elastic properties in a normal walking animal since the transducers yield an uncalibrated record of tendon tension or tendon strain against time. Also both the transducers are bulky and so they can be used only in the study of tendons which are long and readily accessible.There would be some interest in applying both transducers to the same tendon simultaneously however, such a procedure may not prove to be possible because of the large volume of each transducer.

Recent studies have indicated that the averaged rectified EMG signals from the humeral head of the common extensor muscle of the forelimb of a horse can be used to obtain an uncalibrated record which is a remarkably faithfull reproduction of the tendon tension trace (or possibly the tendon strain trace). The data processing circuitry is shown in Figure 5.4-4. Figure 5.6-1 compares the averaged rectified EMG trace with that obtained from a tendon tension transducer attached to the common digital extensor tendon. This clearly indicates that the mechanical behaviour of the tendon may be monitored by simply processing the high frequency information contained in fine wire EMG signals from the appropriate muscle. Hence the electrical activity of a muscle can be considered as a probe for monitoring the mechanical behaviour of the associated tendon.

This is an important development for it obviates the need for the implantation of bulky tendon transducers under general anaesthesia, thereby rendering more tendons susceptable to study. However, experiments need to be performed on other tendon systems to verify that high correlation between the tendon tension record and processed EMG record is a general relationship.

4.7.7

CHAPTER 5

ELECTROMYOGRAPHY

5.1 INTRODUCTION

There is a growing interest in electromyographic kinesiology in which the EMG signal is used to study muscular function and co-ordination. An important part of such research is the estimation of muscular force and its correlation with the EMG signal.

A major difficulty with *in vivo* locomotor studies is that it is usually necessary to calculate the muscular force using assumed positions of the various anatomical parts together with a force measured external to the limb. Partly for these reasons, experiments have often been restricted to isometric muscular contractions.

In the case of experimental animals, the availability of the tendon tension 'buckle transducer' now makes such calculations redundant since it provides a direct measure of the muscle force with which a processed EMG signal may be compared. Furthermore, the experimental animal may perform normal activities without the need to adopt isometric, isokinetic or isotonic modes.

The identification of a satisfactory method for deducing muscular force from EMG data would benefit studies of sports medicine, patient rehabilitation and myoelectric control of prostheses.

5.2 CHOICE OF ELECTRODE TYPE

Electrocardiogram surface electrodes were used in the initial phases of this investigation but it proved difficult to obtain good electrical contact with the skin of the walking horse. It was also difficult to maintain the electrodes in a fixed position on the skin.

However, even if these difficulties had been overcome, two serious problems would have remained. Firstly, since the skin and the underlying muscles move with respect to one another while the animal is walking, the electrodes would record from

FIGURE 5.2-1

Schematic diagram showing the tapered end of the concentric needle electrode used in electromyographical studies of the equine larynx. Both the inner and outer conductors probe electric potentials within a muscle.

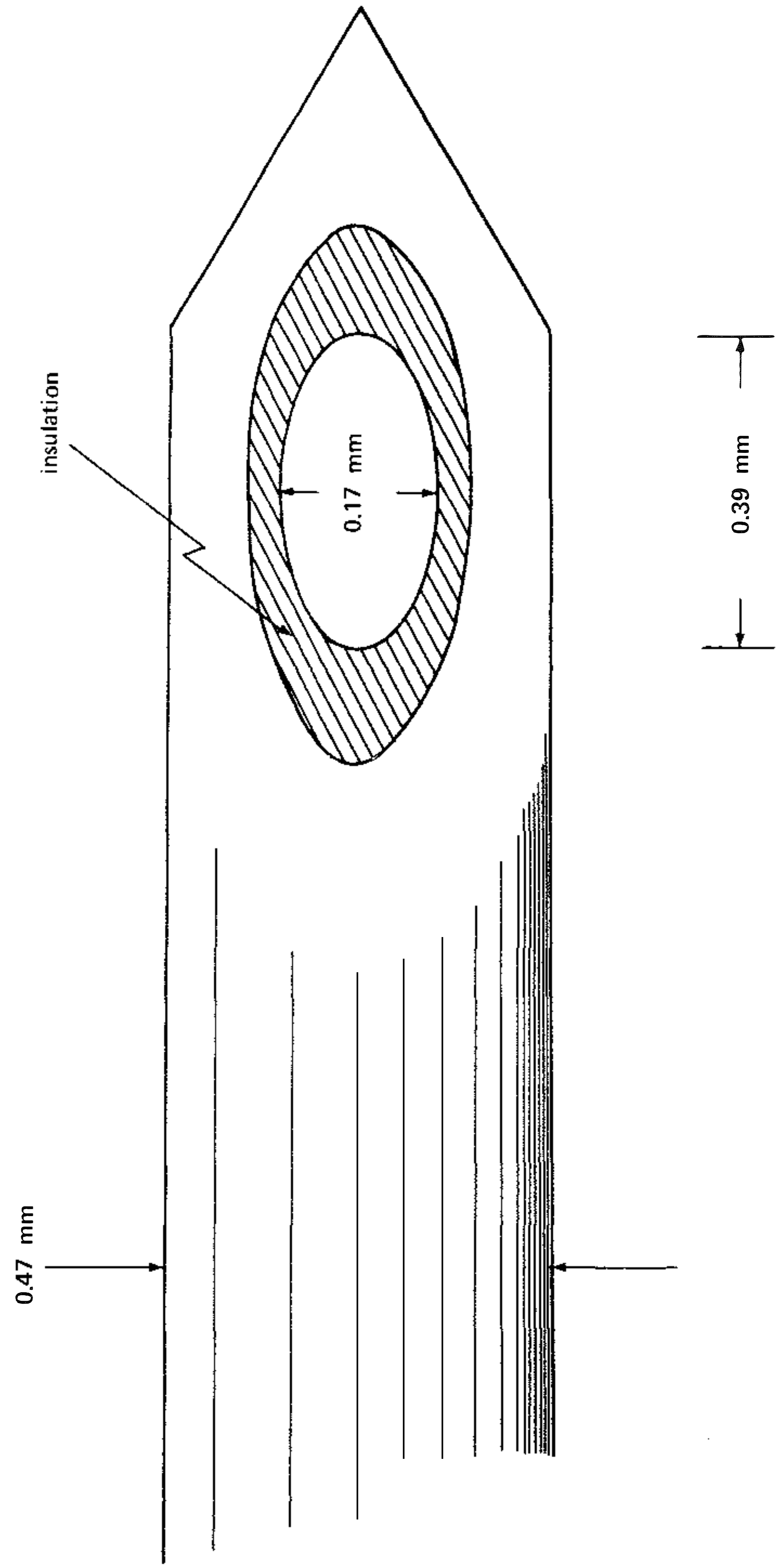
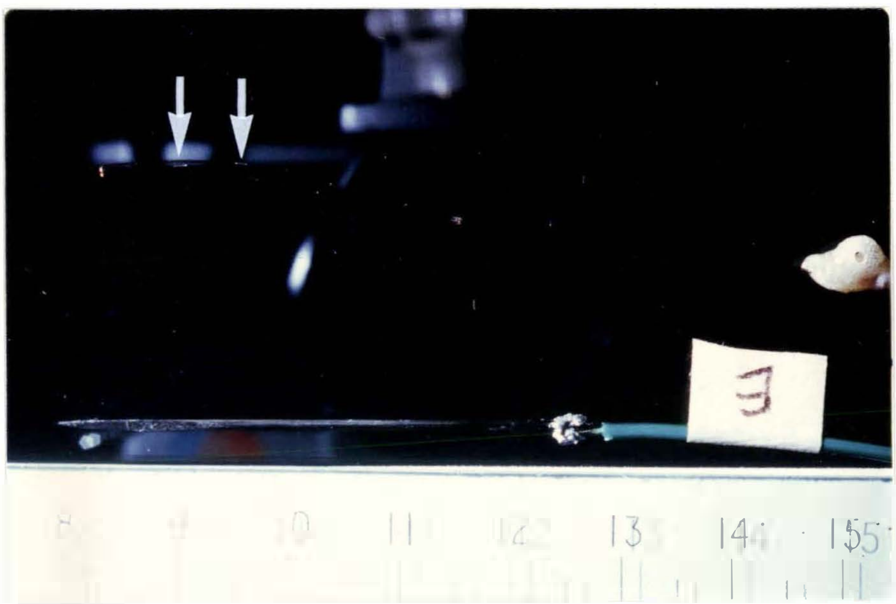

insulation
0.17 mm
0.39 mm
0.47 mm

FIGURE 5.2-2

Photograph revealing the fine wire barbs, each with an electrode (arrowed) at its end.



different parts of the muscle during a stride. Secondly, the surface electrodes could pick up electrical activity from muscles other than the one intended. These additional problems would have made interpretation of the recorded EMG difficult.

A concentric needle electrode (Figure 5.2-1) was used in some trial experiments on electromyography of limb muscles. The electrode was partially inserted into a hypodermic needle which was then used to puncture the animal's skin and muscle sheath. The recording electrode was inserted into the muscle by pushing the concentric needle further into the hypodermic needle.

Unfortunately, the concentric needle electrode was slowly ejected from the hypodermic needle by the contractions of the limb muscle as the animal walked. Moreover, the muscle could have been unnecessarily damaged by movement of the needle as the skin moved relative to the muscle.

The type of electrode ultimately employed for limb muscle monitoring consisted of fine (0.1 mm diameter) stainless steel wire implanted in the muscle. This single strand wire was coated with varnish, a portion of which was scraped off to form an electrode.

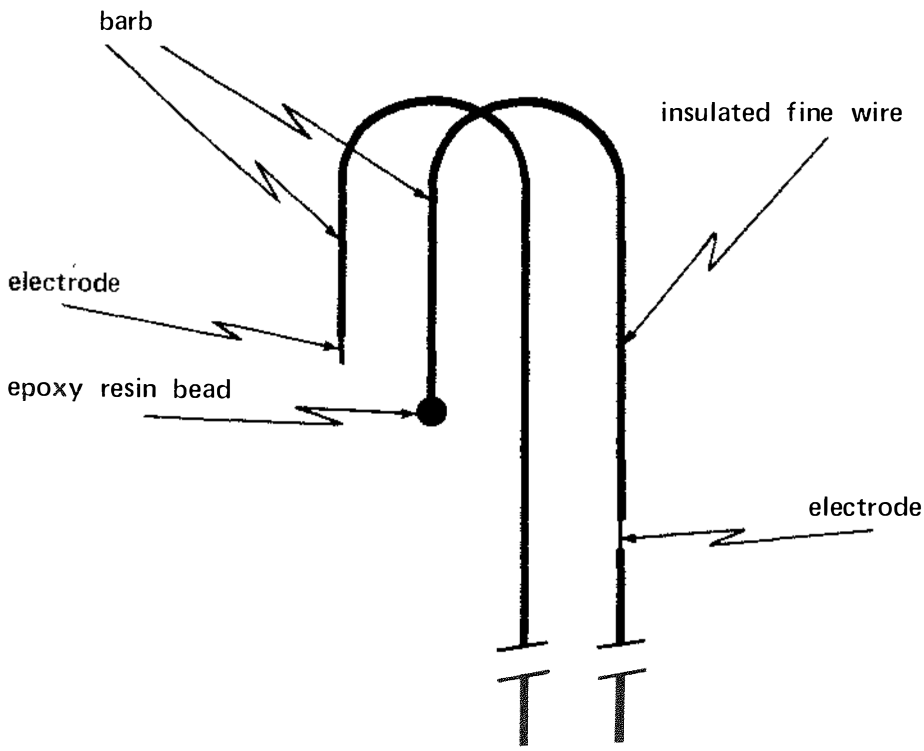
In an initial trial on a sheep, the fine wire was sutured into the muscle as suggested by Kear and Smith (1972). However, it was possible for the wire suture to rotate, so causing the bared wire (electrode) to slide, possibly even out of the muscle. Of equal concern was the failure to record any EMG activity because of fracture of the fine wire where it was sutured into the muscle. This method of positioning fine wire electrodes was subsequently abandoned. In its place the following method was used. A pair of fine wires were hooked over and into the end of a hypodermic needle, the wires lying along the outside of the shaft of the needle. This assembly was then inserted into the muscle and the needle withdrawn to leave the wire barbs (Figure 5.2-2) to secure the electrodes in the muscle.

FIGURE 5.3-1A

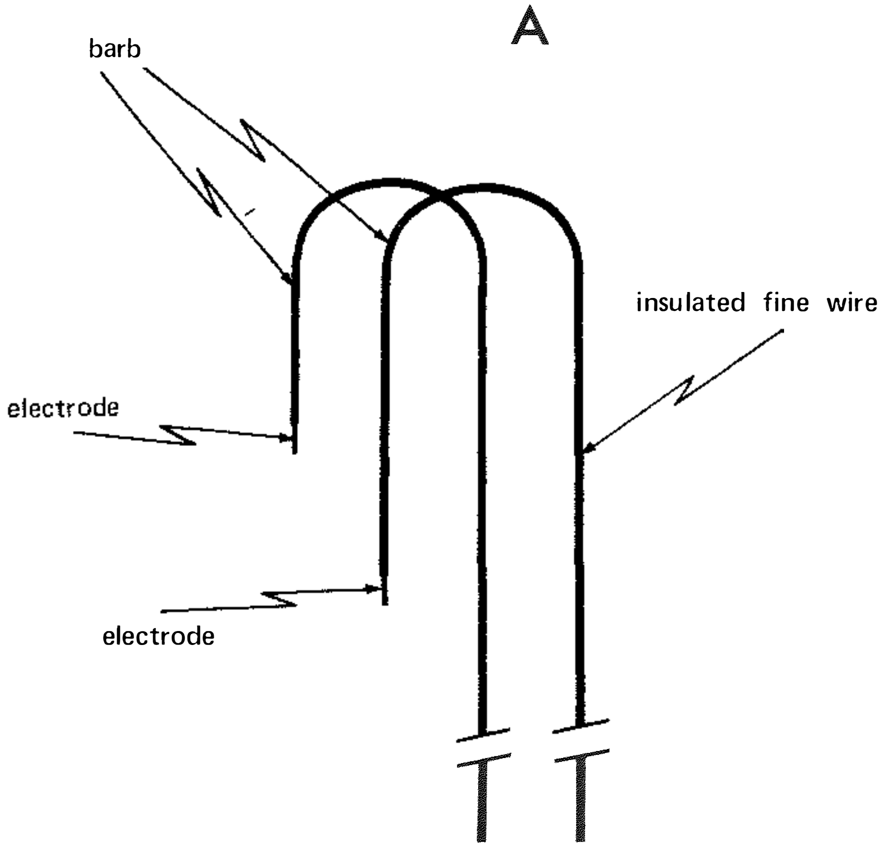
Schematic diagram of the fine wire electrode unit first used. The electrode length and separation were 1 mm and 3 mm respectively.

FIGURE 5.3-1B

Schematic diagram of the type of fine wire electrode unit subsequently used. Electrode dimensions as above.


barb
insulated fine wire
electrode
epoxy resin bead
electrode

A


barb
insulated fine wire
electrode
electrode

B

5.3 MANUFACTURE OF WIRE ELECTRODE UNIT

The unit consisted of two strands of fine wire each soldered to an inner core of light, twin-core, shielded cable (Figure 2.5-1). The shield and cores were soldered to the insert of a three pin DIN plug, as was the single core wire from the reference (or ground) needle.

Polyurethane varnish (Trimmel) coated stainless steel wire of 0.1 mm diameter (Johnson Matthey Metals, London) was used for each intramuscular wire. An electrode surface was formed by removing the Trimmel coating from a small length of this wire. The end of each wire was bent back on itself to form a barb. Both barbs could then be introduced into the tip of a hypodermic needle which, when inserted into a muscle and withdrawn, would leave the wires within the muscle.

It was essential that the bared segments of the wires did not touch when in the muscle, for then no EMG would be recorded as the signal would be shorted.

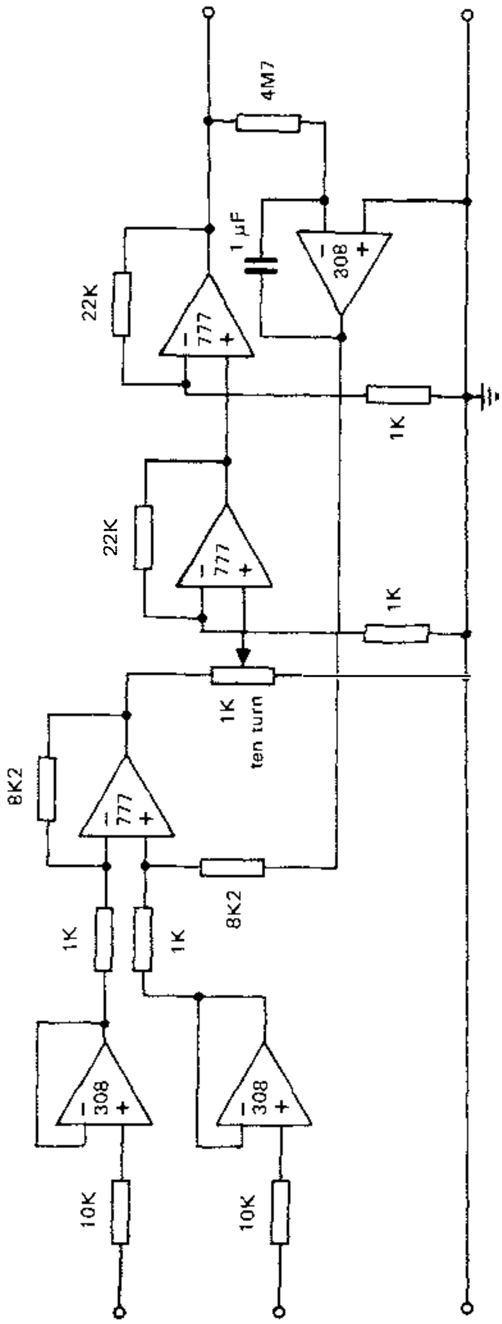
Two wire electrode configurations were designed in an attempt to obviate shorting. Figure 5.3-1 illustrates the two configurations employed. The first design (Figure 5.3-1A) had barbs of almost equal length, with an electrode on the end of one barb, and a minute bead of 'Araldite' epoxy resin on the tip of the other. The other electrode did not overlap either barb. This design was abandoned chiefly because the use of the epoxy resin bead necessitated a hypodermic needle which could otherwise have been smaller. A smaller needle would be expected to cause less muscle damage during implantation of the wire electrodes.

The second design (Figure 5.3-1B) had barbs of different lengths with a 1 mm electrode at the end of each. The barb lengths differed by about 4 mm.

The stainless steel wires were soldered to twin core shielded cable and the joints examined electrically and under a microscope.

FIGURE 5.4-1

Circuit diagram of the differential amplifier built to amplify myoelectric signals. Resistor values are in ohms.


4M7
1 µF
308
777
22K
22K
1K
1K
777
1K
ten turn
8K2
777
8K2
1K
1K
308
308
10K
10K

Joint insulation was accomplished by coating with 'Araldite' epoxy resin which also sealed the shield of the cable. This epoxy resin moulding had a small hole drilled in it, thus facilitating percutaneous suturing of the mould to the skin.

In order to lessen chances of fatigue fracture of the cable where it would emerge from the skin, some surgical 'Portex' sleeving was used to increase stiffness and protect the cable.

5.4 MEASURING AND RECORDING APPARATUS

A high gain, low noise, differential amplifier sensed the differential voltage between the two intramuscular electrodes. The common electrode, a cutting suture needle, was embedded in the neck several decimetres away in a region remote from myoelectrical activity.

A schematic diagram of this amplifier is given in Figure 5.4-1 and shows the gain control (multi-turn potentiometer) which allowed the overall gain of the amplifier to be varied between zero and 4 000. The 3 dB points in the frequency response of the amplifier were at 17 Hz and 12 kHz.

The output of this amplifier was then recorded on a frequency modulated tape recorder (Philips Mini-log 4 Analog Cassette Recorder). This had a frequency response of zero Hz to 5 000 Hz.

The EMG activity was monitored acoustically (using a power amplifier and speaker) and visually (using an oscilloscope).

Simultaneous recordings were made of bone strain and gait (Chapter 3), tendon tension (Chapter 4), and EMG.

The tape recordings were replayed into an ultra violet oscillograph equipped with galvanometers having a frequency response range from zero to 5 kHz. The galvanometers wrote with an 'optical line segment' rather than a spot, causing horizontal lines to be wide but vertical lines to be thin.

FIGURE 5.4-2

Block diagram showing the system used to record data during electromyographical studies of the equine larynx.

DATA ACQUISITION SYSTEM

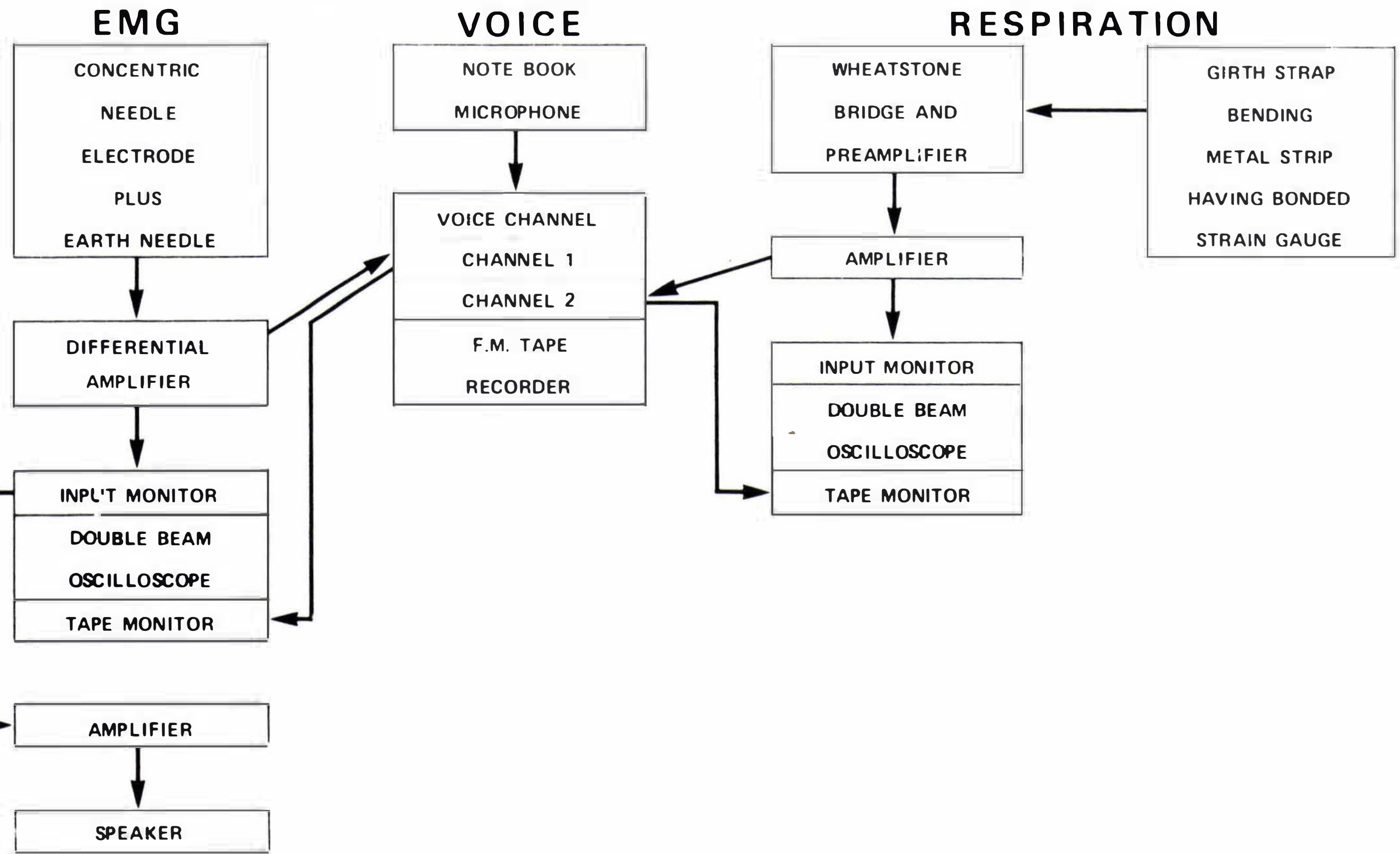


FIGURE 5.4-3

Block diagram showing the data retrieval system used in equine laryngeal studies.

DATA RETRIEVAL SYSTEM

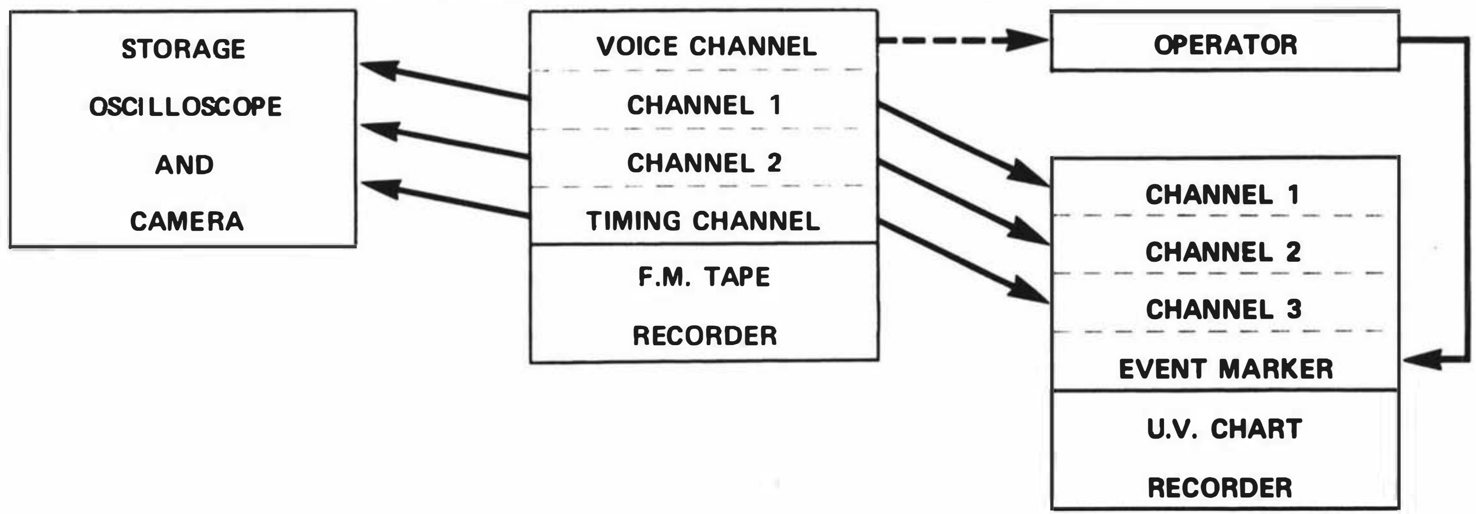


FIGURE 5.4-4

Schematic diagram of the system used to full-wave rectify and average the EMG signal. The averager is that reported by Garland et al. (1972) but with each capacitor chosen to be 0.33 μF.

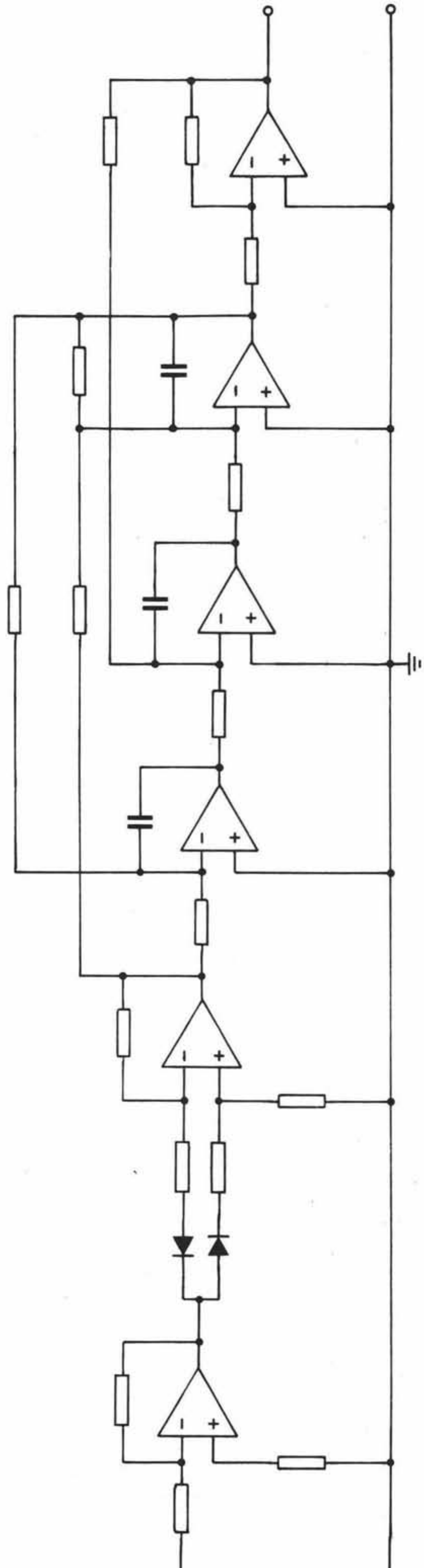


FIGURE 5.5-1

Circuit diagram of the simple potential divider used to calibrate the EMG amplifier. The voltage source was a single Hg cell.

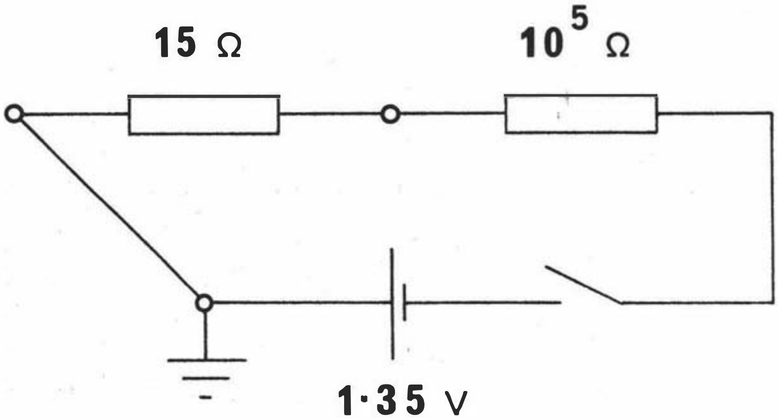
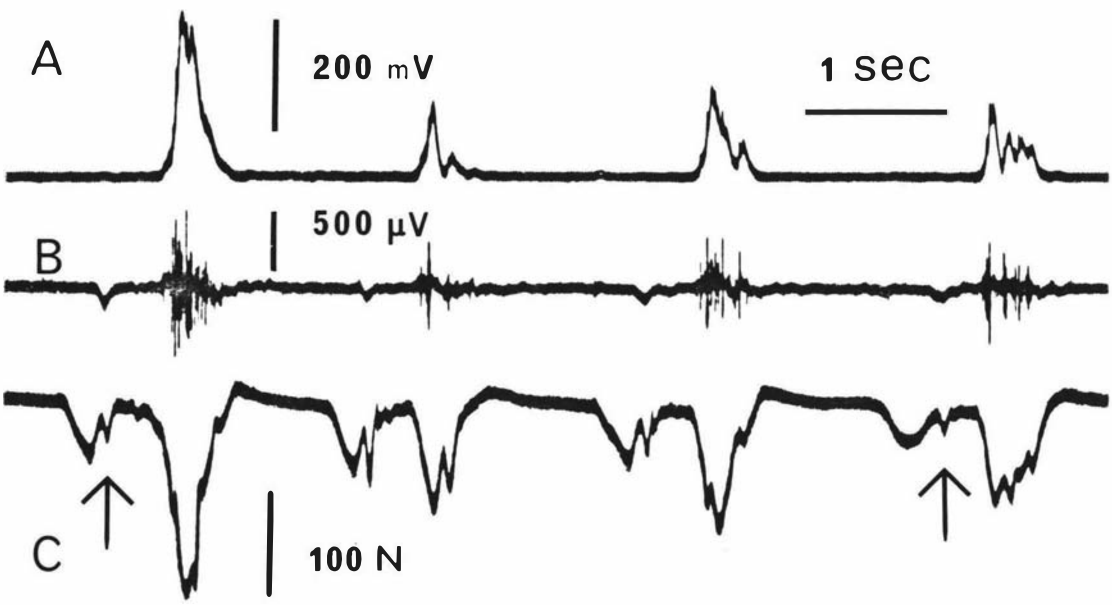

15 Ω
10⁵ Ω
1·35 V

FIGURE 5.6-1

Traces A and C are data previously presented as E and B in Figure 3.7-3. Trace B is an electromyogram recorded from the humeral head of the common digital extensor muscle.

The frequency response of the ultra violet oscillograph used extended from zero to 5 kHz.


A
200 mV
1 sec
B
500 µV
C
100 N

The recording and playback systems used in the laryngeal studies are summarized in Figures 5.4-2 and 5.4-3. The process was very similar for both limb and laryngeal EMG's. The storage oscilloscope and camera were for examination of single motor unit potentials. It is hoped that laryngeal electromyograms may be further characterized by such examination.

In order to quantify the limb EMG, a full wave rectifier and averaging filter were built (Figure 5.4-4). These were preceded by a low gain amplifier to boost the signal from the tape recorder. The tape recorder and amplifier were coupled by a 0.33 μF capacitor which formed part of a high pass filter having a lower 3 dB point of 80 Hz.

The transfer function of the averaging filter (Garland, et al., 1972) was a third order Padé approximation to that of an ideal averaging filter. The settling time was chosen to be 66 ms, about one half the width of the smallest tendon tension peak expected to be resolved. This settling time was implemented by choosing the three capacitors to each be 0.33 μF.

5.5 CALIBRATION

A very simple device (Figure 5.5-1) was used to provide a low amplitude signal for calibration of the EMG differential amplifier. A mercury cell provided a stable voltage which was divided by a resistor network (15 ohm and 0.1 megaohm in series). The 15 ohm resistor connected the common input to one side of the amplifier, the other side was shorted to the common input. The calibration assembly, which was all mounted on a DIN plug, included a miniature microswitch which enabled a square wave of about 200 μV to be generated.

5.6 RESULTS

The EMG signal (Figure 5.6-1B) recorded from the humeral head of the common digital extensor muscle correlates with those peaks of the common digital extensor tendon tension trace

(Figure 5.6-1C) which occur near the end of the swing phase (Figure 3.7-3D) of the leg. The small troughs (arrowed) on the EMG trace are not the result of electrical activity in the humeral head but are caused by movement of the electrodes within the muscle. These troughs correspond with sharp peaks, also arrowed, in the tendon tension trace (Figure 5.6-1C).

The averaged rectified EMG trace (Figure 5.6-1A) was derived from the EMG signal (Figure 5.6-1B) and recorded on light sensitive paper together with the EMG and tendon tension data.

5.7 DISCUSSION

5.7.1 Electrode Wire

The method of insertion of the EMG fine wire electrodes in the muscle was discussed and rejected by Osse et al., (1972). They found that the wires they employed did not readily attach to the muscle when the hypodermic needle was withdrawn. Osse et al., used very fine wire of 20 and 50 µm diameter in addition to the 100 µm diameter wire which was used in the present study. Our surgeon experienced no difficulty in obtaining a secure electrode implant, possibly due to a better technique. However, it is possible that the difficulties described by Osse et al., were mainly associated with the finer wires.

5.7.2 Tendon Tension - Processed EMG Correlation

In any length of record, see Figure 5.6-1, there is good correlation between processed EMG and large tendon tensions but poorer correlation for low tensions. In other words there is good correlation between tendon tension and averaged rectified EMG activity during periods of EMG activity but poorer correlation at other times. It must be remembered that the

EMG trace was recorded from the humeral head of the common digital extensor muscle (Figure 2.2-6), therefore the processed EMG (Figure 5.6-1A) applies only to this head of the muscle. Comparison of Figures 5.6-1A and C will, it is suggested, indicate the contribution to common digital extensor tendon tension (trace C) made by the humeral head of the corresponding muscle. This is a significant step beyond using activity evident in the EMG trace to indicate merely when the particular muscle (or muscle head) was active. Such information would be had by comparison of Figures 5.6-1B and C.

If, when trace B indicates muscular activity there exists good correlation between processed EMG (trace A) and tendon tension (trace C) then it may be assumed that the humeral head of the muscle produces most if not all of the active tension sensed by the buckle transducer on the tendon. To be quite sure on this point, processed EMGs from the other heads of the muscle would ideally be available also. However in this preliminary study it was not deemed essential to monitor all the heads of the common digital extensor muscle. Indeed because of the anatomy of this muscle (see Section 2.2.2) the surgeon would have had some difficulty in placing electrodes securely within the other heads. Such difficulties however would not be insurmountable. In the absence of data from the other heads we must make cautious deductions on the available evidence as presented in Figure 5.6-1.

For the present we will restrict our attention to correlations between traces A and C only during periods when trace B indicates that the humeral head of the muscle is active. Inspection of traces A and C reveals a high degree of correlation between them during such periods. It is encouraging that the gross features of trace C are evident in trace A.

An even greater degree of correlation between tendon tension and averaged rectified EMG would have been possible had the averaging period been slightly greater. It is evident from Figure 5.6-1 that the processed EMG trace is less smooth than the corresponding tendon tension trace, so that the features of the former are enhanced. These features would be smoothed by extending the averaging period beyond 66 ms. The electronic averaging process might then simulate more precisely the effect of the muscle - tendon system in mechanically averaging the tensions developed throughout the humeral head of the muscle.

5.7.3 Correlation at Low Tensions

Comparison of Figures 5.6-1A and C shows that the clusters of (inverted) peaks in trace C span a broader time interval than the corresponding clusters in trace A. Figure 5.6-1 also shows that the onset of tendon tension precedes the onset of the myoelectric activity revealed in trace B. One would expect the reverse to be true (see Bendall, 1969 Figure 6.2), as tendon tension is primarily a consequence of active muscular tension which accompanies myoelectric activity. There are at least two possible reasons for this effect, one concerning the electrodes and the other associated with the EMG processing equipment.

The intramuscular wire electrodes used sensed electrical activity from a relatively small volume of muscle (Komi et al., 1970; Rogoff et al., 1961). Furthermore the electrodes were sensitive to the higher frequency components of the myoelectric activity because of the small interelectrode spacing. It is thus possible that the recorded EMG omitted signals from more remote motor units which may have been functioning at low levels of tendon tension. Such omission would of course deny the processed EMG the opportunity of being non zero at low levels of tendon tension.

The EMG recording system is well suited to respond to the larger motor unit potentials of rapid time course which represent the activity of large, high - threshold motor units. These are recruited only at the higher levels of force generation (Olson et al., 1968); Milner-Brown et al., 1973b, 1973c, 1975).

Even if the recorded EMG did include low levels of activity, the contribution of this to the processed EMG may have been inhibited by a deficiency in the simple full - wave rectifier employed (Figure 5.4-4). The diodes used in this rectifier had finite resistance and non linear temperature-sensitive switching characteristics contributing to a 'rounding' of the turn - on region of their voltage - current characteristics. A forward bias of 0.3 V was required for the germanium diodes used. The amplifier preceding the rectifier was included to increase the signal amplitude so that the forward bias voltage would be only a small fraction of the largest signal processed.

A better solution would have been to use a precision limiter with two diodes in the feedback loops of an operational amplifier (Tobey et al., 1971). The high open loop gain of this amplifier reduces the effect of the diode non-linearity and temperature sensitivity so that the rounding of the turn - on region virtually disappears.

Pressure of time prevented this more satisfactory full wave rectifier from being perfected and used.

5.7.4 The Minor Tendon Tension Peaks

The measurements displayed in Figures 3.7-3 and 5.6-1 are particularly interesting because they are not merely a repetative signal from similar strides, in this sequence each stride is different. The correlation between the tendon tension trace and the averaged rectified EMG signal is significant. Even in this case with non-repetitive signals the correlation is very good for the large tendon tension peaks but there is no correlation for the group of minor tendon tension peaks. This effect is not entirely unexpected in view of the anatomy of the common extensor system. There are at least two possible explanations.

The first is that the minor tendon tension peaks are due to contractions of the small ulnar head of the muscle whose delicate tendon joins the common extensor tendon and is embraced by the tendon tension buckle transducer. Hence the buckle transducer detects the tension developed in the small ulnar head as well as that developed in the humeral head whereas the EMG electrodes are implanted in the humeral head of the muscle and so do not sense electrical activity in the remote ulnar head.

The second explanation is that force developed in the radial head is partially transmitted to the common extensor tendon. Significant mechanical coupling between muscle heads may be possible especially since there are no intramuscular septa forming sheaths for the individual heads of the muscle. However, the EMG wire electrodes are not expected to be sensitive to the electrical activity engendered in the radial head of the muscle. Neither of these two explanations can be discounted, indeed since they are not mutually exclusive both explanations could be operative concurrently.

5.7.5 Movement Artefact

The EMG recording shows movement artefact when the second minor tendon tension peak occurs, see Figure 5.6-1. Possibly this was caused by movement of the insulated sections of the EMG wires as they passed through or near the radial head of the muscle. Thereby indicating that the radial head of the muscle is active during the second minor tendon tension peak. Even a small amount of base line movement in the EMG recording would have a marked effect on the averaged rectified signal if d.c. coupling, were employed. Since there was a degree of base line movement present, the EMG output from the f.m. tape recorder was coupled to the rectifier through a 0.33 μF capacitor. This capacitor helped to constitute a high pass filter with a lower 3 db cut-off frequency of 80 Hz and so obviated the effects of movement artefacts.

5.7.6 Filtering Effects

The use of the high pass filter to remove the effects of electrode movement artefacts must be examined. This filter rejected frequencies below 80 Hz and it is pertinent to consider this in relation to the frequencies present in EMG signals.

Several studies of EMG spectra have been published, but it is important to bear in mind the types and dimensions of electrodes employed. The common surface electrodes and long uninsulated intramuscular wire electrodes explore large volumes of muscle and generally are not very receptive to high frequency components of myoelectric activity. Scott (1967), using such wire electrode, found that the gross myoelectric signal had significant energy only in the frequency range 30 Hz to about 200 Hz, with peak energy at around 80 Hz. Similar results were obtained using surface electrodes (Hayes, 1960).

Electrodes which have small separations generally record from only a small volume and respond to higher frequency components than do those described above. Concentric needle electrodes (as used in the laryngeal studies) and the fine wire electrodes used in the locomotion studies are thus more sensitive to the higher frequencies. Gersten et al., (1965), using concentric needle electrodes inserted into the human abductor digiti quinti muscle, found a mean peak frequency of about 160 Hz, and that the spectrum extended to 840 Hz on average. Fex and Krakau (1957) used bipolar coaxial needle electrodes in spectral analysis of EMGs from the large human skeletal muscles. In each case they found a uniquely large peak at 200 Hz in the EMG spectra.

It is therefore suggested that the frequencies below 80 Hz which were rejected by the high pass filter do not represent a great loss to the processed EMG.

During forcible muscle contractions, Fex and Krakau (1957) found that frequencies in the 40 - 50 Hz region were enhanced, and explained this by assuming a certain degree of synchronous activity in the motor units. It is unlikely that during the walking gait, the digital extensor muscles would need to exert such forcible contractions. In any case, good tendon tension - processed EMG correlation was evident at the higher tension levels.

5.7.6

5.7.7 Inadequacies of the EMG Signal

It is difficult to obtain a true representation of the overall myoelectric activity by using conventional wire electrodes. This is because the motor units which constitute the muscle are of varying size and have properties such as contractile speed, fatiguability and threshold of recruitment which differ. Furthermore, the distribution of the motor units is not entirely random, so that a portion of muscle is not necessarily representative of the whole. Development of a new type of intramuscular electrode might alleviate this problem. Nevertheless, the type of electrode employed in this study has given encouraging results.

Even if it were possible to sample the muscle adequately with EMG electrodes the myoelectric activity does not bear a fixed relation to tension. The relationships among muscle force, speed of contraction and integrated electrical activity have been studied by Bigland and Lippold (1954a). These authors, using surface electrodes applied over human calf muscles, found that the integrated electrical activity in a muscle is directly proportional to the tension it exerts for constant (or zero) rate of change of muscle length.

Bouisset (1973), citing Vredenbregt et al., (1966), showed a curvilinear relation between integrated EMG and isometric force for human elbow flexors, different curves corresponding to different elbow angles. There appears to be some inconsistency between the above results at least for zero rate of change of muscle length (isometric conditions). A further complicating factor is muscle length which affects both passive and active tension (Inman et al., 1954), total muscle tension being the sum of the two.

This evidence seems to suggest that one should monitor the EMG signal together with muscle length and rate of change of muscle length. Schemes for doing so were devised but not developed as we wished to determine how well the processed EMG alone indicated muscle tension during normal unrestrained voluntary movement.

The results obtained (Figure 5.6-1) are sufficiently encouraging for the monitoring of these additional paramaters to appear to be unnecessary in providing good estimations of *in vivo* muscle tension.

The phenomenological approach used in this study appears to be justified in view of the complexities inherent in the electromechanics of muscle. Even if the additional parameters had been monitored it is not immediately obvious how they would have been combined with the processed EMG to (hopefully) yield even better estimates of muscle tension.

A further shortcoming of the EMG signal is that when a sudden, unanticipated flexion or extension is applied externally to a joint, high levels of strain may be developed in bones and tendons during the electrically silent period before the muscles are activated reflexly. This situation, although an important one in relation to the biomechanics of injury, was not considered part of the present study.

5.7.8 Conclusion

The 'buckle transducer' is an adept device for monitoring *in vivo* tension in long accessible tendons during normal voluntary movement. The device is also well suited to assess the usefulness of other methods of indicating tendon tension. In this study the 'buckle transducer' was used to assess the use of the averaged rectified EMG signal as an indicator of muscle force during normal voluntary movement in the horse. Although not exhaustive, the evidence presented indicates that the EMG, processed in the manner described, does give a good reproduction of the relevant regions of the tendon tension record. This is in spite of no account being taken of instantaneous muscle length or rate of change of length. It is therefore possible that the processed EMG may be used to produce a 'muscle tension' record in situations where it would be inconvenient or impossible to attach a 'buckle transducer'.

This is not to say that the processed EMG should supplant the 'buckle transducer' as a method of monitoring tendon tension but may supplement it. Direct methods of monitoring tendon tension are to be preferred to indirect ones when a choice exists.

5.8 LARYNGEAL ELECTROMYOGRAPHY

An opportunity to develop and gain experience with an electromyographic facility arose through research on the equine larynx.

The EMG recording system used was based on an Epsylon Labcorder frequency modulated tape recorder having two data channels, a voice channel and a timing channel which was seldom used. Data retrieval was accomplished by subsequently connecting the tape recorder to an ultra violet chart recorder.

Several joint research projects ensued, chief of which was a study of the normal and abnormal electromyographic patterns of the intrinsic laryngeal muscles in the anaesthetized horse. We wished to increase understanding of the action of the individual laryngeal muscles and to provide a basis on which to compare the normal animal with that suffering from myogenic or neurogenic laryngeal disorders. Of particular interest were horses exhibiting evidence of recurrent laryngeal nerve paresis.

There had apparently been no previous EMG study of the equine larynx, which is surprising since of all the domestic species, the horse is most prone to laryngeal muscular disease (Duncan et al., 1974).

The study of the normal electromyographic pattern used fourteen horses obtained from a local knackery. The intrinsic laryngeal muscles on the right side were selected for examination since they are rarely afflicted with the common pathological changes found in the left intrinsic laryngeal muscles (Duncan et al., 1974). The animals were placed in left lateral recumbency, and a skin incision made immediately ventral and parallel to the external maxillary vein extending forward for 10 cms from the sterno-mandibular muscle of the neck.

An obliquely cut concentric needle electrode (Figure 5.2-1) was used for all of the recordings on the anaesthetized horses.

This had an external diameter of 0.47 mm and central conductor revealing a 0.39 mm x 0.17 mm ellipse at the tip. Since the surgical procedures adopted did not require the needle to pass through one muscle to record from another, the use of a bipolar needle electrode was not required.

The common electrode connected to the ground of the difference amplifier was a suture needle implanted in the skin of the horse over the posterior extremity of the horizontal ramus of the mandible.

Recordings were made of the electromyographic activity in the crico-thyroid and dorsal cricoarytenoid muscles.

The sulcus between the crico- and thyro-pharyngeal muscles was carefully opened by the surgeon so that the concentric needle electrode could be inserted directly into the lateral cricoarytenoid, the transverse arytenoid and the vocalis and ventricularis muscles. Recordings were obtained from up to three separate positions in each muscle.

An important part of the study of equine laryngeal muscle function was to determine the work cycle for each muscle in relation to respiration. To this end, experience was gained with a number of respiratory monitors.

A thermistor placed in the intratracheal tube was used initially. This had two disadvantages :

Firstly, the anaesthetic equipment employed a to and fro rebreathing system allowing the air temperature of this system to stabilize.

Secondly, when the intratracheal tube was removed to facilitate observation of the larynx through a laryngoscope, the respirations could not be monitored by the thermistor.

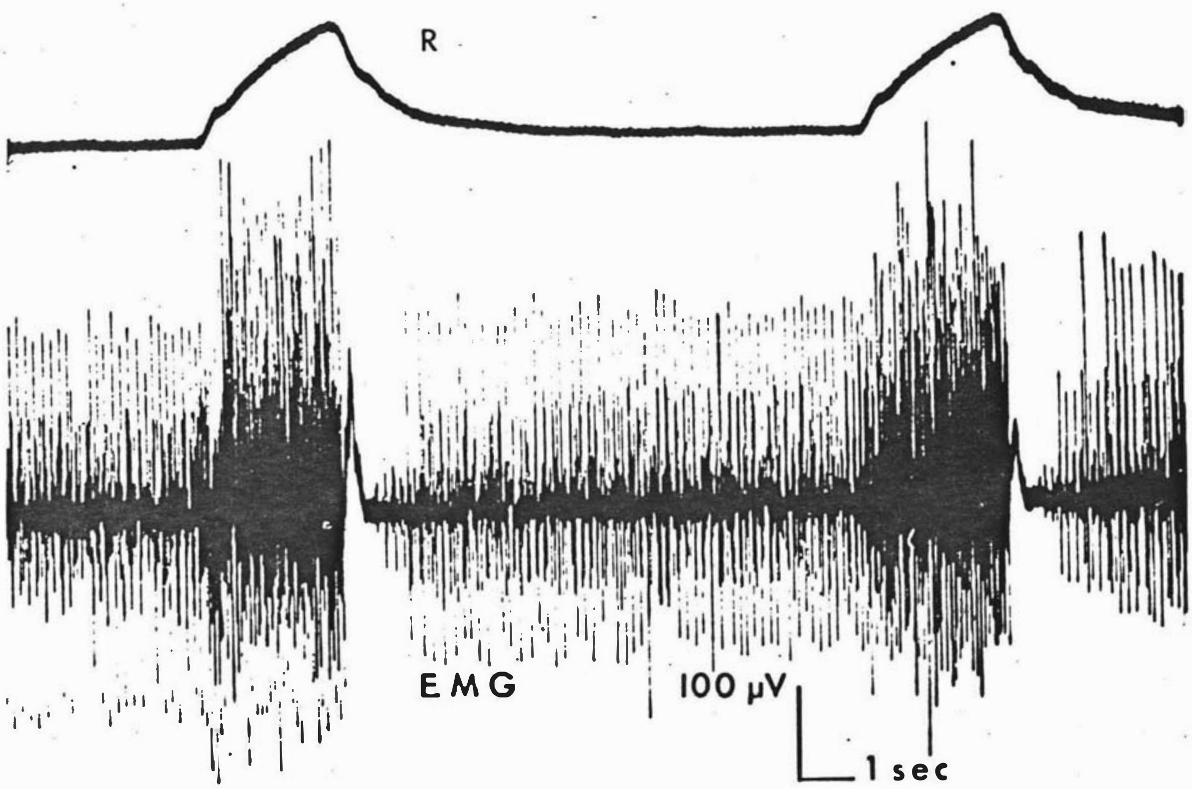
The most successful arrangement for monitoring respiration consisted of a belt surrounding the horse's girth and table. The belt included an elastic section fashioned from rubber

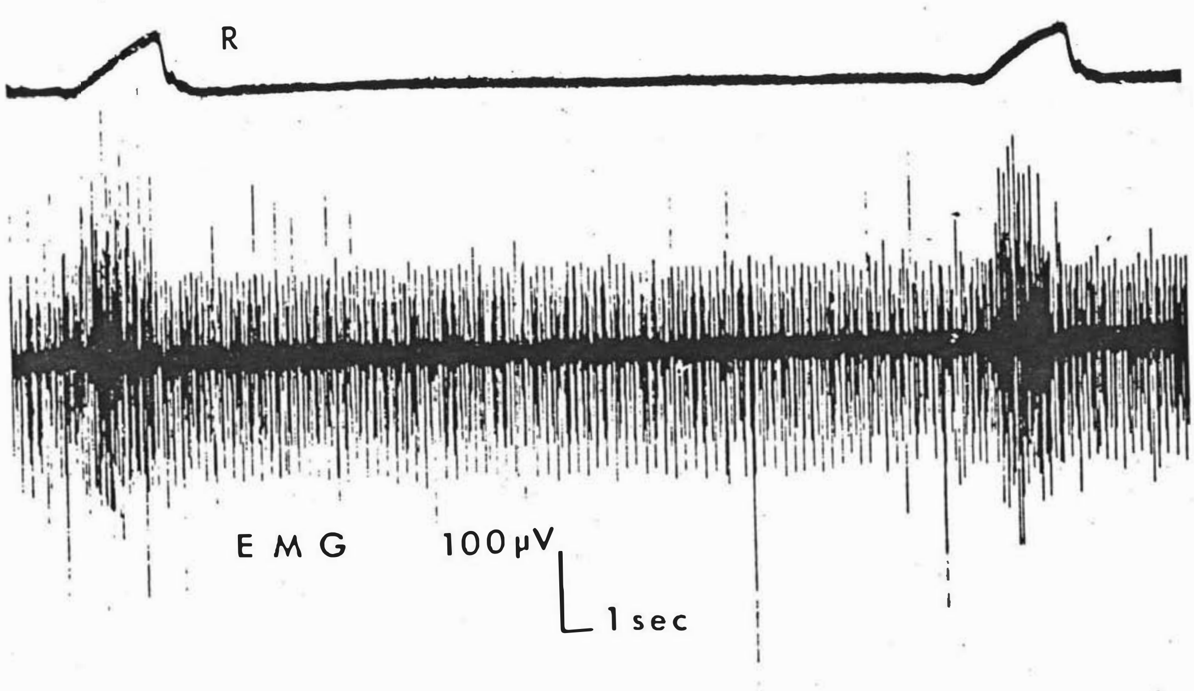
FIGURE 5.8-1

Xerox copy of ultraviolet oscillograph tracings showing respiration (R) and electromyogram (EMG). The EMG represents electrical activity typical of the equine normal left dorsal cricoarytenoid muscle.

FIGURE 5.8-2

Xerox copy of ultraviolet oscillograph recordings showing respiration (R) and an electromyogram (EMG) typical of electrical activity in the equine cricothyroid muscle.


R
EMG
100 µV
1 sec


R
EMG
100µV
1sec

tubing to avoid excessive restriction of thoraco-abdominal wall movement. The ends of the belt were attached to a steel beam in such a manner as to cause four point bending of the beam. This bending increased during inspiration and was sensed with a strain gauge forming part of a Wheatstone bridge. The signal was preamplified as in monitoring bone strain and tendon tension and further amplified before being recorded simultaneously with the EMG signals.

The surgeon was able to view the EMG activity and respiration traces on a double beam oscilloscope, and to listen to the EMG by means of an extra amplifier and speaker. The audio-visual combination was most useful in identifying types of EMG activity.

Speech was also recorded to allow identification of muscle, amplification, tape position and to record the comments of the surgeon.

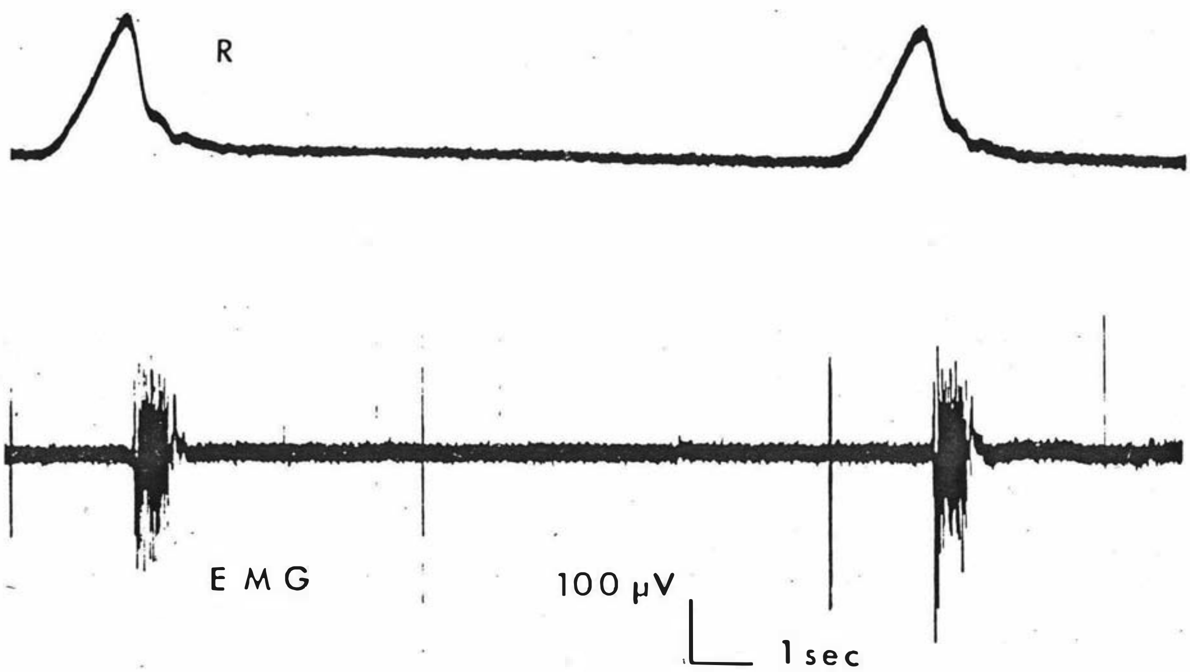
Figure 5.8-1 shows typical activity in the normal left dorsal cricoarytenoid muscle in relationship to respiration. There was considerable motor unit activity during the rest phase between respiratory excursions in 34 of the 38 recordings from this muscle. All recordings showed increased electrical activity during inspiration. In 23 recordings there was an interval on commencement of expiration in which little EMG activity could be observed. In 9 recordings there was motor unit activity throughout the respiratory cycle.

The tonic firing in the dorsal crico-arytenoid muscle probably plays an important role in maintaining the arytenoid cartilages in their usual paramedian position between respiratory excursions. The activity during inspiration confirms that the main function of this muscle is to abduct the arytenoid cartilages.

Thirty two recordings were obtained from the right cricothyroid muscle of 13 horses. In all traces there was tonic activity between respiratory excursions (Figure 5.8-2).

FIGURE 5.8-3

Xerox copy of ultraviolet oscillograph tracings showing respiration (R) and an electromyogram (EMG) typical of the equine lateral cricoarytenoid muscle.


R
EMG
100 µV
1 sec

In 17 recordings from 7 animals there was an increase in motor unit activity during inspiration. The remaining recordings showed either a regular rate of motor unit firing throughout the entire respiratory cycle or a decreased rate during inspiration and an increased rate during exspiration.

The variation in the pattern of electrical activity found in the recordings obtained from the crico-thyroid muscle indicates that whilst the majority of motor units in this muscle fire phasically, others show a continuous train of impulses throughout the whole of the respiratory cycle and are apparently not influenced by the rhythm of breathing.

Four different types of motor unit activity were noted in 24 recordings from the lateral crico-arytenoid muscle (Figure 5.8-3). In 5 of 12 animals, electrical activity was noted during expiration and in some animals continuing for some time after expiration. In a further 3 animals, no respiratory rhythm was noted in the activity. Three animals failed to yield EMG signals and the remaining animal gave bursts of EMG activity, sometimes during inspiration and sometimes during expiration.

Of 10 horses in which the transverse arytenoid muscle was investigated, 13 records were obtained from only 6 animals. In all of these there was increased electrical activity during expiration, there being prolonged EMG activity in 2 of the animals.

The ventricularis and vocalis muscles were sought in 12 horses but only 7 recordings were made from 5 animals. Expiratory EMG bursts were recorded from 4 animals, one of these producing prolonged activity. The remaining animal showed inspiratory activity in these muscles.

The lateral crico-arytenoid, transverse arytenoid, ventricularis and vocalis muscles are all adductors which serve to narrow or close the glottis (Figure 2.7-1). However in a significant number of the adductor muscles examined no electrical activity was detected. Misplacement of the needle is unlikely to explain this apparent inactivity since the surgical procedures used allowed ready access to these muscles. It is more likely that a large number of motor units were inoperative during quiet respiration. A further influence could have been the depth of anaesthesia. Suzuki and Kirchner (1969) found that in anaesthetized cats the depth of anaesthesia influenced the spontaneous electromyographic activity in laryngeal muscles, the adductor muscles being more easily inhibited than the abductors.

These studies, in which an endotracheal tube was usually present in the larynx during data recording, indicate that laryngeal air flow is not required to stimulate contraction in the laryngeal muscles. Evidently some central control must be present.

These laryngeal studies are being continued in the standing horse using fine wire electrodes (Figure 5.2-2) of the type used in recording from the forelimb.

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APPENDICES

APPENDIX I

NELCON 72

OCT 16-18

DEPARTMENT OF UNIVERSITY EXTENSION

P.O. Box 63, Palmerston North.

MASSEY university

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P A P E R 19

AN ELECTRONIC SYSTEM FOR SENSING AND RECORDING THE BIOMECHANICS OF MAMMALIAN LIMBS

G. R. G. Barnes and R. C. O'Driscoll

AN ELECTRONIC SYSTEM FOR SENSING AND RECORDING THE BIOMECHANICS OF MAMMALIAN LIMBS

G.R.G. Barnes and R.C. O'Driscoll,
Massey University.

INTRODUCTION

One may reasonably ask, why study the biomechanics of mammalian limbs ? There are at least five good reasons.

(i) Knowledge for its own sake.
(ii) Mammals, especially racehorses, are prone to leg injury, often resulting in financial loss.
(iii) Safety research - high speed transportation has necessitated research into the forces which the body can withstand.
(iv) Increasing use is being made of prosthetic devices which should be designed in accordance with the biomechanics of the tissue that they replace.
(v) Bone disease and its relationship to bone strain are worthy of investigation.

The parameters that are to be sensed are lateral and medial bone strain, bone bending, and tendon tension. In addition, simultaneous indications of foot-ground contact, and body attitude, are required.

THE SENSORS

Bone strain is sensed with electric resistance foil strain gauges bonded directly to the bone of the living animal using contact cement; strain gauges placed on the medial and lateral aspects of a bone will indicate bending when suitably connected as part of a Wheatstone bridge. The strain gauges used are nominally of 120 ohm resistance with a gauge factor $F = \frac{\Delta R/R}{\Delta L/L}$ of about 2.1 .

Tendon tension is sensed independently by two small strain gauges bonded to a buckle transducer of original design. Tendon tension causes bending of the side members of the transducer. The resulting surface strain may then be detected using the strain gauges.

Foot-ground contact is sensed by mounting heavy duty industrial microswitches in brackets which are strapped to the lateral side of the foot.

To record body attitude a cinematograph of the mammal during locomotion is obtained.

REQUIREMENTS OF THE ELECTRONIC SYSTEM

The system should permit simultaneous recording of at least two parameters, preferably more.

Means of correlation must be provided for data recorded on separate equipment. The data should be readily obtainable in a form compatible with both digital and analogue computer inputs. A permanent record of data should be made in both visual and electrical form.

The system should allow the animal to move with little restriction.

Noise and mains pick-up should be able to be substantially removed from any output signal.

THE SYSTEM

Two systems are used, one D.C., the other A.C. Substantial parts of the two systems are identical. The D.C., system is discussed first.

The active strain gauge forms one of the five arms of a D.C., Wheatstone bridge made wholly from strain gauges and a strain gauge rosette. A stable D.C. bridge supply is provided by mercury cells. A split preset pot across two of the bridge arms permits balance to be attained. A fixed gain low noise preamplifier is included on the same printed circuit board as the Wheatstone bridge, both being in a small aluminium box mounted on the animal's leg.

Coaxial cable connects the power supply box mounted on the animal's back to a multichannel chart recorder and also to an amplifier feeding a voltage controlled oscillator (V.C.O.). The V.C.O. output is fed to a four track tape recorder. The other tape channel records from a reference V.C.O. which may be 'punctuated' by a monostable multivibrator activated by a button.

Two event markers on the chart recorder are operated by the foot switches. The foot switches also activate a light emitting diode within the field of view of a movie camera. A seconds time signal is also recorded on the chart.

The A.C. system uses a stable 1 KHz Wien bridge oscillator as bridge supply. The preamp output is fed via coaxial cable to a tuned input amplifier, detector, and Butterworth filter, whose output is fed to the chart recorder and the amplifier for the V.C.O.

The tuned input amplifier discriminates against mains pick-up, as does the Butterworth filter. The remainder of the A.C. system is the same as for the D.C. system.

Although not used, provision was made for placing a four track tape recorder on the back of the animal (horse). This would have allowed the animal even greater freedom and the system would then have been portable and completely self-contained.

DATA RETRIEVAL

Data recorded on magnetic tape may be retrieved in a number of forms. A decoder can produce an analogue voltage which may be displayed on a chart recorder or film strip. It is also possible to obtain punched cards directly from the magnetic tape using a data processing facility operated by the University of Otago Physics Department. Thus both digital and analogue methods are available for analysis of data and for investigations of mathematical models of bone.

CALIBRATION

For the strain gauges bonded directly to the bone, calibration is done by substituting a variable calibration resistor combination for the active strain gauge, and observing the output of the system as the calibration resistor is varied.

The tendon tension transducer is calibrated by retrieving the limbs of the slaughtered animal and hanging known weights from the tendon. The output of the system is recorded on the chart recorder and on magnetic tape.

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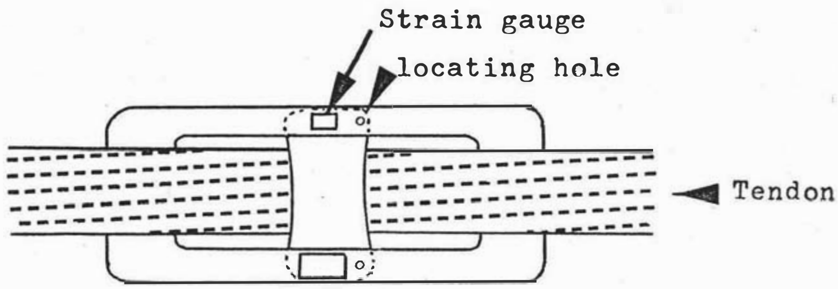
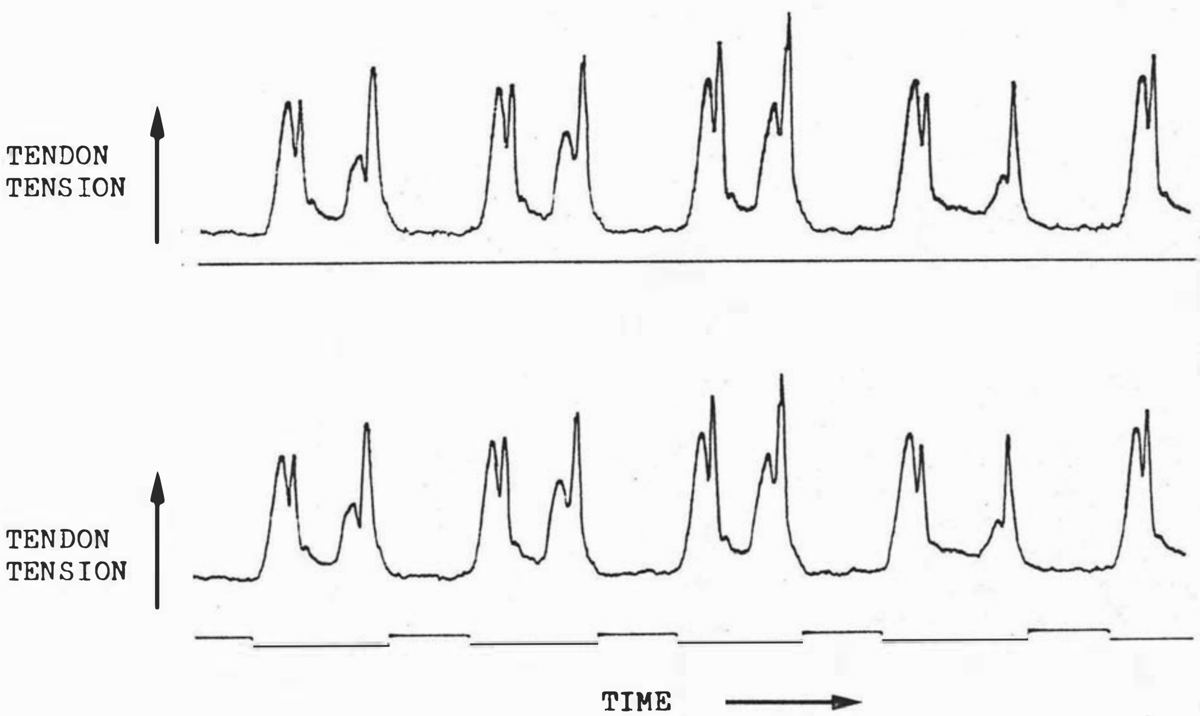


<u>Figure 1.</u> Tendon Tension Buckle Transducer
reverse side.

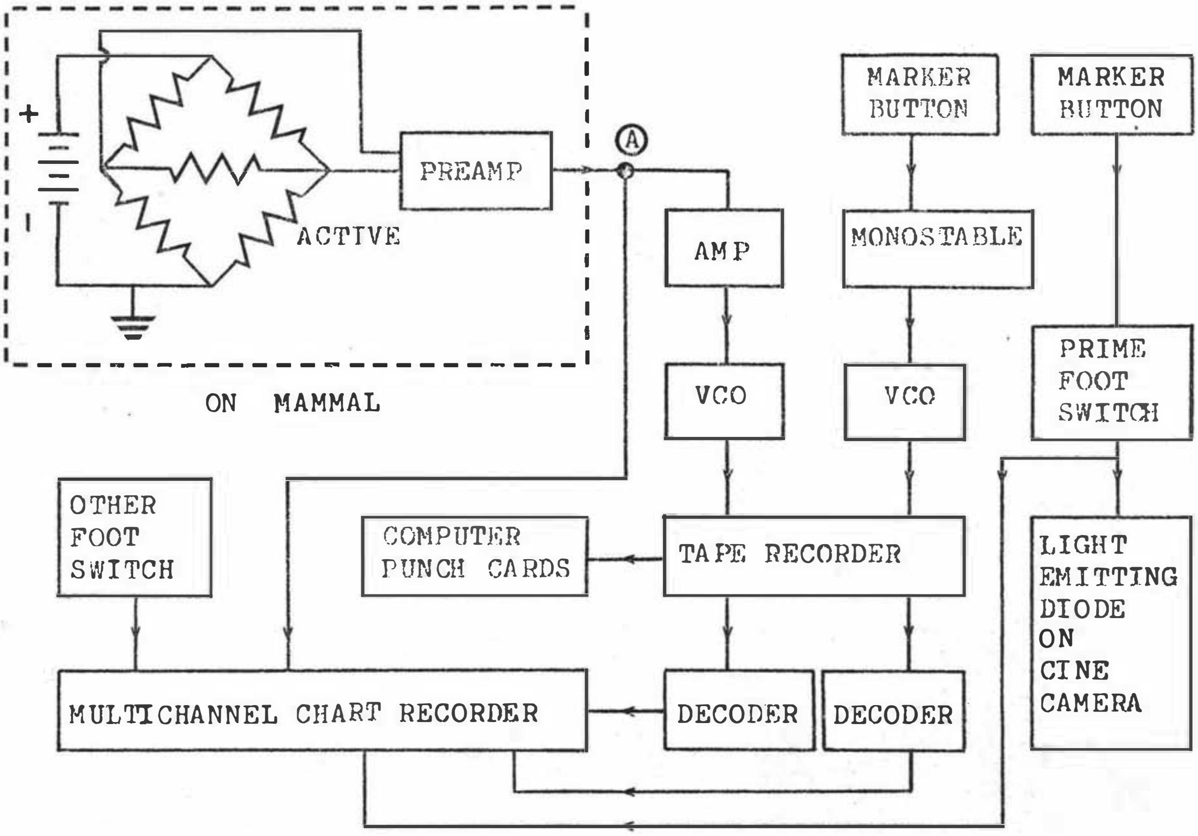


<u>Figure 2.</u> The top trace is a reproduction of the bottom trace made by decoding the magnetic tape.

The bottom trace was recorded directly by the chart recorder.

Fig. 3 THE SYSTEM

(a) D.C. SYSTEM



(b) A.C. SYSTEM