DYNAMIC RESPONSE CHARACTERISTICS OF THE HUMAN VERTEBRAL COLUMN

An experimental study on human autopsy specimens

By

LEON KAZARIAN

STOCKHOLM 1972

"For the life of all Flesh is the Blood thereof."
Levicitus XVII

Tryckeri Balder AB, Stockholm 1972
Copyright © 1972 by Leon Kazarian

ISBN 87 16 01224 0

To my Wife and my Parents

Contents

Preface

It was at the invitation of Professor Carl Hirsch and through the endeavors of my Chief, Dr. Henning E. von Gierke, the late Dr. Fred Berner, Chief scientist, and Col. Clinton Holt, M.D., Commander of the Aerospace Medical Research Laboratory, that I began the investigations reported here. I am greatly indebted to the Aerospace Medical Research Laboratory, Aeromedical Division, Air Force Systems Command and the DOD-USA, for awarding me this opportunity.

I would like to express my sincere appreciation to:

Professor Sune Bergström, Rektor of Karolinska Institutet, who spared neither time nor trouble to make the presentation of this thesis possible.

Professor Carl Hirsch, Head of the Department of Orthopaedic Surgery of Karolinska Institutet, who opened his clinical facilities to me and allowed me to set up the test equipment in his department. He was my adviser and took great interest in my project.

Dr. H. E. von Gierke, for his unhesitating support and encouragement throughout this period of time, as he has done in my past research endeavors. He was responsible for making funds available to construct and forward the impedance tester to the Department of Orthopaedics. The topic for this thesis was suggested by him.

Professor Henry Wallman, Chalmers Institute of Technology, who provided valuable guidance and gave me the benefit of his stimulating comments and constructive criticism.

Professor Carl Wegelius, Wenner-Gren Research Laboratory — Norrtulls Hospital, for providing me with a multitude of technical and clinical resources and critically evaluating portions of my thesis.

Docent Sven Scheller, Chief Radiologist at the Department of Orthopaedics, who interpreted the roentgen films.

Docent Urban Ungerstedt for appreciated discussions and assisting with the formalities of presenting this thesis.

Dr. C. Holt for his generous financial support through Commander's Funds.

The late Dr. F. Berner, who encouraged me by his interest in my work.

The Defense Attaché Office, American Embassy in Stockholm, and especially G. Flanagan, for his unreserved support.

Engr. L. Muhic, who put his unexcelled initiative and knowledge at my disposal and integrated the mechanical-electric components.

Engrs. G. T. Collins and W. Hovey for their design and fabrication of the impedance computer.

Engr. B. West, who designed and performed the structural analysis of the mechanical hardware.

Civ.ing. Bo Klasson and his staff R. Andersson, P. Eriksson, G. Garbergs, G. Söderlund, for the construction and modification of various test jigs, and fabrication of the creep boxes.

Catherine Sundström for correcting my spelling, weeding out inconsistencies and proof-reading my manuscript.

Ingrid Lettström, for her patient forbearance in grappling with the intricacies and manifold formalities involved by the financial administration of this project.

Karin Johansson, for her preparation of tables and illustrations.

Nina Norris Hudson, student at Konstfackskolan, for the pencil illustrations of the spinal columns.

Last but not least, I wish to thank my wife for her unreserved support and encouragement throughout the time I have been engaged in this work and for singly shouldering our parental responsibilities.

This work was supported financially by Karolinska Institutet (Professor S. Bergström), the Swedish Medical Research Council (Professor C. Hirsch), and Commander's Funds (Colonel C. Holt, M.D.).

Stockholm, December 1972 Leon Kazarian

1. Introduction

The human vertebral column has been under scrutiny from a mechanical viewpoint for nearly a century. Throughout the past 25—30 years, its form, function, pattern, and magnitude of reaction have been examined, utilizing a host of mechanical strength techniques adopted from materials science and structural mechanics.

The results of these experiments have furnished some understanding and insight into the engineering properties, mechanostatic nature and performance of this system to a particular level and manner of excitation. Using the information derived from these static tests, clinical and operational standards have emerged which are reported to describe spinal column response to dynamic mechanical forces. The object of much of this research has been to postulate the various mechanisms, modes and patterns of reaction, to develop mathematical formulations for describing these, and to compute general behavior patterns as a function of geometry and frequency.

Although there has always existed the underlying thought that the peculiarities of the materials which constitute the human spinal column might be dependent on dynamic loading, dynamic experiments have not yet been carried out. Therefore, a scientific basis for the correlation of static and dynamic tissue reponse data is nonexistent. Furthermore, the role of vertebral body preload has been completely ignored.

The purpose of this study is to shed some light on a few of the dynamic response characteristics of the human vertebral column.

This investigation deals with the application of a sinusoidal vibration to the segmented cadaver vertebral column under a compression bias. Its objectives are to analyze the dynamic resistances and mechanics of deformation and to provide a phenomenological description of the reactions of the cervical, thoracic, and lumbar spine to finite amplitude deflections, using mechanical driving point impedance relations. In addition to these primary objectives, experimental data concerning phenomenological and mechanistic trends associated with the creep and relaxation characteristics of the human spinal column are presented and discussed, the role of the disks, vertebrae, and apophyseal joints regarding their dynamic reaction to compression loading is further delineated, and a hypothesis introduced on the warping actions and function of prestress with respect to the vertebral unit.

2. Theory of Impedance

INTRODUCTION

The human spinal column is a dynamical system. It involves the motion of elements and the action of forces in producing or changing the motion of the structure. Dynamical systems may be classified as mechanical, electrical, magnetic, nuclear, acoustical, etc. For this discussion, the analysis describing the performance of this dynamical system is primarily mechanical and adapted from electrical circuit theory. The concept of applying solutions from one engineering field to the problems of another by use of analogous transformations is well known. Olson (1943). Extensive studies of electrical-mechanical analogs have been carried out and there exist specific methods for setting up analog circuits to simulate and analyze the dynamical behavior of specific mechanical systems. Firestone (1938, 1957), Soroka (1961).

THE CLASSICAL THEORY OF IMPEDANCE

The concept of mechanical impedance has evolved from the similarity between the differential equations governing the behavior of linear electrical and mechanical systems. (Kirchhoff's laws for electrical circuits correspond precisely in form to Newton's laws of motion for a mass.) The analogy is valid, within certain limitations, because the basic equations describing the electrical and mechanical peculiarities are of the same form. The basic shift from mechanical to electrical systems is attained by using the analogies summarized in *Figure 1* from Soroka (1961).

MECHANICAL QUANTITY		ELECTRICAL QUANTITY		RELATIONSHIP [a]
NAME	SYMBOL	NAME	SYMBOL	
MASS	m	INDUCTANCE	L	$m = S_1 L$
DISPLACEMENT	x	CHARGE	Q	$x = S_2 Q$
TIME	t_m	TIME	t_e	$t_m = S_3 t_e$
VELOCITY	V	CURRENT	I	$V = S_4 I$
FORCE	F	VOLTAGE	E	$F = S_5 E$
DAMPING CONSTANT	c	RESISTANCE	R	$c = S_6 R$
SPRING COMPLIANCE	$\frac{1}{K}$	CAPACITANCE	C	$\frac{1}{K} = S_7 C$
FREQUENCY	f_m	FREQUENCY	f_e	$f_m = S_8 f_e$

(a) $S_4 = S_2/S_3$; $S_5 = S_1 S_2/S_3^2$; $S_6 = \frac{S_1}{S_3}$; $S_7 = \frac{S_3^2}{S_1}$; $S_8 = \frac{1}{S_3}$

Figure 1. Scaling factors for mechanical-electrical analogs. Soroka (1961).

Mechanical impedance is the analytical tool which allows the dynamic consideration of a complex structure by the use of a quantitative parameter; this parameter is the ratio of force to velocity.

In its classical form mechanical impedance or impedance (Z) applies to a sinusoidal force driving a lineal elastic mechanical system. The steady state response of the mechanical system exhibits sinusoidal acceleration, velocity and displacement at the frequency of excitation, and with a fixed phase relationship with respect to the force of excitation. Because many structures are considered as linear (in a particular small amplitude frequency range), this ratio is considered to be independent of the vibration amplitude and is a function of sinusoidal frequency only. However, the relationship between the magnitude of force and velocity may not be in phase with respect to each other, in which case their ratio, Z, can be expressed as a complex number with a magnitude and phase angle, or, alternatively, with real and imaginary parts.

The impedance may be derived in the following manner and illustrated as in *Figure 2*. The account presented here is adapted from Hixson (1957, 1961). The force and motion of the system are defined in the following:

Force: A sinusoidal force can be represented as a rotating phasor on the complex plane, as shown in *Figure 2*. The phasor has a magnitude F_0, it rotates counterclockwise with an angular velocity ω. At any particular instant in time t, it subtends an angle ωt with the real axis. The instantaneous force is a projection of the phasor on the real axis represented by $F_0 \cos \omega t$. In terms of real and imaginary components it is expressed as

$$F = F_0 \cos \omega t + jF_0 \sin \omega t$$

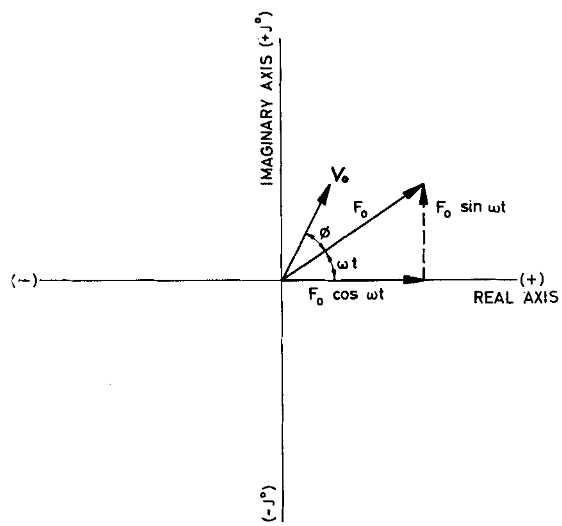


Figure 2. Complex-plane representation for sinusoidal force and velocity. Hixson (1961).

Velocity: A sinusoidal velocity can also be represented as a phasor, as shown in *Figure 2*. When V_o is rotating at the same angular velocity ω as F_o, it can be represented by the expression

$$V = V_o\,[\cos(\omega t + \varnothing) + j\sin(\omega t + \varnothing)]$$

where $\varnothing$ is the phase angle between F and V.

Mechanical impedance is defined as the complex ratio of an applied sinusoidal force F to the resulting sinusoidal velocity V, and is given by the equations:

$$F = F_o \cos(\omega t + \varnothing)$$
$$V = V_o \cos \omega t$$

Thus the impedance by definition becomes

$$Z = \frac{|F_o|}{|V_o|} \angle \varnothing$$

where $\varnothing$, the phase angle between the applied force and resulting velocity, is a function of the frequency ω.

The impedance can also be expressed as five other ratios, between force, acceleration, velocity and displacement; all are used to express the same information but in different forms, see Hixson (1961). In the following discussion, only mechanical impedance (Z) or simply "impedance" is used. Impedance may be further differentiated into ***driving point impedance***—where the force and velocity are taken in the same direction and at the same point (driving point), and the *transfer impedance* where the forces are being taken at one point with the structure and the velocity at another, as shown in *Figure 3*, Molloy (1957, 1958).

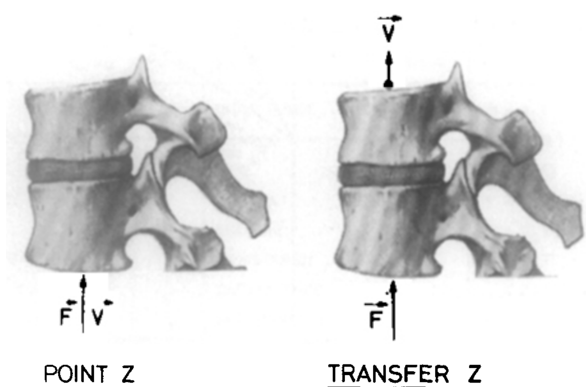


Figure 3. Illustration of driving point and transfer impedance.

THE CONCEPT OF IMPEDANCE AS APPLIED TO A PURE MASS, SPRING, AND DAMPER

Mechanical impedance measurements are similar to those of any generalized frequency response plot in that vector ratios are derived from sinusoidal motions. For a complete specification of a system's behavior both the magnitude and phase determinations as functions of frequency are necessary.

The methods of measuring impedance are well established. Impedances associated with the three basic mechanical elements — a mass, a spring, and a damper, are given by:

mass:	$Z = j\,\omega\,m$
spring:	$Z = -\dfrac{jk}{\omega}$
damper:	$Z = \beta = constant$

where
Z = *impedance*
F = *applied force*
V = *velocity*
k = *spring constant*
ω = *frequency*
m = *mass*

The impedance response plot is described graphically. The logarithm of the impedance magnitude and the phase are displayed as functions of the logarithm of frequency. A pure mass is represented on the magnitude plot by a line with a positive slope. A spring is shown as a line with a constant negative slope, and a damper by a line with a slope of zero. A constant line of impedance foı increasing frequencies is defined by a decreasing apparent mass, increasing apparent stiffness and constant damping. The phase shift for driving point impedance measurements on the mass, spring and damper are constant at +90 degrees, —90 degrees and 0 degrees, respectively. Osgood (1965), Hixson (1961). The general variations with frequency of the impedance of these three elements are shown in *Figure 4*.

1. A mass in motion does not dissipate energy but stores it in kinetic form which is recoverable.
2. A damper dissipates energy in the form of heat which is not recoverable.
3. A spring does not dissipate energy but merely stores it in potential form, which is recoverable.

Any physical system may be described in terms of mechanical impedance by considering it as an assembly of linear passive elements with lumped constants. Firestone (1931), Coermann (1961), Osgood (1965). Hixson (1961) concisely introduces the concepts on impedance, derives the above equations and gives tables of impedance formulas for many simple and elaborate mechanical models involving combinations of lumped elements.

Of these three structural properties, damping is the one least understood and most difficult to predict, and unlike mass and stiffness, damping usually cannot be inferred from simple static measurements. Any oscillating structure, including the vertebral unit, at any instant in time contains some kinetic energy and some strain (or potential) energy: the kinetic energy storage is related to its mass and the strain energy to its stiffness. In addition, the vertebral unit dissipates some energy as it deforms. It is this energy dissipation process or, more precisely, this conversion of ordered mechanical energy into thermal energy that is called damping. Unlike the mass and stiffness, damping does not refer to a unique physical phenomenon. There are in all likelihood as many damping mechanisms as there are ways of transforming mechanical energy into thermal energy. These mechanisms may include interface friction, mechanical hysteresis, and fluid viscosity. Manley (1941), Olson (1958), Mutch (1962). To analyze and/or predict damping mechanisms within the vertebral column all conceivable damping mechanisms ought to be accounted for. Perhaps two, three or four of these will predominate and others may be neglected. Throughout this study, a few of these mechanisms and phenomena will be brought to light.

THE IMPEDANCE RESPONSE PLOT

At a particular frequency the impedance (Z) of a system is a complex number commonly expressed as the sum of a real and an imaginary component: its resistance *R* (real part) stipulates the frictional component of the vertebral unit and coincides with its resistance characteristics. The reactance Q (imaginary part) is equivalent to the energy storage in the unit and is correlated to the presence of mass and elasticity. Both terms are represented by the equation

$$Z = F/V = R + j\,Q$$

$$|Z| = (Z_R^2 + Z_j^2)^{1/2} \qquad \varnothing = tan^{-1} \frac{j\,(Z)}{R\,(Z)}$$

The impedances of the three "ideal" elements are functions of frequency. It is convenient to plot impedance data on graph paper with log-impedances and log-frequency scale or as an impedance nomograph which mathematically relates the apparent weight (force/acceleration), dynamic stiffness (force/displacement) and mechanical impedance (force/velocity) to the excitation frequency. Wright (1961), Bouche (1961), McGregor (1967). Damping at resonance may be directly read off the nomograph, or at any point of resonance calculated by simply multiplying *Z (Cos Ø).* Keller (1967).

The curve of the impedance of an ideal mass has a slope of +1 and crosses the line f = 1 *Hz* at $Z_m = 2\pi m$. The value of *m* for an unknown mass can be obtained.

The curve for the impedance of an ideal spring has a slope of —1 and intercepts the line *f* = *1 Hz* at $Z_k = k/2\pi$. Resistance is a constant value with frequency and represented by a horizontal line. *See Figure 4.*

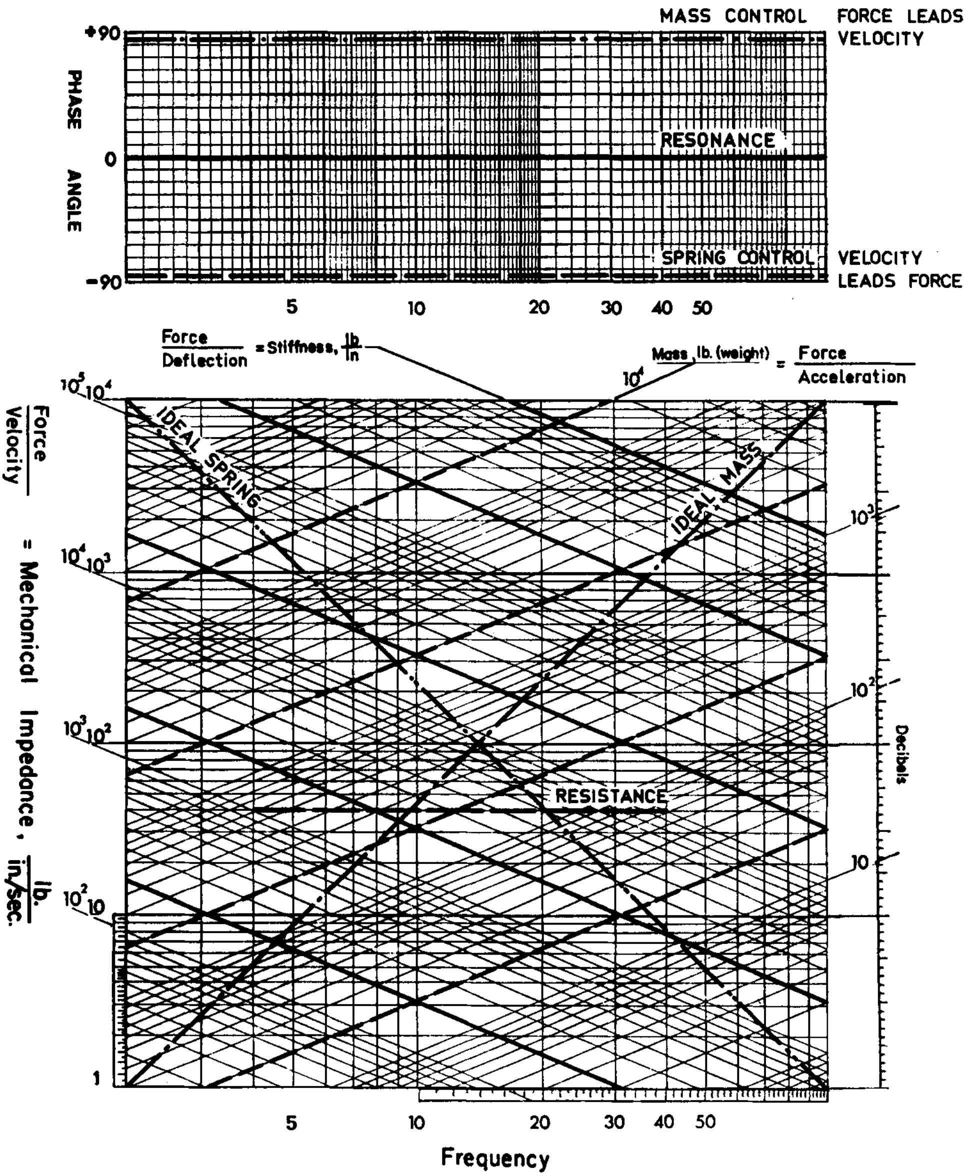


Figure 4. The impedance magnitude as a function of ideal mass, spring and damper, plotted to a Log-Log Scale.

TERMINOLOGY AND UNITS

The terminology used for the various ratios of impedance has not yet been standardized. The ASA has recommended certain terminology which will be used throughout this study. The Structural Impedance Panel, Ships Noise Committee of the Bureau of Ships, has proposed graph forms based on the British (in-lb-sec) system of units. Wright (1961). This was the format selected for presentation of these data. Conversion factors to the cgs system are given in section 1 of the Appendix.

PHYSICAL ELEMENTS AND THE HUMAN SPINAL COLUMN

In general, the characteristics of "real" masses, springs, and mechanical resistance elements differ from the ideal elements' response and its representation on the impedance plot in four respects:

1. The masses which comprise the spinal column have elasticity and all spring elements have mass; this being so, a spring may exhibit a nonlinear force deflection characteristic, a mass may undergo phenomena such as hysteresis and plastic deformation with motion and the force shown by a resistance may not be precisely proportional to the velocity.

2. All the elements making up the spinal column possess mass. As such, they cannot be expected to react as ideal springs or resistances. Energy can be dissipated in each element of the system in distinct modes: friction, creep, yielding etc.

3. The elements of the vertebral column can differ from ideal elements at high frequencies of excitation. This is so, because elemental length becomes comparable to the wavelength of the stress waves propagated within the structure.

4. Geometric considerations. The shape and geometry of the human spinal column are clearly of major importance. The human vertebral column is a bilaterally asymmetrical structure, anisotropic in development, construction and mechanical behavior. Its response mechanisms must be considered as an idiosyncratic one.

To utilize the method of impedance analyses, the following simplifications were assumed to be satisfied.

1. The vertebral unit was subjected to uniaxial loading; therefore only forces and velocities effective in this mode of reaction were analyzed.

2. The oscillatory displacements were regarded as sufficiently small to allow motions experienced due to vertebral unit deflections to be regarded as linear.

3. Review of Literature

INTRODUCTION

The purpose of this chapter is to review the literature involving impedance techniques used to explore the active and passive mechanodynamics of the human body, its organs and tissues.

An extensive survey of the pertinent literature relating to mechanical impedance and the human body revealed no prior test results concerning impedance in connection with the human spinal column. Consequently the peripheral findings regarding impedance and biological tissues are provided.

This review correlates to this study only in the sense that the authors reported utilized impedance analysis to gain further knowledge of the effects of various mechanical force patterns on the human subject. The latter section reviews the literature findings relating to the isolated cadaver spinal column subjected to controlled mechanical loading.

IMPEDANCE ANALYSIS AS APPLIED TO THE HUMAN BODY

The impedance has been measured for small areas of the body surface by von Gierke (1949, 1950), Franke (1951), Dadson, Robinson and Greig (1954), and for the sitting and standing human subject exposed to whole body vibrations by von Bekesy (1939), Magid and Coermann (1960), Coermann (1961), Wittmann and Phillips (1969). Impedance has also been used to study the vibration response of the human cadaver head, Hodgson (1968), Gurdjian, Hodgson and Thomas (1970), Stalnaker, Fogle and McElhaney (1970); the mechanical vibration properties of the skull, Corliss and Koidan (1955), Franke (1956), Hodgson (1967 and 1970), and the clinical mobility of teeth, Noyes, Clark and Watson (1968). The non-linear impedance characteristics of the human body exposed to vibrations and increased gravitational force are reported by Vogt, Coermann and Fust (1968), Vykukal (1968).

Impedance studies have been used in connection with vibrating hand tools. Vibration transmission from the tool to the hand, to the arm and body has been reported by von Bekesy (1939), Agate and Druett (1947), Dieckmann (1948) and Axelsson (1968).

The first investigator to apply mechanical impedance techniques with regard to the human body was von Bekesy (1939), who analyzed the diffusion of mechanical vibrations and the isolation of resonance in the human body with respect to the sitting and standing position. The frequency dependence of the vibration threshold was continuously monitored and through impedance measurements collected via the skin surface. It appeared that the sensation of vibration was not governed by the alternating pressure but rather by the magnitude of deformation. In a later publication von Bekesy (1939) reported on the sensitivity of standing and sitting man to sinusoidal vibrations on a vertically vibrating platform in the frequency range from 0.1 to 100.0 Hz. The pressure threshold of vibration sensation for standing man was found to be independent of frequency in the test frequency range. The first resonance peak occurred between 12—15 Hz. Von Bekesy suggested that this reaction may be a result of soft and hard tissue interactions between the upper vertebral column and the thoracic chest. He also brought to light that for constant amplitude excitation, the amplitude of the vibrations transmitted to the head continually decreased with increasing frequency, disclosing that vibrations are rapidly attenuated from the foot to the head. With resting, standing or sitting men, periodic waveform deviations were detected which evidently coincided in part to body pulsation. To calculate the transmission of vibration energy throughout man, the impedance function was ascertained.

Barany (1938) studied the threshold of hearing by bone conduction utilizing mechanical impedance analysis. He found that at a frequency level of 435 Hz the stiffness and viscosity values for the human head were:

$$k \text{ (stiffness)} = 1\text{—}2 \times 10^8 \text{ dynes/cm}$$
$$r \text{ (viscous)} = 1\text{—}2 \times 10^4 \text{ dynes-sec/cm}$$

Von Gierke (1950) measured the sound absorption by the body surface of man for a frequency bandwidth from 100 to 12,000 Hz. Acoustical impedances and absorption coefficients of the surfaces were calculated from the resonance characteristics of an air-filled tube (for low frequencies), of an air column excited by sound pulses (middle range frequencies), and an aluminum rod pressed against the body surface for higher frequencies. He reported that the absorption coefficient of the human body surface varies, but decreases uniformly with increasing frequency and with the size of the area of tissue tested.

In 1952, von Gierke et al. reported on the vibrational response of human tissue in the frequency range from 0 to 20k Hz. The results of this study revealed that vibratory energy may be absorbed by the body surface in several ways. At low frequencies (100 Hz) nearly all the energy was found to be dissipated as damped shear waves. Above 100 Hz tissue viscosity and shear reactions were found to influence the behavioral response, while at higher frequencies (up to 5000 Hz), viscosity alone governed soft tissue resistance. Above 5000 Hz, compressional waves were observed to increase rapidly, while for frequency levels up to 100k Hz, the vibratory wave propagation energy became increasingly important. At this level of excitation, it was found, the wave patterns generated were strongly influenced by boundary and geometry. Above 100k Hz compressional waves were said to predominate and possess 85 per cent of the total energy. These findings suggested a theory wherein the propagation of vibratory energy in soft human body tissue may be comparable to wave propagation in an elastic-viscous compressible medium. Applying this postulation to specifically the experimental results, it was discovered that a simplified model of vibrating sphere in an elastic-viscous compressible medium entirely described the mechanical behavior of the tissue. The quantitative agreement between calculated and measured peculiarities in terms of the impedance function were said to be sufficiently good to enable one to approximate values of shear elasticity and viscosity of body tissue.

Franke (1951), resolved the absorption characteristics of sound and other vibratory energy by the surface of the human body, using three styli pressed against the body surface in the frequency range from 10 to 2000 Hz. He showed that it was possible to calculate the stiffness constant of muscle utilizing impedance measurements. He established that a non-linear relationship existed between stiffness and applied pressure.

Oestreicher (1951) furnished a phenomenological classification of wave propagation in a medium with elastic-viscous and relaxational properties, founded on the impedance measurements of Franke and von Gierke. He revealed that for frequencies up to 30 Hz, the reactance was negative and the system behaved like a spring. At 30 Hz, there occurred a resonance and the reactance became positive. At low frequencies the muscle tissue reacted like an incompressible medium and up to 100 Hz viscosity and shear elasticity were of influence, but from 500 to 5000 Hz, only the viscosity forecasted the resistance.

Measurements on the human mastoid were carried out by Franke (1956), in the frequency bandwidth from 200 to 1800 Hz. He utilized a driver tip area of 1 cm^3 applied with a 250-g coupling force on the mastoid and concluded that the reactance of the tissue was purely elastic and suggested that it was justified to consider the resistive component to be a constant value. In an effort to assess the nature of bone conduction receivers, Dadson, Robinson and Greig (1954), conducted a series of investigations into the driving point impedance of the mastoid process for the frequency range from 125—6200 Hz. Their experimental data shed further light on the mechanical distinctions of the head surface and showed them to be predominantly governed by the soft tissue stratum overlying the cranium.

Corliss and Koidan (1955), reported similar findings as the above, while collecting data for the design of an artificial mastoid. Franke (1955) revealed a few principles regarding the application of engineering mechanics to the human body. He noted that the mechanical behavior of human tissue could be treated as a semi-infinite viscoelastic medium and that of the whole body handled as a simple harmonic oscillator. In 1956, Franke further discussed his impedance measurements and the resonant frequencies of the human skull. These studies were carried out by means of an electrodynamically actuated small contact area piston, in the frequency range from 200 to 1,600 Hz. The data were collected from living human subjects, a dry skull preparation and a human cadaver. The modulus of elasticity of skull bone was found to be consistent with the value obtained by static measuring techniques. The propagation of bending waves in the skull bones, calculated from the resonance frequency, agreed satisfactorily with the experimentally determined propagation velocity. He also proposed that a vibrating spherical shell is a suitable model for the skull and described its vibration patterns with reasonable approximation.

To obtain the parameters of the human body so that it may be handled as a mechanical system, impedance experiments were carried out to assess the physical and physiological effects of vibration and impulsive forces on various areas of the human body by Coerman, Ziegenrucker, Wittwer, and von Gierke (1960). Abdominal wall displacements, oscillating changes in the chest wall circumference and periodic air flow through the mouth were measured on human subjects under periodic, longitudinal vibration excitation. The point of resonance for these processes was found to lie between 3 and 4 Hz.

Coermann (1961) applied the theory of mechanical impedance utilizing one or more lineal degrees of mechanical freedom with respect to the human body. He developed a method for measuring impedance and disclosing the complex parameters of the system under controlled vibrations. Impedance response curves for longitudinal vibrations in the sitting and standing positions were collected in a frequency range from 1—20 Hz. He reported that up to 2 Hz the impedance of the human body was similar to that of a pure mass. At around 4 to 6 Hz he uncovered a resonance peak which produced the greatest whole body deformation along with a second peak at around 10—12 Hz, which he said was a subsystem resonance. The elements in the human body attributed with the apparent resonances were also identified. Impedance studies have also been conducted by Clark, Lange and Coermann (1962), White, Lange and Coermann (1965), Ziegenrucker and Magid (1959), Coermann (1963), Edwards and Lange (1964, 1969), to establish human response to mechanical energy and give some idea of the biological effects and subjective responses that may be expected when the human body is vibrated at various frequencies.

Noyes, Clark and Watson (1968), attempted to establish a relationship between mechanical impedance and clinical mobility of teeth. Vogt, Coermann and Fust (1968), measured the impedance of sitting subjects under sustained acceleration in an effort to uncover the degree of non-linearity in the human body. They reported that the stiffness of the human body increased from 69×10^6 dynes/cm under normal gravity to a value of 164×10^6 dynes/cm under $+3$ G_z. Vykukal (1968) determined the effects of vibration combined with linear accelerations on human body dynamics and reported that the effects of increased linear accelerations on human body dynamics are primarily increased stiffness, reduced whole body damping and greater energy transmission to the abdominal and thoracic organs. Wittman and Phillips (1969) measured the reaction of erect human subjects seated in several controlled impact and vibrational environments. They noted that non-linearities were observed to take place in the structure both in the transient and steady state environment and believed the nature of the non-linearities could be sufficiently great to make the assumption of human body linearity impractical. Campbell and Jurist (1971) performed impedance measurements on the proximal heads of intact excised human femurs and speculated on the feasibility of utilizing impedance analysis to assess the mechanical integrity of this structure. They noted that the intact femur reacted as a pure mass up to a frequency level of approximately 70 Hz.

In summary, the overall results of these studies have more or less shown that impedance techniques may serve as a convenient tool to gain additional information with respect to the behavioral response of the human body to mechanical inputs. Reviews of impedance and transmission parameters of man are available: the reader is referred to the surveys of Goldman and von Gierke (1960), Coermann (1962).

PHYSICAL AND MECHANICAL CONSIDERATION OF SPINAL COLUMN ELEMENTS IN CONTROLLED ENVIRONMENTS

Experiments concerned with the physical and mechanical behavioral response of the human spinal column have been carried out on segments or sections of hard and/or soft tissues. The investigations of Rauber (1876), Messerer (1880), Lange (1902), Göcke (1926, 1928, 1931), Petter (1933), Geertz (1939), Ruff (1940), Furst (1940), Hirsch (1951), Virgin (1951), Ingelmark and Ekholm (1952), Hirsch and Nachemson (1954), Hirsch (1955), Yorra (1956), Bartelink (1957), Perey (1957), Brown, Hansen and Yorra (1957), Hardy, Lissner, Webster and Gurdjian (1958), Virgin (1958), Evans and Lissner (1959), Nachemson (1960), Patrick (1962), Hirsch (1965), Crocker and Higgins (1966), Rolander (1966), Galante (1967), Moffatt and Howard (1968), White (1969), Yamada (1969), King and Vulcan (1970), Markoff (1970), dealt with ascertaining the compressive static, quasi-static or dynamic load deformation response peculiarities of the human spinal column.

ON THE PHYSICAL AND MECHANICAL NATURE OF THE INTERVERTEBRAL DISK

The primary elements of the spinal column are thought to be the intervertebral disks, the vertebral bodies and the intrinsic longitudinal and shorter ligamentous structures.

The physical nature of the intervertebral disk has been surveyed and examined by Norlen (1927), Schmorl (1927, 1928, 1929), Andrae (1929), Calve and Galland (1930), Smith (1931), Mixter and Barr (1934), Beadle (1939), Donohue (1939), Hirsch and Schajowicz (1952), Nachemson (1963), Galante (1967), among others. The natural history and structural peculiarities of the disk are reviewed by Keyes and Compere (1932), Saunders and Inman (1940), Gordon (1961).

The intervertebral disks are suspected to be the primary deformable mechanical linkages between the adjoining vertebrae and are frequently thought of as hydraulic-like shock absorbers since they posses an inherent capability to absorb mechanical energy. Hirsch (1955), Evans and Lissner (1965). The intervertebral disks are noted by Hirsch (1951), to be made up of four components, the nucleus pulposus, the annulus fibrosus and the cartilaginous end plates.

The nucleus pulposus is the enclosed substance situated, contained, and encased within the central part of the intervertebral disk and recorded to be of decisive importance for the proper function of intervertebral disk reactions. Petter (1933), Joplin (1935), Charnley (1952). It embodies a colloid solution and is formed by a minute dense network of collagen which extends from the fibrocartilaginous (annular) envelope and microscopically forms a wavy branching argentophilic, three-dimensional, fibrous, translucent lattice. The collagen is homogeneously immersed in a liquid-saturated, amorphous, mucoid, gelatinous, water-rich matrix, confining a small number of cell bodies of variable morphology, and a comparatively large measure of intercellular material. Hirsch, Paulsen, Sylven and Snellman (1952). It contains 30—36 % dryweight of polysaccharide chondroitin sulphate and mineral salts, in addition to its water content. Sylven (1950), Meachim and Cornah (1970).

At about the third decade of life, a sequence of events follows which in effect are said to be a maturation process of the collagenous network, a process involving an increase in fibrillar structure and a modification in its geometric arrangement. The nuclear fibrils are microscopically observed to become thicker, and uni-directionally oriented. Dense fibril arrays are systematically generated out of the original randomly distributed configurations. The process simultaneously involves a progressive dehydration of the mucoid material. Coventry, Ghormley and Kerohan (Part I, II, III, 1945), Hirsch, Paulsen, Sylven and Snellman (1952), Hallen (1962). The presence of the nucleus pulposus, according to Virgin (1951), is an indication of the immaturity of a disk, and not related to mechanical function. He also proposed that the disk attains its greatest mechanical efficiency in adult life, a time when the nucleus pulposus has disappeared as an entity.

The morphology and the natural history of the annulus fibrosus was described by Beadle (1931), Hirsch and Schajowicz (1952), Schmorl and Junghanns (1958), Van de Hooff (1964) and Galante (1967). The annulus fibrosus forms as a series of intricate fibrocartilaginous bands to surround, encase and impede the motion of the nucleus pulposus. Hirsch and Schajowicz (1952). It is oriented in a perpendicular fashion to the longitudinal axis of the vertebral column. Happey, MacRae, Naylor (1953). The density of the cartilaginous bands is a function of the annular region. The lamellae tend to be closely packed anteriorly and posteriorly and less closely laterally. The ventral and lateral portions of the annulus fibrosus are similar in their appearance and present an orderly herringbone-like arrangement, The posterior lateral regions of the annulus fibrosus are reported to have marked irregularities. The lamellae bands are observed to split or merge to interlock with other bands, gradually to attenuate and finally cease, or they may bifurcate and interlock, to close a termination. Beadle (1931). Each annular band composes a characteristic angle relative to its adjoining band. Subsequent lamellae make up an asterisk-like interwoven array, whose directional obliquity alternates in subsequent lamellae. The interstriation angles are reported to form an angle of approximately 30° relative to each other. Labalt (1835), Horton (1958).

Variations in the interstriation angles have been reported by Naylor (1962) to occur in the periphery and near the regions of the annulus and nucleus. Macroscopically, the obliquity of the annular bands is greatest in the outermost lamellae of any given disk, at a higher level. Horton (1958), Rabinovitch (1961). The collagen walls form multiple concentric spirals and contraspirals crossing each other in opposite directions. The concentric layers of collagenous fibers, with their characteristic interstriation angles run in an oblique helicoidal fashion at any particular intervertebral disk level and when contrasted to the succeeding level in such a manner that the fibers in contiguous lamellae lie at about 100 degrees relative to each other. Horton (1958), Naylor (1962), Galante and Hirsch (1966), Galante (1967). The number of lamellae, their size, obliquity of arrangement as well as thickness and height, have been shown to possess wide variations at any given band within the annular circumference, at any particular anatomical level, and morphologically vary from individual to individual. Joplin (1935), Inman and Saunders (1947).

Rouvierre (1921) found that the inclination of the annular bands was proportional to their individual length and increased proportionally to their degree of movement at a particular vertebral level. Labalt (1835) pointed out that the herringbone arrangement (interstriation angles) was comparatively indistinct in the cervical region, more evident in the thoracic region and well defined in the lumbar region. In 1958, Schmorl and Junghanns drew attention to observing loose spanning fibres between the lamellae bands and suggested that these fibers provided a cohesive force between the adjacent walls.

The mechanical action of the individual annular bands was reported by Naylor and Smare (1933), Naylor, Happey and McRae (1954), Happey, Horton, MacRae, Naylor (1955), Naylor, Happey and MacRae (1955), Naylor and Horton (1955), Horton (1958), Naylor (1958), and Naylor (1962). These investigators observed that the initial extensibility of the annular walls occurred as a result of collagen fibril alignment. Horton (1958), emphasized the point that an increase in the annular cross-sectional wall area could also be achieved through relative movements of adjoining uniaxial annular walls which brought about a decrease in the comparative angular directions of the biaxially oriented sheets. Based on his findings he postulated that the functional nature of the annulus was not altered with regard to age. Schmorl and Junghanns (1958) and Caillet (1962), thought the annular walls could partially rotate with respect to each other (shearing action between the annular walls) upon application of a longitudinal load.

The cartilaginous end plate has been structurally characterized (in the earlier years of life) as a homogeneous, soft, pliable material possessing a translucent glass-like, chondromucoid matrix.

Cleland (1889) described the young end plate to consist of an array of intertwining, connected and matted white fibers. Within this array were spaces which housed cartilaginous elements. An end plate is shown in section 2 of the Appendix. McCutcheon (1964) assumed that the resilient nature of the vertebral end plates provided some sort of hydrostatic lubrication to the adjoining vertebral bodies.

Beginning at about the third decade, the end plates begin to show signs of ossification accompanied by an increased deposition of minerals within their matrix. Coventry, Ghormley and Kerohan (Part I, II, III, 1945). Fibrillation becomes evident and ranges from a marked thinning of the central end plate zone, to complete destruction of cartilaginous material. The hardness of any enduring end plate material increases with age. Bradford (1921), Eckert and Decker (1947), Schmorl and Junghanns (1958).

When a static compressive load is applied to the intervertebral disk, a number of phenomena are reported to take place. In 1951 Virgin observed that the disk exhibits hysteresis which he said may influence its physical response. Hirsch and Nachemson (1954) measured the resulting deformation (creep response) for a disk subjected to a constant 100 kg load as a function of time, and made the following observations:

1. The nature of the creep curve was independent of the applied load.
2. Most of the deformation occurred within the first 30 seconds following loading.
3. The creep curve was essentially level after 5—10 minutes subsequent to the application of the load.
4. The intervertebral disk *completely* recovered following the removal of the load.

Brown, Hansen and Yorra (1957), supported by their own experimental experience reported that when a disk was placed into a universal testing machine and a compressive load applied, there was observed to be a tendency for the applied load to gradually decrease, and that frequent adjustments of the test machine were required in order to maintain a prescribed value of load.

On longitudinal compressive tests of thoracic vertebrae, White (1969), turned up evidence indicating that more than 98 % of the creep deformation occurred within the first minute following the application of load and after 5 minutes all, or nearly all deformation tendencies were inconsequential.

According to Göcke (1932) intervertebral disks excised from young individuals showed considerable elastic deformation on loading. He reported a complete elastic response for externally applied loads up to 70 kg and reported that intervertebral disks removed from older cadaver material exhibited a greater stiffness value, along with a diminishing tendency for residual deformations, when contrasted to younger cadaver material. Virgin (1951) repeated and verified Göckes experimental findings, and found that "complete disk recovery" could be dependent upon and modified by the duration of the applied load.

Rolander (1966) made the following observations with regard to a longitudinal compressive load axially applied to a vertebral unit. There existed a viscoelastic stress-strain relationship, with increasing rigidity of the disk and with a measurable deformation of the vertebral body at about 100 kp. At a loading magnitude of 5—6 kp/cm^2 the compression of the normal disk is 2.6 % and for the degenerated disks 3.4 % (for loads up to 100—120 kp). He believed that the vertical deformation for degenerated disks was larger at all loading intervals. Furthermore, the greater compressibility of the degenerated disks resulted in greater remaining deformation. However, following the application of cyclic loadings, the residual deformation was equally large for the normal and degenerated disk.

On the removal of a compression load, Göcke (1932) reported that a vertebral specimen disclosed some residual deformation. Its magnitude varied with respect to the applied load and the age of the test specimen. Virgin (1951) found hysteresis to be less in the lower thoracic and lumbar disks than in the lower lumbar disks.

Hysteresis and residual deformation under cyclic loading were also evident in the biomechanical studies as published by Ingelmark and Ekholm (1952), Hirsch and Nachemson (1954), Brown, Hansen and Yorra (1957), Galante (1967), Markoff and Steidel (1970) and Evans (1970). Evans and Lissner (1965) measured the energy-absorbing capacity of the disk when loaded in compression to failure. The ultimate compressive strength of disk T_{12} was reported to be 7730 lbs/in^2 (543.48 kp/cm^2). The stiffness values of the disk at the L_4 level ranged from 6410 to 6670 lbs. per in^2 (450—468 kp/cm^2). At the L_5 level the stiffness values were reported to be in the neighborhood of 3790 lbs/in^2 (266 kp/cm^2). They also found that the mean total energy absorbed to failure for the last thoracic and first lumbar disks from people less than 60 years of age, was greater than that of the corresponding disks from people more than 60 years of age. The posterior arches were removed from the vertebral units used.

Hysteresis was said by Virgin (1951) to be greater in young cadaver specimens, less in degenerated disks from older cadaver subjects, and least in disks from people in the middle decade and in undegenerated disks in the sixth and seventh decade of life. Ingelmark and Ekholm (1952) stressed the fact that with age the compressibility of the intervertebral disk decreased. Healthy disks were found to be less compressible than degenerated disks. Ingelmark and Ekholm (1952), Hirsch and Nachemson (1954). According to Markoff's (1970) studies, the load deformation behavior of the disk is non-linear. He reported increasing stiffness values for axial compression loads. Göcke (1932), Virgin (1951), Galante (1967), pointed out the interregional mechanical and physical variations of the anterior and posterior annulus. Galante's (1967) experimental results demonstrated that the anterior annulus was much stiffer and showed better recovery characteristics than the posterior annulus. He also produced some evidence with regard to the viscoelastic nature of the disk in static compression.

The rheological response of the disk has in part been attributed by some clinical authors to the osmosis theory put forth by Keyes and Compere (1932), Inman and Saunders (1947), Naylor and Smare (1951), Lewin (1955), Armstrong (1958), among others. This theory is thought to be somewhat substantiated by the "in vivo" observations of Bencke (1897), cited in Lewin (1955), Keyes and Compere (1932), DePuky (1935), on the diurnal and nocturnal variations in an individual's height, and clinically correlated by Cloward and Buzard (1952), who described that following injection of a water-soluble radiopaque solution into the disk space, the radiopaque substance "disappeared" rapidly, which suggested there may exist a fluid transfer and exchange across the cartilaginous end plates.

DePuky (1935) measured 1,200 persons between the ages of 5 to 90, and observed oscillations in total height averaging one to two centimeters. The younger subjects reportedly showed greater variations than did the older ones. The subjects examined were taller in the morning (on rising from bed) but showed a steady decrease in stature during the course of the day. The mechanisms by which these actions are thought to occur are reported as follows: due to the dynamic longitudinal loads acting on the spine, throughout the day, the osmotic pressure within the disk allows fluid to be driven out of the nucleus pulposus and into the spongiosa of the adjoining vertebrae. During a period of sleep, the distributed loads on the vertebral bodies allow the reverse to occur, or the increased osmotic pressure in the disk allows "water" to be drawn from the spongiosa through the end plate into the interior of the intervertebral disk.

The mechanical response of the sectioned intervertebral disk has been reported by Barr (1937), MacNab (1969) and others. If a mid-sagittal section is made on two vertebral bodies and their intervening disk, the nucleus pulposus immediately protrudes. When an axial compressive load is applied to the unit, the nucleus responds by an increase in protrusion; upon removal of the load the nucleus partially retreats. Rabinowitch (1961). Harris and MacNab (1954) made the observation that when an intervertebral disk is divided horizontally, the nucleus pulposus does not protrude, and proposed that perhaps the distension occurring in the sagittal section may be in part due to the unopposed contraction of the remaining annular walls.

In compression loading tests on vertebrae with adjoining disks, Perey (1957), Brown, Hansen and Yorra (1957), Decoulx and Rieunau (1958), Hardy, Lissner, Webster and Gurdjian (1958) found the end plate to be the weakest element in the system and showed fractures involving the central end plate to be a common response mode.

ON THE MECHANICAL AND PHYSICAL NATURE OF VERTEBRAL BODY BONE

The mechanical nature of fresh intact vertebral bodies and of cancellous cubes of vertebral bone has been under investigation for nearly a century. Studies have been carried out by Rauber (1876), Messerer (1880), Lange (1902), Göcke (1926, 1931), Weaver and Chalmers (1966), Rolander (1966), Bell, Dunbar and Beck (1967), Rockoff, Sweet and Blenstein (1969), Evans (1970).

Rauber (1876) conducted compression tests on cubes of fresh cancellous bone excised from adult lumbar vertebrae and found a breaking strength value of 84 kg/cm^2.

Messerer (1880) reported breaking strength values between 22 kg/cm^2 and 78 kg/cm^2. Messerer's (1880) breaking strength data for a few vertebral levels are reproduced in part by Perey (1957). These data still remain as the only measured data available with regard to vertebral fracture between the levels of the low cervical and high thoracic vertebrae. Lange (1902) conducted compression tests of vertebral bodies; each specimen consisted of three vertebral centra and their intervening disks, but their posterior arches were removed. The breaking strength of the various vertebral units was shown to deviate between 15—56 kg/cm^2, with an elastic limit in the range of 5—30 kg/cm^2.

Göcke (1926, 1928, 1931), compressed human lumbar vertebrae in a testing machine, and compiled stress/strain response plots for each specimen. He reported a breaking strength for the lumbar spine in the range of 57—70 kg/cm^2. He explained that the normal intact vertebral body of an 80 year old cadaver specimen has a compressive strength of 22.3 kg/cm^2 and a compressive strain deformation of 7.3 per cent.

According to Chalmers and Weaver (1966) there appears to be an increase in the compressive strength of vertebral cancellous bone up to about the third decade. However, after this time the hard tissue turnover rate is dynamically characterized by a period wherein net bone reabsorption exceeds bone production. This decrease in remodelling rate is mechanically characterized by an exponential decrease in compressive hard tissue strength. Bell, Dunbar, Beck, Gibb (1967), Rockoff, Sweet, Blenstein (1967). Arnold (1970) illustrated the fact that up to the latter portion of the third decade, a direct correlation existed between vertebral body strength and ash content; after this period, the relationship ceased to hold true. He asserted that the vertebral centrum is one of the earlier organs in the body to undergo "degenerative type" changes with the aging cycle. This is macroscopically related to a fenestration in the bony plates and trabecular geometry, along with an involution of the spongiosa and trabecular architecture. Hirsch (1965), Bick and Copel (1952). Galante, Rostoker and Ray (1970), have established a relation between compression strength and the density of human vertebral bone.

The effects of prestressing vertebral bone were studied by Göcke (1928), who preloaded two intact vertebral bodies with a static load (within their elastic range), then applied a compressive axial load with a testing machine. He found that *prestressing vertebral bone (below the point of failure) increased its density and reduced its elasticity by decreasing its ultimate strength.*

In summary, biomechanical studies have been conducted to ascertain the elastic nature, energy relationships, and failure modes for the intervertebral disk and vertebral body. These results were found by externally loading vertebral units and measuring the magnitude of deflection and/or deformation. A general finding concerning the nature of the intervertebral disk and vertebral bone was that the intervertebral disk shows elastic-like properties which are retained throughout life. The load deformation or stress-strain time histories obtained from these studies indicate, that energy is stored during the loading cycle and may or may not be released in full during or following unloading. The physical mechanisms operative, the type and order of magnitude of rheological effects were not explored.

The bulk of these experimental findings have afforded sufficient information to conclude that on the application of a compressive longitudinal static and/or cyclic load, the intervertebral disk excised from a cadaver subject showed creep tendencies.

The recorded residual deformation was larger in the first cycle of loading than in subsequent cycles. Stiffening load deflection curves of various degrees were usually the end product and further, there was some experimental evidence to indicate that energy dissipation in the unit increased concurrently, reflecting a decreased hysteresis and conditional deformation when the loading rate was increased.

Even at very low stresses the elements of the spinal column did not behave in a perfectly elastic manner. Inelasticity was always present under all types of loading conditions. It manifested itself in many different ways: creep, relaxation, deformation, flow, damping. This inelastic behavior led to energy dissipation.

The static and dynamic impact loading response of the disk has been reported by Hirsch (1951), who observed that following the application of a static load the dynamic load necessary to attain the elastic limit of the disk is considerably reduced (similar to Göcke's results). According to Hirsch and Nachemson (1954), Patrick and Evans (1962), Crocker and Higgins (1966), Moffatt (1968, 1970), the strength of the intervertebral disk was found to be greater than that of the vertebral body bone and the behavioral response of the vertebral unit was dependent upon the imposed rate of deflection.

Interdiskal pressure measurements have also been carried out by Nachemson (1960), who described the degenerated disk to be mechanically deranged. He expressed the opinion that if the nuclear pressure was reduced with respect to the annular wall surface, the intrinsic hydrostatic nature of the structure was diminished or might not even prevail. This meant that forced longitudinal loads on the annulus increased and jointly the tangential loads decreased. Comparative pressure measurements of "young" and "old" intervertebral disks have shown that intradiskal pressures in individuals under 16 years of age are lower than in the older people for the same external load excitation. The results of Brown, Hansen and Yorra (1957), and Nachemson (1969) reflected that with aging, the disk was subjected to direct compressive loading rather than hoop tensions. They expressed similar results by stating that the disk structure was incapable of normal absorption and redistribution of external loads.

The Vertebral Arch

A number of investigators have conducted axial compression tests with the posterior arches intact, but maintained that the articular surfaces function to sustain a certain degree of weight-bearing ability. Other investigators have postulated that the axial stiffness of the system was due to the nature of the intervertebral disk and that the role of the vertebral arches was negligible and therefore these elements were removed.

In combing through the clinical literature one finds many speculations and contradictory opinions put forth with regard to the role of the posterior arch and its articulating joints as load sustainers. Fick (1904), Spurling (1953), Hadley (1935) expressed the opinion that the articular facet joints have no loading capability under normal conditions, while Ghormley and Kirklin (1934), Frykholm (1951), Nachemson (1960) and White (1969) have regarded the elements as partial weight-bearing structures and still others have said that these elements were important as a buttress. Ghormley (1933), Guntz (1934), Shore (1935), Severin (1943), Gianturco (1944), Lewin (1955), Kelly (1958). Ingelmark (1959), assumed the possible importance of the articular facet joints to be their load-carrying functions.

According to a study by Åkerblom (1948), if an isolated complete series of lumbar vertebrae were mechanically loaded in the planes of the series of arches, the force-deformation curve obtained was similar to that obtained by loading only anterior aspects of the vertebral column.

Fick (1911) observed that following the separation of the spinal muscles from a cadaver subject the vertebral column assumed its natural form (corresponding to its attitude if it were in the erect standing posture). He attributed this typical response pattern to the fact that the unloaded spine was subjected to tensile loading on account of the inherent behavior of its common ligaments and musculature. Petter (1932) reported that when two or more vertebral bodies and their intervening disks were excised from cadavers, there occurred an expansion in the length of the vertebral unit. The thickness of each disk was found to increase on an average of 0.7 mm. A further expansion in length was observed when the longitudinal ligaments were cut at the level of the disks and an additional 1.2 mm in disk thickness was gained when the annulus was circumferentially cut. He reported that the magnitude of the load necessary to drive the vertebral unit back to its original length (as it was in the cadaver) was approximately 32 pounds (13.5 kg).

The experiments of Ruge, cited by Åkerblom (1948), showed that the vertebral column lengthened about 3.7 cm when its common ligaments were cut (so that the vertebral bodies were only held together by the intervertebral disks). When the posterior arches and interarcuate ligaments were separated from the laminae and arches by cutting through the pedicles, the posterior arches were found to lose approximately 14 per cent of their original length. Steindler (1955). Conversely, the vertebral bodies were found to straighten out and assume a new position in a "natural form". Meyer (1873), Petter (1933), Åkerblom (1948), Hagelstam (1949) and Rolander (1966).

Nachemson (1960), measured intradiskal pressures in isolated vertebral units, with and without the posterior arches. He reported an average increase of 18—20 per cent in intradiskal pressure to occur when the posterior processes were removed. He speculated that the magnitude of intradiskal pressure variation may be dependent upon spinal region. Hirsch and Nachemson (1964) were unable to find that the removal of the facets at the pedicles either laterally or bilaterally had any appreciable influence on the magnitude of compressive load carried at the various disk levels. White (1969) summarized that upon removal of the posterior facets with the arches, the motion of the vertebral unit was significantly affected.

The articular facet joint planes in each of the characteristic spinal regions and transregions have been recognized to differ in their geometric orientation. Struthers (1875), Bardeen (1905), Willis (1923, 1929), von Lackum (1924) Danforth and Wilson (1925), O'Reilly (1927), Brown (1932), Brailsford (1928, 1934), Rodgers (1933), Odgers (1933), Mitchell (1936), Rockwell et al. (1938), Bagley (1941), Capener (1944), MacNab (1950), Francis (1955), Ingelmark (1959), and Davis (1959). Their topographical relationships have been reported to vary at any given level of a spinal region. Farmer (1936), Oppenheimer (1938), Taylor (1939), Brailsford (1929), Southworth and Bersack (1950), and Hanraets (1959). Their degree of directional freedom has been known to vary as a function of vertebral number and region and may deviate from individual to individual, and most certainly varies as a function of age and race. Keller (1924), Hagelstam (1949), Schmorl and Junghanns (1959).

Biomechanical and clinical studies regarding the kinetic and kinematic nature of the vertebral column, both with and without its posterior aspects, have demonstrated that upon removal of these elements, spinal mobility increases with respect to three mutually perpendicular grid planes, vertebral stability decreases while the characteristic envelope of motion for the vertebral segment and its respective regions vanishes. Based on these experimental observations it may be said that while relative motions between adjacent vertebrae are principally a result of intervertebral disk deflections, the direction, comparative magnitude and mode of response at any given intervertebral articulation are fastidiously related to the geometric orientation and general shape of the articular facet joint planes associated with the specific vertebral level. These small joints act as guides to control vertebral column kinetics. Humphrey (1958), Fick (1904), Strasser (1913), Braus (1921), Steindler (1955).

Kellogg (1932) experimentally found that the strength of the vertebral symphyses was greater than that of the vertebral bodies. Loading tests have also been conducted on individual processes of the vertebral arch by Tylman and Ramotowsky (1961), to obtain further quantitative knowledge with respect to internal fixation procedures.

VERTEBRAL COLUMN ASYMMETRY

That bilateral morphologic asymmetry and its various manifestations can be demonstrated in the human vertebral column was ascertained by Miles in 1944, who carried out comparative measurements and performed statistical analyses of the right and left sides of thoracic and lumbar vertebrae. His results showed that minute bilateral differences (not greater than 2 mm) existed between individual vertebrae, along with a tendency for the lateral height dimension to be greater on the same side through several consecutive vertebrae. This was followed by an equal, less or greater number of succeeding vertebrae whose opposite lateral height dimensions revealed similar bilateral anomalies. This shift took place most frequently between a group of 2—3 vertebrae, but was often found to occur between alternate vertebral groups, or a group of vertebrae could extend to 4—5 without any interruption. The influence of torsion was not considered in his reported findings.

Farkas (1941), came to the conclusion that up to about the age of five, the normal human spinal column matures in a bilaterally symmetric manner. From this age onward, Farkas noted, an asymmetry develops in the form of the spinal column, which steadily increases throughout life.

Sullivan and Miles (1959) studied AP and LAT spinal radiographs of 49 subjects ranging in age from 19 to 45 years. They reported vertebral body torsion to be a common occurrence with regard to the lumbar spine.

Estes (1920) radiographically examined the vertebral columns of 1,856 college age men. He noted that a very mild type of functional vertebral column asymmetry existed in from 10 to 20 per cent of this study group. He also observed that a left-sided thoracic column asymmetry predominated in 70 per cent of his subject population.

Prives (1960) utilized radiographic techniques to monitor radioactive substances introduced into the skeletal system. He reported that a redistribution of radioactive matter occurred when the labelled skeletal system was subjected to mechanical loading. He asserted that this asymmetrical reaction indicated that there exists a relationship between hard tissue remodelling dynamics and mechanical loading and expressed the opinion that physical work favored growth in bony length, and further that certain occupations and sports may promote unequal development of the upper extremities.

ASYMMETRY OF THE ARTICULAR FACET JOINTS

Asymmetry of the articular facet joints has been recognized for a number of years throughout the scientific literature. Whitney (1926), Putti (1927), Rendick and Westing (1933), Farmer (1936), Oppenheimer (1943, 1944). MacConaill (1953), referring to general joint structure, said that there exist no flat hard tissue surfaces but that each articulation has an anatomical curvature which functions to govern joint kinematics and kinetics. He also reported that for any particular joint, the comparative topographic curvatures at corresponding points vary from individual to individual. Similar viewpoints have been expressed by Kraft and Levinthal (1951). Boswall, cited in MacConaill (1953), pointed out that helical intra-articular deformations must take place for maximal joint efficiency, whereas MacConaill (1953) reported joint action to occur along paths of minimal shear stress.

Hanraets (1959) expressed the opinion, that even if the articular facet joint surfaces were not symmetric with respect to their right and left median neural arch plane, they were still bound to function synchronously. Von Lackum, cited in Bagley (1941), indicated that when asymmetry existed in the articular facet joint planes, asymmetrical motion resulted. Whitney (1926), and Taylor (1939) noted the individual variations of the posterior arches and facet joint planes in prepared hard tissue specimens.

Putti (1927), conducted a comprehensive survey on the size, shape, and inclination of the posterior facet joint planes and claimed he found as much variation in these surfaces as in the asymmetry of the entire facet.

Upon analysis of 3,000 roentgenograms Brailsford (1929) pointed out that 69 per cent of the facet planes within the lumbar spine were symmetrically oriented, 57 per cent obliquely oriented, 12 per cent vertically oriented (faced inwards) and 31 per cent one side oblique, and the other side vertical.

Badgley (1937, 1941) examined the shape of the lumbar articular processes, their facets' size, shape and inclination, and noted their variations. Friedman, Fischer and van Demark (1946) studied the lumbosacral radiographs of one hundred soldiers and reported observing asymmetrical facets in 39 per cent of the spines. Francis (1955) described the relative geometry and structure of the articular facet surfaces in the cervical spine and reported on the asymmetrical variations in the right and left processes, the inclination of the articulation angles, and on the physical size of the articular facets in dry prepared specimens.

Kuhns (1935) reported on the developmental changes in the articular processes, their variations in shape and directional slant throughout growth and decay of the spinal column. Kimberley (1937) noted that when joint asymmetry existed, eccentric spinal movement was present.

Asymmetry of the articular processes and their facet planes at the level of the eleventh and twelfth thoracic vertebrae was reported by Struthers (1875), Ward (1876), Smith (1902), and Whitney (1926).

Struthers (1875) and Ward (1876) remarked on the inconsistency of the level at which the change from thoracic to lumbar type of articulation occurred. Farkas (1941) and Ingelmark (1959) suggested that the asymmetry of the articular facet joints seemed to be a type of functional adaptation. In a lumbar facet study by Robert (1959), the variations and distribution of articular facet planes and their functional significance were investigated. He pointed out that asymmetrical facet planes in these bilateral structures are a frequent occurrence.

There have been numerous morphological and physiological studies of posterior arch asymmetry. The knowledge gained and the experimental data reported in the medical, anatomical and anthropological literature give the impression that there exists significant bilateral asymmetry in the posterior arch and its processes, as in the whole spinal column.

4. Experimental Methods

THE MEASUREMENT OF MECHANICAL DRIVING POINT IMPEDANCE—ELECTRONIC INSTRUMENTATION AND MECHANICAL HARDWARE

FORCE GENERATION—THE VIBRATION TEST MACHINE

To measure impedance directly, a sinusoidal force generator with an excitation frequency covering the range of interest was required. The magnitude of the force generated need only be large enough to produce measurable motion.

For this study a Calidyne Model C-44 vibration test machine was used. It converts electrical energy into a mechanical vibrating force. *Figure 5* illustrates the electrodynamic vibration machine—referred to as a shaker.

The model C-44 shaker consists of a cylindrical, magnetic field structure which supports and encloses a movable armature assembly. The field structure itself is supported upon trunnions in a cast frame.

The field structure of the shaker contains a d.c. coil which excites the magnetic circuit through the mild steel field casting, pole piece and cover plate. The armature assembly consists of a coil mounted upon a cylindrical spider. The spider is suspended upon fiberglass flexures mechanically attached to the field

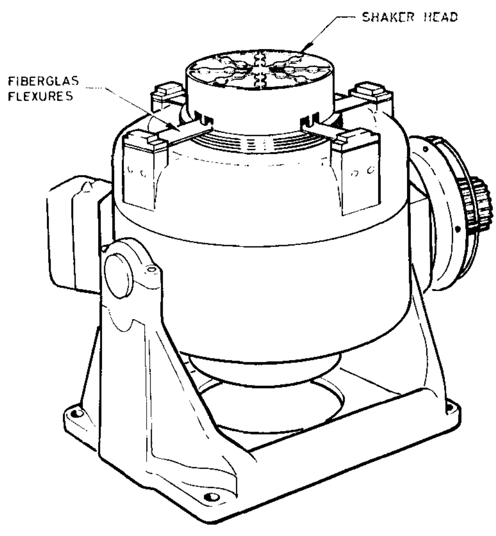


Figure 5. The electrodynamic vibration machine.

structure cover. The fiberglass flexures suspend the armature assembly so that the coil is positioned within the circular air gap of the field structure. The almost frictionless flexures permit free axial travel of the armature assembly up to one inch peak to peak displacement, while restraining the armature assembly with respect to radial movement. The end of the spider, which extends beyond and above the field structure, is referred to as the table head. The table head is drilled and tapped with 16[1]/4 inch concentric and symmetrically located holes, which are used to attach and secure the test head to the armature assembly. Directly beneath the shaker table, and secured to it, is a small coil. This coil enters the air gap of a permanent magnet which is seismically mounted in and near the top of the field structure. The coil and magnet together make up the signal generator of the shaker. Motion of the armature assembly relative to the field structure causes a voltage to be induced in the signal generator coil. If the field structure is motionless, the voltage is proportional to the motion of the shaker table. The electrical output of the signal generator is brought to a connection where it may be used to determine the force output of the shaker table, observe the wave form of the armature, or provide a feedback signal for the operation of a servo system to control the output of the shaker. *Figure 6* is a schematic of the vibration machine.

A degaussing coil is placed concentric with and below the table mounting surface. The coil is energized with direct current when the shaker is electrically on, and functions to cancel the leakage flux from the primary magnet circuit

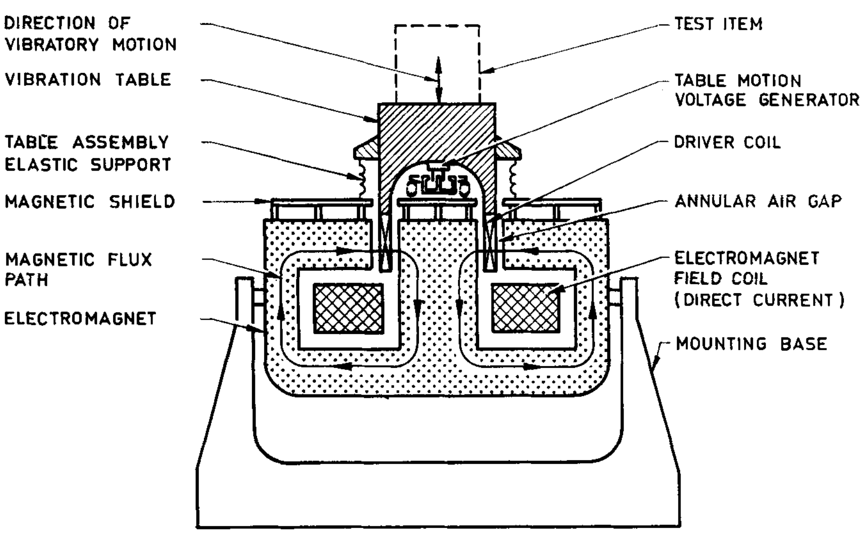


Figure 6. A schematic of the electrodynamic vibration machine. Adapted from Hixson (1961).

at the table. A motor-driven blower is mechanically fixed to the field structure by a larger air duct, and provides forced draft cooling by convection air currents for the field and armature coils.

SHAKER OUTPUT AND THE MEASUREMENT OF VELOCITY

The shaker is equipped with a velocity type pickup, which is installed in the end of the table head proximal to the armature assembly. This pickup, as mentioned earlier, is called the signal generator and produces a voltage while the shaker is in operation. This voltage is proportional to the velocity of the armature with respect to the shaker body, hence it is proportional to the armature velocity when the motion of the shaker body is zero. The output of the signal generator is brought to a connector in the side of the junction box of the shaker, from which it is applied to a voltage sensitive device to observe the output of the shaker.

For effective control of the shaker, a Calidyne Signal Monitor Model 54-C is integrated with the shaker and operates from the output of the signal generator. The signal monitor converts the velocity signal from the signal generator to either displacement or acceleration, which may then be read directly on a meter in terms of peak to peak inches displacement, or vector-g units acceleration. In addition to providing a continuous visual indication of vibration output, the signal monitor is also used for shaker system control.

THE MEASUREMENT OF FORCE

Force is measured at the point of force application using a Strainsert Universal Load Cell: Serial Number Fl-IU-0224. The load cell is a device in which the elastic deformation of a centrally located rod (in this case, the rod of the lower loading head) transmits a force that is measured by a series of strain gages. The strain gage is a variable electrical resistance element that produces a voltage in an electrical circuit which is used to measure the value of the applied force. The load cell has a peak output limit of 1000 lbs (453.0 kgs), this value corresponds to a strain gage electrical output of 2 millivolts/volt. The calibration graph of the load cell is shown in section 3 of the Appendix.

THE INTEGRATED SYSTEM

To measure the relationship between the force and velocity phasor, an analog computer was specially designed and developed to process the output force and velocity signals. The output for the impedance computer consisted of three separate data channels, force, velocity, and their phase relationship. These data were fed into a Brush Mark 260 analog recorder with variable chart speeds of 5, 10, 25, 50, 100, 200 mm/second plus divide-by-100. Two Altec Lancing Corporation Model A 127 and A-287W power amplifiers were integrated with the system. A B/N Model 24-200 solid state power supply was used to drive the system, together with an Exact Function Generator Model 251. The interrelationships between the various components in the instrumentation panel are shown in *Figure 7*.

It is desirable to measure the value of the excitation force without adding any mass to the vertebral unit. Because usual methods of mechanical fixation do not allow this, compensation was made for the additional mass. Using the electrical circuitry in the impedance computer, a portion of the acceleration waveform was fed through a wideband inverting amplifier in the computer package to the force channel of the measurement system for cancellation of the unwanted mass. Keller (1967). The electrical schematic of the computer package is shown in section 4 of the Appendix.

THE TEST FRAME

The reader is referred to *figures 8 and 9* which illustrate the following description. Mounted and mechanically fastened onto the degaussing coil and cast shaker frame was a two-inch thick anodized-aluminum mounting platform, with a nine-inch circular cutout at its geometric center.

The shaker head **K** protruded above the mounting platform and was surrounded by a one half-inch air gap between it and the mounting platform. Mechanically fastened onto the mounting platform were two parallel $2'' \times 2'' \times 3/4''$ square steel tubes **L**, each $30''$ in length and referred to as the primary rails. The $2'' \times 5'' \times 1/4''$ upper rectangular steel cross member **I** has four posi-

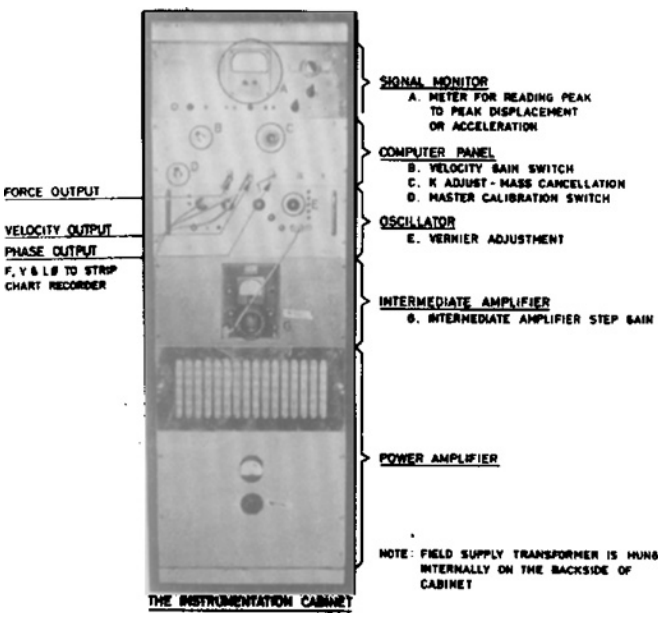


Figure 7. The location of the various components in the Instrumentation Panel.

tions of vertical adjustment along the primary rails. Located within the geometric center of the upper horizontal cross member and with respect to the axis of the shaker head and mechanically mounted within it was a one-inch diameter T-handled captive screw drive **G** with a double lock nut. The movement of **G** occurred in a plane perpendicular to the shaker head.

On the lower end of **G** is the loading head **F E A. A** was the actual loading jaw and was mechanically fastened to **E. A** and **E** were fastened to **F** via three compression springs, which act against a 1.5 inch ball bearing. The overall drawings for **F** and **E** are shown in *Figure 10.* The upper loading head was designed in a manner so that a "pure" compression load could be applied. Mechanically mounted onto the shaker head was an aluminum triangular plate, identified as **C.** In each corner of **C** was a cylinder closed in one end and open in the other, and referred to as a cartridge. There was one such cartridge **D** on each corner of **C.** The cartridges were mounted parallel to each other and to the thrust axis of the shaker head. Within each cartridge **D** was a helicoil spring and a sliding cylindrical bearing block. The spring was in contact and partially enclosed by the base plate **J** and at its upper end in

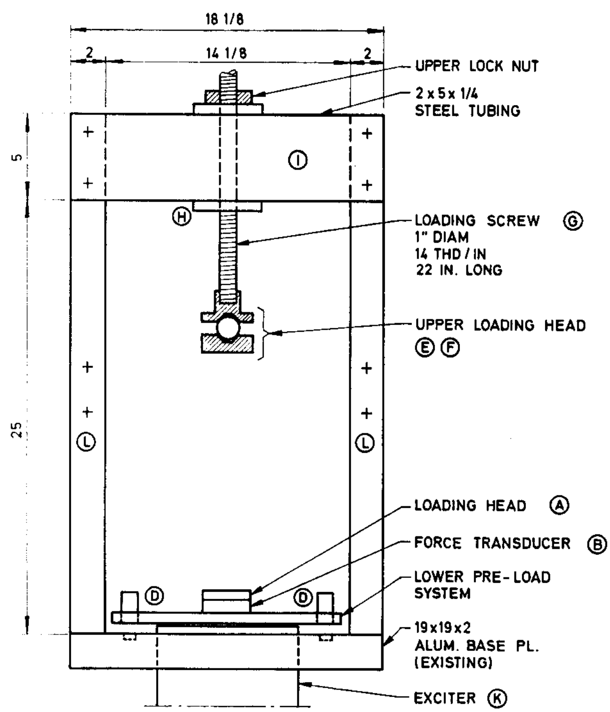


Figure 8. The loading frame.

contact with the sliding bearing block. *Figure 11* is a detailed drawing of the cartridge assembly. The upper end of each cartridge **D** was fitted with an adjusting screw, in axial alignment with the longitudinal axis of the helicoil spring. Tightening or loosening this screw increased or decreased spring compression against the base plate **J** and concurrently increased or decreased the height of the shaker head.

Mechanically mounted onto the triangular platform and in axial alignment with the shaker head and loading screw was the compliance element **B.**

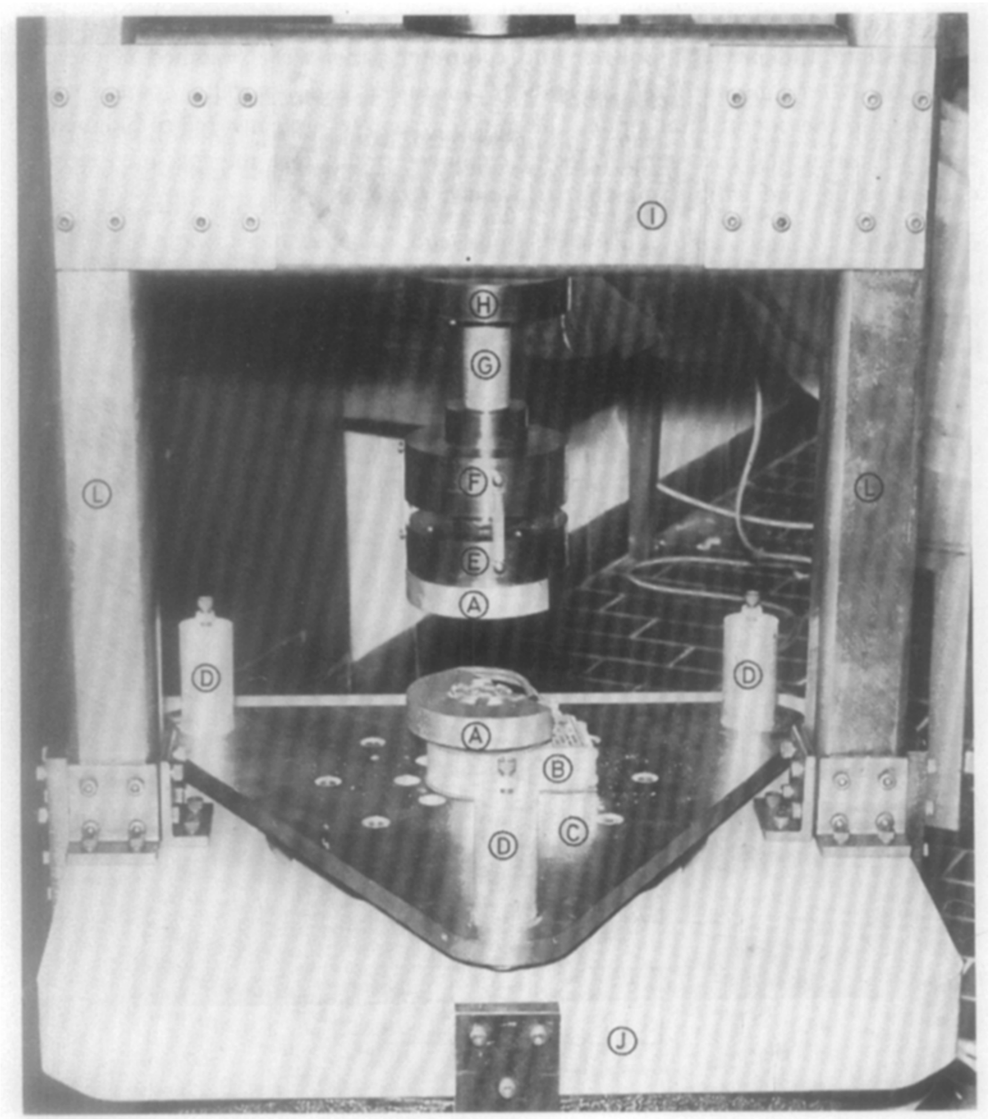


Figure 9. Close-up photograph of preload table, loading head and steel frame.

The upper and lower flat loading jaws were machined from 1.6 inch stainless steel stock. The lower jaw was coupled to the compliance element using a single stainless steel double end threaded rod. The upper jaw was mechanically attached to the upper head fixture using machine bolts.

The engineering specifications for the impedance system are listed in section 4 of the Appendix.

SYSTEM SET-UP

Calibration of force and velocity channels and the strip chart recorder.

The field supply transformer was turned on, the signal monitor, oscillator, intermediate and power amplifier and computer panel were switched to the on position. After one minute, the power amplifier and computer panel were switched to the operational mode. The initial system set-up consisted of balancing and calibrating the force and velocity channels, preloading the specimen to the desired value, and calibrating the strip chart recorder.

Electronic

The following format was followed in the setting up of the impedance computer circuitry. For the initial set-up the following full scale ranges were used for all tests.

Force Channel	— 200 millivolts
Velocity Channel	— 200 millivolts
Real Power Channel	— 200 millivolts

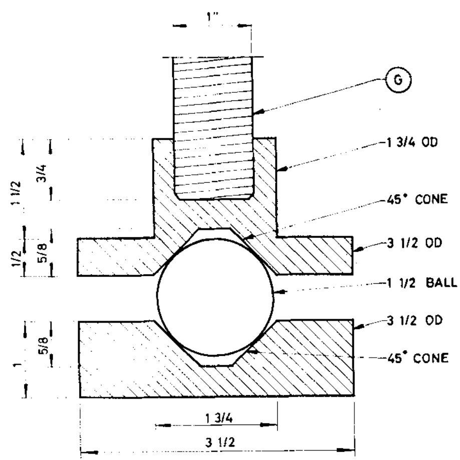


Figure 10. Engineering specifications of the upper loading head.

Zero base lines were verified for each channel. The K-ADJUST potentimeter was set to a value of 8.4 units and locked in position, the recorder started at a slow speed, and the velocity gain set. The master calibration switch on the front panel of the computer package was actuated and produced three D.C. (rectangular pulses), corresponding to the following calibration signals:

Force = 150 lb
Velocity = —5 ips
Real Power = 750 lb-in/sec.

These values were the maximum allowable peak full scale values.

Method for Applying the Compression Bias

The following descriptions, criteria and format were used in applying a compression bias for all impedance tests. The electronic instrumentation was on and the calibration factors for each of the respective outputs were recorded. The strip chart recorder was operating at a slow paper speed (10 mm per second). The value of the desired preload was decided in advance. The prepared test specimen, enclosed in its translucent polyethylene cylindrical stocking (to keep it moist) was placed onto the lower loading jaw. The upper preload head screw was rotated until its loading jaw was in contact with the superior aspects of the specimen. The force channel was carefully monitored. The loading screw was slowly rotated until the designated preload value was registered on the strip chart recorder. Upon achieving the steady state bias, the upper locknut was rotated and locked to secure the location of the preload screw and take up any

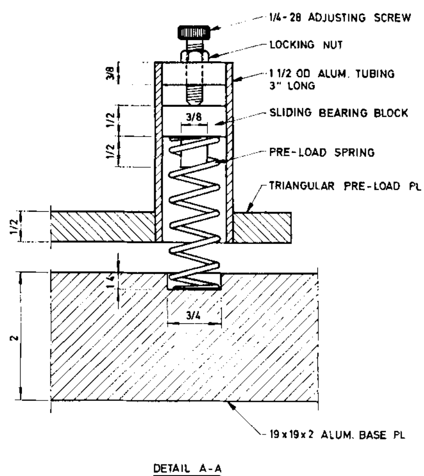


Figure 11. Detail drawing of the preload cartridge.

undesirable slack in the system. Fine force adjustments were then made by turning each screw located on the three cartridges an equal amount.

Three different sets of calibrated helicoil springs were available which could be inserted (prior to the test) within the individual cartridges located on the triangular table. These were identified by color as the bronze springs, the blue springs and the silver springs. A rotation of 15 degrees on the individual preload screws located on top of these cartridges corresponded to a tension load on the mounting platform and compressive load on the specimen of approximately 0.6 lbs (.272 kgs), 1.0 lbs (.454 kgs) and 2.5 lbs (1.14 kgs) respectively for each spring within any particular spring set. See *Figure 12.*

Following the vernier screw adjustments, the corresponding lock nuts were tightened. A cylindrical plexiglas spacer was then inserted between the triangular platform located on the shaker head and the aluminum mounting platform to insure that the shaker armature was in its center equilibrium position. During an impedance test, the specimen oscillated about this equilibrium position while maintaining a compression bias. The value of the compression bias was subject to the following two limitations.

1. For $0 \leq$ preload ≤ 75 lbs (34.05 kgs), the peak oscillating force had to be less than the designated preload (mechanical limitation).
2. For a condition of 75 lbs (34.05 kgs) $\leq$ preload $<$ 150 lbs (68.1 kgs), the peak oscillating force had to be less than or equal to 150 lbs (68.1 kgs) (electrical limitation).

Following application of the compression bias the tangent plane of the upper

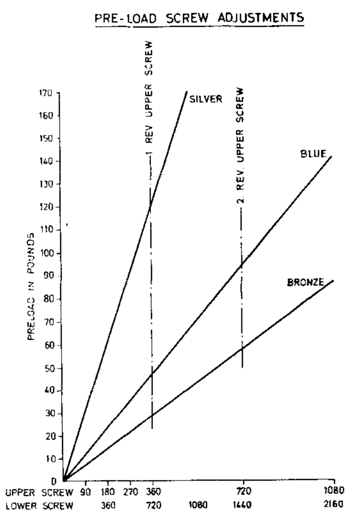


Figure 12. Calibration graph for preload springs.

loading jaw with respect to the ball bearing was ascertained using a circular spirit level. If the tangent plane inclination exceeded 5 degrees, the compression bias was removed, the specimen repositioned and the preload reapplied.

RANGE OF FREQUENCY SWEEP

The human body experiences sequential oscillatory disturbances in the acts of walking, running, riding in various modes of transportation, in heavy equipment operation and also in non-conservative environments such as pilot ejection, parachuting and helicopter flight. Laboratory studies concerned with human response to various biodynamic environments have shown that man is sensitive to mechanical oscillations from below 1 Hz up to a least 100k Hz (some biological effects are not related to frequency at all). The frequency range from about 3 to 30 Hz has been shown to be characteristic of a wide variety of oscillatory disturbances man experiences in the "everyday sense". This constituted the primary rationale to select the particular range of sweep.

COLLECTION AND RECORDING OF IMPEDANCE DATA

A static sweep technique was used to collect the impedance data. The bandwidth of interest was from 5 to 50 Hz. The sweep began at 5 Hz, and within the frequency range between 5 to 10 Hz, data were collected at 1 Hz intervals, and in the frequency range between 10 and 50 Hz at 2 Hz intervals, at the even numbered frequencies 10, 12, 14, 16, 18 . . , 50 Hz.

The initial system set-up was performed as described earlier. The oscillator was set to the desired frequency range, and amplitude control on the oscillator was positioned to the center of drive rotation, with the velocity gain set to the G level. The intermediate amplifier step gain switch was advanced until the desired peak to peak oscillatory motion was reached on the force channel. The oscillator amplitude control was utilized as a vernier adjustment. The peak to peak amplitude on the velocity channel was noted. The velocity gain was increased, using the selector switch on the computer panel, to the maximum value possible but not to exceed 5 ips peak or strip chart sensitivity. The power channel was set for a DC offset. If necessary, the strip chart sensitivity was adjusted for optimum signal resolution.

The gain setting on the intermediate amplifiers was noted, then returned to zero output position to obtain a static reference, serving as the base line for the particular test. The amplifiers were then returned to their previous settings, the values of force, velocity and real power were obtained at a slow paper speed setting and the maximum excursion noted from the displacement meter (this was not done in all tests). The speed of the recorder was increased (200 mm/sec) so that the recorded waveform could be analyzed later. Following up to 500 milliseconds of paper running the recorder was returned to its original slow speed setting. The intermediate amplifier gain was switched to zero, and the strip chart recorder stopped. The test frequency, amplitude of excursion, velocity gain setting, and the strip chart sensitivities were noted and recorded for all three channels. This format and description was the maximum overall hardware set-up for any particular test frequency. A photograph of the overall mechanical hardware and electronic instrumentation is shown in *Figure 13*.

DATA REDUCTION SCHEME

The peak to peak values for both the Force (F_{p-p}) and Velocity (V_{p-p}) were measured for each frequency. The magnitude of the impedance was calculated using the formula,

$$Z = \frac{F_{p-p}}{V_{p-p}}$$

and the calibration values.

To obtain the phase angle $\varnothing$, it was necessary to construct the power factor diagram which is actually a right triangle with the "Real Power" as one side and the "Apparent Power" as the hypotenuse, and the angle $\varnothing$ between the two. The value for the "Real Power" P is equal to the product of the RMS value of the force and velocity. The relationship between peak to peak force and velocity values and RMS values for sinusoidal function are:

$$V_{RMS} = \frac{V_{p-p}}{2\sqrt{2}}$$

$$F_{RMS} = \frac{F_{p-p}}{2\sqrt{2}}$$

From this it follows that the "Apparent Power" can be expressed in terms of the peak to peak values of force and velocity as

$$Apparent\ Power = \frac{(F_{p-p})\ (V_{p-p})}{8}$$

The angle $\varnothing$ can be calculated from the following equations,

$$\varnothing = Cos^{-1} \left[\frac{Real\ Power}{Apparent\ Power}\right]$$

$$or$$

$$\varnothing = Cos^{-1} \left[\frac{8P}{(F_{p-p})\ \ (V_{p-p})}\right]$$

$$or$$

$$\dot{Z} = Z \angle \varnothing = \frac{F_{p-p}}{V_{p-p}} \angle Cos^{-1} \left[\frac{8P}{(F_{p-p})\ \ (V_{p-p})}\right]$$

for a pure sinusoidal function.

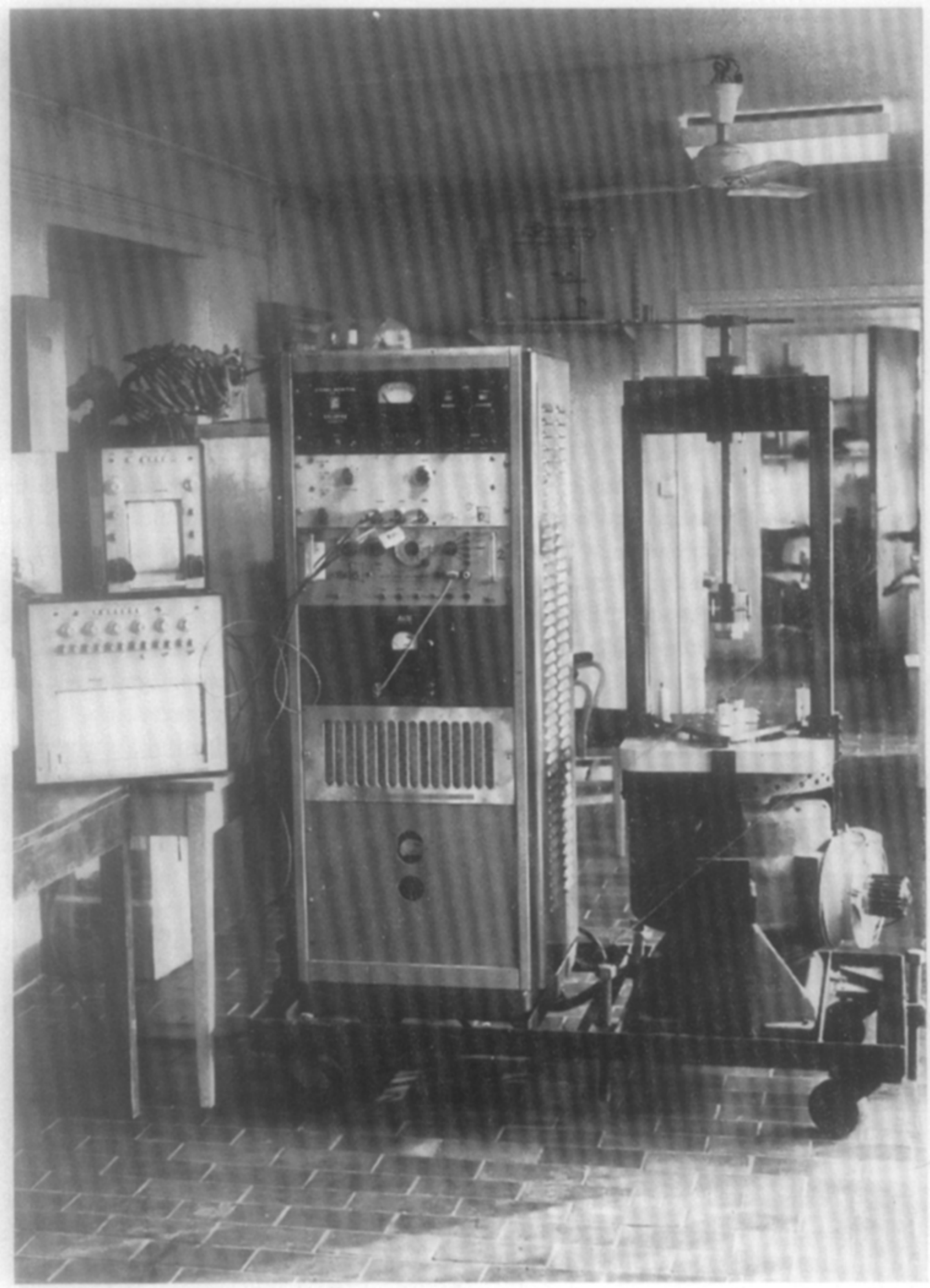


Figure 13. Overall view of mechanical and electronic hardware.

NON-SINUSOIDAL/PERIODIC MOTION

At frequencies less than 15 Hz, the velocity waveform would sometimes deviate from being sinusoidal. This result was attributed to the poor signal to noise ratio of the velocity transducer at small displacements and to mechanical and electrical restrictions placed on the shaker system by the nature of the particular test being performed, and other factors.

Using the procedure just discussed, the value for the impedance and the phase angle were calculated. In determining the scalar portion of the impedance Z, it was found that considerable error could result (100—300 %) if one used the observed value of peak to peak velocity. A Fourier analysis of several waveforms revealed that a considerable amount of harmonic content was present. In particular, the third harmonic was found to contribute 45 % of the total energy compared to the fundamental (5 Hz). The extended velocity peaks were thought to be most likely to contribute to the third harmonic. It was found, by trial and error, that if the velocity waveform was carefully inspected, the fundamental energy content of the vibrations could be located. This was done and a "new" peak to peak distance was found and used to calculate the value of V_{p-p}. As the frequency steps increased, the harmonic content decreased. Above 15 Hz harmonic content was negligible, below 10 Hz it was worse. It was felt that as one used this system and became familiar with the characteristic traits of the waveform, an accurate estimate of the fundamental wave form could be determined by measuring the value of peak to peak velocity. (A Fourier analysis could be performed to determine the fundamental frequency spectrum of the particular waveform in question, but this was not considered necessary.) With regard to the phase angle Ø for non-sinusoidal periodic motion, it was mathematically found, using a Fourier analysis of several randomly selected waveforms, that the value of Ø was apparently not affected by the distorted velocity waveform. The analog multiplier filter circuitry of the impedance computer sufficiently filtered the waveform so that the fundamental frequency could be analyzed.

Typical waveforms are shown in section 5 of the Appendix along with electronic calibration, and waveform analysis information.

STATISTICAL METHODS

Impedance Response Plots

The values of Z and Ø were computed and the relationship of either force leading velocity or velocity leading force visually ascertained at each frequency. In all there were 26 values of Z and 26 values for each phase angle Ø or 52 individual computations for each impedance test. Rather than use all 52 values, the impedance and phase data points at the frequencies of 5, 10, 12, 16, 20, 26, 30, 36, 40, 44, 50 Hz, a total of 11 frequencies denoted as the primary frequencies, were used to construct the impedance nomographs.

The test parameters were

Values of Compression Bias	*Line Identification*
$P_1 = 36$ pounds (16.3 kgs)	———————
$P_2 = 48$ pounds (21.7 kgs)	———●———
$P_3 = 72$ pounds (32.6 kgs)	———×———
$P_4 = 144$ pounds (65.3 kgs)	———o———

	Line Identification
Group 1 = G_1 = less than 29 years of age	———●———
Group 2 = G_2 = between 30 and 50 years of age	———×———
Group 3 = G_3 = more than 51 years of age	———————

The respective impedance values and phase values for the primary frequencies were averaged and plotted according to the following cases.

Case 1. The impedance response was plotted as a function of vertebral unit and compression bias for the respective age groups.

Case 2. The impedance response was plotted as a function of vertebral unit and compression bias, without regard to age. Identified with an (*) in the identification box of each nomograph.

The same symbols were used to plot the values for the phase angles. The actual crossover points through resonance for the phase angle were also averaged.

In many cases only one line is shown for the phase, which means that the F, V deviation from mass control was less than 10 degrees, and the values were lumped together. In each respective experimental section and on each impedance nomograph appropriate symbols are used which should prove helpful to interpret each graph.

5. Experimental Material

THE CADAVER MATERIAL

This investigation is based on quantitative and qualitative observations made on a group of isolated, segmented cervical, thoracic, lumbar, and sacral spines. The specimens were removed from 90 cadavers during routine autopsy procedures conducted at the Karolinska Institute. The vital statistics for each specimen are given in *Figure 14.* The age range of the subjects is from 17 to 90 years. The length of the sacrum excised and tested throughout all these tests varied by 1—3 cm but usually included S_1.

Following death, a vertebral block was grossly removed, sealed in an opaque polyethylene bag, the bag labelled and fresh frozen at —25 degrees C.

EVALUATION OF MATERIAL

Prior to testing, each specimen was removed from the deep freeze and radiographed (while sealed in its bag). The radiographic plates were systematically examined in the form of a general survey during which alignment of the vertebral bodies, the outlines and internal structure of the vertebral bodies, pedicles, apophyseal articulations, intervertebral disks and the condition of the spinal ligaments were considered. Two specimens were found to be in a state where osteophytes of two adjacent vertebrae had fused the vertebral bodies together, thereby forming a bony bridge across the adjoining intervertebral disk and immobilizing the corresponding intervertebral joint. All other specimens were judged as normal with regard to individual age.

PREPARATION OF MATERIAL FOR TEST

Following 12 hours of thawing at room temperature, the vertebral specimen was removed from its opaque polyethylene bag, and dissected free from any loose tissues. No attempt was made to "clean up" the vertebral block. The remaining rib sections in the thoracic column (about 3 cm on each side) were not dissected away. Muscle fibers and tendinous structures were not removed.

The ends of the vertebral blocks were cut parallel to each other through the intervertebral joint space (except for saw cuts through the sacrum), using an electric bandsaw (BKW 1.3 horsepower). Any remaining loose disk or soft tissue perpendicular to the saw cut was scraped clean. The vertebral block was sealed into a translucent polyethylene bag. Elastic bands were stretched, situated around each vertebral end, positioned at the level and point of maximum bony wasting, and snapped into place. The polyethylene-covered ends of the vertebral block

Study number	Initials	Age/sex	Group number	Karolinska number	Cause of death	Unit removed
66	L.D.	22M	1	483	suicide	T_1 — S
69	J.N.	24F	1	423	suicide	T_1 — S
23	A.L.	25M	1	1200	strangulation	T_1 — S
60	K.S.	29F	1	368	fract. vertebra C_1	T_1 — S
68	H.G.	38M	2	1432	strangulation	T_1 — S
13	D.A.	45M	2	2478	suicide	T_{10} — S
39	B.E.	47F	2	415	suicide	T_{10} — S
67	M.L.	49M	2	1060	suicide	T_1 — S
26	E.W.	76F	3	—	suicide	T_1 — S
87	H.B.	17M	1	2510	suicide	T_1 — L_5
71	R.S.	18M	1	1984	suicide	T_1 — S
19	J.T.	22M	1	1037	head injury	T_1 — S
17	L.L.	24M	1	560	suicide	T_1 — S
37	N.H.	25M	1	2430	suicide	T_1 — S
51	F.R.	25M	1	917	suicide	T_1 — S
88	E.T.	26M	1	2500	suicide	T_1 — L_5
59	E.I.	26F	1	364	cardiac failure	T_1 — S
34	M.	30M	2	521	suicide	T_1 — S
61	E.E.	33F	2	2573	suicide	C_4 — T_{12}
86	B.M.	36M	2	2489	suicide	T_1 — L_5
15	S.K.	36M	2	2498	suicide	T_1 — S
13	D.A.	45M	2	2478	suicide	T_{10} — S
4	I.K.	55F	3	394	suicide	C_4 — T_{12}
85	B.J.	56M	3	2481	suicide	T_1 — L_5
7	T.E.	63M	3	2897	suicide	T_1 — S
22	B.F.	76M	3	1040	strangulation	T_1 — S
89	T.L.	77M	3	2492	strangulation	T_1 — L_1
20	H.O.	78M	3	1042	cardiac failure	T_1 — S
90	M.P.	79M	3	2493	suicide	T_1 — S
79	N.N.	84M	3	326	suicide	C_4 — T_{12}
81	E.W.	86M	3	1223	cardiac failure	T_1 — S
54	J.B.	21M	1	352	suicide	T_1 — S
58	C.L.	29M	1	332	suicide	T_1 — S
10	S.T.	46M	2	2939	suicide	T_1 — S
3	P.K.	49M	2	382	chron. alcoholism	T_6 — S
70	H.B.	18M	1	1983	suicide	T_7 — L_3
72	T.M.	27M	1	2099	suicide	T_7 — L_3
24	L.A.	35M	2	1217	cardiac failure	T_7 — L_3
73	L.L.	40M	2	2096	suicide	T_7 — L_3
83	C.B.	85M	3	—	suicide	T_7 — L_3
84	H.A.	90M	3	—	suicide	T_7 — L_3
31	R.E.	25M	1	568	suicide	T_1 — S
50	P.O.	26M	1	903	strangulation	T_1 — S
48	W.M.	27M	2	746	suicide	T_{10} — S
41	G.E.	29F	1	417	suicide	T_{10} — S

Figure 14.

Study number	Initials	Age/sex	Group number	Karolinska number	Cause of death	Unit removed
47	J.S.	29M	1	572	strangulation	T_{10} — S
53	J.F.	30F	2	322	cardiac failure	T_{10} — S
29	K.S.	31F	2	668	strangulation	T_{10} — S
43	K.J.	33M	2	411	head injury	T_{10} — S
33	J.K.	36F	2	502	suicide	T_{10} — S
40	T.R.	36M	2	413	drowning	T_{10} — S
56	H.O.	37M	2	344	strangulation	T_{10} — S
42	N.K.	38M	2	414	perf. ulcer	T_{10} — S
9	T.B.	45F	2	2936	suicide	T_1 — T_{12}
14	B.I.	47F	2	2509	strangulation	T_{10} — S
64	L.H.	50F	3	2571	suicide	T_1 — S
52	P.T.	59M	3	323	pneumonia	T_1 — T_{12}
16	D.J.	60M	3	2523	suicide	T_1 — S
27	F.G.	74M	3	1221	cardiac failure	T_1 — S
82	W.E.	76M	3	1223	cardiac failure	T_1 — S
80	O.St.	78M	3	—	drowning	T_1 — T_{12}
49	B.L.	17M	1	756	head injury	T_{10} — S
75	W.K.	24M	1	1517	suicide	T_1 — T_{12}
76	C.	24M	1	—	drowning	T_1 — S
25	S.P.	24M	1	1220	strangulation	T_{10} — S
77	E.	24M	1	—	suicide	T_{10} — S
74	A.L.	25M	1	1200	strangulation	T_{10} — S
45	V.	25?	1	442	head injury	T_{10} — S
46	K.H.	25F	1	570	head injury	T_1 — S
35	M.K.	25M	1	537	strangulation	T_1 — S
12	L.B.	27M	1	2476	head injury	T_{10} — S
57	A.K.	27M	1	353	suicide	T_{10} — S
78	O.S.	27M	1	—	—	T_{10} — S
55	A.R.	29M	1	348	suicide	T_{10} — S
28	J.B.	29M	2	6260	suicide	T_{10} — S
18	J.H.	34M	2	1023	drowning	T_1 — S
32	A.N.	35M	2	479	suicide	T_1 — T_{12}
30	M.R.	37M	2	672	cardiac failure	T_1 — S
1	J.O.	39F	2	385	strangulation	T_1 — S
2	R.A.	39M	2	375	suicide	T_1 — T_{12}
65	L.A.	42M	2	1408	suicide	T_{10} — S
62	N.E.	42M	2	2577	rupt. aorta	T_1 — T_{12}
44	L.K.	45M	2	419	suicide	T_{10} — S
38	L.O.	47M	2	2437	suicide	T_1 — S
11	K.H.	50M	2	2963	suicide	T_1 — S
63	W.I.	52M	3	2581	suicide	T_1 — T_{12}
36	W.M.	52F	3	2427	strangulation	T_{10} — S
6	L.M.	57M	3	236	cardiac failure	T_1 — S
8	P.K.	69M	3	2926	suffocation	T_1 — T_{12}
5	G.L.	69M	3	243	mult. injuries	T_1 — S
21	A.F.	74M	3	1046	cardiac failure	T_1 — S

Figure 14.

were cut away, allowing the raw vertebral bony surfaces to be in direct contact with the parallel surfaces of the loading jaws.

DISCUSSION OF EXPERIMENTAL METHODS AND MATERIAL

Post Mortem Changes

As with all living systems, when the heart stops beating, circulation is impaired, the internal volume and pressure of the vertebral arteries and plexuses decrease, the system experiences obstruction, retrograde congestion and finally equilibrates for zero volume circulation. The alterations which came about immediately following death and up to the time of post mortem examination evade the control of this investigator. Nevertheless, it was surmised, for the aims of this protocol, that within a fixed period of time subsequent to death and up to post mortem examination, the character of the spinal column impedance had not deteriorated. A 36-hour death-post mortem sampling cutoff time limit was speculatively stipulated.

There are no doubt mechanical and physical properties of the intact system which cannot be comprehended by dissecting the vertebral structure from the cadaver subject and quantitating the passive elemental reaction of this inert tissue. But these problems have yet to be investigated. A scattering of these topics has been looked into by Galante (1967) and Tkaczuk (1968), who carried out some tests pertaining to the variations in water content and environmental influences on small samples of spinal column tissue subjected to unidirectional mechanical loading.

FREEZE STORING OF CADAVER MATERIAL

Prior studies of Åkerblom (1948), Evans (1957), Viidik and Lewin (1966), Sedlin and Hirsch (1966), Rolander (1966), Hirsch and Galante (1967), Galante (1967), Tkaczuk (1968), pursued the role of freeze storing on vertebral bone, intervertebral disk, and ligamentous structure. Their findings assert that freeze storing has little or no influence on the physical-mechanical qualities of these biological materials.

RATIONALE FOR THE MAGNITUDE OF THE COMPRESSION BIAS

No former investigations have been carried out employing a compression bias to explore the response pattern of the vertebral unit. Ruff's (1950) information contains some knowledge regarding the portion of weight supported by the individual vertebral bodies over the length of the spinal column.

Upon considering the broad variations in magnitude of the weight sustained at the individual levels and the limitations of the machine, it was decided that preload values of 36 pounds (16.4 kgs), 48 pounds (21.7 kgs), 72 pounds (32.6 kgs) and 144 pounds (63.5 kgs) would be utilized throughout these series of tests. The selected values are in all probability high in so far as the superincumbent static weight carried by the vertebral bodies in the upper thoracic column is concerned and conceivably low upon reckoning the weight carried by the lower lumbar vertebrae.

In each test, the minimum compression bias was applied initially, subsequent tests were pursued at the successive increasing preload level and so on. The specimens were all dynamically measured with their caudal vertebra in contact with the lower compression jaw.

All tests were performed in such a manner that a slice of vertebral bone was on either side of an intervening disk, a vertebral unit or a region of the spinal column. For each case, the two end vertebrae, or their intervening hard tissue cross-sectional surface area constituted the load bearing surfaces. The geometric centers of the lower and upper vertebral centrum were placed into position under visual control with regard to the loading jaws, after which a compression bias was applied. Immediately, three sources of error were discernible:

1. The bending cross-section of the vertebral unit was not symmetric with respect to the loading head. The magnitude of the principal load could vary at any point while the principal direction remained fixed.
2. The position of the vertebral arches may or may not contribute to the bending rigidity of the vertebral unit.
3. The distribution of compressive load from one vertebral body to the next may or may not be uniform.

The compression bias and dynamic oscillatory loading of the specimen took place by means of the simply supported hard tissue bearing surfaces, which were also able to excite a number of unharmonious modes of vibration including lateral, torsional, and bending modes and/or some combination of these. Since the specimen being measured was tightly clamped between the shaker head and the preloading head, its primary mode of reaction was rationally postulated to be that of the vertebral unit compressing and relaxing. The nature of the damping was likely to be less than critical in some of these excitation modes; consequently differential motions with resonant amplification could occur.

In the living state the spine is under the influence of the many trunk muscles. Each vertebral body has muscle forces acting directly on it and a portion of the body weight directly transmitted to it. With regard to this study, the forces required to produce the primary range of motion were applied in an extremely crude manner, a manner that can in no way compare to the forcing functions and the neuromuscular apparatus in the living state. The application of an external load through the lowermost and topmost vertebra has been pointed out by Novogrodsky (1911) to be anatomically unrealistic. Nevertheless, it was felt that this approach offered a practical way of simulating a functional system that may be approached by analytical means. (The shorter the specimen length, the more realistic the test and the less the influence of section geometry on the measured impedance. Individual fundamental units, vertebrae, and disks were also tested using this technique but these results are to be published elsewhere).

ON THE METHOD OF FREQUENCY SWEEP AND CONSIDERATIONS REGARDING REPETITIVE LOADING

The frequency sweep utilized began at the 5 Hz lower frequency limit and increased to the 50 Hz limit. There is no information available, known to this writer, which correlates steady state excitation and peak response. The interval within each frequency domain was arbitrarily stipulated to be 15 seconds. This time duration could be modified as experience deemed apropriate in order to achieve a steady state response and repeatable and reproducible waveforms.

The literature contains some experimental knowledge of repetitive loading characteristics of the vertebral elements. Göcke (1928), Virgin (1951), Brown, Hansen and Yorra (1957), Hardy, Lissner, Webster and Gurdjian (1958), Rolander (1966), and Galante (1967). However, their tests have been more or less static in nature and their results cannot be applied directly or contrasted to these studies.

6. Observations and Experiments

INTRODUCTION

Initial static force deflection data were accumulated using the impedance tester. The following phenomena were qualitatively observed.

When a static compressive load was applied to the vertebral unit via the captive drive and the force channel of the strip chart recorder monitored, the magnitude of the compressive load was observed to decrease gradually. If an additional load was superimposed upon the vertebral specimen, the force channel deflection proportionally increased but within a period of time (about 30—180 seconds) a similar time-dependent behavior happened, except that the decrement of compressive load was reduced, and the time period to equilibrate to the new level was also less, when compared to the previous loading history. If an additional increment of load was applied to the compression bias a similar pattern of behavior was noted (the load increased, then decreased, but again the load loss was less and the equilibration time decreased). Following a static load application of 145 lbs. (65.8 kgs), 5 lbs. (2.3 kgs) less than the limitations of the tester, equilibration occurred within 5—8 minutes of load application. These generalized results were noted from a static test incorporating the $L_1L_2L_3$ vertebral segment. Similar observations were made for other vertebral units. In each case the magnitude and time of the exposure to load governed the reaction of the vertebral unit at any particular vertebral level.

Preliminary repetitive impedance tests were carried out using these vertebral units (each made up of three vertebral bodies and their intervening disks). The following observation was made. When a single impedance test was conducted and its results contrasted to the second impedance test, the reduced oscillograph tracings were noted to shift. Presumably, this reaction may be attributable to some sort of transient adaptive vertebral unit behavior. Comparative examination of oscillograph recordings between the first, second, third and fourth and subsequent tests suggested that the behavior may equilibrate. The oscillograph tracings also showed that an acceleration of transient actions may occur under dynamic loading, which appeared noticeably greater at certain frequencies of excitation and less at others.

With the highly limited supporting facts available, any decisions concerning the definition and manner of test procedures was difficult. It was therefore decided to conduct preparatory investigations with special attention given to the following questions:

1. Does there occur a loss in height of a vertebral specimen as a function of

constant load application? It was felt this question should be answered before tackling any dynamic aspects of the problem.

2. If so, does this loss in height vary when contrasted to other vertebral units taken from the same level, by how much, in what manner, and how is the resistance of the transregional zones affected?

3. Are the reactions of a vertebral unit subjected to constant load influenced by the factor of age or other age-related morphological changes?

4. Can the posterior facets of the vertebral unit be separated without affecting the mechanical behavior of the total vertebral unit?

5. In what fashion is the impedance response plot altered with regard to questions 2, 3 and 4?

The fourth question was included as a matter of peripheral interest to clear up any questions that may be posed with regard to this experiment and to shed some light on the inconsistencies postulated in the literature.

Five experiments were carried out in an attempt to answer these questions and were identified respectively as:

Experiment A — The response of the vertebral unit to constant compressive loading.

Experiment B — The dynamic response of the unconditioned and conditioned vertebral segment.

Experiment C — The impedance response of the unconditioned vertebral disk.

Experiment D — Creep response at the various vertebral levels.

Experiment E — The effect of the posterior arch on the anterior spinal column.

EXPERIMENT A — THE RESPONSE OF THE VERTEBRAL UNIT TO CONSTANT COMPRESSIVE LOADING

Experimental Design—Cadaver Material

To provide further illumination of the problems posed, the test apparatus shown in *Figure 15* was made operational. It essentially consisted of a plexiglas container, a weight box, two flat platens (one mechanically attached to the underside of the weight box, the other mechanically fixed onto the wooden floor of the plexiglas container; each plate was parallel to the other), a block and tackle, and a portable humidifier with its attachments. The entire unit was referred to as a Creep Conditioning Chamber (CCC).

The static preload including the weight of the box equalled 48 lbs. (21.7 kgs). The temperature and relative humidity were maintained at approximately 34 degrees C and 99 percent relative humidity. Cadaver numbers 66, 69, 23, 60, 68, 13, 39, 67, 26 from *Figure 14* were utilized.

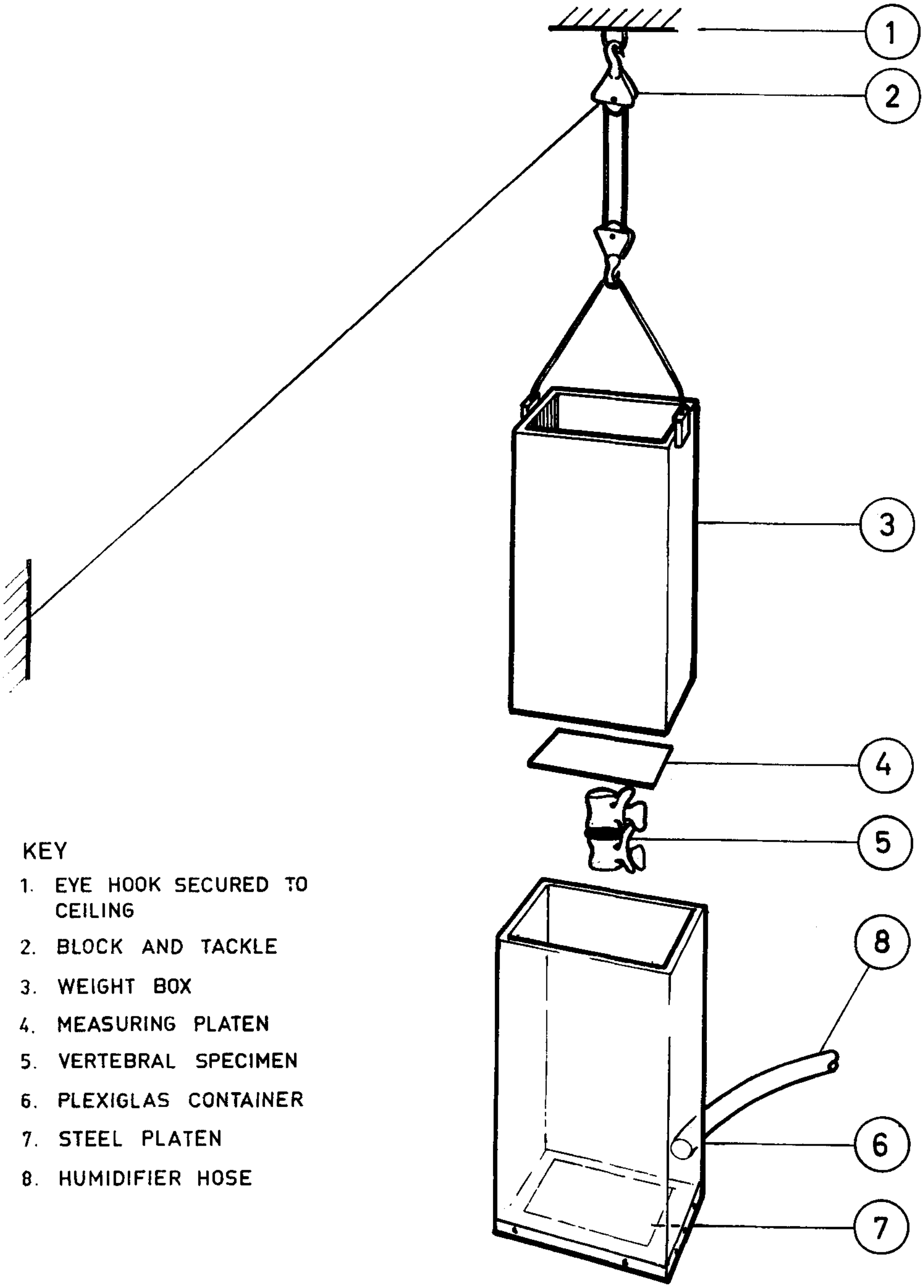


Figure 15. The creep conditioning chamber (CCC).

EXPERIMENTAL PREPARATIONS AND PROCEDURES

Initial tests were carried out to evaluate the magnitude of the load essential to initial creep. For instance 5 lbs (2.3 kgs), 10 lbs (4.5 kgs), 15 lbs (6.8 kgs) or 20 lbs (9.1 kgs) etc. were placed within the weight box and positioned onto the axially aligned specimen. The time duration for any particular test was of the order of 15 to 120 minutes. These low load values proved ineffective to initiate any well-defined creep reaction in a lumbar spine. On the other hand, it was found that a 20 lbs (9.1 kgs) load applied to a high thoracic unit did initiate creep. Repeating this procedure with several vertebral units brought to light that a critical level of load was necessary to commence with the creep. (But, the magnitude of creep was also dependent on age and vertebral level.)

The specimens were divided in accordance with the scheme shown in *Figure 16*. The vertebral section to be utilized was removed from its polyethylene bag, and longitudinally placed onto the lower platen of the CCC. (The remaining spinal units, not to be immediately tested, were enclosed in a cellophane bag and placed inside a refrigerator). The weight box was fitted within the plexiglas chamber and eased toward the specimen. At the moment the weight box made physical contact with the specimen, the position of the weight box relative to the plexiglas housing was scribed onto four strips of gummed paper strategically located near the four corners of the weight box. These lines were identified as the baseline for each gummed strip. The weight box was gently lowered until its full dead weight came to rest onto the axially aligned specimen and its position relative to the plexiglas housing was promptly marked and identified as point B. A fifth gummed paper strip was placed onto the weight box parallel and next to one of the four gummed strips. Its purpose will be discussed later. The portable humidifier was activated. The position of the weight in contrast to the plexiglas housing was manually scribed onto the four gummed paper strips at time intervals of 30, 60, 120, 180, 240 and 300 minutes or until the rate of creep was sufficiently reduced to the level where the adjacent scribed lines (about 0.3 mm wide) were noted to overlap and/or the total variation of three successive measurements could not be differentiated by eye. Whenever either action occurred, the fifth gummed strip was scribed, the dead weight eased off the vertebral unit. When the interface between the platen and vertebral unit was separated another line was scribed onto the fifth gummed strip. The vertebral unit was now defined as "*conditioned*". The conditioned vertebral unit was removed from the CCC. The four gummed strips of paper were eased off the surface of the weight box. The distance between the base line and the line B was identified as ΔB, or the change in height of the vertebral unit with respect to the applied compressive load. The distances between the scribed lines for each gummed strip were measured and recorded, their values for the respective time intervals summed, then averaged. The identical procedure was followed for the refrigerated specimen, removed approxi-

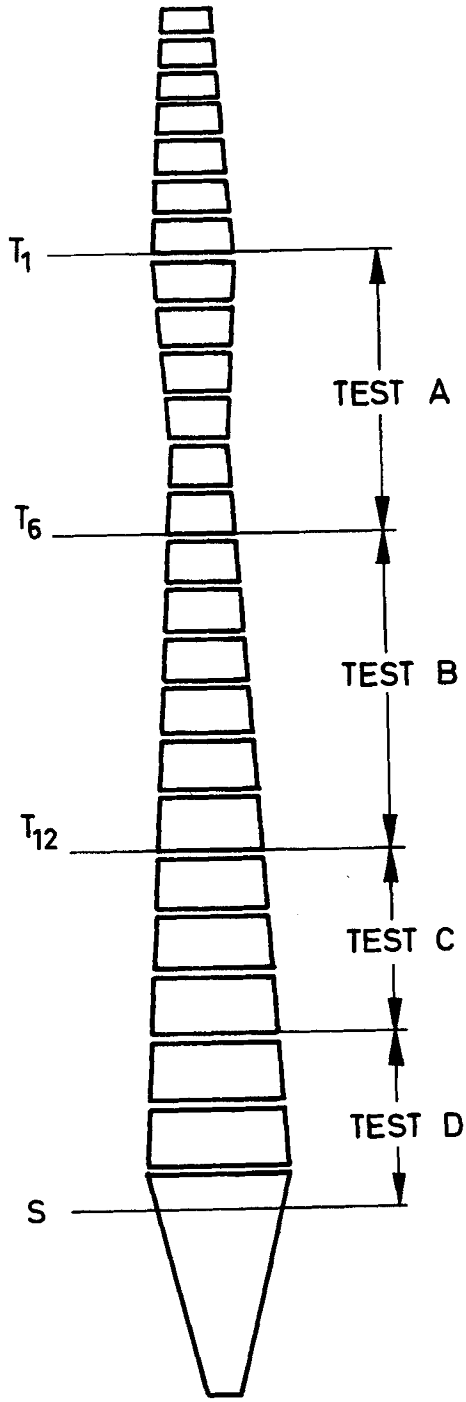


Figure 16. The division scheme for creep tests.

mately two hours prior to its test time. The fifth gummed strip was removed, the "recoil" was measured, and recorded.

RESULTS AND DISCUSSION

The results of this experiment are shown in tabular form in *Figure 17*. It is evident that the excised defrosted vertebral unit subjected to uniaxial dead weight load exhibits creep tendencies. The process is of a mechanical nature and is clearly time-dependent.

The data reveal that a considerable measure of creep occurs within the first 30 minutes of any particular test and that within the first three hundred minutes the creep subsides to a minimum rate. The younger age group cadaver material appears to need a greater time interval to approach a level of stability, whereas the "older" material conditions to the load in a relatively shorter period of time. However, this time-dependent action also appears to be a function of vertebral unit level. Test A (T_1—T_6) conditioned to the constant load in a much shorter time period than did the material for Test D (L_3—S), but here again an additional biomechanical factor must be taken into account, the lumbosacral angle. (According to Hagelstam [1949], the shearing component is greatest at the lumbosacral intervertebral disk where the longitudinal axis of the last lumbar and first sacral vertebrae under normal circumstances form an angle of aproximately 135 degrees. Perhaps if this creep test had been designed to accomodate for both the shear and axial components of the "normal" load carried, the results presented for this vertebral level might have been different.)

Test B (T_7—T_{12}) persisted to creep for a longer period of time than did Tests A, C, or D. The cause for this response was uncertain, as were the results reflected in Test C.

Figure 18 is a plot of the average and range of total axial height decrease for the respective test specimens. The vertebral units used for Test B showed the greatest loss in axial height, followed by the vertebral units in Test C, A, and D, in that order. Test D reflected the least average height decrease.

A generalized schematic representation of the deformation accumulation and load redistribution is exhibited in *Figure 19*. The nature of the creep was such that the creep rate for a constant load decreased markedly during the initial 30 minutes of the experiment. This was followed by a period of decreasing creep rate which may continue for some hours. Eventually the creep tendency ceased and the vertebral unit maintained itself at a particular level of deformation, a function of the applied critical static load.

Two factors with regard to the conditioning process seem to be predominant:

1. *Load Dependence:* Here there is no general conclusion. Each vertebral unit interface has variable loading magnitude which varies throughout the length of the spinal column and seems to be related to the body weight carried at a particular vertebral level.

	ID-AGE	ΔB	30 MIN	60 MIN	120 MIN	180 MIN	240 MIN	300 MIN	TOTAL LOSS (mm)
TEST A	23 - 24	3	6	–	–	–	X		7
	60 - 28	2	4	–	–	–	X		5
	66 - 22	2	4	–	–	–	X		5
	13 - 45	3	3	2	–	–	–	X	6
	67 - 49	3	4	1	–	–	X		6
	68 - 38	1	8	–	–	–	X		8
TEST B	23 - 24	7	6	3	2	–	–	–	11
	60 - 28	6	5	2	1	–	–	–	9
	66 - 22	6	3	1	2	1	–	–	8
	67 - 49	6	2	2	2	1	–	–	7
	68 - 38	4	2	2	1	1	–	–	6
	39 - 47	4	2	2	1	2	–	–	9
	26 - 45	7	3	1	3	–	–	–	8
TEST C	23 - 24	9	8	–	–	–	X		9
	60 - 28	4	5	1	–	–	–	X	7
	69 - 24	3	2	2	1	–	2	1	8
	67 - 49	6	3	1	–	–	–	X	5
	13 - 45	6	2	3	1	–	–	–	6
	39 - 47	5	3	2	3	–	–	–	8
	26 - 75	8	2	1	–	–	–	X	4
TEST D	23 - 24	6	4	–	–	–	X		4
	60 - 28	5	3	1	–	–	–	X	4
	68 - 38	3	1	1	–	–	–	X	3
	67 - 49	3	1	1	–	–	–	X	3
	13 - 45	2	1	1	–	–	–	X	3
	39 - 47	2	2	1	1	–	–	X	4
	26 - 75	1	1	–	–	–	X		3

NOTE:
1. ALL VALUES GIVEN IN MILLIMETERS AND ROUNDED OFF ($<5=0$; $>5=1$)
2. ΔB IS NOT SUMMED WITH TOTAL LOSS COLUMN
3. VALUE OF TOTAL LOSS BASED ON TOTAL MAGNITUDE OF CREEP
4. STATIC LOAD LONGITUDINALLY APPLIED 21.7 kg (48 pounds)

KEY:
– = LESS THAN .5 mm CHANGE PER 60 MINUTES
X = CONSIDERED AS CONDITIONED - TEST TERMINATED

Figure 17. Creep test results.

2. *Time Dependence:* Each vertebral unit has its characteristic equilibration time for the applied load which may be related to age.

Contrasting the magnitude of deformation under constant load, the various spinal units exhibit different rates as well as magnitudes of deformation for the different vertebral levels. These combined factors suggest that an excised vertebral unit has a memory for adaptation with regard to its previous loading history and envelope of deformation.

If the initial loading and subsequent reactions introduced vertebral deflections, the following phenomenon took place when the dead weight load was removed.

1. There occurred an immediate slight lengthening (recoil) of the vertebral unit, leaving the specimen with a length somewhat shorter than its original value, however.
2. If the vertebral unit (free of load) was allowed to rest undisturbed in a humidified chamber, it gradually extended in length, and might or might not eventually go back to its original length. (The energy input was larger than the energy released—as may be expected).

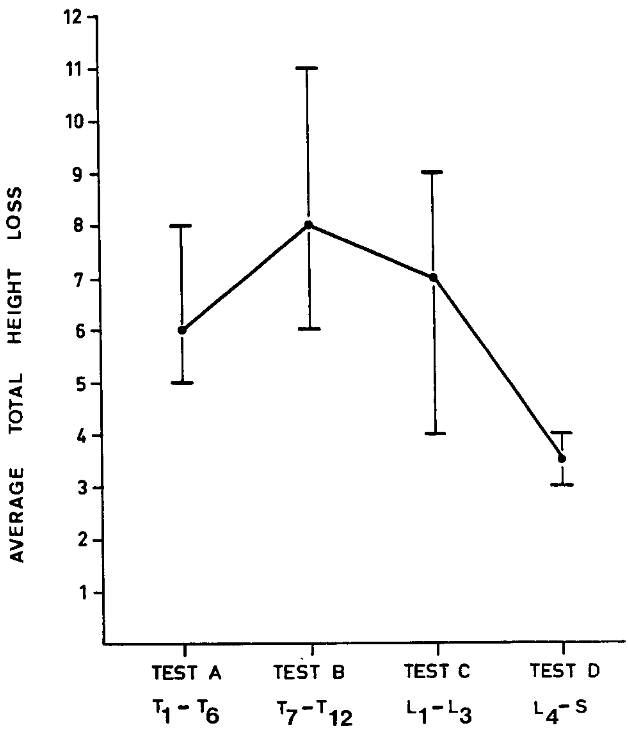


Figure 18. Average total height loss for the individual vertebral units.

The magnitude of the recoil was found to be dependent upon the length of the specimen and averaged about three to four millimeters.

It may be speculated that this responsive action is related to the axial height of the disks in this region which may provide greater deflection capability and/or morphological changes of the vertebral unit. Farfan and Sullivan (1967), Farfan (1969), Farfan, Cossette, Robertson, Wells, Kraus (1970).

Based on data to be published elsewhere as well as data presented here, there seems to be some indication that with increasing age the magnitude of the creep for any given vertebral unit remains approximately the same, but that the vertebral unit conditions itself to external loads in a much shorter period of time.

The physical interpretation for this effect may be that with increasing age the common soft tissue structures are no longer able to resist external loads. For instance, in the case of static compressive loading, the limiting deflection is reached as the external load on the vertebral unit is increased. But in this case the internal resistance was increased to a "conditioning" plateau by the in-

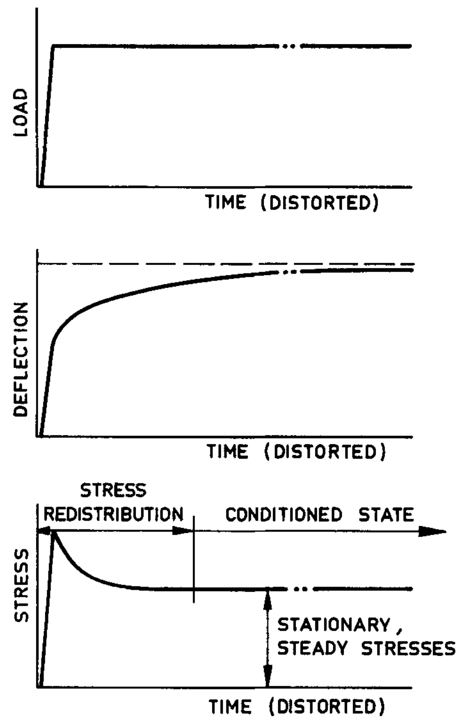


Figure 19. Schematic representation of strain accumulation and stress redistribution following loading.

creases in deflection alone, as the external load remained constant or until the internal load had become exhausted.

Based on these data, and a follow-up study on the creep response (to be published) at several values of constant load, the load deflection response of the vertebral unit is generally portrayed in *Figure 20* and elucidated in the following:

Upon application of a dead weight load, the vertebral unit undergoes an immediate deflection and subsequently a loss in vertebral unit height, followed by an equilibration after some time.

Each vertebral unit has a particular normal range of loading which differs as a function of its level. Consequently each disk within a vertebral unit and its common ligamentous structures at that vertebral unit level possess a characteristic envelope of load response or domain orientation with regard to load saturation. With increasing age the slope of the load deflection curve becomes steeper and equilibration occurs in a much shorter period of time (for the same level of vertebral unit). Beyond this level it is postulated that the disk material itself and the posterior joints interact to produce a final steep upward trend in the curves.

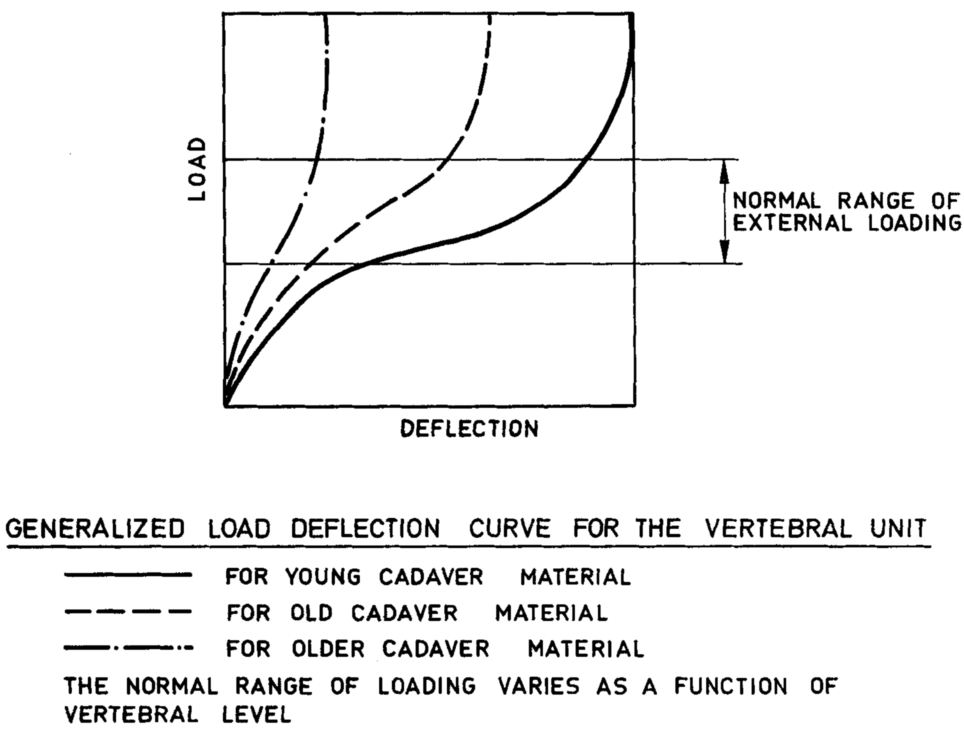


Figure 20.

EXPERIMENT B — THE DYNAMIC RESPONSE OF THE UNCONDITIONED AND CONDITIONED VERTEBRAL SEGMENT

Objectives and classification and preparation of material.

In order to corroborate creep conditioning and impedance response magnitudes, cadaver specimens listed below from *Figure 14* were impedance tested, creep conditioned, and again impedance tested, after which the posterior arches were removed and the anterior column was again impedance tested.

The cadaver material was divided into the following vertebral units

Experiment	*Vertebral Unit*	*Cadaver No.*
1.	Transregional Cervical-Thoracic Unit C_4—T_4	79, 61, 4, 81
2.	Thoracic Unit T_1—T_6	20, 22, 17, 19, 51, 37, 15, 34, 59, 71, 7
3.	Thoracic Unit T_7—T_{12}	20, 22, 17, 19, 51, 37, 15, 34, 59, 71, 7
4.	Transregional Thoracic-Lumbar Unit T_9—T_{10}—T_{11}—T_{12}	79, 61, 4, 81
5.	Lumbar Unit L_1—L_3	20, 22, 17, 19, 51, 37, 15, 34, 59, 71, 7
6.	Lumbar Unit L_4—S_1	20, 22, 17, 19, 51, 37, 15, 34, 59, 71, 7
7.	Transitional Unit T_{11}—T_{12}—L_1	85, 86, 87, 88, 89, 90

The adjacent units of the spinal column not to be promptly tested were sealed in a polyethylene bag and placed in a refrigerator. The unit within the spinal segment to be utilized was impedance tested, withdrawn from its polyethylene bag and tested in the CCC, as delineated in the previous experiment.

The static load applied and the magnitude of the compression bias for each case was 21.7 kgs (48 lbs). Upon equilibration of the creep rate, the vertebral unit was placed back into its plastic bag and promptly recycled between the jaws of the impedance tester. The unit was removed from the tester and stripped of its polyethylene cylindrical stocking, the anterior and posterior vertebral column were separated using an electric band saw, and the change in height for each structure noted. The anterior vertebral column was placed back into its cylindrical polyethylene stocking, secured between the jaws of the testing machine, a preload was applied, and then impedance was tested. The overall flow diagram for each experiment is shown in *Figure 21*. For instance, if a T_1—T_{12} segment was obtained at autopsy, it was divided at the intervertebral disk between T_6 and T_7. If a T_{10}—S_1 was removed, T_{10}—T_{12} was separated, then

L_1—S_1 was impedance tested and sawed at the intervertebral disk between L_3—L_4.

The data are shown in accordance with the statistical scheme shown earlier. The various letter symbols employed on the nomographs are to be interpreted as follows:

R = raw vertebral unit in the defrosted state.
C = conditioned vertebral unit, a static preload of 21.7 kgs (48 lbs) was used for all tests
WO = without posterior arches

		Total number of subjects
G_1 = Group 2	<29 years of age	6
G_2 = Group 2	between 30—50 years of age	4
G_3 = Group 3	>51 years of age	6

The final nomograph of each experiment illustrates the average impedance implications for that individual vertebral level excluding the age factor and as a function of the preload value. It is identified by an asterisk (*) in the identifi-

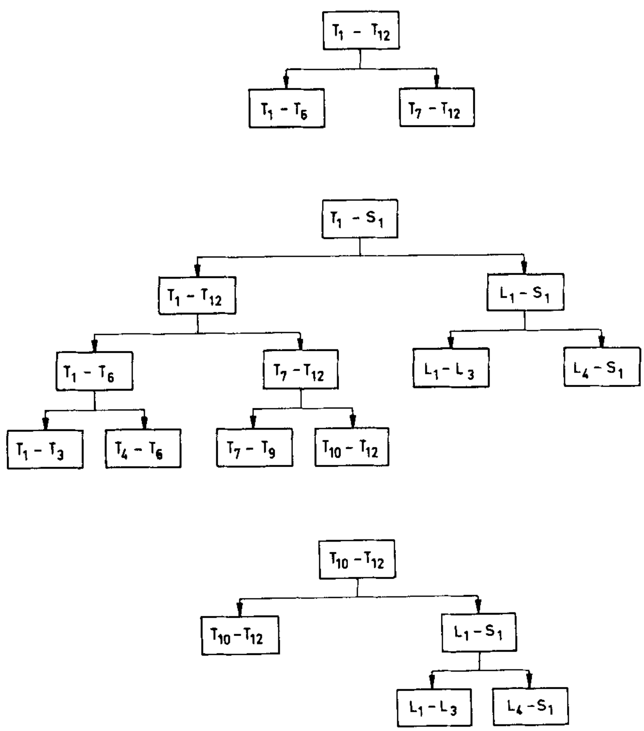


Figure 21. Flow diagram for Experiment B.

cation box and the letter D trailing the figure number. The number of subjects used to construct each impedance plot and their age is provided in the subsequent summary:

Experiments 2, 3, 5, 6
Total Number of Cadaver Subjects-11
Number of Specimens in Each Age Category
$G_1 = 6$
$G_2 = 2$
$G_3 = 3$

Experiments 1, and 4
Total Number of Cadaver Subjects-4
$G_1 = 0$
$G_2 = 1$
$G_3 = 3$

Experiment 7
Total Number of Cadaver Subjects-6
$G_1 = 2$
$G_2 = 2$
$G_3 = 2$

Results and Discussion

The primary impedance values were summed, averaged and plotted for experiments 2, 3, 5, 6. Experiments 1, 4, 7 were not divided according to age groups. Rather, the impedance values for each vertebral unit were lumped together and presented as raw **(R)**, conditioned **(C)**, and without facets **(WO)** and a **D** curve was then compared.

The results are best visualized by examining *figures 22, 23, 24, 25, 26, 27* and *28*. In general, following a period of conditioning, the vertebral unit was found to possess different impedance characteristics as compared to its unconditioned state. The vertebral unit could exhibit a softening or hardening characteristic with respect to the degree of conditioning, the compression bias, and the vibration environment. (If a vertebral unit became stiffer with increasing frequency it was termed a hardening unit, while if it became softer with increasing frequency it was termed a softening unit.) In the case of a hardening vertebral unit the impedance value was found to increase with vibration frequency, thus causing any resonance response to move towards the higher frequencies (longer segments). Conversely, if the impedance value was found to decrease with increasing vibration frequency, the resonance was found to move toward the lower frequencies. At the lower frequency levels the vertebral unit remained at approximately the same Z values, but shifted upward or downward as the frequency increased with respect to the various age groups.

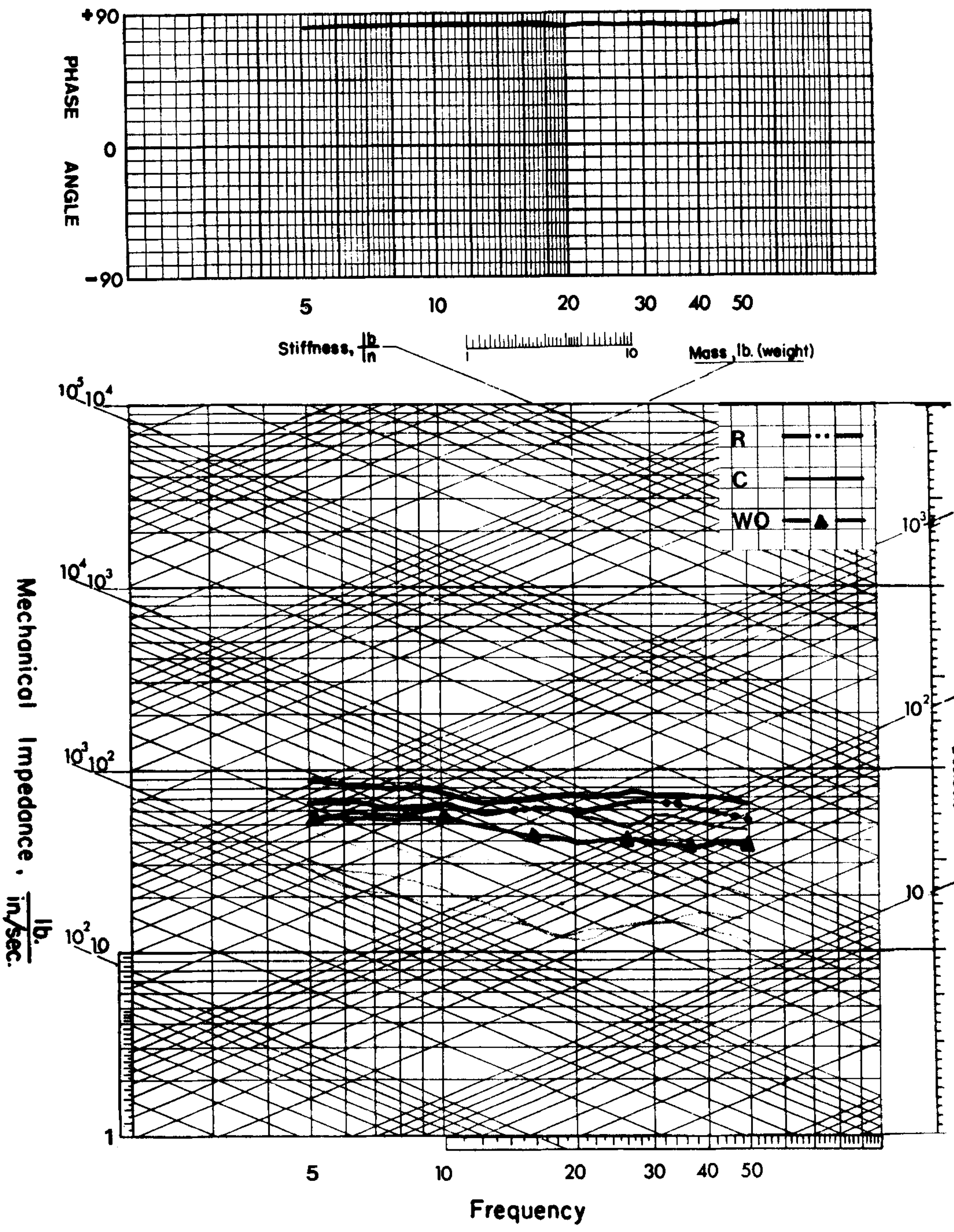


Figure 22. C_4—T_4 / R—C—WO.

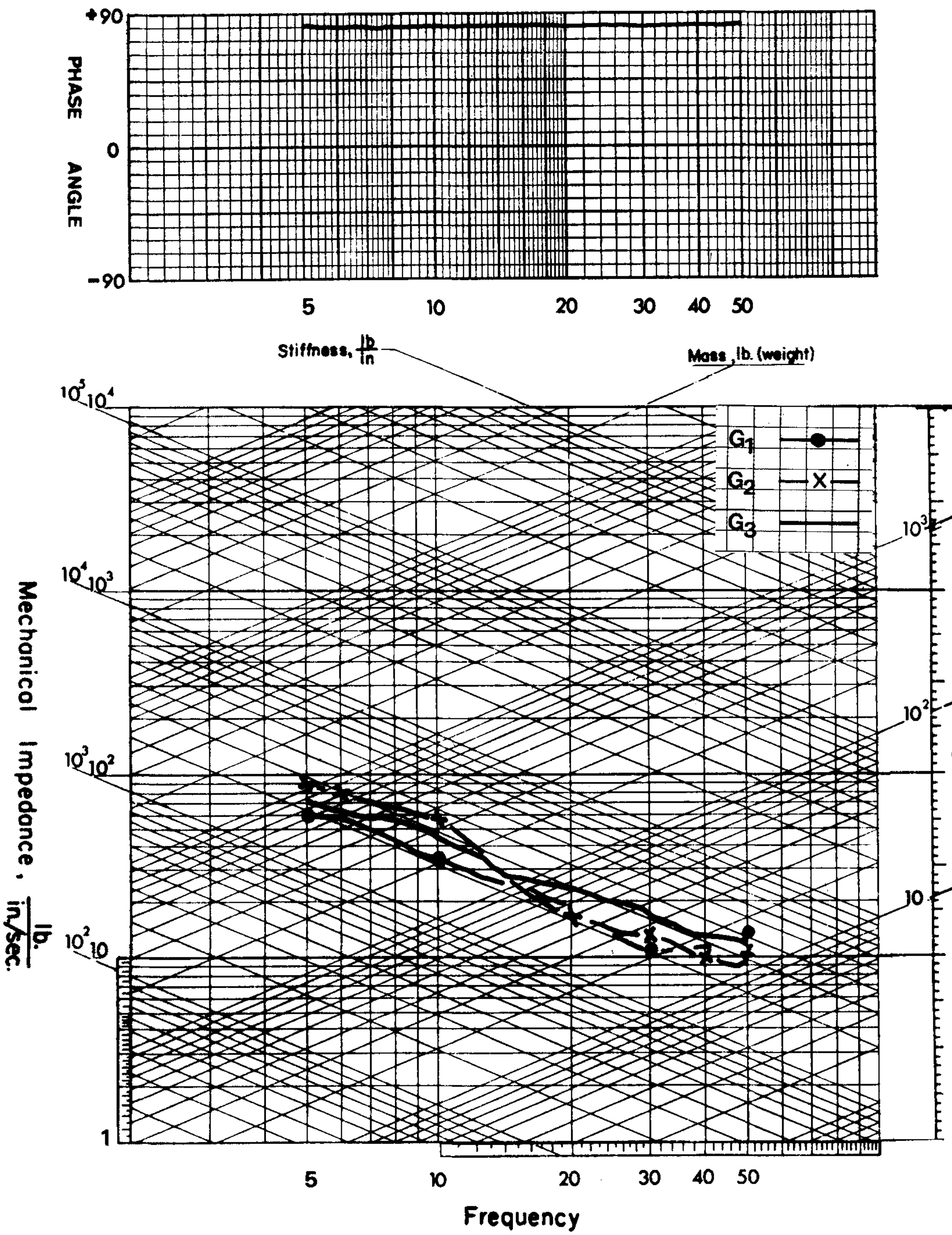


Figure 23 A. T_1—T_6 /G_1—G_2—G_3 / R.

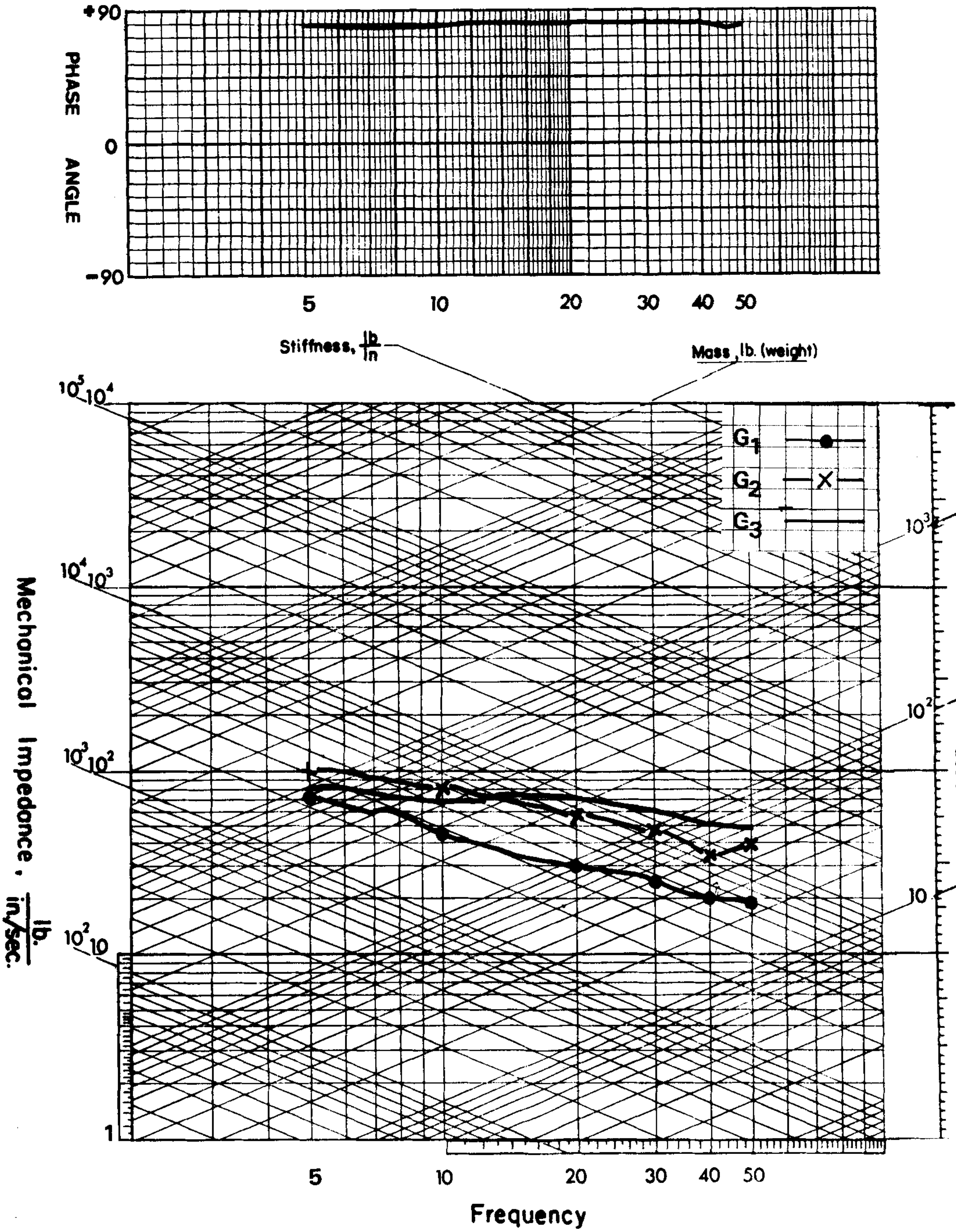


Figure 23 B. T_1—T_6 / G_1—G_2—G_3 / C.

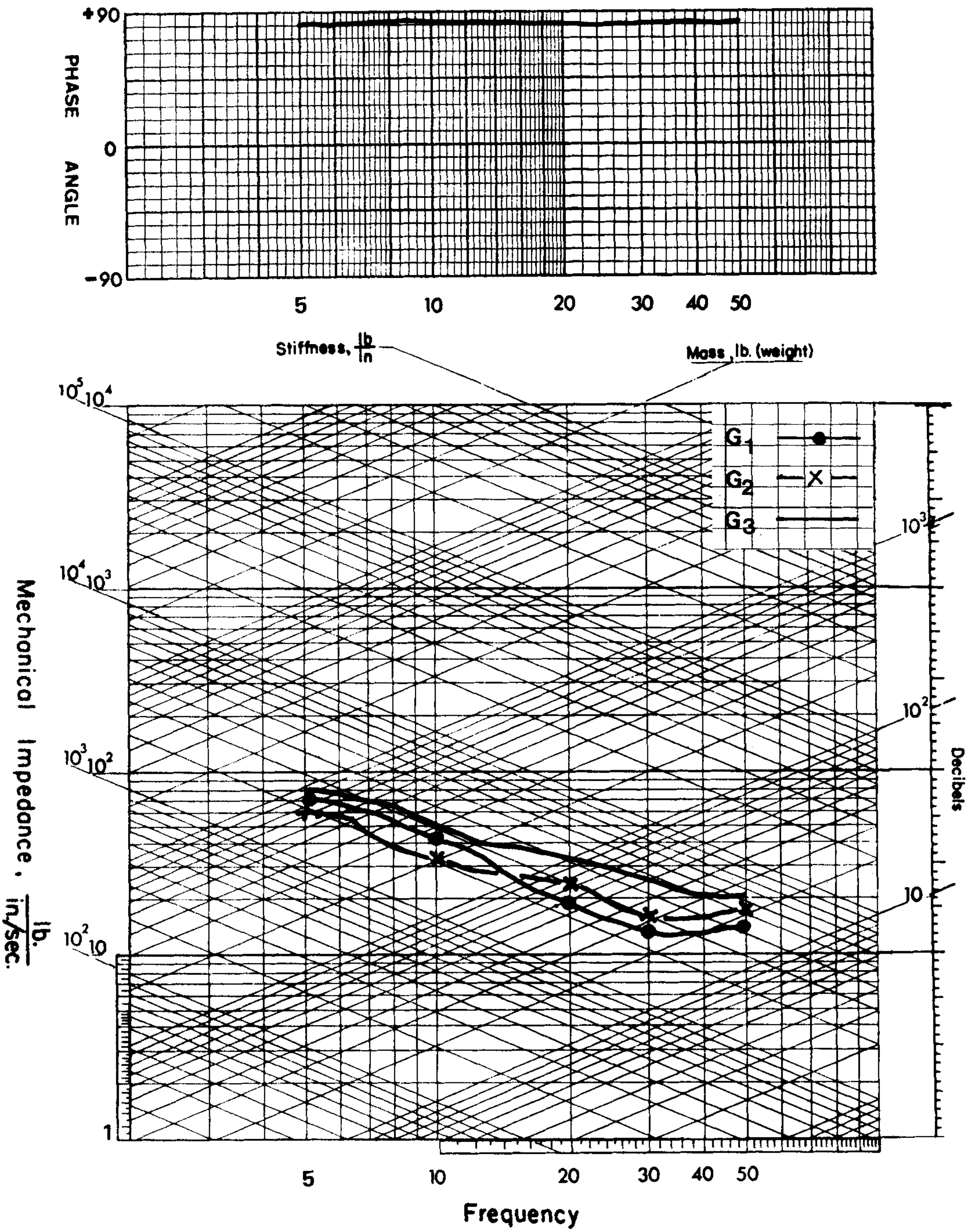


Figure 23 C. T_1—T_6 / G_1—G_2—G_3 / WO.

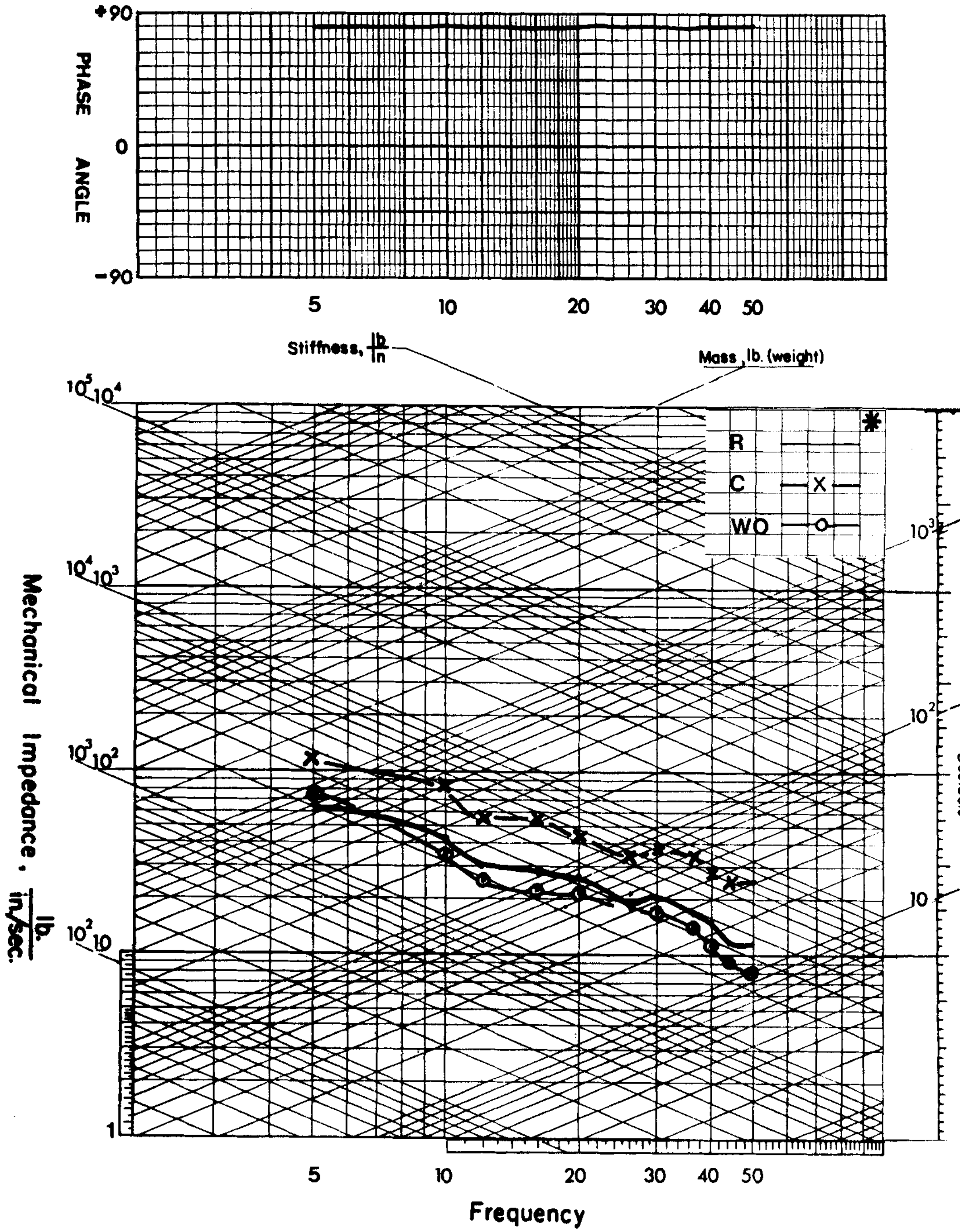


Figure 23 D. T_1—T_6 / R—C—WO / D.

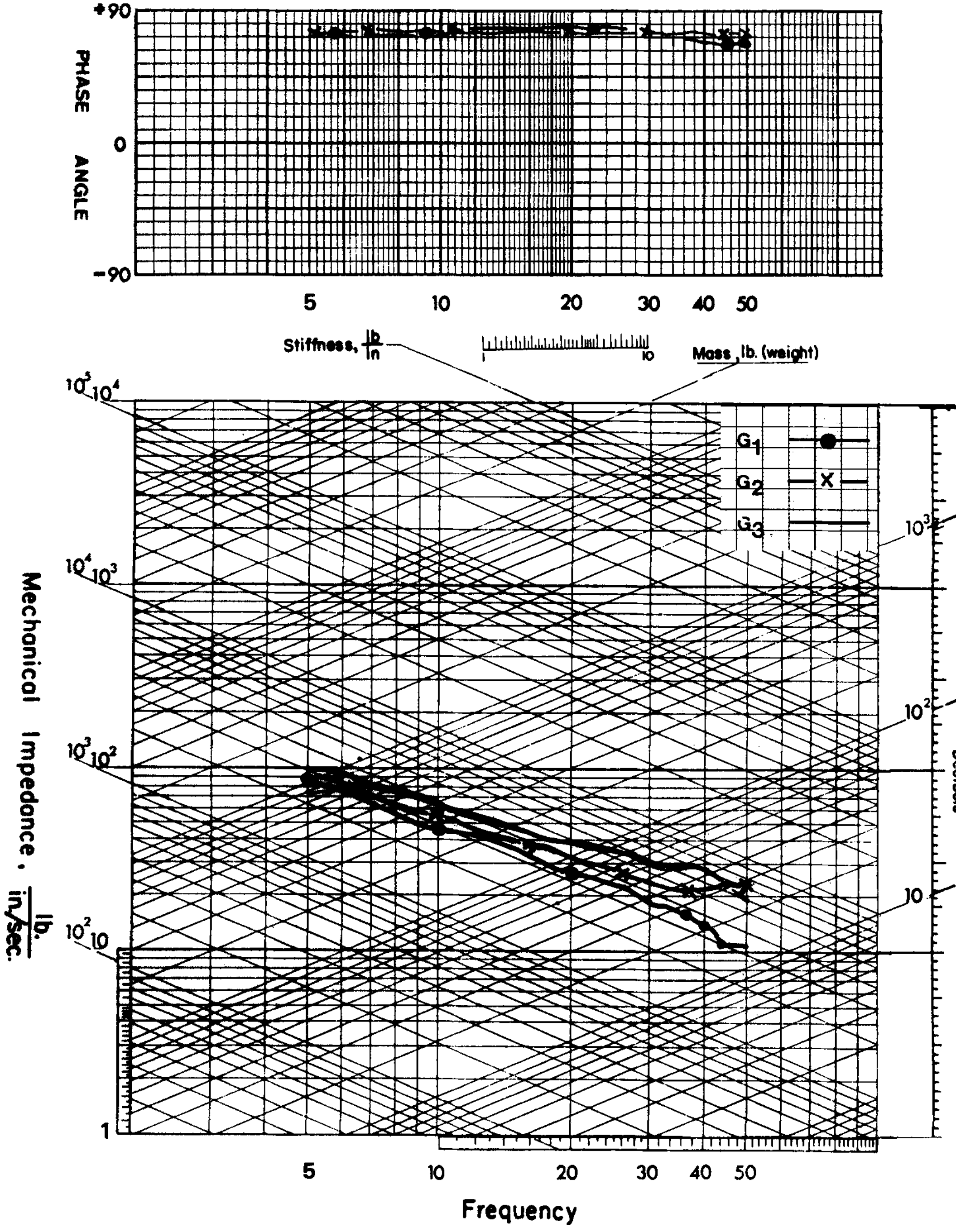


Figure 24 A. T_7—T_{12} / G_1—G_2—G_3 / R.

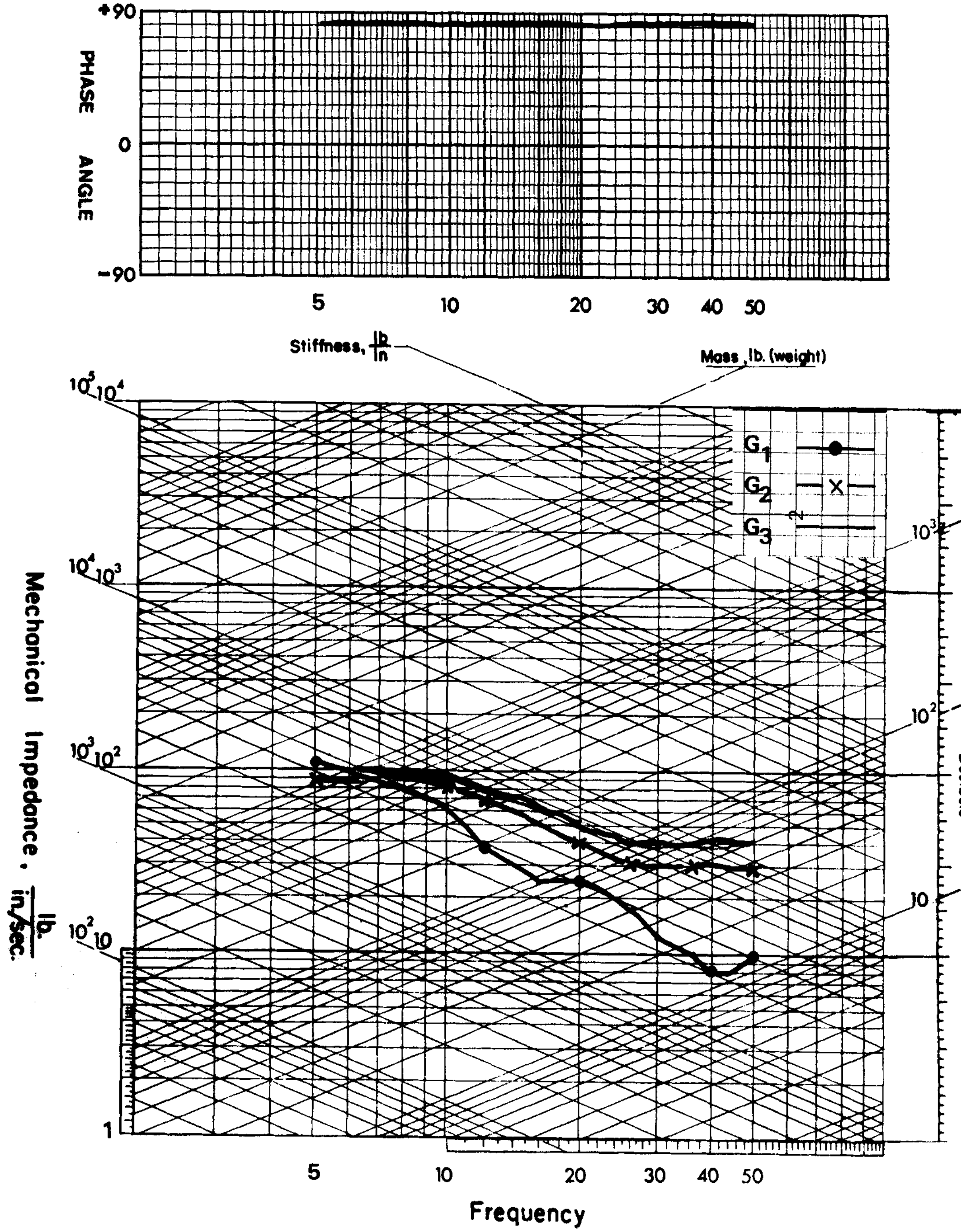


Figure 24 B. T_7—T_{12} / G_1—G_2—G_3 / C.

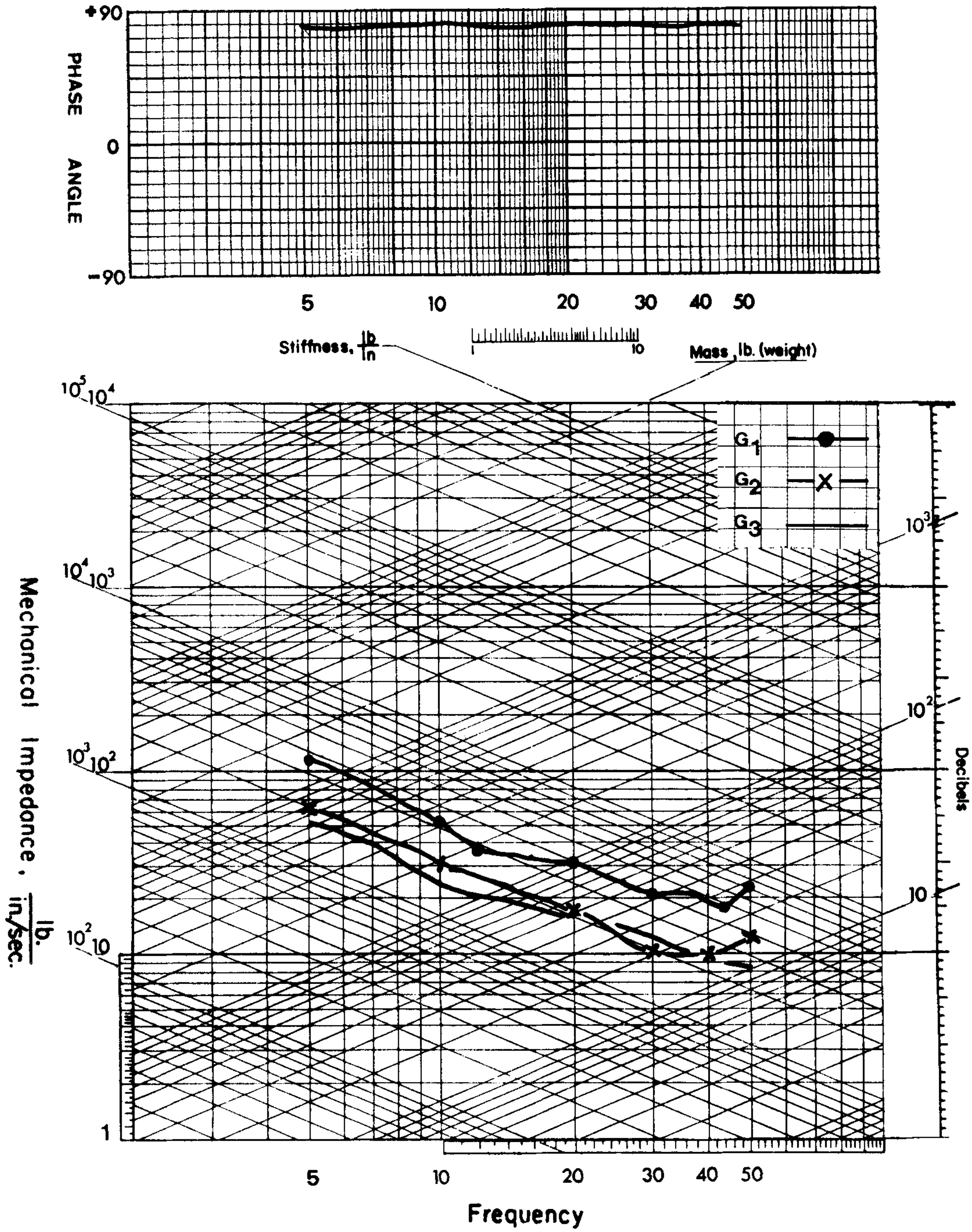


Figure 24 C. T_7—T_{12} / G_1—G_2—G_3 / WO.

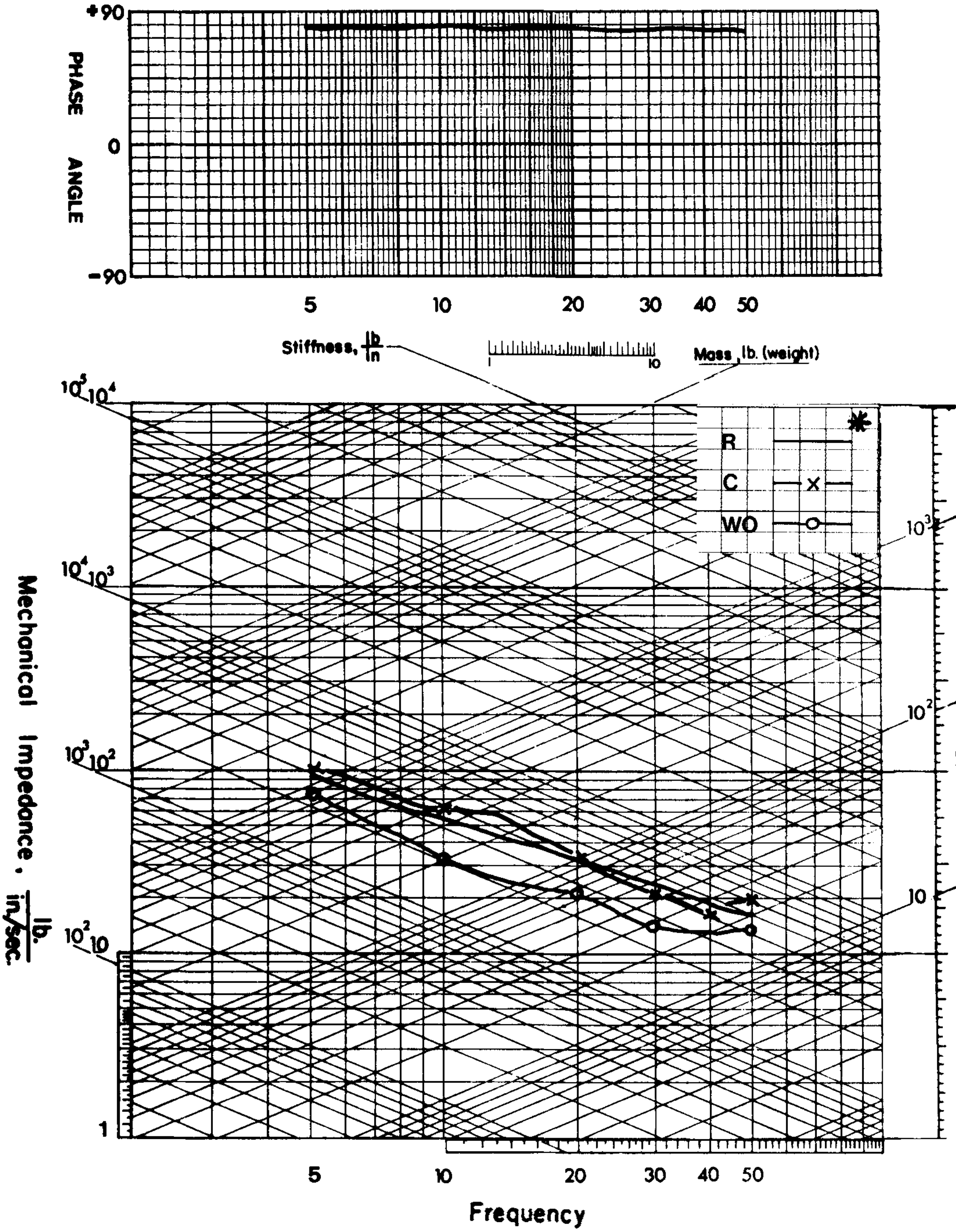


Figure 24 D. T7—T12 / R—C—WO / D.

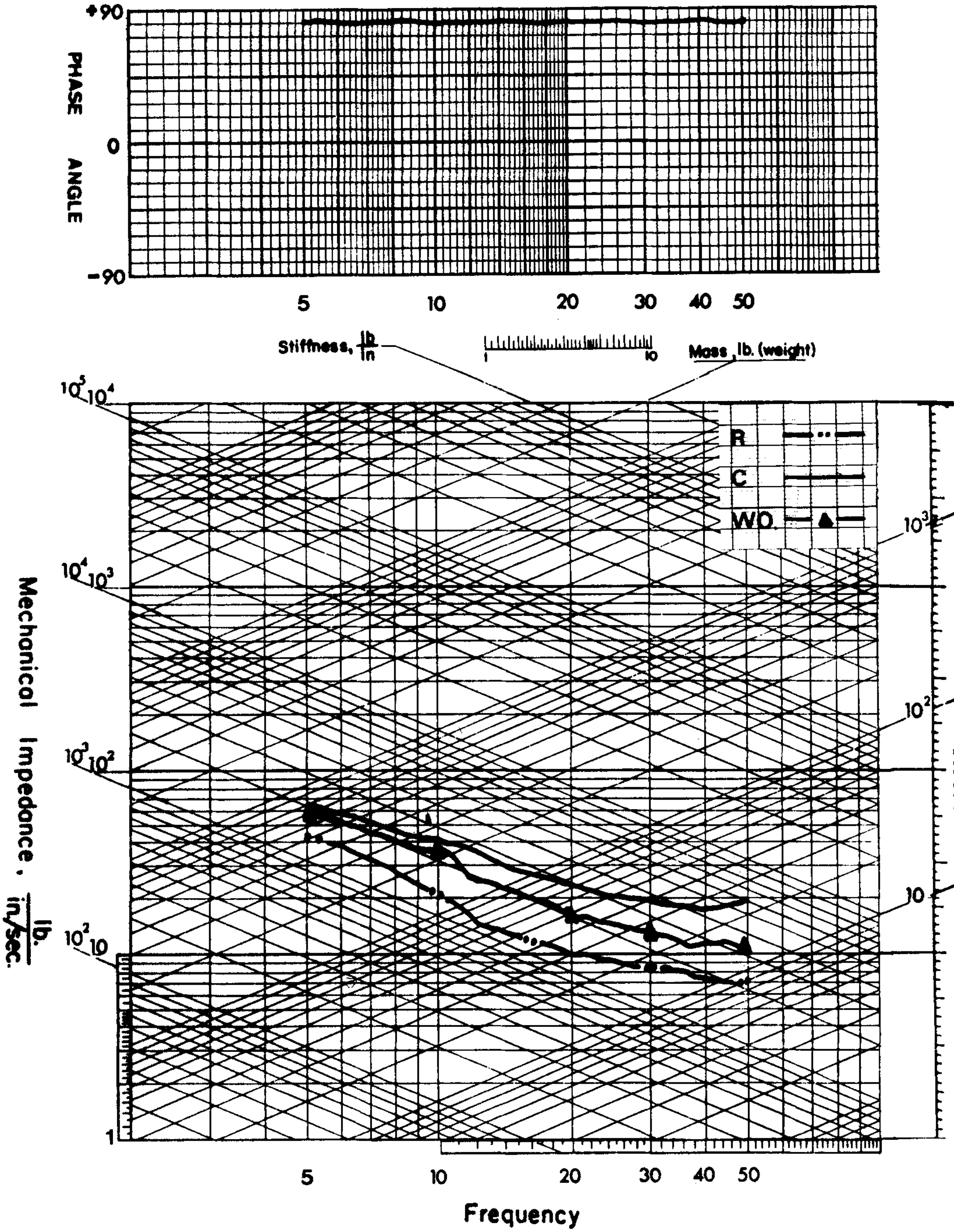


Figure 25. T_9—T_{12} / R—C—WO.

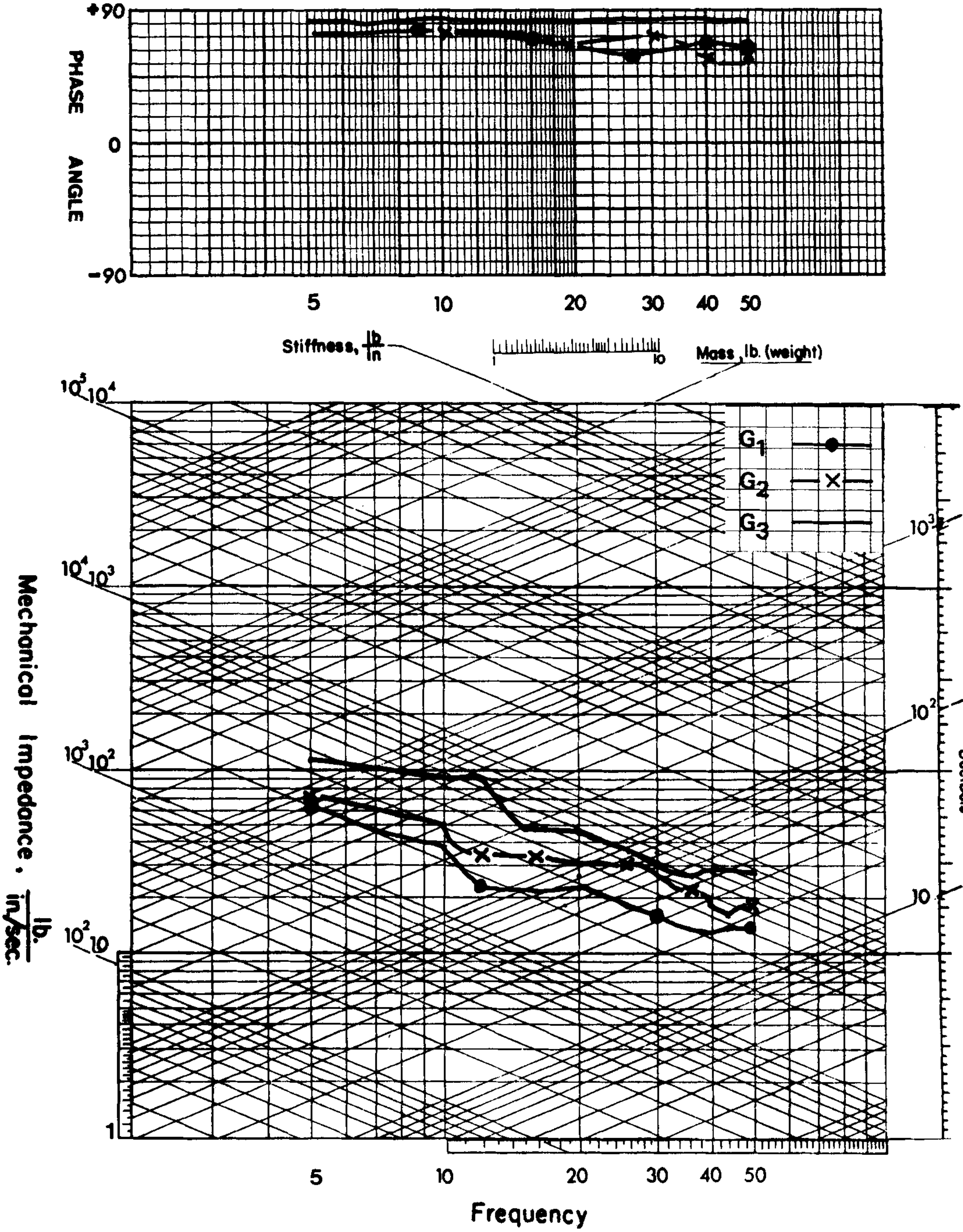


Figure 26 A. L_1—L_2—L_3 / G_1—G_2—G_3 / R.

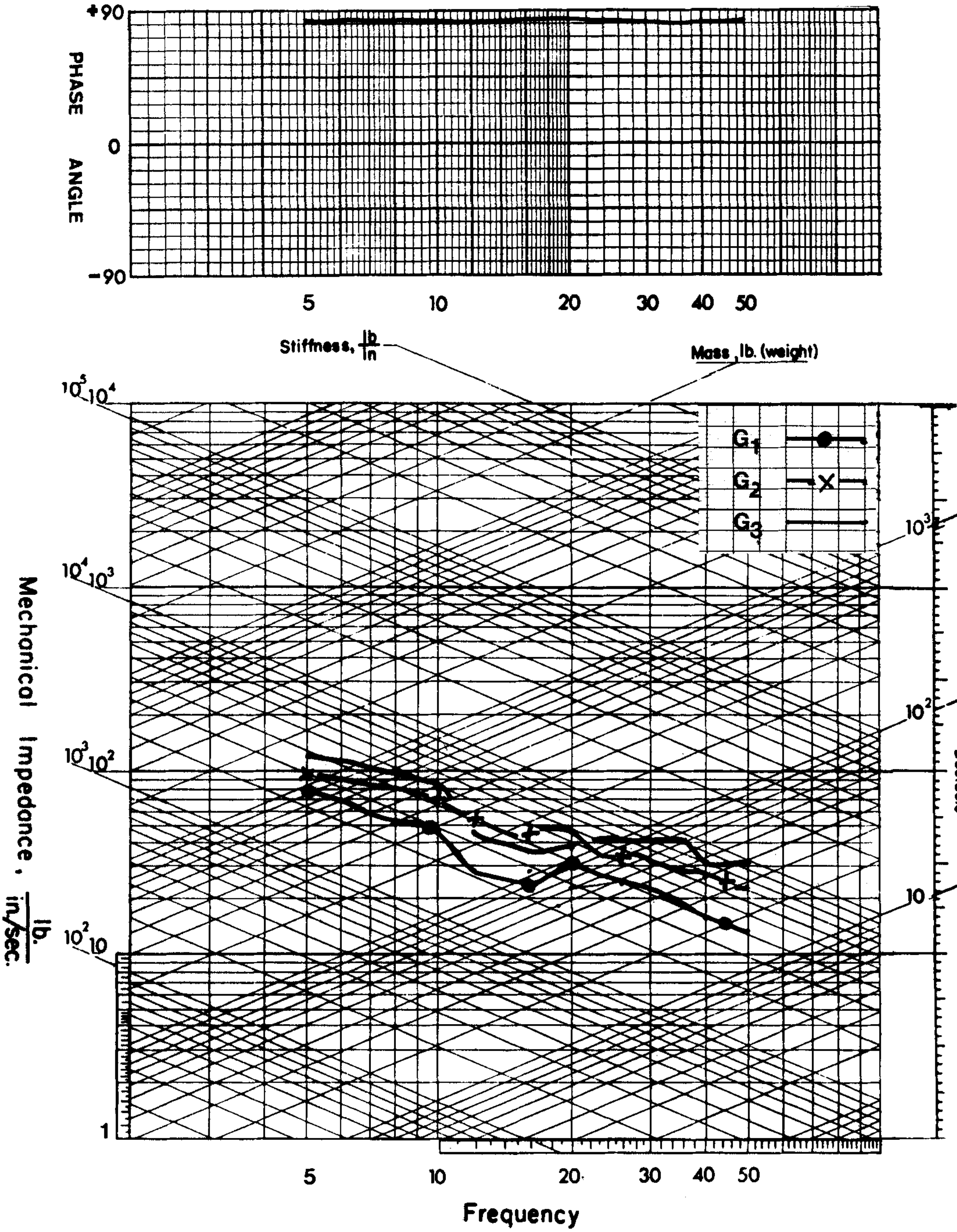


Figure 26 B. L_1—L_2—L_3 / G_1—G_2—G_3 / C.

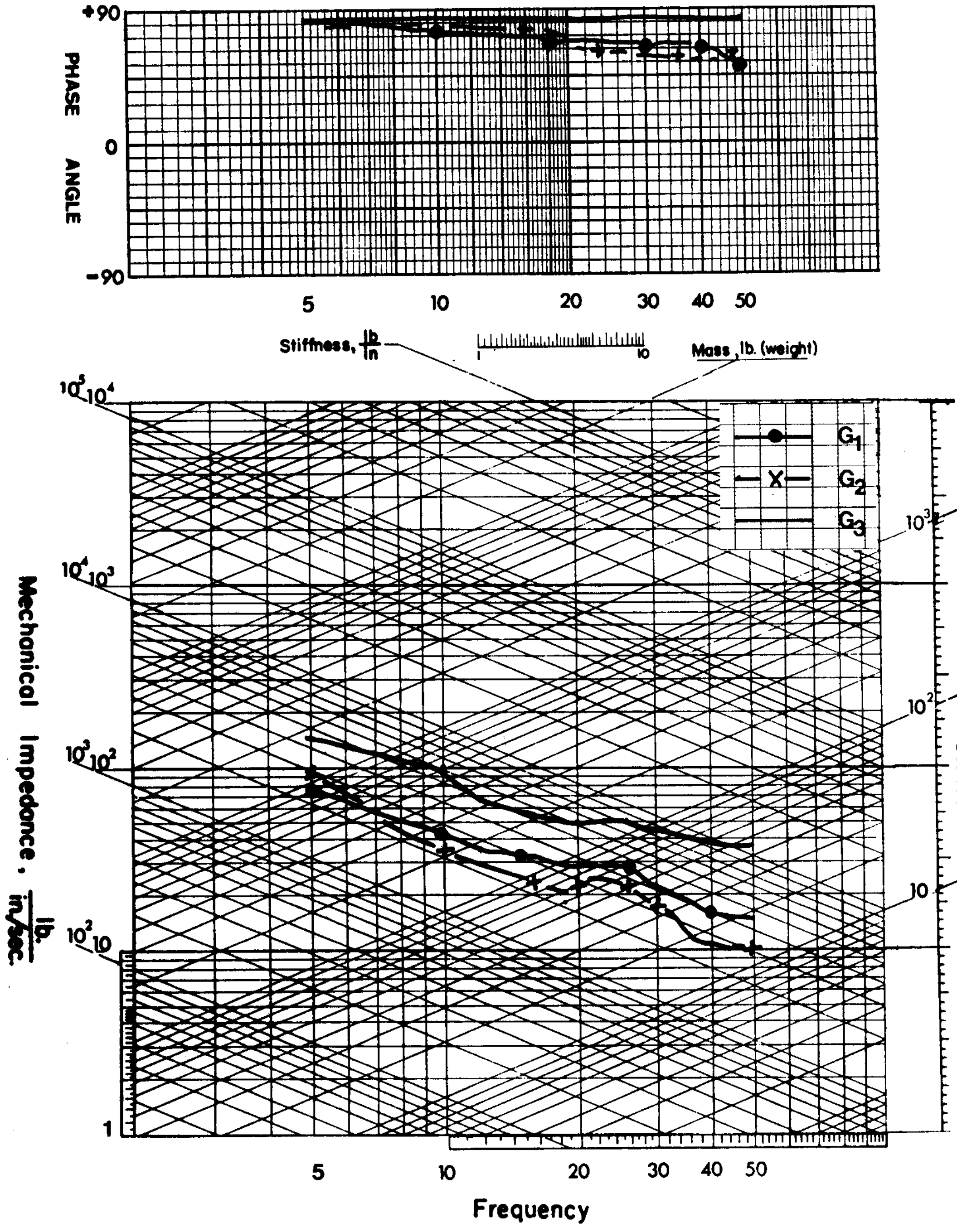


Figure 26 C. L_1—L_2—L_3 / G_1—G_2—G_3 / WO.

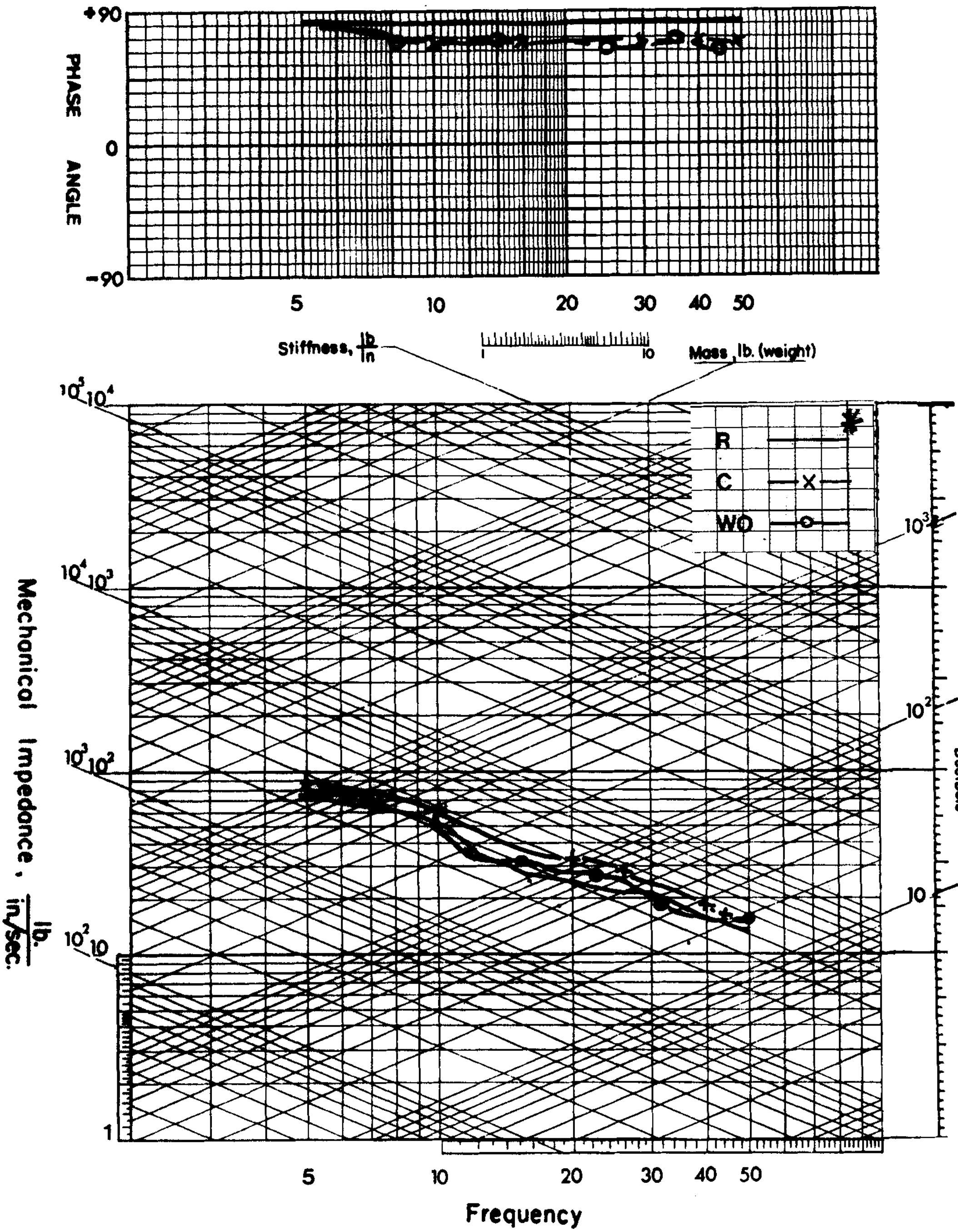


Figure 26 D. L1—L2—L3 / R—C—WO / D.

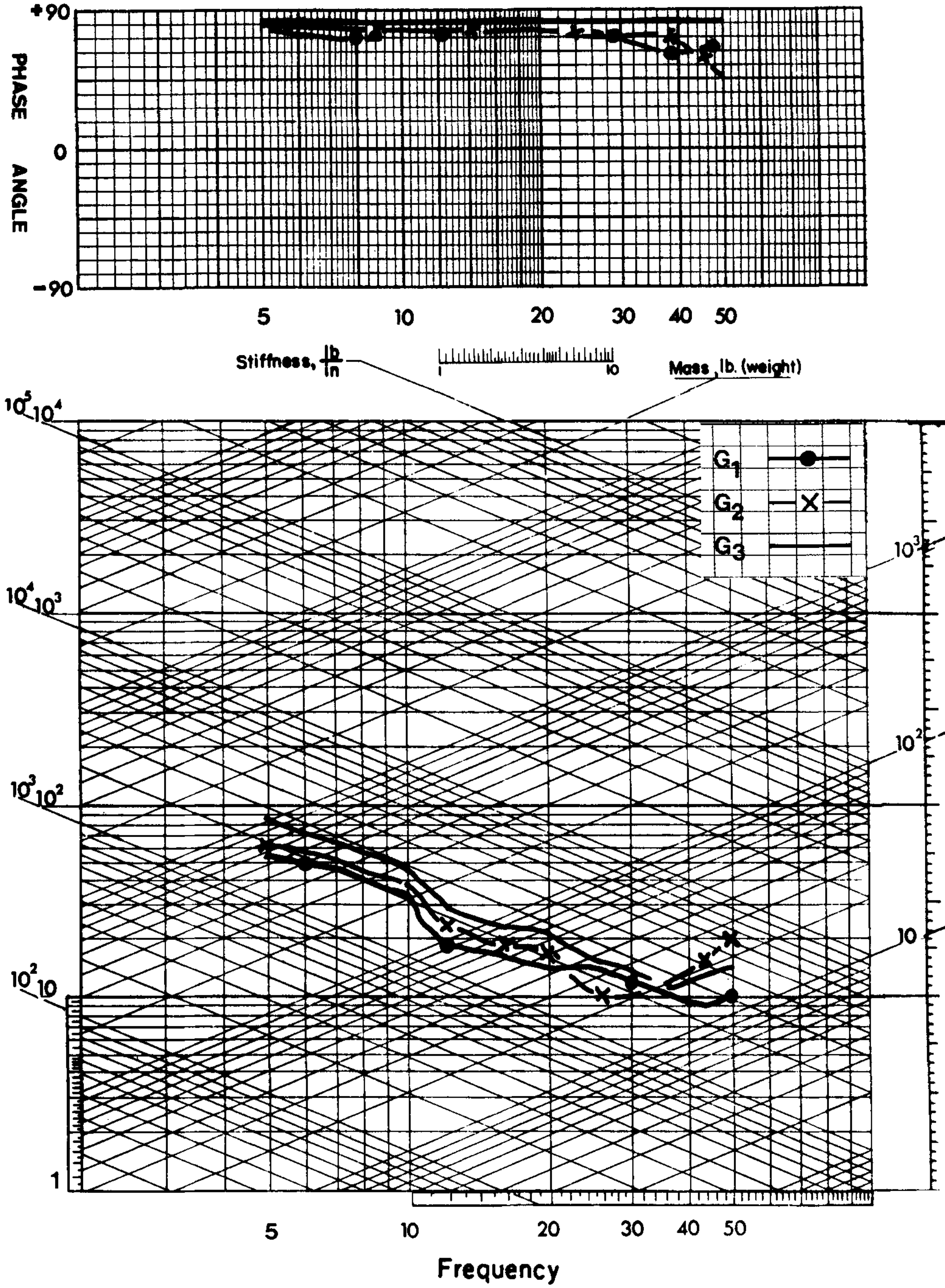


Figure 27 A. L_4—L_5—S / G_1—G_2—G_3 / R.

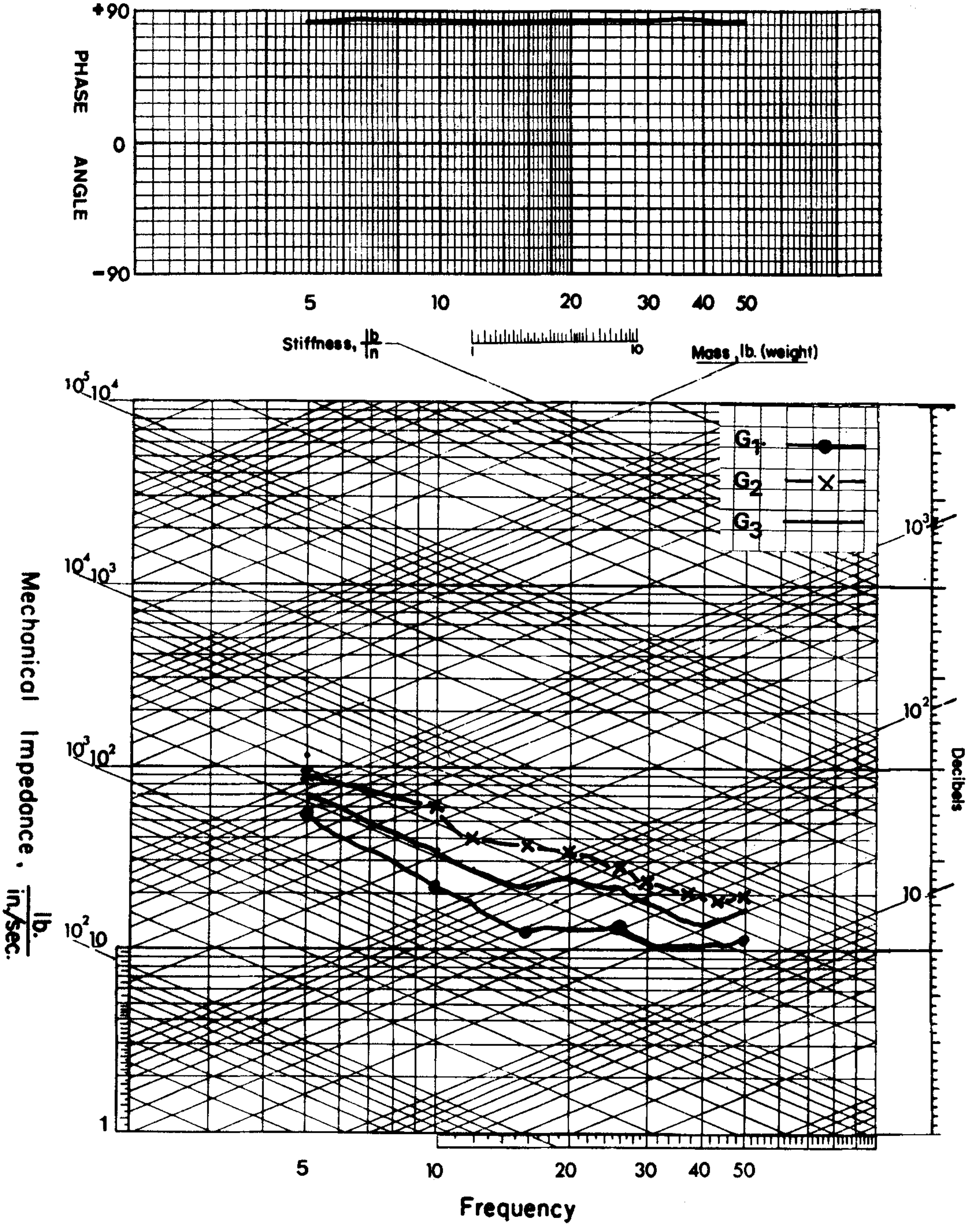


Figure 27 B. L_4—L_5—S / G_1—G_2—G_3 / C.

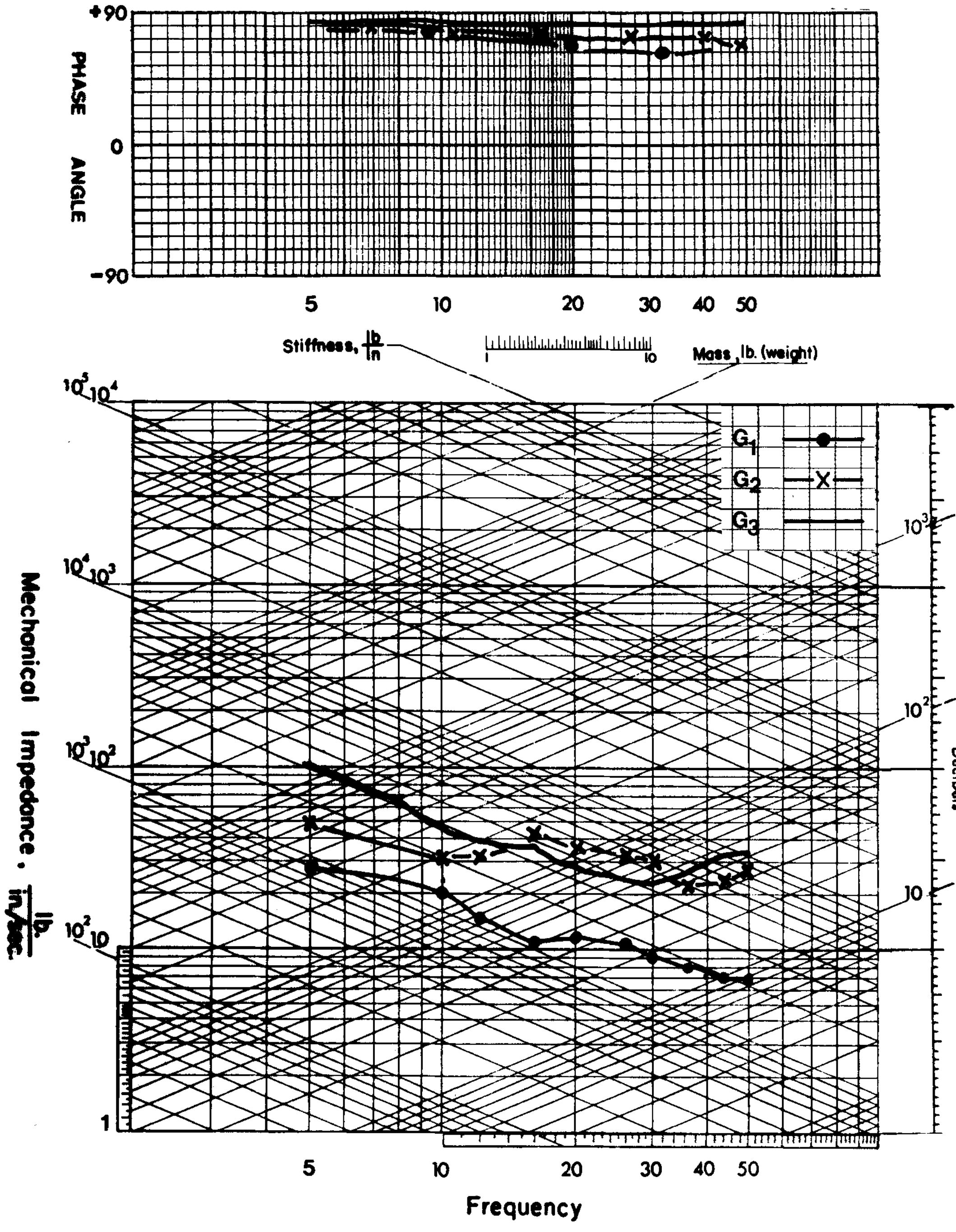


Figure 27 C. L4—L5—S / G1—G2—G3 / WO.

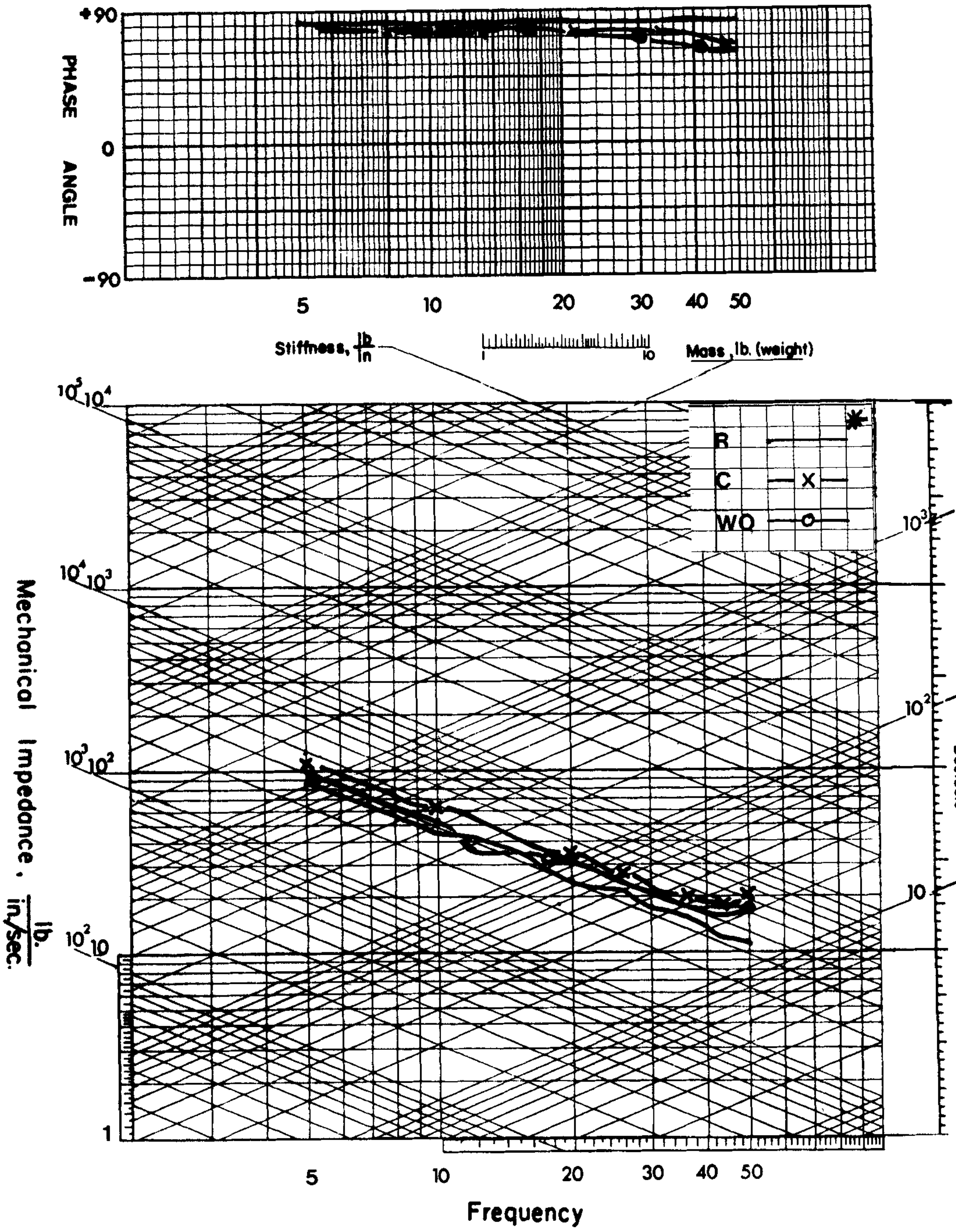


Figure 27 D. L_4—L_5—S_1 / R—C—WO / D.

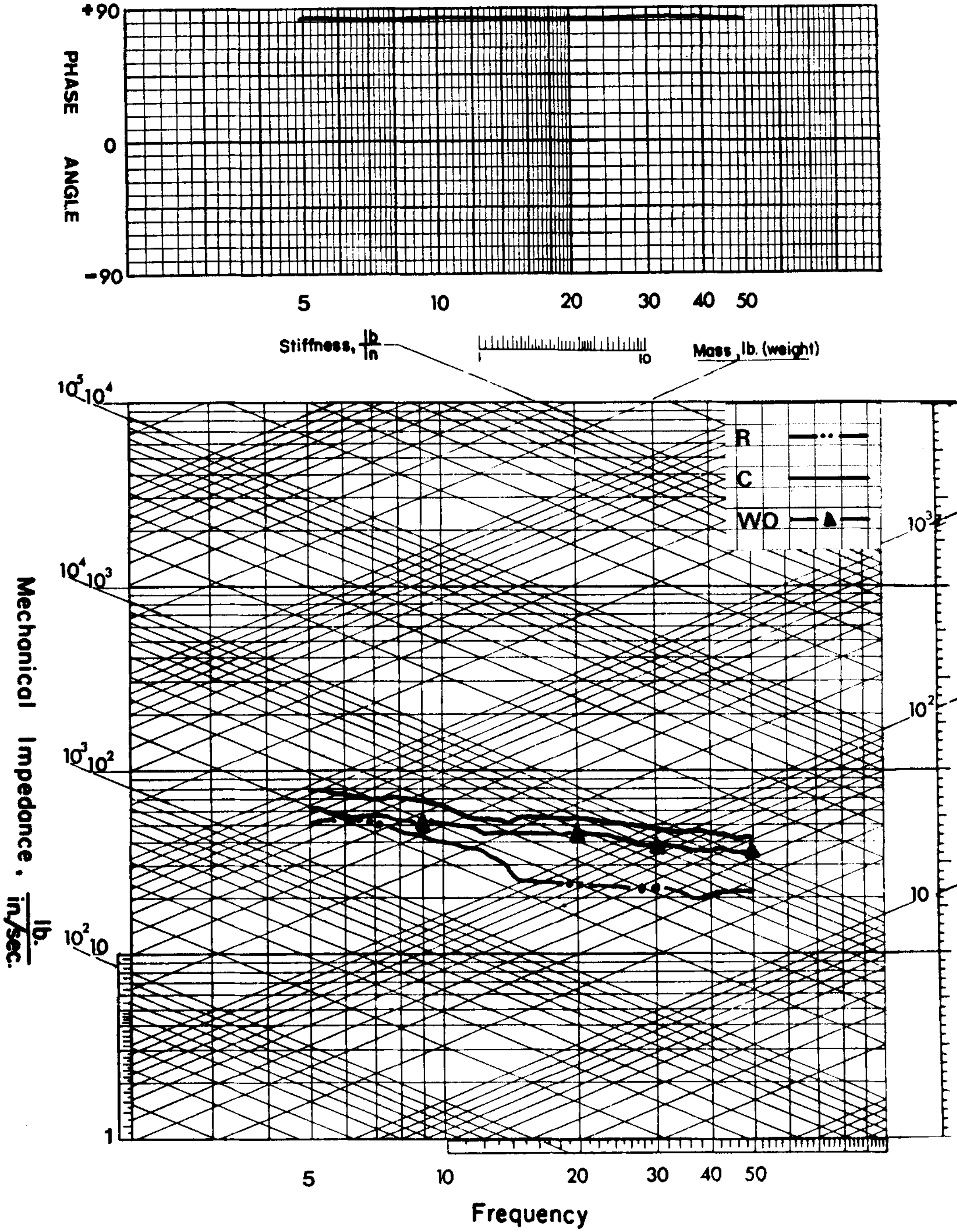


Figure 28. T_{11}—T_{12}—L_1 / R—C—WO.

By combing through the impedance waveforms it was found that the vertebral unit under test could "distort" (but in some cases improve) the waveform shape of the input signal, i.e. even if the force driving the system was purely sinusoidal in its wave shape, the response of the system could or could not be sinusoidal. This typical pattern of behavior was attributed to the nonlinearities within the vertebral unit, its multidegree freedom of action and the impedance tester. The degree of "distortion" was a function of the compression bias, and the frequency of excitation, and level of the test unit.

For instance, the stiffness range of vertebral unit T_1—T_6 (*Figure 23* A, B, C, D) with respect to case **R** lies between 2 and 3×10^3 (335—534 kp/cm). When this value is compared with case **C** the envelope of stiffness increases to an approximate magnitude between 2 and 8×10^3 (355—1426 kp/cm). When the posterior arches are removed from the conditioned unit, the stiffness values encompass a yet broader latitude which ranges from 2×10^3 (355 kp/cm) to 1.5×10^4 lb/in (2674 kp/cm).

The stiffness values for vertebral unit T_7—T_{12} for case **R** extend over a range of 2.6 to 7.4×10^3 lb/in (463—1319 kp/cm). The envelope shifts to a low value of 2.8×10^3 (499 kp/cm); then to a high value of 1.4×10^4 (2496 kp/cm) in the conditioned state, after which the range of stiffness changes to a value between 1.5×10^3 (267 kp/cm) and 7×10^3 (1248 kp/cm) for case **WO**.

The range of stiffness values for vertebral unit L_1—L_2—L_3 for case **R**, **C**, and **WO** were 1.6—9×10^3, 2—10×10^3 and 2.2×10^3—1.2×10^4 (285—1604 kp/cm, 355—1783 kp/cm, 392—1378 kp/cm) respectively.

L_4—L_5—S reveals a stiffness latitude of 1.6×10^3 to 6.2×10^3 lb/in (285 kp/cm to 1105 kp/cm) for case **R**, which remains in approximately the same scope for case **C**. When the posterior arches are removed, the stiffness domain changes to a low of 9×10^2 (160 kp/cm) and a high of 10×10^2 lb/in (178 kp/cm). A comparison of cases **R** and **C** reveals that with conditioning the stiffness values for the vertebral unit increase. Contrasting cases **WO** and **R** shows that the values of stiffness for the vertebral unit decrease.

If a vertebral unit exhibited a phase angle value other than +90 degrees before conditioning, following a period of conditioning the phase angle shift (if any) or resonance moved into the higher frequency range and out of the envelope of capability of this machine (see diagrams L_4—L_5—S and L_1—L_2—L_3).

The comparative stiffness values for the various test cases **R**, **C**, **WO**, deviated with respect to each other. The greatest divergence is seen to occur between cases **WO** and **R**. With respect to an analysis of the impedance mode shape in comparison to a detailed age analysis, the reader is referred to individual nomographs. It seems that the role of the posterior arches varies as a function of vertebral unit level. Undoubtedly the magnitude of the applied dead weight load, compression bias and structural configuration also interact as variables.

In all instances, case D showed the least divergence with respect to age.

A comparative examination of the experimental findings for the three age groups furnished some data to hypothesize that with increasing age the posterior arches undergo a transition in their function from being primarily a guiding element to a force-transmitting, force-attenuating, rotational, snubber-like element. The individual peculiarities are very much dependent upon:

1. The level and length of the vertebral unit.
2. Whether the vertebral unit is mechanically conditioned or not, if it is *not* conditioned the arches play little role in compressive vertebral mechanics.
3. The magnitude of the static compression bias and the exposure time in relation to load and dynamic loading of the vertebral unit.
4. Age of the specimen.

   Of these mentioned factors, the second, third and fourth play a dominant role.

In the cases where a softening response was observed with the posterior arches removed (WO), perhaps this action may be attributed to an additional override protection provided by the disks because the conditioning load proved ineffective to attain a conditioning level within a normal operational domain.

Before proceeding further, a few comments are in order concerning the forceful response of the L_4—L_5—S units, which hold true for the remainder of these experiments. The L_4—L_5—S vertebral complex proved to be a most difficult unit to insert, firmly restrain within the loading jaws and dynamically analyze within the scope of the test parameters. This was attributed to the fact that this vertebral unit is bow-shaped. To place two parallel axially aligned hard tissue surfaces between the loading jaws required up to two-thirds of any particular sacrum and even then the hard tissue bearing surfaces, although parallel to each other, often were not in uniaxial alignment to the applied load (because of the natural curvature of the sacrum with respect to the intervertebral disks between L_4—L_5 and L_5—S). Upon the application of the designated value of preload, the convexity of the bowing increased anteriorly. When this occurred the bearing surface contact area on the lower loading jaw was reduced, and the vertebral unit snapped out of place. It was discovered by trial and error that this reaction could be minimized by reducing the degree of parallellity between the hard tissue surface areas. This was done for necessary tests. One interesting observation was that the older the specimen, the greater the tendency to forcefully snap out of position from between the loading jaws.

The conditioning and subsequent shift in energy absorption and transmission characteristics may be better understood by considering a vertebral unit reaction like that of a series parallel array of springs (*e.g. Figure 29*). Each spring represents a means of storing energy. For any given value of constant loading there occurs some displacement of the top array with respect to the base. The displacement causes each spring to deflect to some degree. **With increasing deflection, the deformation increases and at each increment of deflection, the** impedance changes. Assuming that each spring dissipates a fraction of the energy it stores, the overall storing mechanism can probably dissipate a significant amount of energy and have a reasonably high value of impedance. However, this factor is continually shifting as the deflection changes. The important question thus becomes just how much conditioning is necessary in order to be in the normal envelope of operation for any particular disk and to attain "anatomically accurate" impedance values and mode shapes for any given vertebral unit? This so-called "range of normal operation" for any disk ostensibly varies widely with reference to the impedance test parameters and the peak to peak force output of this tester. If the force output of the machine were greater, it would seem very likely that the impedance values would deviate even more.

As discussed earlier, there exists a relationship between the impedance curve shape and stiffness. Hetenyi (1950) defined stiffness as the inherent capacity to

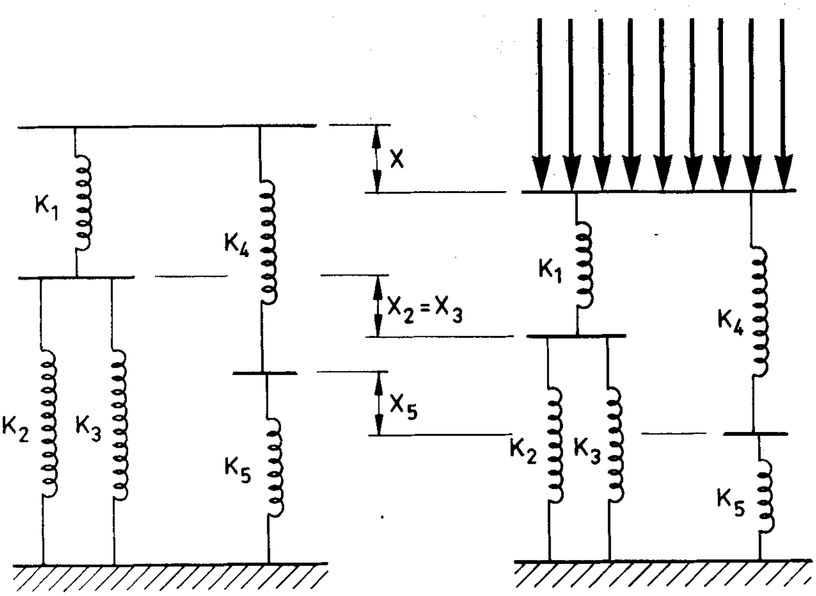


Figure 29. The damping action of a disk may be better understood by considering a series-parallel array of springs, representing the energy storage mechanisms of the vertebral unit in the undeflected state (left)—as removed from the cadaver, and of the deflected (right) and/or conditioned state. The characteristic domain of any particular disk's energy transmission, and attenuation peculiarities are clearly a function of its deflection within a defined geometry. However, deflection is dependent upon the vertebral level, specimen age, and preload. (Deflections $X_4 = X - X_5$, $X_1 = X - K_2$ are not indicated).

resist elastic displacement under load. Assuming that the values of stiffness are related to the load-carrying capacity of the vertebral unit, a difference is noted (its magnitude varies with regard to age and vertebral level) between cases **R**, **C**, and **WO** and when **WO** is contrasted to case **R**. Therefore, it was decided that the posterior arches do play a role in load distribution and affect the energy transmission characteristics of the conditioned vertebral unit, and they must remain on the unit to fulfil the intentions of these studies.

The results of test 1, 4 and 7, required additional examination. These mode shapes were unique in that unexpected vacillations were observed in the Z mode shape and its magnitude. In test 7, it emerged that the conditioned T_{11}—T_{12}—L_1 unit possessed a comparatively high impedance magnitude but that T_9—T_{10}—T_{11}—T_{12} (Test 4) was comparatively low. With regard to this experimental discrepancy, Wilson and Cochrane (1928) asserted that spinal motion was most marked at the joints of the thoracic-lumbar portion of the spine. In 1955, 1958 Davis postulated that the range of motion at the level of the thoracic-lumbar mortice joint was such that vertical loads of sufficient magnitude could "lock the mortice". It was felt that this finding could account for the high impedance values for the T_{11}—T_{12}—L_1 unit, but could not throw light upon why the impedance values descended with regard to the T_9—T_{10}—T_{11}—T_{12} unit. In an attempt to resolve this disparity, five T_7—T_8—T_9—T_{10}—T_{11}—T_{12}—L_1—L_2 vertebral units were water-macerated, after which the wet vertebral units with disks intact were grossly examined. A structural discontinuity was located to exist amidst the superior articular facet planes of T_{11} as opposed to its inferior articular facet planes (in two cases it was well defined, in two cases not so obvious and in one case its existence was questionable).

These morphological variations are illustrated in *Figure 30*. This finding signified that there probably occur two abrupt discontinuities within the human thoracic-lumbar column: 1. the abrupt changeover of the thoracic-lumbar mortice, and 2. the facet joint planes at the level of T_{10}—T_{11}, and their dynamic interactions contrasted to the adjoining stable mortice and relatively flexible thoracic unit should be further searched into. (To the best of this writer's knowledge morphologic differences are not described in clinical tests or previous studies of this nature.)

Upon considering the softening and hardening action of the vertebral unit and the abrupt morphological gradients between the articular facet joint planes at the vertebral body T_{11} and the mortice, it was postulated that perhaps with conditioning, rotation brought about by vertebral unit twisting could come about in the upper thoracic column (above the level of the superior facet planes of T_{11}) with regard to the stable mortice joint. It was considered that this articular facet joint interface level ought to react in some cases as an anterior and lateral swivel-like union (but restricted posteriorly and in extension) with respect to the adjoining vertebral bodies.

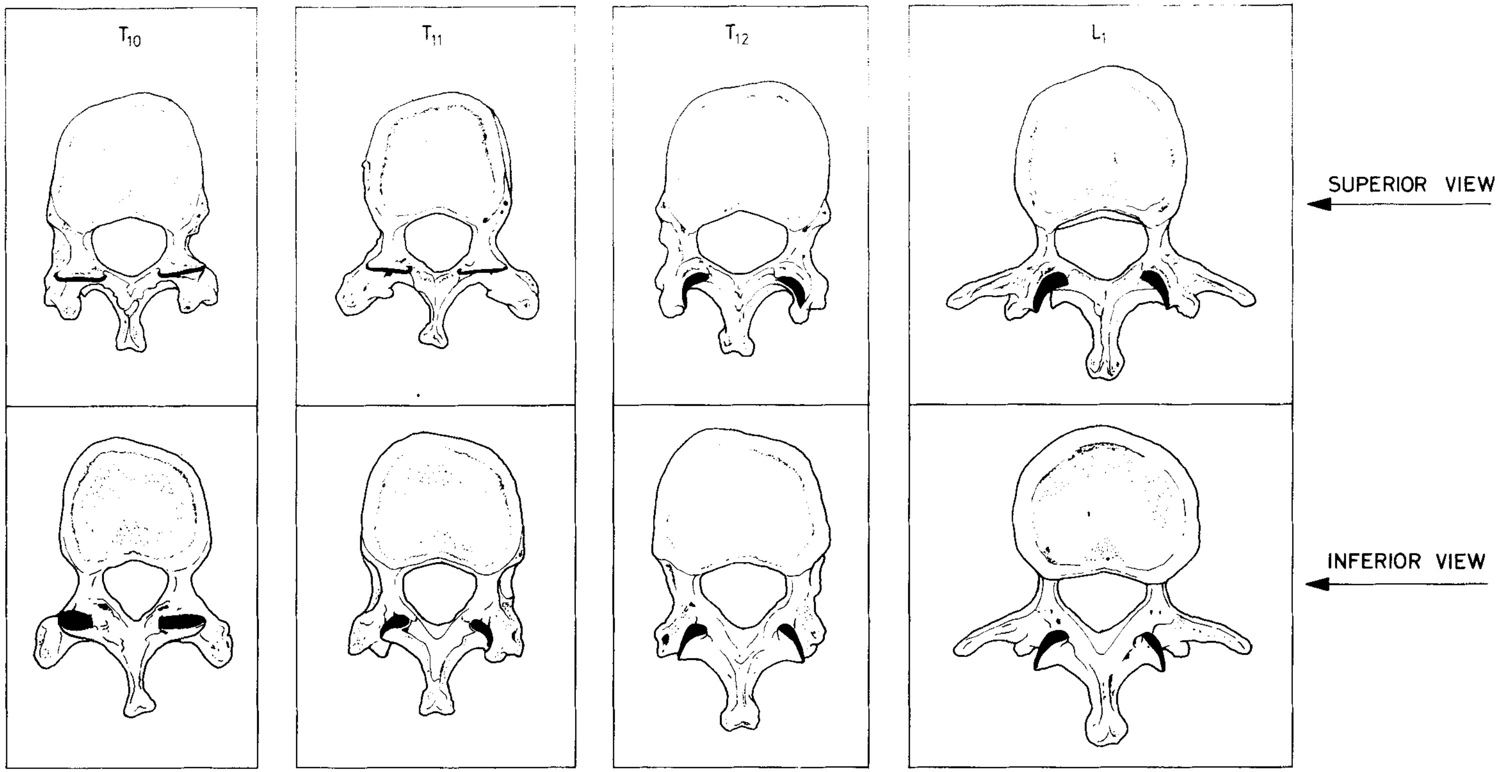


Figure 30. The comparative morphology of the superior and inferior articular joint surfaces at vertebral level T_{10}—T_{11}—T_{12}—L_1.

If such a twisting reaction were to take place at this vertebral level, it was stipulated it must transpire at the level of the intervertebral disk. Its magnitude of response must be governed by several factors, amongst them the ligamentous apparatus, the posterior joints, and even more important, the resilient nature of the intervertebral disks.

A water maceration process was employed for the cervical-thoracic transregional joint. A comparative observation of these articular facet planes revealed that normally the relative structural differences in orientation are quite subtle. However, there is also an abrupt changeover in the directional planes of the facet joints. *Figure 31* illustrates the critical transregional vertebrae at the cervical-thoracic junction. This abrupt structural gradient is glossed over in the clinical literature.

These critical morphological factors are matters for further consideration with regard to the clinical vulnerability of the cervical and thoracic-lumbar spine to injury. Jefferson (1927), Nicoli (1949), Kazarian et al. (1971). Perhaps the discussion of synovial joints under weight bearing could be considered as a functional adaptation to forces. Thereby the pattern and manner of creep may be a prerequisite as an underlying mechanism and translated to affect the observed patterns of spinal injury. At present there exists no generally accepted view of the biomechanical factors which underlie this high frequency phenomenon.

In summary, the primary effects of critically conditioning a vertebral unit are:

1. There comes about a stress redistribution process which results in a new state of stress. This action is assumed to be a function of the past history of strain (or stress).
2. The redistribution process produces a new geometric shape for the vertebral unit.
3. There results a more efficient propagation of energy, and
4. A shift in resonances (if any were originally encountered).
5. Morphological and geometric factors also play a definite role with regard to the impedance.

EXPERIMENT C — THE IMPEDANCE RESPONSES OF THE UNCONDITIONED INTERVERTEBRAL DISK

Objectives—Cadaver Material

Of the previously mentioned elements, affecting the transient action of the vertebral unit, it was felt that the intervertebral disk should primarily reflect the passive resilient nature of the vertebral unit. Four normal vertebral columns, numbers 54, 58, 10 and 3, were prepared to test the relative impedance mode shapes, but principally the comparative lumped impedance values of the intervertebral disks.

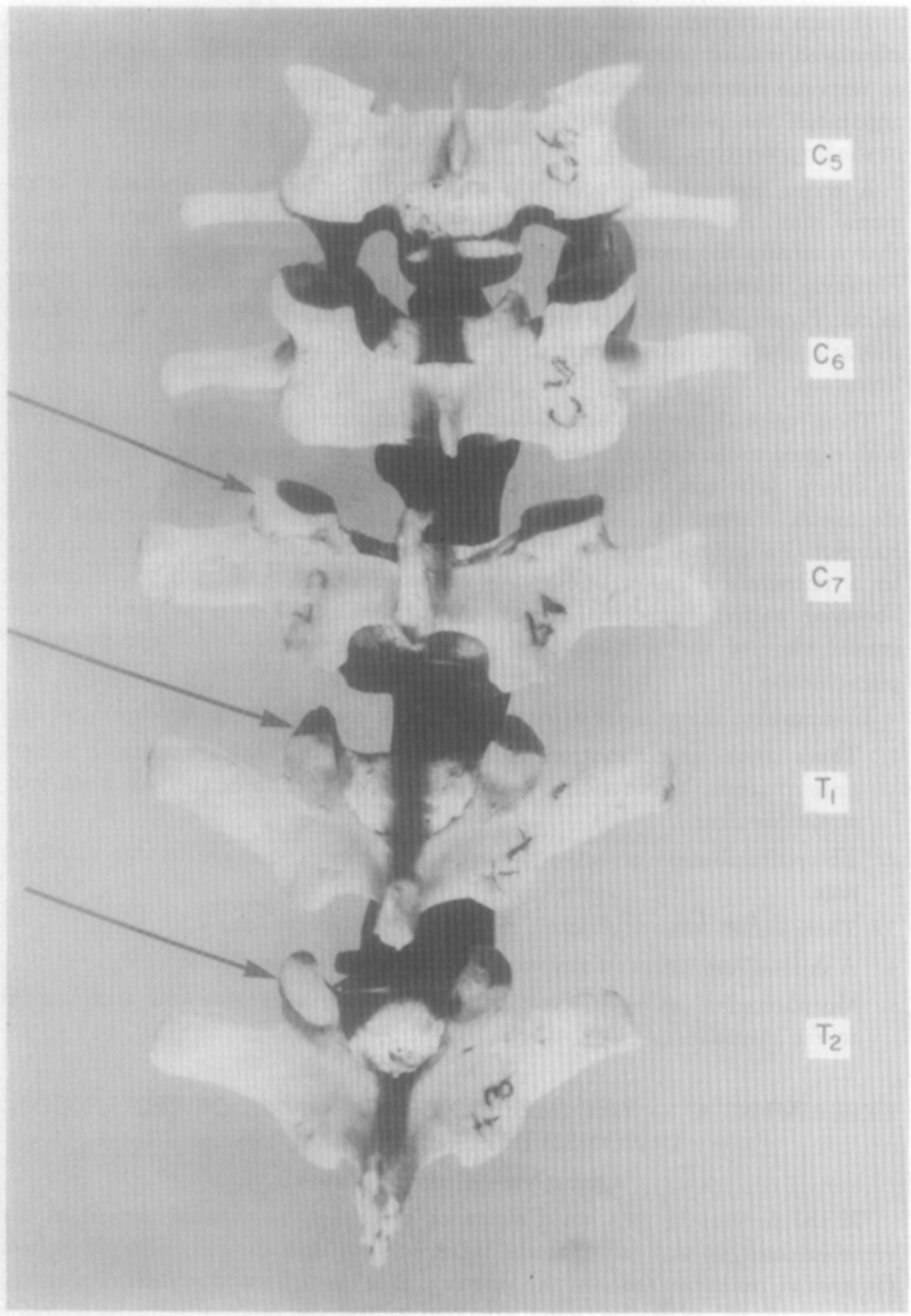


Figure 31. Illustrates the comparative morphology of the cervical—thoracic regions. Note the abrupt changeover between levels C_7—T_1—T_2.

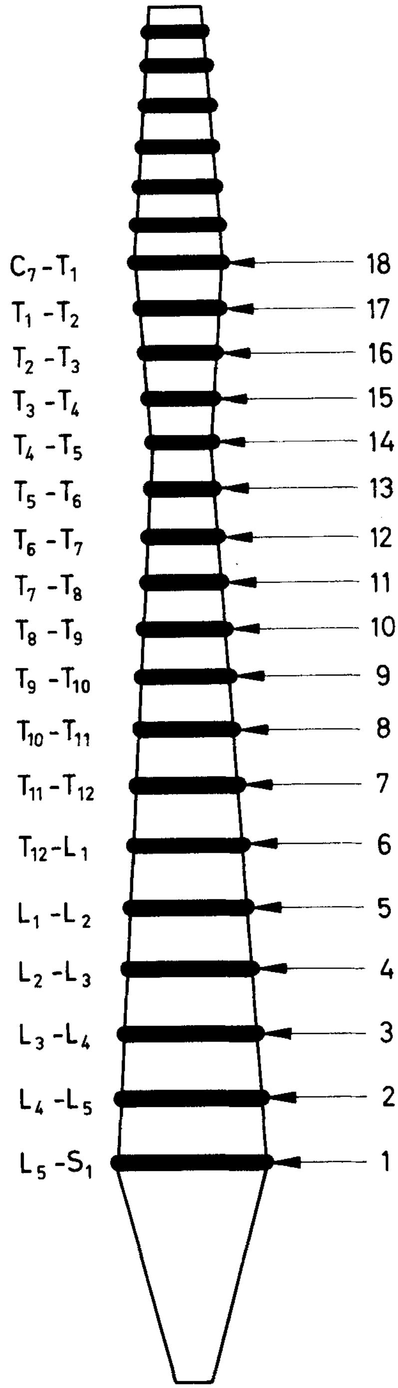


Figure 32. Intervertebral disk numbering system.

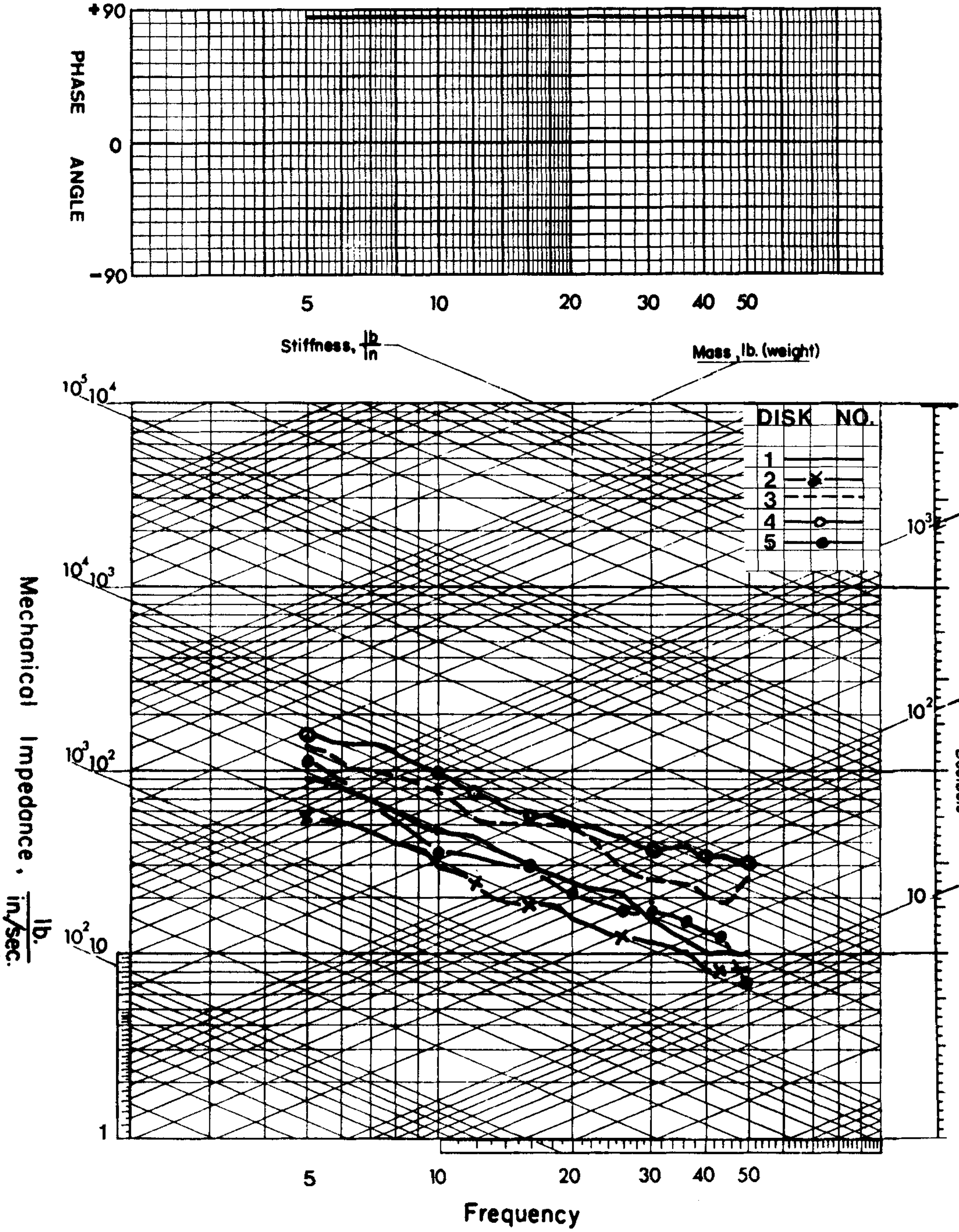


Figure 33. Impedance curves for (unconditioned) intervertebral disks 1 through 5.

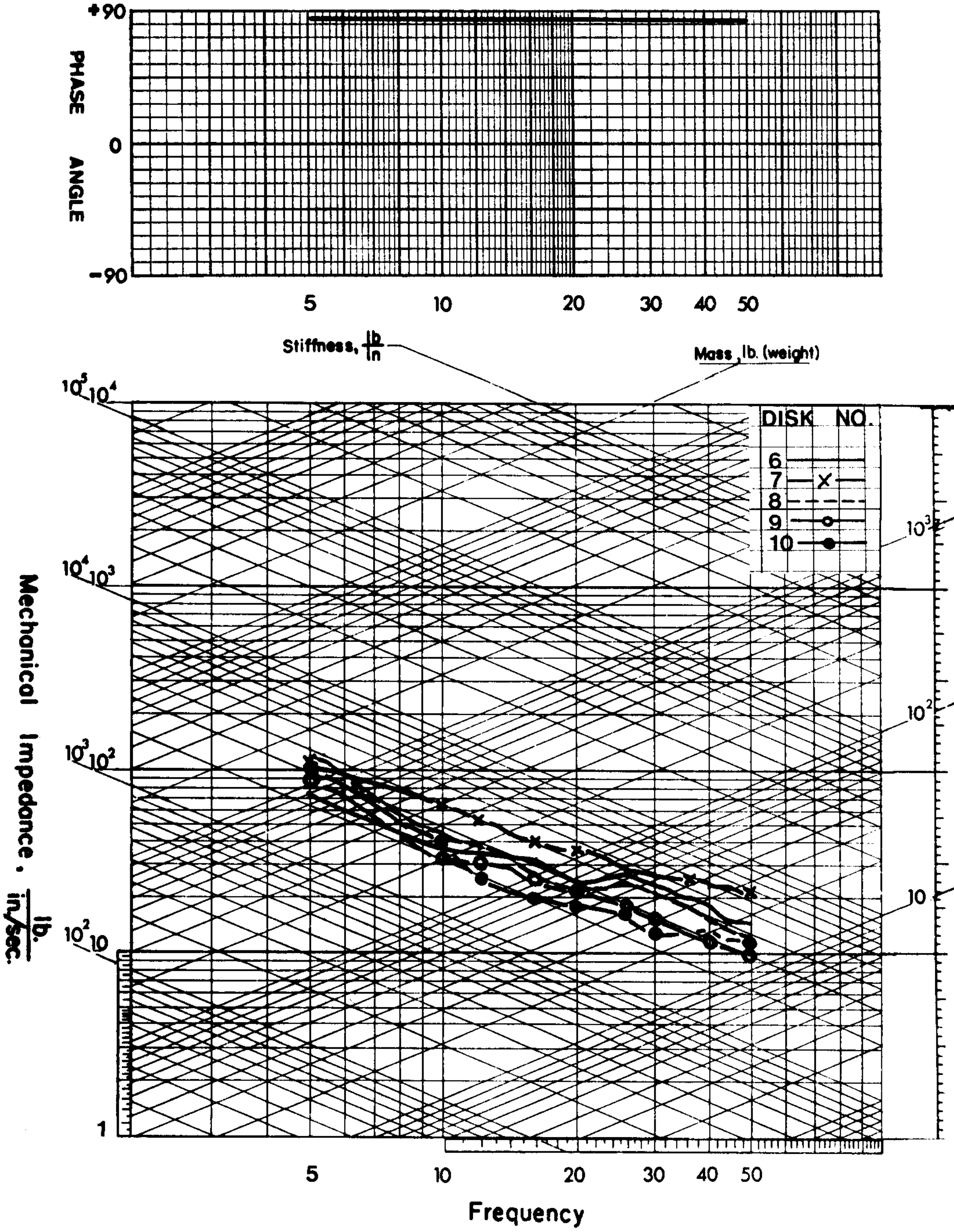


Figure 34. Impedance curves for (unconditioned) intervertebral disks 6 through 10.

Methods

Each vertebral column was stripped of its posterior aspects by separation through the neural arches, using an electric band saw. Each intervertebral disk was excised away from the vertebral column together with a thin slice of bone on either side, then placed between the loading jaws of the calibrated impedance tester. Each disk was preloaded to a value of 21.7 kgs (48 lbs.). The thickness of the vertebral bone varied according to vertebral level. *Figure 32* identifies the intervertebral disk numbering system.

Results and Discussion

When the intervertebral disk was placed under compressive load and vibrations were impressed by the mechanical oscillator, the intervertebral disk was observed to undergo a deformation. It bulged out of the sides of the bony ends and a complicated non-uniform stress field was probably set up within the disk. Undoubtedly, shear stresses resulted from the bulges and this mechanism allowed for higher loads to be carried by the structure.

Figures 33 and 34 disclose the composite impedance mode shapes for disks 1 through 9. The impedance mode shapes for the individual intervertebral disks

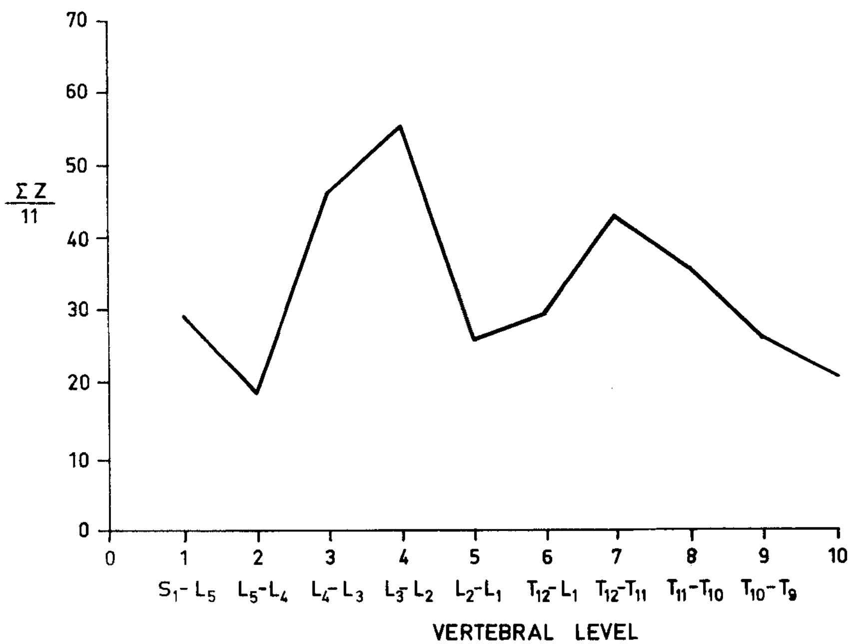


Figure 35. Comparative impedance values for the intervertebral disks.

remained somewhat parallel to each other up to the level of intervertebral disk 6. At this level, the resistance of the joint to mechanical loading comparatively begins to differ.

In Figure 35 the averages of the impedances are calculated for the primary frequencies. A comparison of impedance values for disks 4 through 10 reveals that a decrease in Z values takes place between disks 4 and 5, whereupon a gradual increase in the Z values occurs between disks 5 and 7; this is followed by a steady decline in the impedance as far as disk level 10.

These relatively low impedance values suggested that the intervertebral joints at these levels (between 7 and 10) are easier to excite, move, have flexibility, are capable of large motions and of storing large amounts of energy. Intervertebral joint 2 also has a comparatively low value of impedance.

Remark:

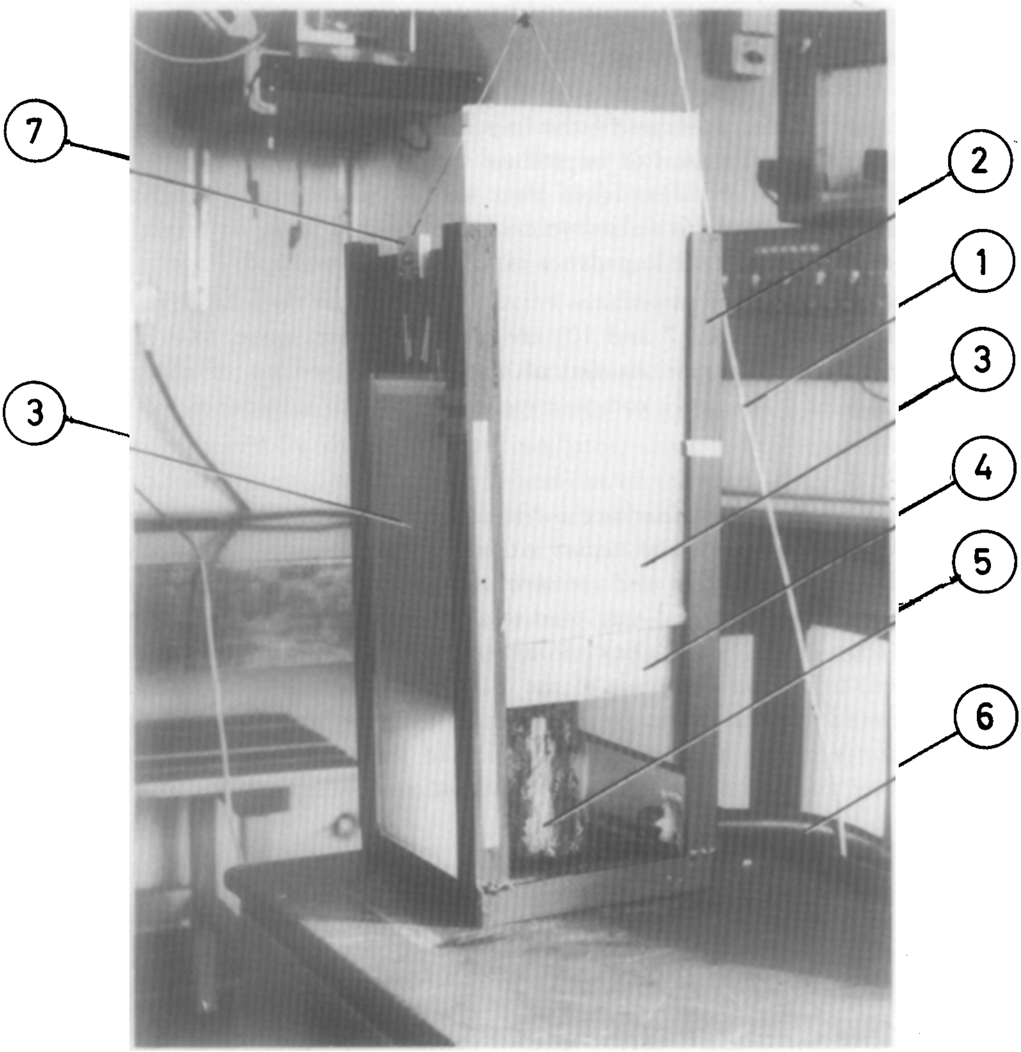
It must be recognized that as the frequency with which the energy is applied changes, there probably also occurs a change of both the magnitude and phase of motion. These reactive and resistive components of impedance are complex functions of frequency and may tend to alternate in their respective dominance as a function of the system's behavioral response. Tests were also performed (unpublished) to delineate the role of the disk with and without its nucleus intact. It was found that the mechanism which leads to energy dissipation was probably the viscosity of the nucleus and relaxation of the annular walls—two factors which may also operate to bring about vibrational energy losses.

EXPERIMENT D — CREEP RESPONSES AT VARIOUS VERTEBRAL LEVELS

Methods

The Creep Chamber in Experiment 1 was modified and identified as CCC-1. An external rectangular angle-welded steel frame was fabricated to fit snugly around and enclose the plexiglas chamber. The screws which held the plexiglas chamber together were removed. This allowed the three plexiglas panels to slide up and down; the fourth panel, which had the humidifier hose attached to it, was cut parallel to its base and approximately five centimeters from it. This smaller section was cemented in place. The upper panel was free to slide perpendicularly. *Figure 36* shows the modified Creep Chamber.

When the weight box was placed within the rectangular steel frame and its dead weight came to rest on the axially aligned specimen, any or all of the four plexiglas panels could be raised at any one time for purposes of photographing the specimen. The electric humidifier was turned on and remained on for the entire test duration.



KEY

1. LIFT LINE
2. STEEL FRAME
3. SLIDING PLEXIGLAS PANELS
4. WEIGHT BOX
5. VERTEBRAL SPECIMEN
6. HUMIDIFIER HOSE
7. REMOVABLE CABLE ANCHORS

Figure 36. CCC—1.

Cadaver Material and its Preparation

Six transregional vertebral units (T_7—L_5), specimen numbers 70, 72, 24,73, 83, 84, were used. They were used to resolve the manner, relative magnitude and partitioning of creep deflection and deformation over the length of the unit. Each transregional vertebral unit was prepared according to the following procedure. Sharpened steel pins, approximately 1 mm in diameter and 10 cm in length, were lightly hammered into each vertebral body centrum, spinous process, and as near the transverse process as possible.

The vertebral unit with its pins in position was situated onto an X, Y, Z grid and photographed in the Anterior, Posterior, Left Lateral, Right Lateral, Caudal and Cephalad Planes. The unit was radiographed, after which it was placed into the apparatus and positioned onto the base plate of the CCC-1. The weight box was perpendicularly positioned, introduced into the apparatus, then lowered until its dead weight rested on the axially aligned unit. The relative positions of the steel pins were photographically monitored in four planes every 30, 120, 180, and 240 minutes. Following the full time period, the static load was hoisted off the specimen. Next the specimen was promptly photographed in six planes and radiographed.

Results and Discussion

The deflections of the various pins and their respective vertebrae offered a number of qualitative observations. The elements of the conditioned vertebral unit had a different geometric configuration than those in the defrosted, unweighted, relaxed state. A comparison of the points pre and post-conditioning revealed that there ensued relative motions amongst the pins. To define the geometry of the distortion, the displacement vector is being investigated, and the findings are to be reported in a subsequent publication. However, a number of findings relevant to this effort will be presented. The amount and direction of distortion was unique for each spinal section. Each vertebral unit arrived at its ultimate configuration in a distinctive manner. It was photographically and radiographically perceptible that following uniaxial loading the superior aspects of the superior facet surfaces would proximate, and impinge upon the pedicles of the superior vertebral body in an asymmetrical fashion. The asymmetry was unique for each vertebral unit and level. In no instance was there observed to be symmetrical loading. The long term reaction of the constant compressive axial load on the vertebral unit in addition induced intrinsic clockwise or counterclockwise torsional or spiraling and bending motions. The direction and magnitude of rotation varied as a function of vertebral level. Relative spiraling patterns were present at all vertebral levels when the position of the pins was contrasted to an absolute reference point fixed on the grid. Manually imposing a temporary torsional load on adjacent vertebral bodies to twist the intervening intervertebral disks in a direction opposite to that of their rotation, with the anticipation that the vertebral unit would rotate and set in this new direction of spiraling, proved unsuccessful. Instead, the vertebral unit returned to its equilibrium position dictated by its initial direction or pathway of spiraling. Repeated creep tests were carried out using the same specimens (which were allowed to relax in a humidity chamber) and in each case the pattern of rotation, as well as it magnitude remained unique to the specimen. This provided additional evidence to support the concept of memory effects.

Pre and post-conditioning, interpedicular widths (IW's) were measured and compared from the radiographs. The post-conditioning radiographs showed that the average bilateral IW's of T_7—T_8 were approximately equal (19 mm). The value increased to 23 mm at the level of T_9, then decreased to 20 mm at the level of T_{10}, but was found to increase again to a value of 24 mm, continuing to do so down to the level of L_5. The rotational component of the creep at the T_7—T_8—T_9 level was greater than the axial compressive component. The intervertebral disk (number 7) at the thoracic-lumbar junction showed the least motion, and the joint above this level showed an increased tendency for rotational motions.

A Torsional Warping Mode of the Human Spinal Column

Based on these observations, and the data reported in the medical, anatomical and anthropological literature, a provisional hypothesis is put forth for a mode of action of the human spinal column.

Up to the age of about five years, the human vertebral column is a rather symmetric structure. From this age onwards, it begins to manifest morphological variants. These modulations are oriented directly toward conditioning the spinal column to interact with environmental forces in accordance with the manner and character of locomotor habits, or to modifications in these habitual patterns. The vertebral column gradually acquires a time-dependent characteristic posture, which is not fully established for some years. With aging and the incessant domination of gravitational and environmental forces, the individual elements of the column proceed to deflect and remodel in an ingrained manner assumed to be in accordance with the individual's established habit. On account of its noncircular cross-section, axial asymmetry, and anisotropic development and behavior, the spinal units undergo distortions in their cross-sectional areas when subjected to longitudinal loads. These time-dependent distortions may be localized or extend throughout the length of the column. Stress gradients develop within the individual hard tissue elements and at various points of discontinuity and irregularity, but are slowly dispersed as bony remodelling concurrently takes place. The oscillating external forces produce a monotonous, meticulously regulated deflection at the various disk levels, but are continually resisted by the vertebral ligaments and rationally resisted by the musculature and posterior joints. The varying deflections induce force modulations within some "normal envelope of operation", which on account of ligamentous and disk reaction decreases with age in an attempt to form an inherently stable structure. During a period of rest the individual disks, ligaments and musculature unwind and the vertebral bodies relax. When a complex but primarily longitudinal load is again placed on the column, the force is gradually transmitted to each intervertebral disk which then proceeds to deform in its characteristic manner. The vertebral centra gradually rotate and transverse shear and warping of the cross-sectional intervertebral disk area occurs, and the dynamic process is repeated again, again and again. The spine becomes culturally conditioned.

Gradually adjacent vertebrae throughout the length of the spinal column join in forming interlocking and stable motor units. The spinal column distorts laterally, for greater stability, and seems to alternate its twisting tendencies until the structure ist rotationally and flexurally stressed at certain points and counterstressed (at other points) over its entire length following "full-true" conditioning. With increasing age, habitual position, intervertebral disk thinning and bottoming, contiguous articular facet joint surfaces interlock, either symmetrically or asymmetrically, to assist in forming stable vertebral units. With time and habit, the spiral turning becomes fixed, a particular path of "warping" is established. The normal adaptive torsional warping about its longitudinal axis provides increased stability to the vertebral unit and the spinal column.

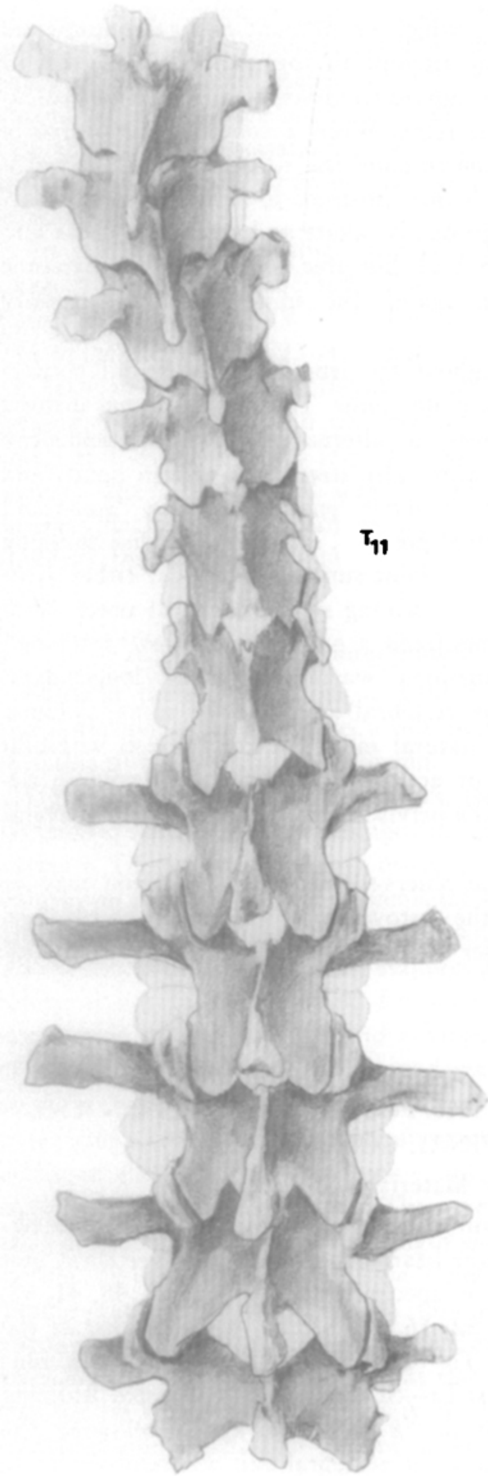
Figures 37, 38, and 39 illustrate 3 natural modes of conditioned vertebral column warping. A postulated mode of action for the vertebral column deformation is illustrated in *Figure 40*. The pelvis is included and also has several rotational degrees of freedom.

Of course, for the living subject the observed mechanical rotation may or may not occur, and structures such as the costovertebral joints and the rib cage, as well as sternum interference, shoulder and pelvis interactions may minimize or accentuate the observed distortions.

EXPERIMENT E — EFFECT OF THE POSTERIOR ARCH ON THE ANTERIOR SPINAL COLUMN

Objectives—Cadaver Material and Methods

To ascertain the contributory role of the posterior vertebral arch, its ligaments and joints on the anterior weight bearing portion of the vertebral unit in the unconditioned (raw) state, cadaver specimen numbers 31, 50, 48, 41, 47, 53, 29, 43, 33, 40, 56, 42, 9, 14, 64, 52, 16, 27, 82, 80, were divided at the intervertebral disks between T_6—T_7; T_{12}—L_1; L_3—L_4. This resulted in four vertebral units $T_{1—6}$; T_7—T_{12}; L_1—L_3; L_4—S_1. Each unit was tested utilizing the impedance machine at a preload of 48 lbs (21.7 kgs). Following the impedance test, the bias was taken off and the vertebral unit was removed from



Figures 37, 38, and 39 reveal three different patterns of vertebral body deflection caused by axially aligning a constant compressive load on vertebral unit for not less than 480 minutes. The illustrations are based on time lapse photographs and radiographs. When a vertebral unit is subjected to compressive loading it is assumed that uniform forces exist at any given cross-sectional area and the projection of a straight line on a cross-sectional plane remains straight after twisting. When a prepared spinal column is examined its hard tissue surfaces are distorted and warped so that a given cross-sectional area no longer lies in the original straight plane (if it ever did at all). Therefore, when a load is applied to the structure it deforms and sets in an idiosyncratic pattern. In each case of the three cases, the disks displayed a dominoe effect (so that the disks nearer the ends of the vertebral unit were deflected first. After a period of time (not less than 240 minutes) all the disks in the column had undergone various degrees of deflection.

At each vertebral level unique adaptive traits were observed. The vertebral unit T_9—T_{11} exhibited a rotational mode of response, the mortice joint locked, the L_1—L_3 unit formed a stable column, while the L_4S unit exhibited shear tendencies.

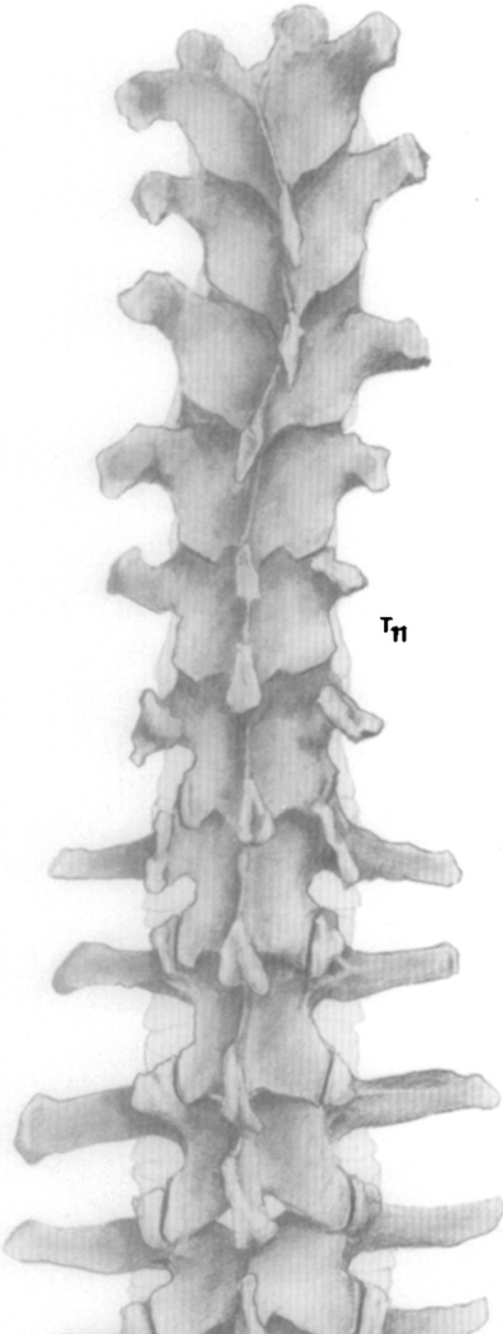


Figure 38. See Caption in Figure 37.

Figure 39. See Caption in Figure 37.

Figure 40. An oblique illustration of a prepared human spinal column and pelvis. The relative positions of the posterior joints are emphasized. The counterclockwise spiraling of the high thoracic vertebral unit when contrasted to the lumbar column is apparent.

the loading head, its posterior segment and interarcuate ligaments were separated from their laminae and arches through one side, then through the other side of the pedicles using the electric band saw. The machine was calibrated, the columnar vertebral centra, with the anterior and posterior longitudinal ligaments intact were placed back into the loading jaws, and the same magnitude of preload applied and the unit was recycled in the same frequency range, and at approximately the same amplitude of excursion. After the test was completed the bias was removed and the anterior column was taken away from the test machine. *Figures 41, 42, 43,* and *44* are the composite impedance and phase modes for the vertebral units. In general, when the posterior arches were removed, the impedance magnitude showed a slight shift downward. If no weight was carried by the posterior aspects the Z curve would shift upwards rather than downwards (assuming the deflection remained constant).

Results and Discussion

The whole anterior and posterior column possesses an intrinsic equilibrium wherein the nucleus pulposus and the annulus fibrosus exert a tensile force and attempt to keep the vertebral bodies apart, while the common ligamentous structures, the vertebral arches and their processes are continually exerting a force in the opposite direction to keep the structure together. When the posterior vertebral arches are removed from the vertebral unit, the intrinsic tripodal-like state of equilibrium of the whole spinal unit is *reduced*, (especially in the lower lumbar region). The vertebral centra straighten out and assume another position of equilibrium. Conversely, the tensile loads on the vertebral arches caused the structures to contract. These observations confirmed the earlier reported findings of Myer (1873), Petter (1933), Åkerblom (1948) and Hagelstam (1949). This typical response pattern implied that the anterior and posterior segments of the human cadaver vertebral unit inherently possess different degrees of mechanical impedance. When an unconditioned vertebral unit was cycled through the impedance tester, then physically removed from the jaws of the machine, its anterior and posterior columns separated, and the anterior column recycled at the same compression bias in the impedance machine, the impedance values of the vertebral unit were generally found to decrease and its creep characteristics to increase. In view of these results it again appeared desirable to conduct this experiment with the posterior arches intact.

GENERAL OBSERVATIONS, DISCUSSIONS AND CONCLUSIONS RESULTING FROM EXPERIMENTS

Creep and the Prestressed Vertebral Unit

Based on these studies, a number of general conclusions may be drawn relevant to the mechanical behavioral response of the excised vertebral unit subjected to constant compressive loading.

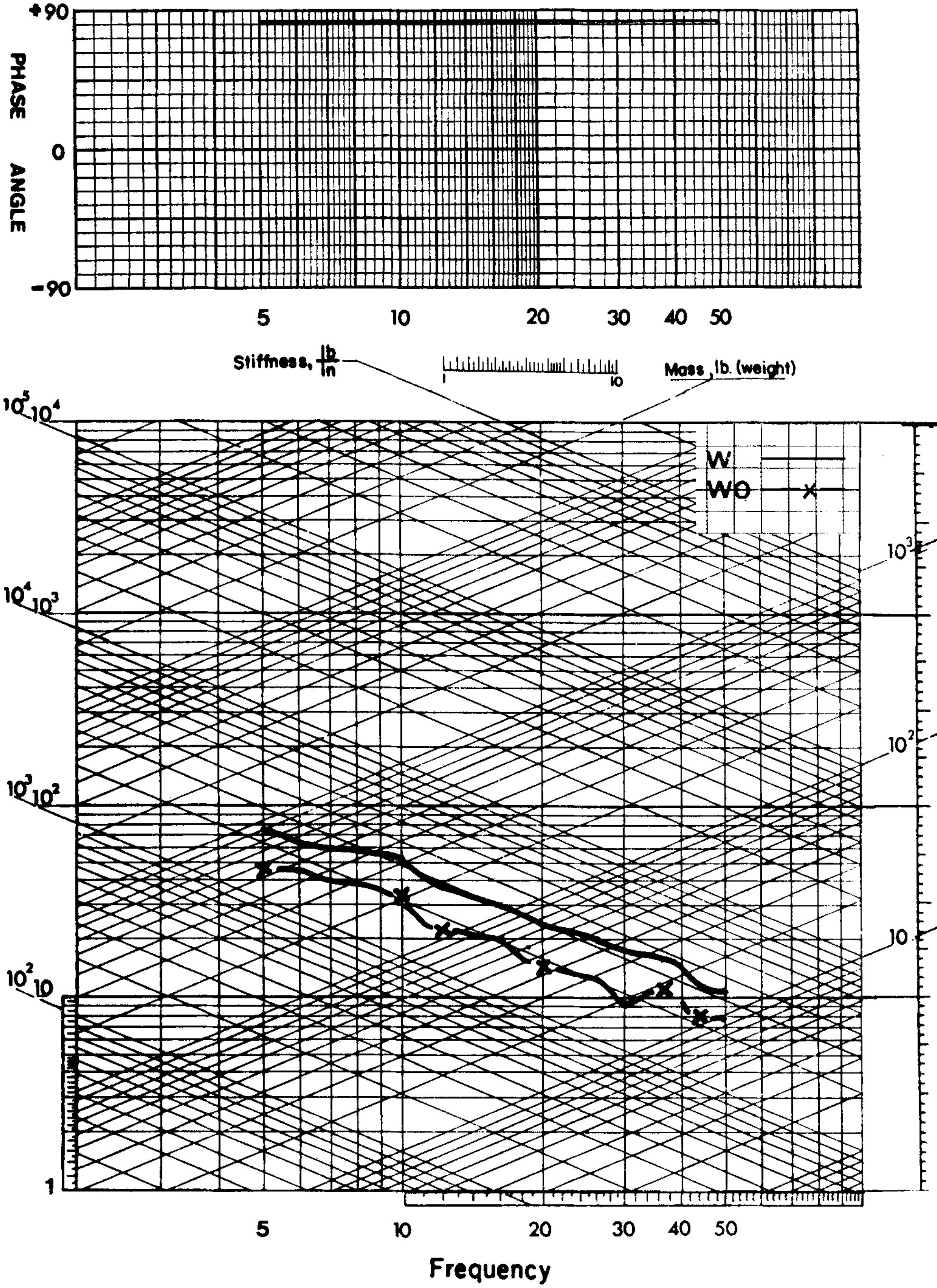


Figure 41. T_1—T_6 / W—WO / R / P_2

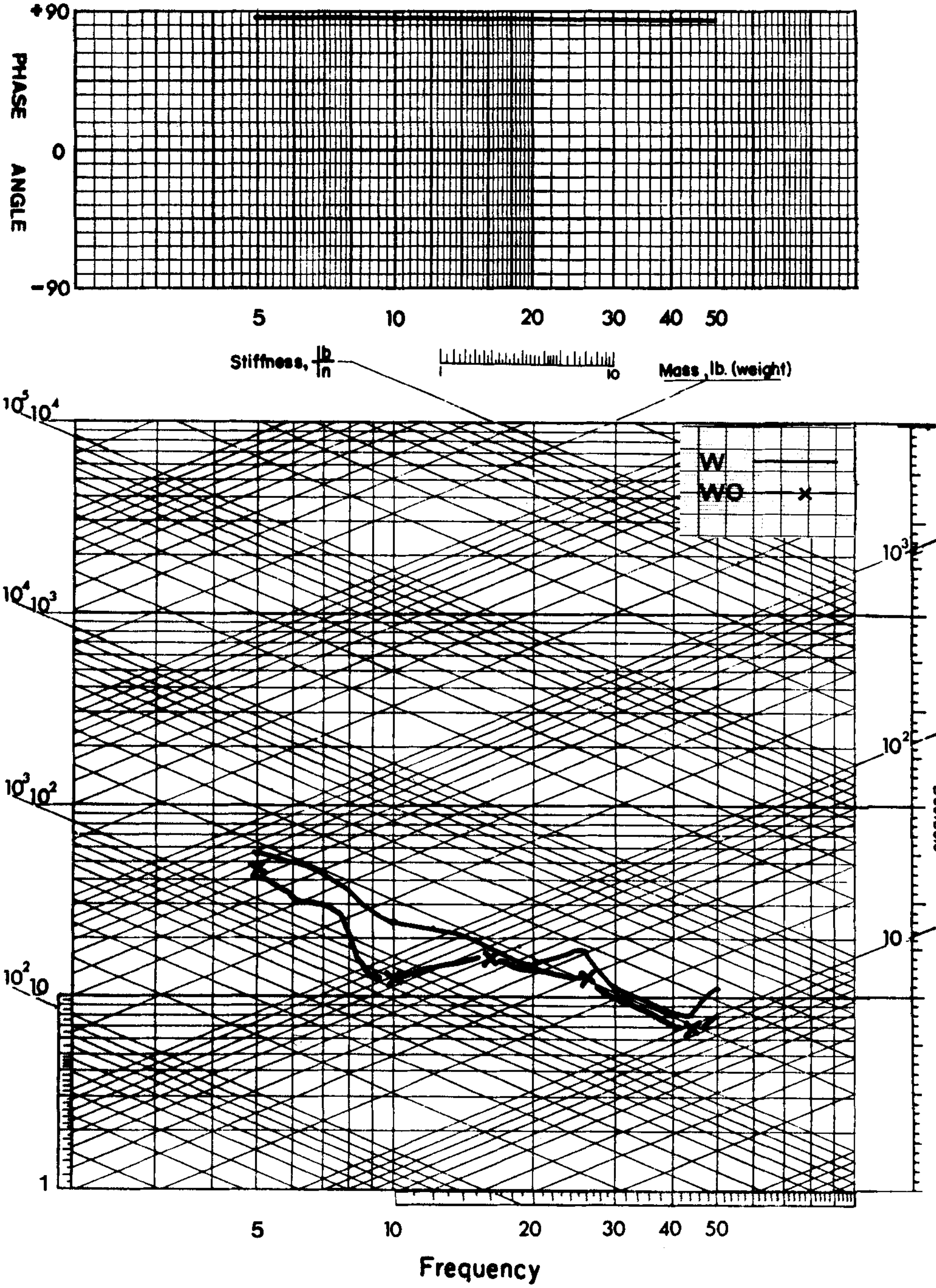


Figure 42. T_7—T_{12} / W—WO / R / P_2.

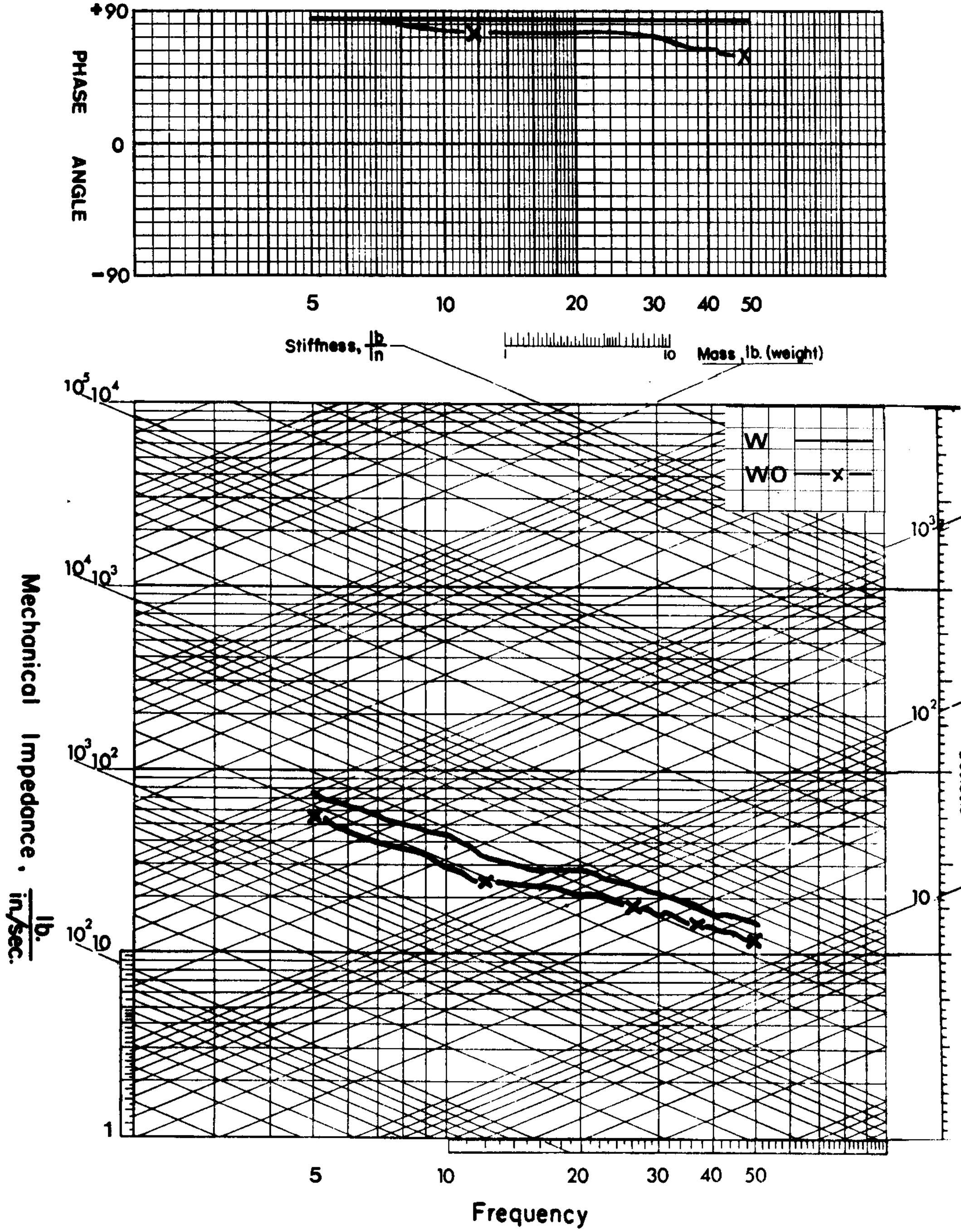


Figure 43. L_1—L_2—L_3 / W—WO / R / P_2

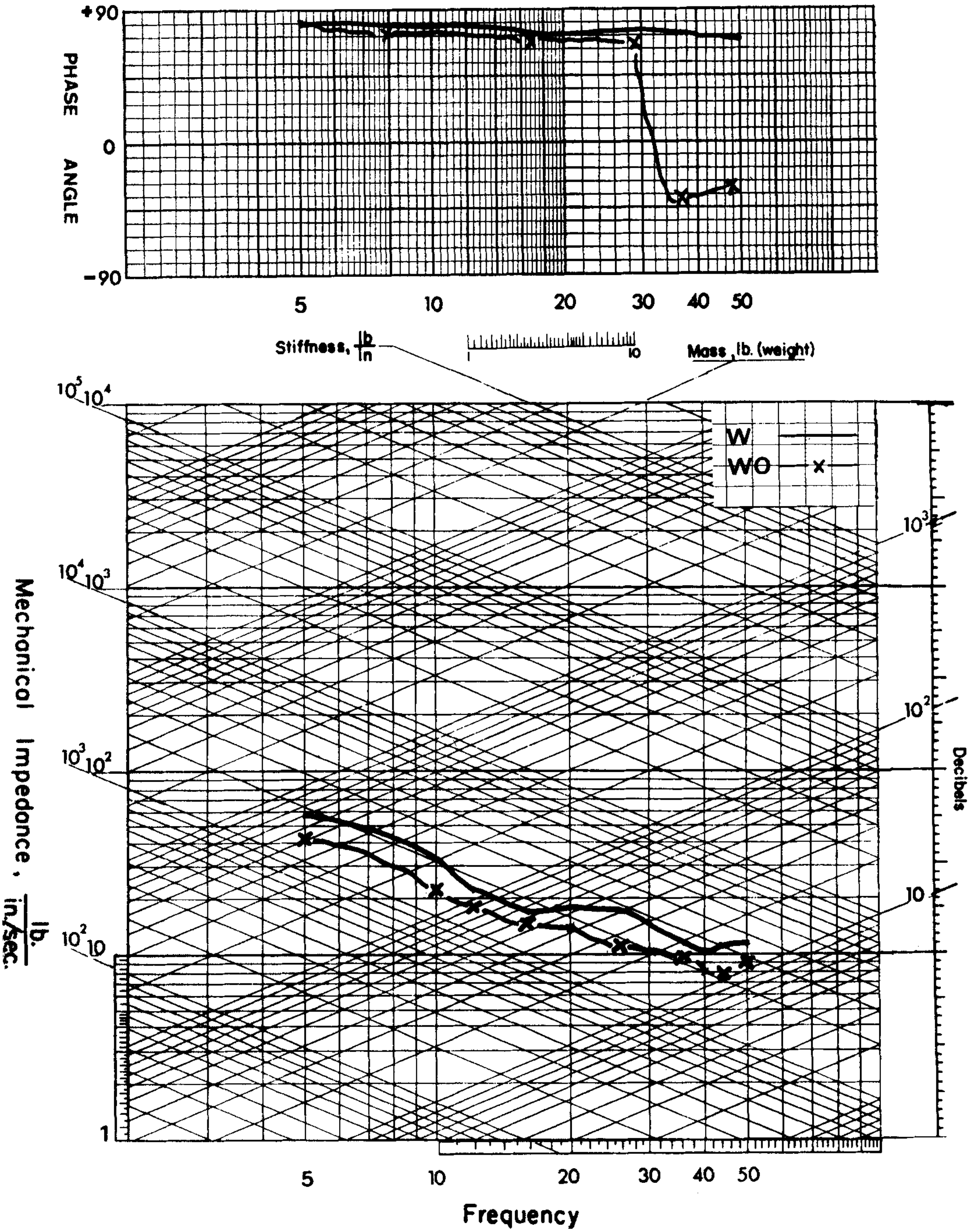


Figure 44. L_4—L_5—S / W—WO / R / P_2

One of the innate peculiarities governing the action of a vertebral unit under incessant load are its creep traits. The factors contributing to its function include the amount of the compression preload, length of time exposure, vertebral region, age of specimen, and in all likelihood temperature, humidity level, and any morphological variants, e.g. disk prolapse.

Upon subjecting the vertebral unit to a critical domain of constant loading, the structure did not lose its actual height immediately but did so over a period of time, usually several hours. The magnitude of the delayed reaction was dependent upon the material properties of the system. The alteration of its shape, under the application of the external force system was clearly a time-dependent process. Ostensibly, the physical process was one of compression, then rotation accompanied by bending. The typical pattern of flow appeared to be controlled by prescribed mechanisms of response. The comparative trigonometric relationships of the individual vertebral bodies within the unit were asymmetrically modified. The intervertebral disk material and its reactions in all probability entered into the physical process which brought about the time-dependent mechanical deflections and deformations of the individual vertebral array. The magnitude of the alteration in shape of the total vertebral unit was contingent upon the orientation of the unit with respect to the external load and its magnitude and whether or not these two constituents rearranged the structure into a state of complex loading. The extent of deformation was probably inhibited by the feedback of the intervertebral disk material, its configuration and its state, as well as the resiliency of the common spinal ligaments.

Following the conditioning period, and upon removal of the external force system, the vertebral specimen always manifested a continuing asymmetry which slowly subsided at a diminishing rate with increasing duration. For short intervals of time under static load (3—4 hours), restitution of the vertebral unit could be disproportionately long (6—8 hours) but in addition, this time factor varied greatly upon collating young and old cadaver material.

The magnitude of the external load and the duration of time exposure of the vertebral unit to uninterrupted weight must be contemplated simultaneously to assess the intricate reactions of the vertebral unit. Generally speaking, for high extrinsic loads and long time exposures, distortions were apparent not only within the cross-sectional area of the disk, but also in the articular facet joints. The overall effect was a time-dependent flow, creep and relaxation of cartilaginous material within the facet joints, annulus fibrosus and ligamentous apparatus which were the source of the rheological behavior. This behavior was a reaction to predominantly shear loading. The shearing forces in turn produced reciprocal cross-coupling effects among the axial, tilting and lateral modes but, most importantly, actuated mutual elemental rotations. As a consequence of the axially applied compressive load, a complex mode of response followed. This so-called torsional mode of perturbation was accompanied by deformation and flow of material comprising all soft tissue elements. In addition, this reaction contributed to an intrinsic contrasting clockwise or counterclockwise torsional response of the individual vertebral bodies, whose magnitude and direction of response was varied from specimen to specimen, vertebral region, and level, but whose direction of rotation was unique for each vertebral level as well as unit.

It may be postulated that in the presence of a constant compressive preload, the deformed vertebral network was probably closer to its accustomed "in-vivo configuration". This "conditioned" geometrical form possessed distinct impedance qualities when analogized to its relaxed unconditioned state. However, the articular facet joint surfaces could or could not exhibit a rotational degree of freedom or joint clearance while in the conditioned state. This mode of response was tempered by the *magnitude of loading and the age of the specimen.* Further qualitative results seemed to connote that concurrently with increasing age and whenever interacting deflections were too great within or over the normal operational range of a particular vertebral level, individual articular facet joints engaged their mutual cartilaginous surfaces and reacted as an override snubber, either symmetrically (by compressing the disks, increasing hoop tension and driving the joint surfaces together) or asymmetrically (by torsional reaction), and in this manner the interlocking controls and restricts the relative envelope of performance attributed to a particular vertebral level. In cases where the aging process and skeletal decay was more advanced, individual facet joint planes were radiographically observed to automatically and sequentially "set", and interacted to function in the distribution of complex loads. This observation brought to light the fact that at least in a prepared cadaver specimen, the thoracic and lumbar articular joint surfaces have acquired the ability of interconnecting in the rotational mode and transmitting mechanical energy. Their degree of interplay was governed by the inherent prestress and compression bias and morphological factors. Each vertebral unit attained a characteristic dynamic stability and possessed varied degrees of override protection within a characteristic normal domain of operation which seemed to decrease with increasing age.

Granting that the initial loading and subsequent creep reaction brought about viscous flow, the impedance characteristics and ingrained mobility of the conditioned vertebral unit were modified with respect to its relaxed defrosted state. The conditioned vertebral unit could exhibit a higher or lower impedance mode shape, a factor which was again dependent upon the vertebral unit, the magnitude of the preload applied, the length of time exposure to load, and age of the specimen.

The creep rate was found to increase when the posterior facets were eliminated. If only a posterior lamina was removed, the vertebral unit asymmetrically twisted under the constant load. Because of constructional heterogeneity, local strains probably varied from point to point within any particular unit. It followed then that the recovery mechanism also varied from place to place.

In considering the laws of reaction which influence the mechanical performance of the various vertebral units, structures and state of anisotropy, one must include creep and relaxation. The formulation of a theory of mechanical action of the vertebral unit must embody these concepts and no doubt many others. For this writer one of the more important concepts is spinal column prestress.

Tkaczuk (1968, 1970), calculated the magnitude of the prestressing on standardized samples of the longitudinal ligaments. He reported preload values for the interior ligament to range between 200—250 grams. For the posterior ligament the range was from 225—300 grams. He surmised that with increasing uniaxial loading the collagen fibers experience a process of mechanical adaptation by undergoing alignment with regard to the path of the applied force. He found that the strength for any distributive specimen was approximately 30 % higher in the younger material than in the older material. Nachemson and Evans (1968) have devoted some attention to the mechanical response characteristics of the ligamentum flavum.

In this study a number of qualitative observations were made with regard to the anterior longitudinal ligament. When the anterior longitudinal ligament (ALL) in the lower lumbar region was perpendicularly severed at a single level across the disk surface, the degree of relative lateral convexity markedly increased. If this was done at several levels, it was learned that not only did the magnitude of deflection change but so did the pattern of deflection, which clearly varied along the length of the column. For example, with regard to the lower lumbar units, if the ALL was severed its convexity of bowing was well defined, whereas in the lower thoracic unit it was less defined, in the mid-thoracic even less. The bowing was again noted to increase and continued to do so up to the level of the lower cervical vertebrae. This furnished some evidence to show that the prestress was distinct along the length of the spinal unit and its magnitude may be in relationship to the ALL. The ligaments accountable for the particular mechanisms of action were not distinguished.

When a compression bias was imposed on a defrosted lower lumbar unit, its bowing was observed to increase, but on the other hand, the bowing was relatively less for the upper lumbar column, and still less at the mortice and even less up to about the level of T_6. When the T_1—T_6 unit had an axial compression bias applied through its hard tissue bearing surface its *concavity* decreased (convexity increased).

The picture of the C_4—T_6 unit under load was complex and resembled that of a column about to buckle. It was reasoned that perhaps the manner in which the compression bias was applied, could nullify the reaction of the ALL with respect to the T_1—T_6 units, whereas in the case of the lower lumbar unit, tensile loads would probably be found to increase on the ALL. Under such conditions, the ALL in between these units in all likelihood were under various degrees of tensile loading. If this be the case, the particular adaptation characteristics of the vertebral units were most probably governed to some extent by the disks, and the counteracting ligamentous structures in the posterior vertebral arches in both these observations. However, in the latter findings, there could be the possibility of hard tissue interaction. The question then asked was, in what manner does the external force system excite the unit? Obviously, the "live" vertebral unit was not subjected to continous stress or strain but contrarily to a multiformity of loading conditions. This factor, linked with the realization that the vertebral unit creeps and relaxes were contradictory findings, because the creep would no doubt offset some of the prestress. The reason why the vertebral unit creeps was still left open to conjecture.

To shed some light on this problem the subject of prestressing was studied.

Prestressing can be defined as the application of predetermined force or moment to the vertebral unit in such a manner that the combined internal deflections resulting from this force or moment from any anticipated condition of external loading will be confined within some normal envelope of action. The *combined effects of the prestressing force and the externally applied loads* will result in a distribution of vertebral stresses in the unit that may be thought of in the following manner. The locus of the points of application of this force in the vertebral unit can be called for lack of a better term the pressure line.

It seemed likely that the pressure line in the vertebral unit was dependent upon the magnitude and direction of the externally applied load as well as the magnitude and deflection of stress due to the prestressing. This implied that a change in the external loading within the prestressed vertebral unit could produce a shift in pressure line and less of an increase in the resultant force within the vertebral unit.

From this deduction it follows that the pressure line no doubt varied as a function of vertebral unit level and deflected its geometric configuration with regard to range of motion, the magnitude and manner of external loading and specimen geometry. This could probably explain the cause as well as the manner by which the system was excited.

However, the external forces acting on the vertebral unit are constantly changing. There probably occurs a time reduction in the prestressing forces due to the time-dependent creep characteristics of the disk material and relaxation of the common ligamentous apparatus. But this leaves the question, "why does the creep take place to begin with?" This is best answered by additional experimental data, but based on the present experimental findings perhaps its role was to alter the deflection for cases where the resultant forces are significantly eccentric, because the rotational warping changes at certain vertebral levels due to the creep were greater than the creep shortening effects on reducing the prestress, especially with regard to the T_7—L_3 unit.

During the time of this writing additional experiments were going on to further substantiate this proposed explanation for this particular mechanism of action.

Vibrocreep

Vibrocreep is defined as the acceleration of creep under a compression bias and an additionally superimposed vibratory load. In the beginning of this section it was reported that under the action of a vibratory load, additionally imposed upon the basic compression bias, the rate of creep was in some cases observed to accelerate at certain frequencies. Despite the practical importance of and fundamental interest in this phenomenon, attempts to describe, isolate and explain this action proved unsuccessful. The literature in this area contains no information.

The role of mechanical stress and its effects on the shape and physical properties of hard tissue is well recognized. The influence of oscillations on the creep pecularities of the vertebral unit suggests that not only is the mechanical force environment important but the nature of this environment in terms of dynamic loading and frequency spectra ought to be defined. Perhaps, the energy content within a force-time history triggers a characteristic mode of cellular interactions. As the frequency changes a gradual modification occurs in base line electrical potentials and initiates specific cellular and collagen turnover. Thus the observed hysteresis may be a mechanism for the induction, differentiation, memory and maintenance of a chemical and cellular homeostasis. Based on these experiments a number of qualitative observations were made.

1. Creep under vibration was different for different vertebral units within the spinal column.
2. The larger the excursion peak to peak amplitude the greater the creep.
3. Upon removing the posterior vertebral arches the tendency to creep was greater but very much age-dependent.

Vibrocreep is probably an intermediate response rather than the subsequent response between the applied compression bias and dynamic oscillatory loading. For purposes of this study, it was assumed that the deformation of the vertebral unit under the basic static load plus the additional vibratory load exceeded the creep deformation produced by the constant load corresponding to the maximum value of the alternating force.

The analysis of force waveforms showed that the vertebral unit adapted readily to the excitation and that the time period of 15 seconds was sufficient to achieve a steady state response.

Further qualitative information is needed, before quantitative information is available which may prove useful in reducing test variables. In section 6 of the Appendix the creep response for a vertebral unit is illustrated.

This conglomeration of reactions and probably others further complicate the problem at hand because factors such as:

— individual geometric variations,
— load eccentricity,
— non-uniform distribution of compressive load and cross-sectional properties, radial inertia, and shear deformation,
— warping of spinal cross-section,

may produce significant variations from the conditions of applicability earlier outlined in this experiment.

It was clear that there are a large number of interacting processes at work, including a host of variables.

The difficulties of elucidating these questions are manifold. The testing machine together with the test specimen forms a very complex dynamic system. Perfection is not achievable in the loading system, nor, of course in the geometric form or in the material characteristics of the vertebral units. The question to be answered now is, "To what extent do these various factors *in combination* influence the nature of the response phenomena of the vertebral unit?" However, there will always arise questions that cannot be answered with complete assurance unless a complete analysis in the above sense is undertaken.

An understanding of such processes is thought to be of importance in interpreting and possibly improving current attitudes with regard to spinal mechanics.

7. Studies on the Impedance Response of the Segmented Human Spinal Column

OBJECTIVES, CADAVER MATERIAL, AND METHODS

The objectives of this experiment were to determine the impedance variations encountered with regard to the applied compression bias and age group. Specimen numbers 49, 75, 76, 25, 77, 74, 45, 46, 35, 12, 57, 78, 55, 28, 18, 32, 30, 1, 2, 65, 62, 44, 38, 11, 63, 36, 6, 8, 5, 21, were divided into their respective age groups andsectioned according to the scheme shown in *Figure 45*.

Each vertebral unit was cycled from 5 to 50 Hz, and repetitively loaded at four different magnitudes of compression bias — 36 lbs., 48 lbs., 72 lbs., and 140 lbs. (16.3, 21.7, 32.6, 65.3 kgs. respectively).

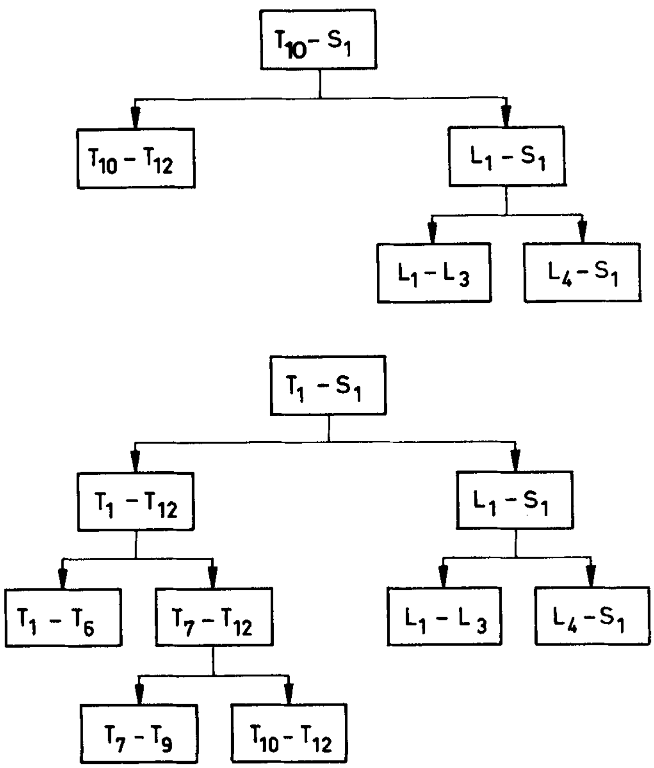


Figure 45. Flow diagram for final experiment.

DISCUSSION OF RESULTS

The results of these experiments are shown in graphical form numbered to correspond to the overall scheme illustrated in *Figure 45* and corresponding to *Figures 46—77.* The purpose of the graphs is to illustrate the mode shapes, any reflected resonances, as well as the phase relationships of force and velocity. A general picture of the data of the various tests compared with one another is best seen by examining each composite impedance plot.

In the phase plot, which was recorded simultaneously during the impedance test, the actual phase relationship between the force and velocity is displayed versus the frequency. Its purpose is to delineate the energy exchange characteristics of the vertebral unit. There are several pertinent aspects to the energy characteristics: the distribution of the dissipated energy over the frequency range, the distribution of the stored energy over the frequency range and the total amount of each. The impedance also defines energy transfer patterns.

With regard to the vibrocreep, in many cases the magnitude of the originally applied compression bias decreased by 6—12 lbs (3—6 kgs) for the first value of preload applied and decreased less on the second, third and fourth impedance runs. This repetitive process of preloading, impedance testing, preloading, impedance testing etc., seems to put the vertebral unit into a mechanically conditioned state.

The results of this study have shown that the isolated vertebral unit is a frequency-sensitive element. Its mechanical behavior is initially dominated by viscoelastic phenomea—that is, it possesses combinations in various proportions of energy storage (elastic) and energy dissipation (viscous) capabilities which in particular depend on a number of variables: the state of the material (conditioned or not), preload, the state of stress (uniaxial or multiaxial), frequency and amplitude of load/deformation (frequency being directly related to load/deformation rate) along with an increase in the impedance function when amplitudes are large—therefore an amplitude dependence.

Factors such as prior stress history, as well as temperature, humidity, circulation, and other factors are without doubt influential in affecting the properties of the specimen under vibration. A variety of rather complicated damping phenomena are synergistically operative. Their range and mechanisms seem complex so that more than one mechanism probably should be considered to explain the phenomena observed.

In the compilations of impedance mode shapes examples were observed in the raw impedance data, where the "same" vertebral unit under the same test conditions did not lead to the same results. The primary causes for such discrepancies were thought to be:

a. the actual variability in the vertebral unit,
b. deviations from the nominal test conditions,
c. unsuspected loss in the loading jaws and head, and other mechanical

connections between the vertebral unit and testing machine and the transducer,
d. creep and relaxation,
e. The mass and inherent stiffness of the system.

The vertebral unit could react as an amplifier or an attenuator to mechanical loading. The inherent characteristics which governed its mechanodynamic response to the excitation energy were dependent on a host of factors. These intrinsic and transmissibility controlling characteristics were of two natures: geometric (structural) and physical (material or tissue behavior). For the vertebral unit proper, the primary geometric characteristics are:

1. The geometric shape of the unit—anterior, posterior, lateral etc.
2. The section shape of the three primary regions; their degree of conditioning and their morphologic variations.
3. The mode and/or direction of loading and the characteristic response.

Items, 1, 2 and 3 are all inextricably connected and may be treated only in concert.

These inherent geometric and physical factors of the unit interact to generate still another set of parameters typically, the resonant frequency resulting from the shape, mode of loading, and the elastic modulus of the particular unit under consideration. The resonance observed in the vertebral unit was a direct consequence of its mass stiffness relationships whereby, at certain frequencies of excitation and values of compression bias, the vertebral unit exchanges kinetic and potential energy in synchronism, thereby storing energy as it was received from the excitation, and dynamically loads the system abruptly. Independent of the magnitude of damping, the phase lag between response and excitation at resonance was in the neighborhood of 90 degrees.

These physical factors affected the elastic modulus of the structure, which was an important factor in setting the resonant frequency through the EI product, the deflection and the spring constant. Undoubtedly, the single most critical parameter in controlling transmissibility of energy within the unit was the lowest resonant frequency, for this condition imposes the maximum response of load on all elements within the unit (but very much dependent on specimen geometry) and creates localized dynamic environments. With increasing values of compression bias, there occurred a shift of the resonance toward higher frequencies. The shape of the overall impedance curve remained more or less constant while a contrasting displacement took place in its magnitude. The shift could be up or down, but seemed to be dependent upon:

1. The past history of the vertebral unit.
2. The magnitude of the conditioning.
3. The level of the vertebral unit.
4. Morphological factors.

Finally, in these data it is clear that the behavior of the vertebral unit cannot be explained by a theory in which constant characteristics are assumed. The stiffness of the vertebral unit is not constant, it varies with deflection and its deflection varies as a function of many factors. In any system with constant characteristics the amount and distribution of mass, damping, and stiffness is fixed. In other words, the restoring force between any two points in the vertebral unit due to stiffnes is proportional to their relative displacement. The damping forces resisting the relative motion of any two points of the system are also taken to be proportional to the relative velocity of these points. No allowance is made for changing damping forces which may change discontinuously with a particular mode of excitation. Nevertheless, when the assumption of constant characteristics is made to represent the vertebral column, then certain general engineering properties emerge which are often taken for granted. For instance, the system has an inherent natural frequency and mode of response. When such a system is subjected to sinusoidal excitation it possesses two known factors; it has an input excitation and its response is in sinusoidal form (though it may be out of phase). In this dynamic state, the form of the fluctuating distortion is dependent upon the peculiarities of the vertebral unit and the input frequency. The trouble with such a theory is not just a question of numerical accuracy, but rather, that vital and obvious factors such as geometry, nonlinearities, etc., are often approximated right out of existence. The question therefore arises whether or not anything vital is lost when constant characteristics are assumed. This factor is dependent upon the viewpoint one is trying to put forth.

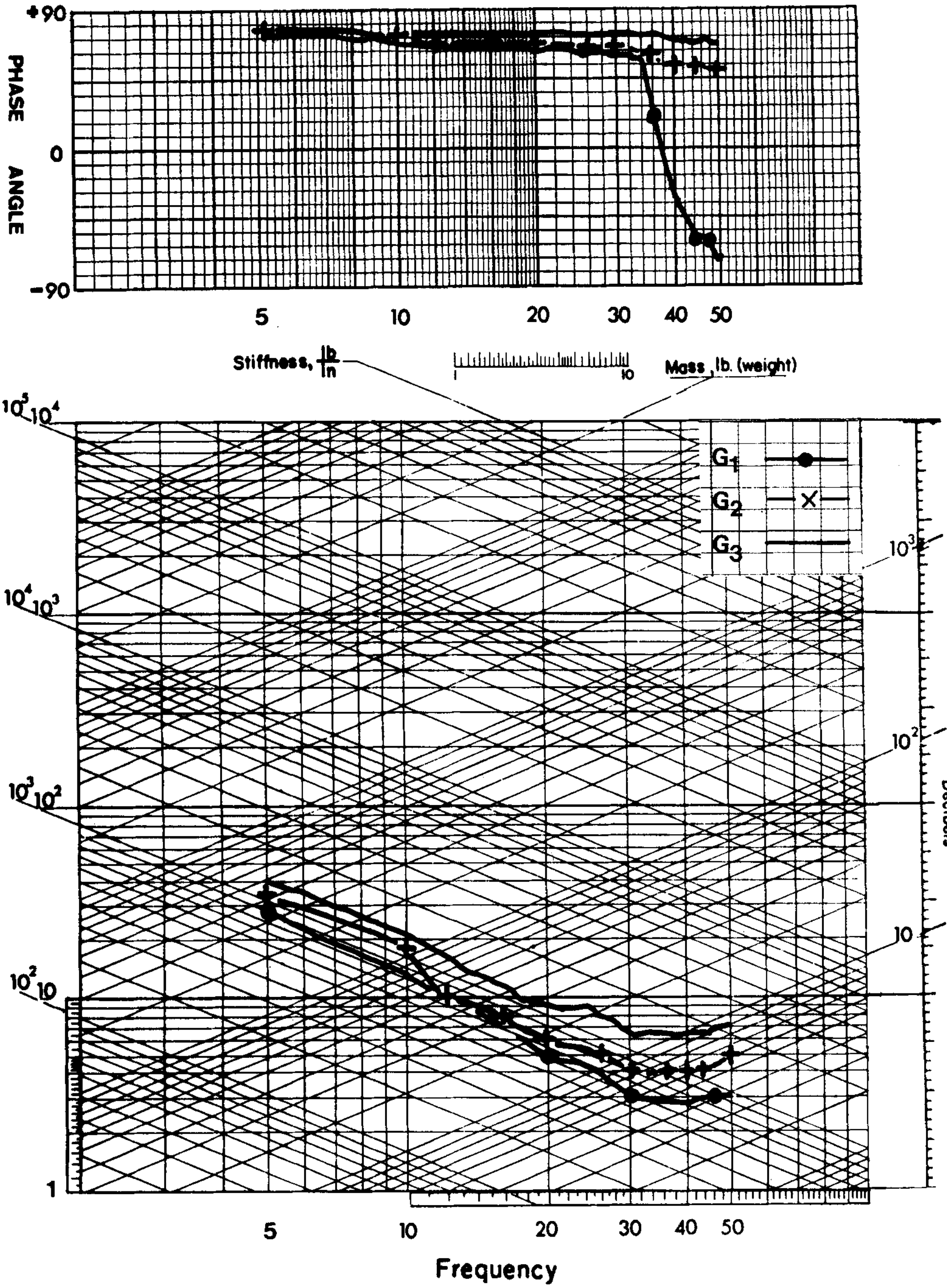


Figure 46. T1—T12 / G1—G2—G3 / P1

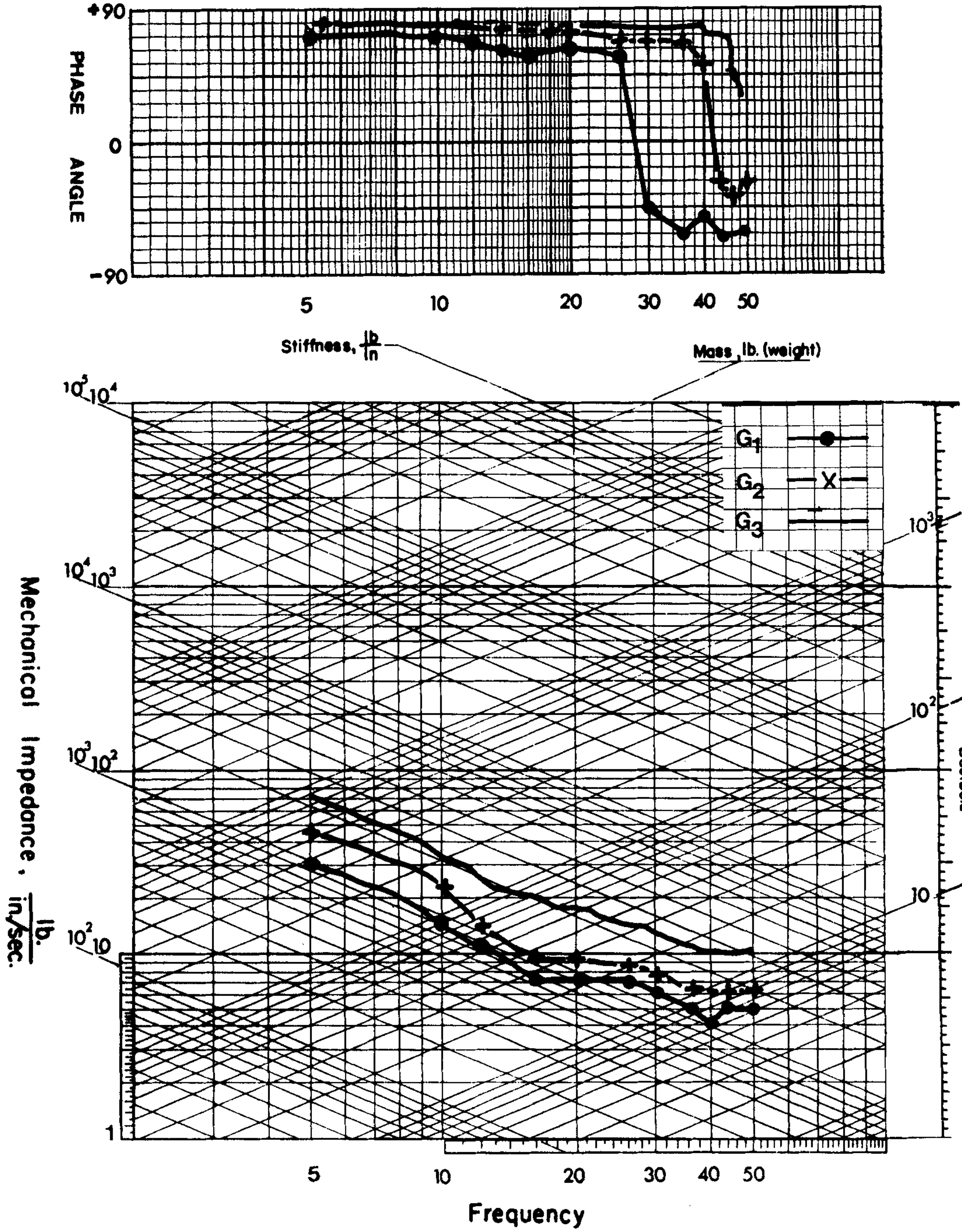


Figure 47. T_1—T_{12} / G_1—G_2—G_3 / P_2

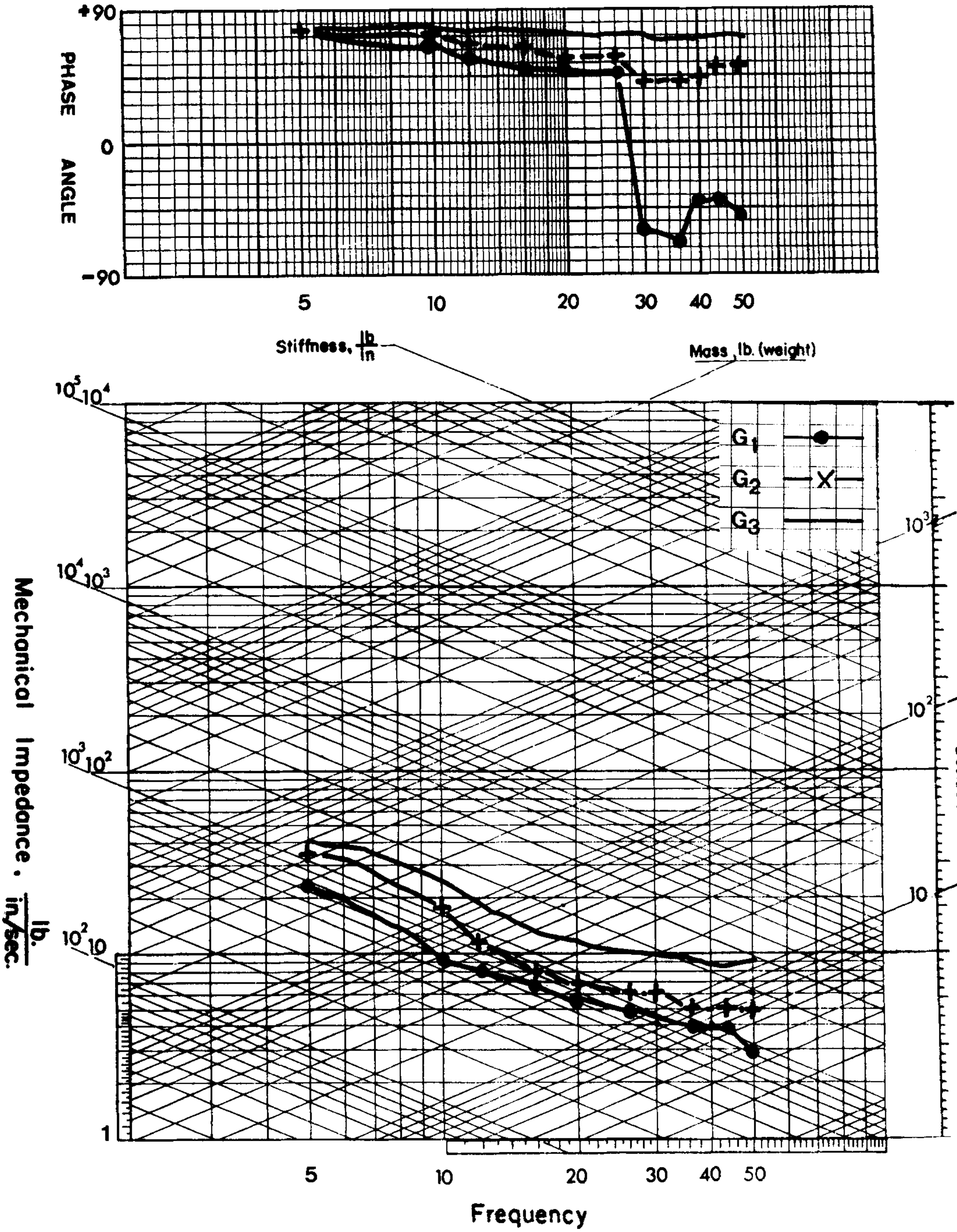


Figure 48. T_1—T_{12} / G_1—G_2—G_3 / P_3

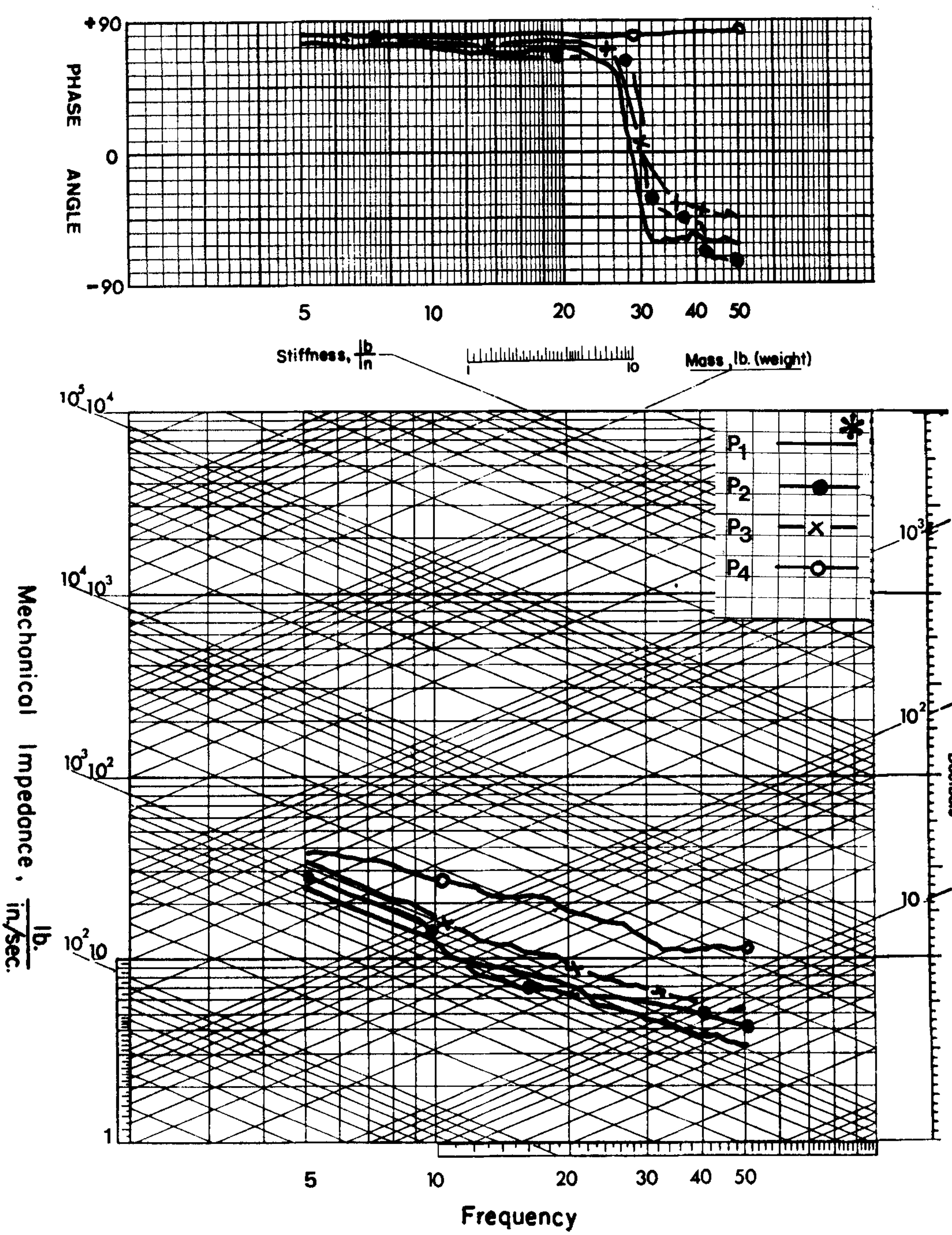


Figure 49. T_1—T_{12} / P_1—P_2—P_3—P_4

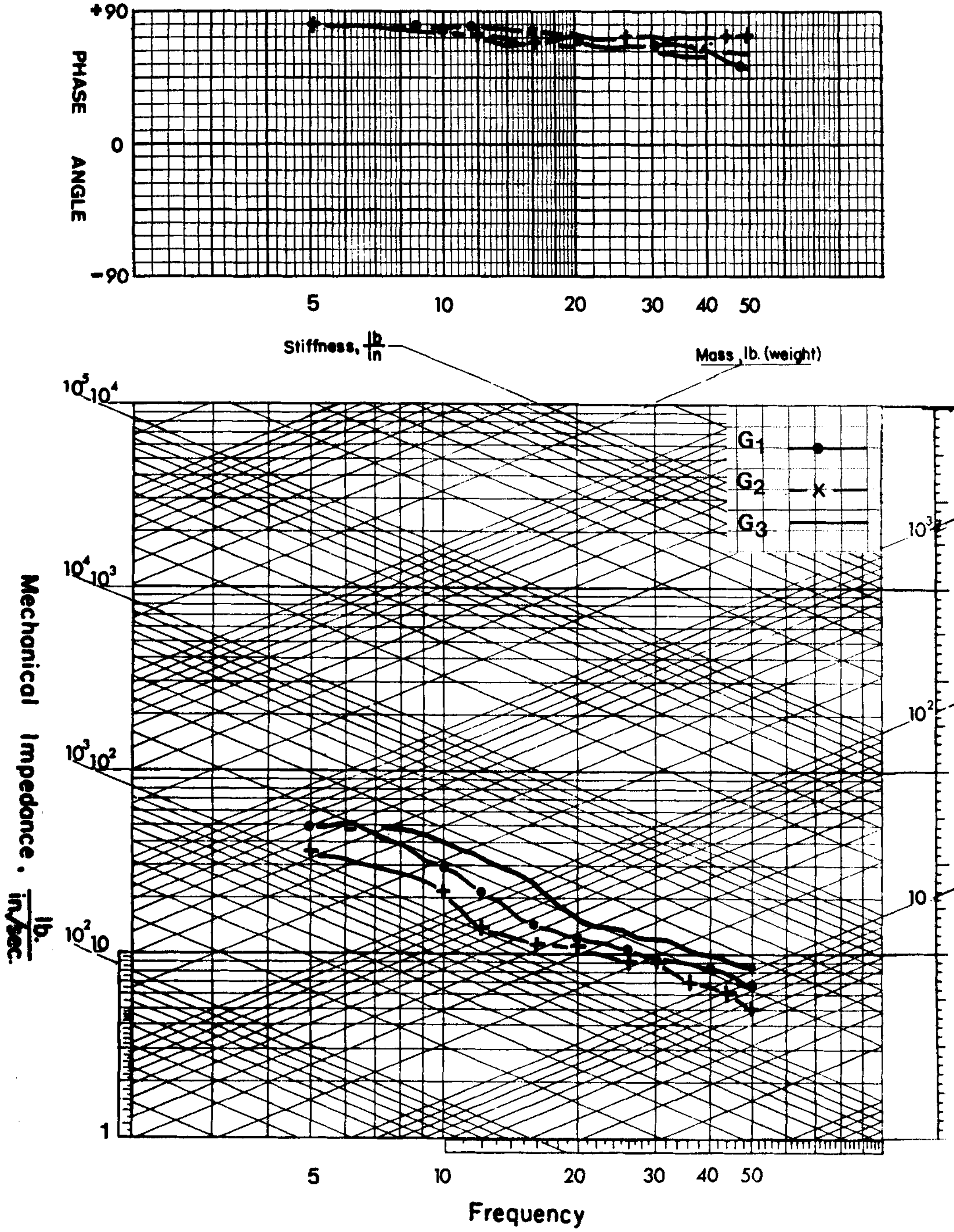


Figure 50. T_1—T_6 / G_1—G_2—G_3 / P_1

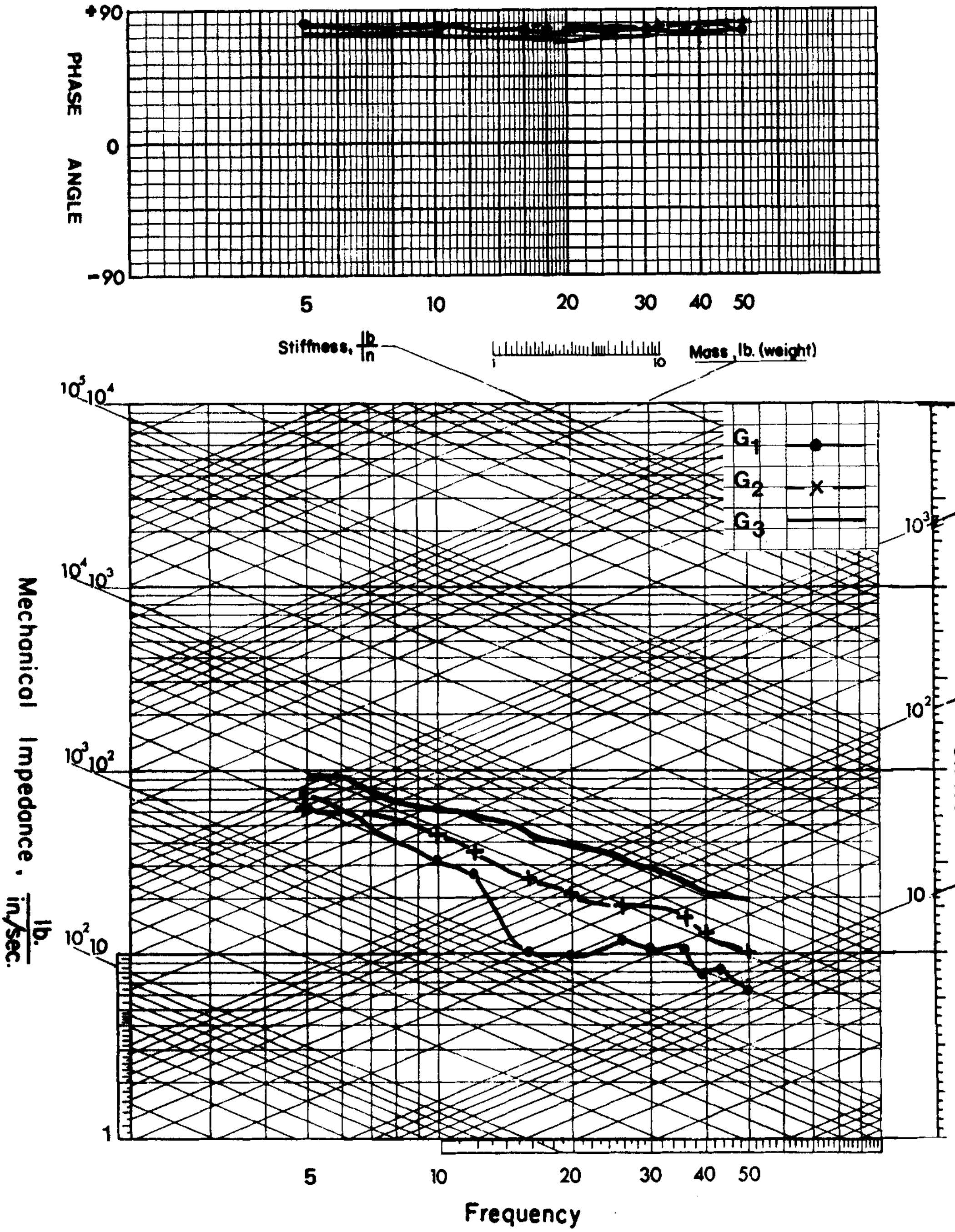


Figure 51. T_1—T_6 / G_1—G_2—G_3 / P_2

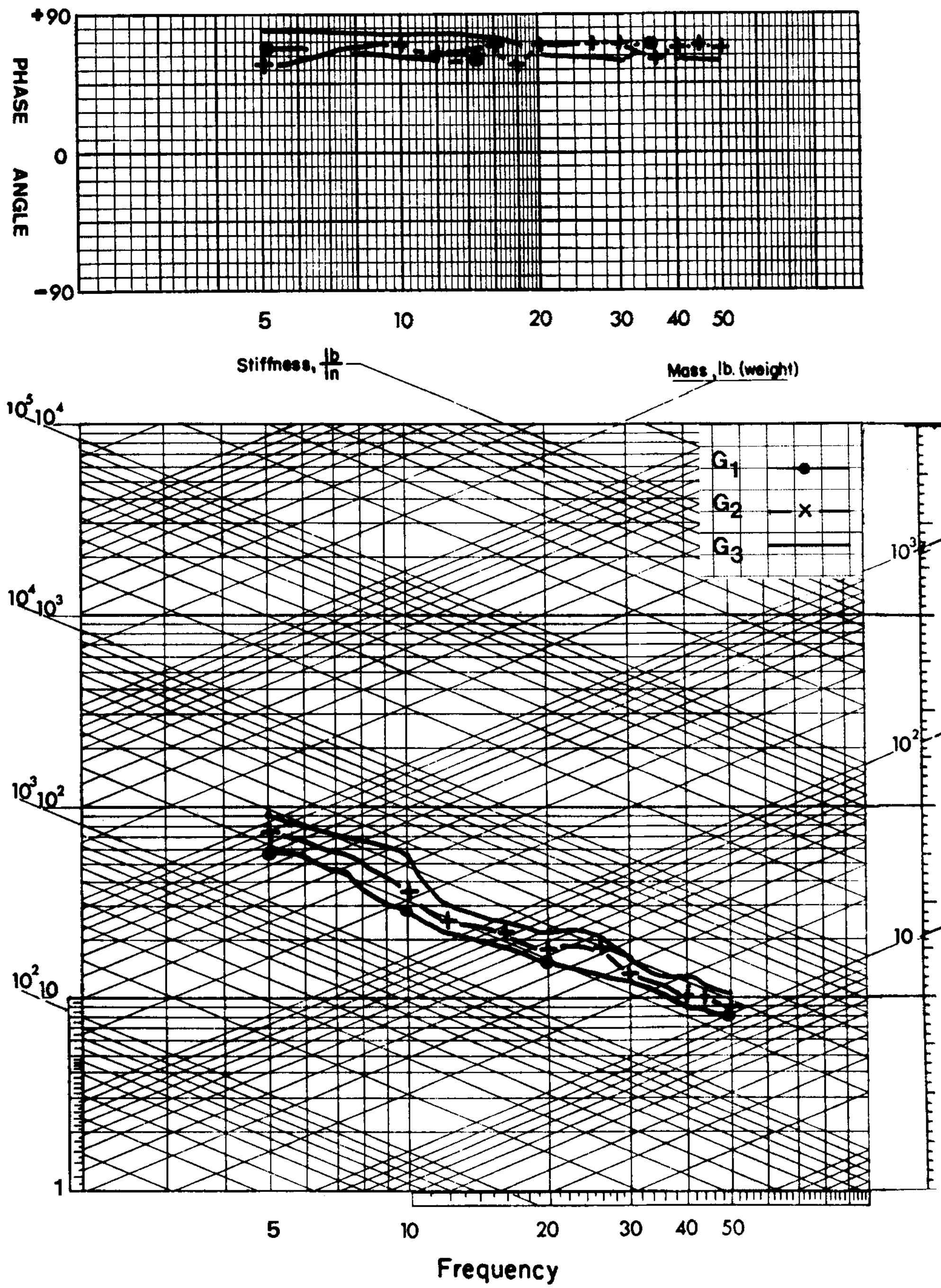


Figure 52. T_1—T_6 / G_1—G_2—G_3 / P_3

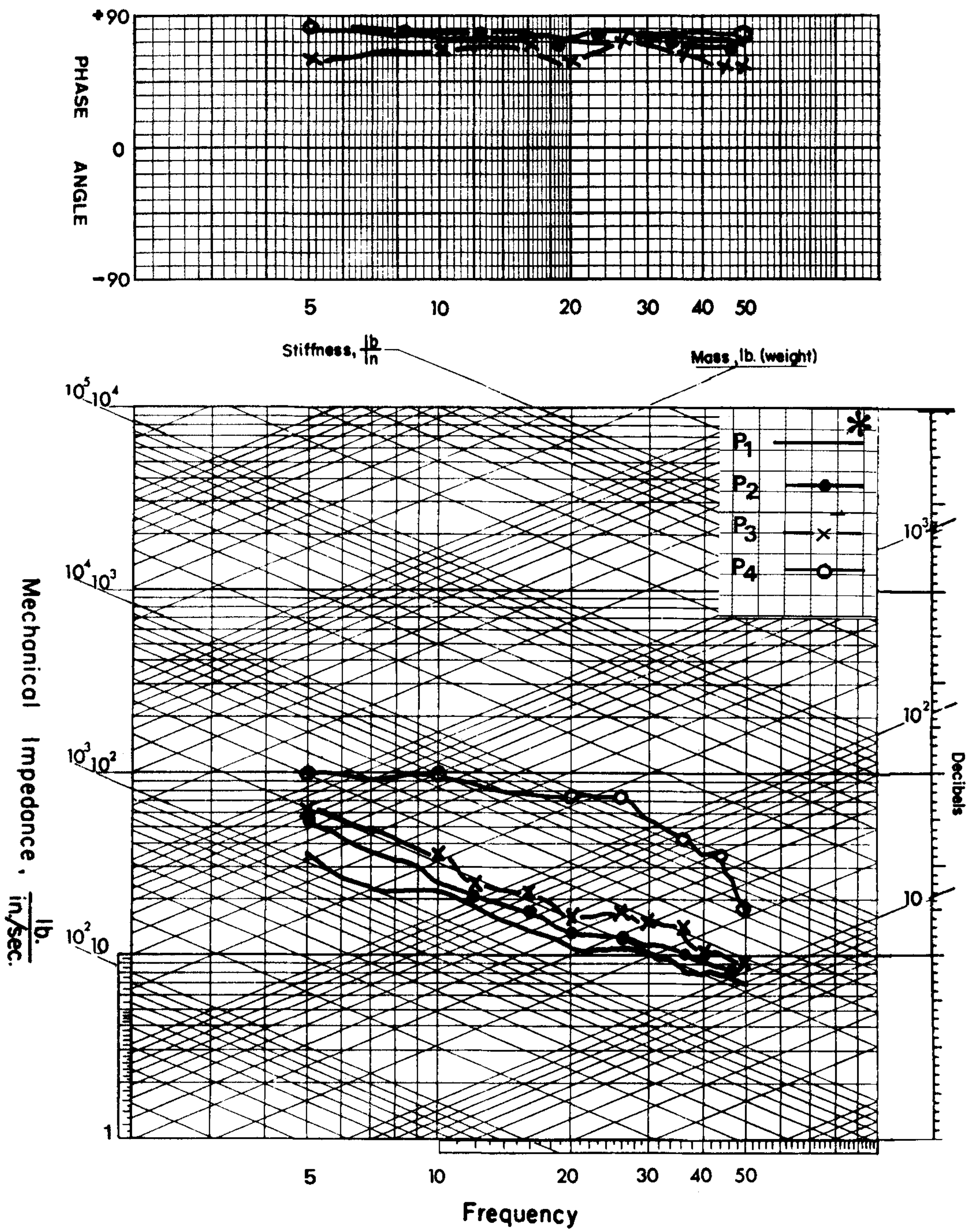


Figure 53. T_1—T_6 / P_1—P_2—P_3—P_4

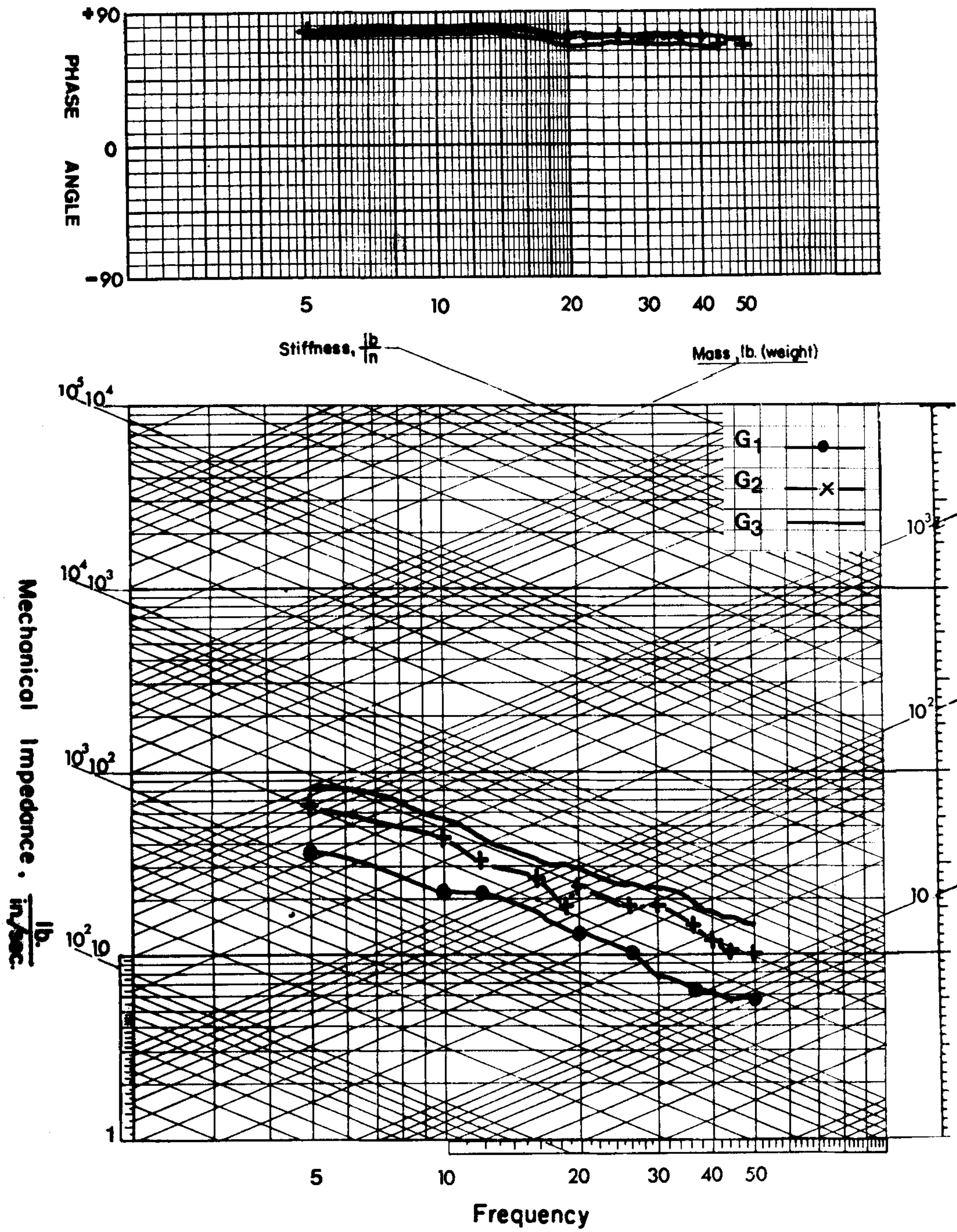


Figure 54. T_7—T_{12} / G_1—G_2—G_3 / P_1

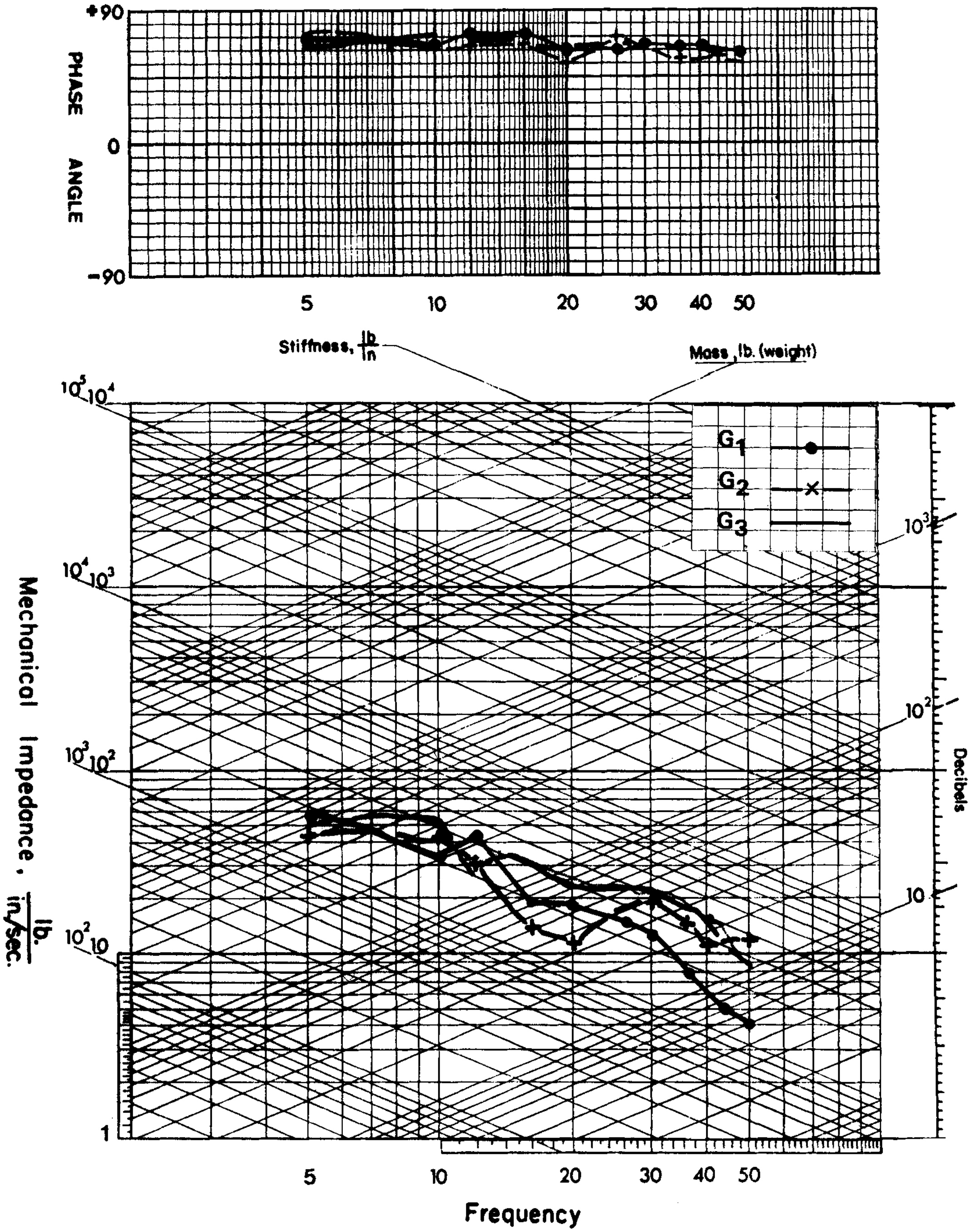


Figure 55. T_7—T_{12} / G_1—G_2—G_3 / P_2

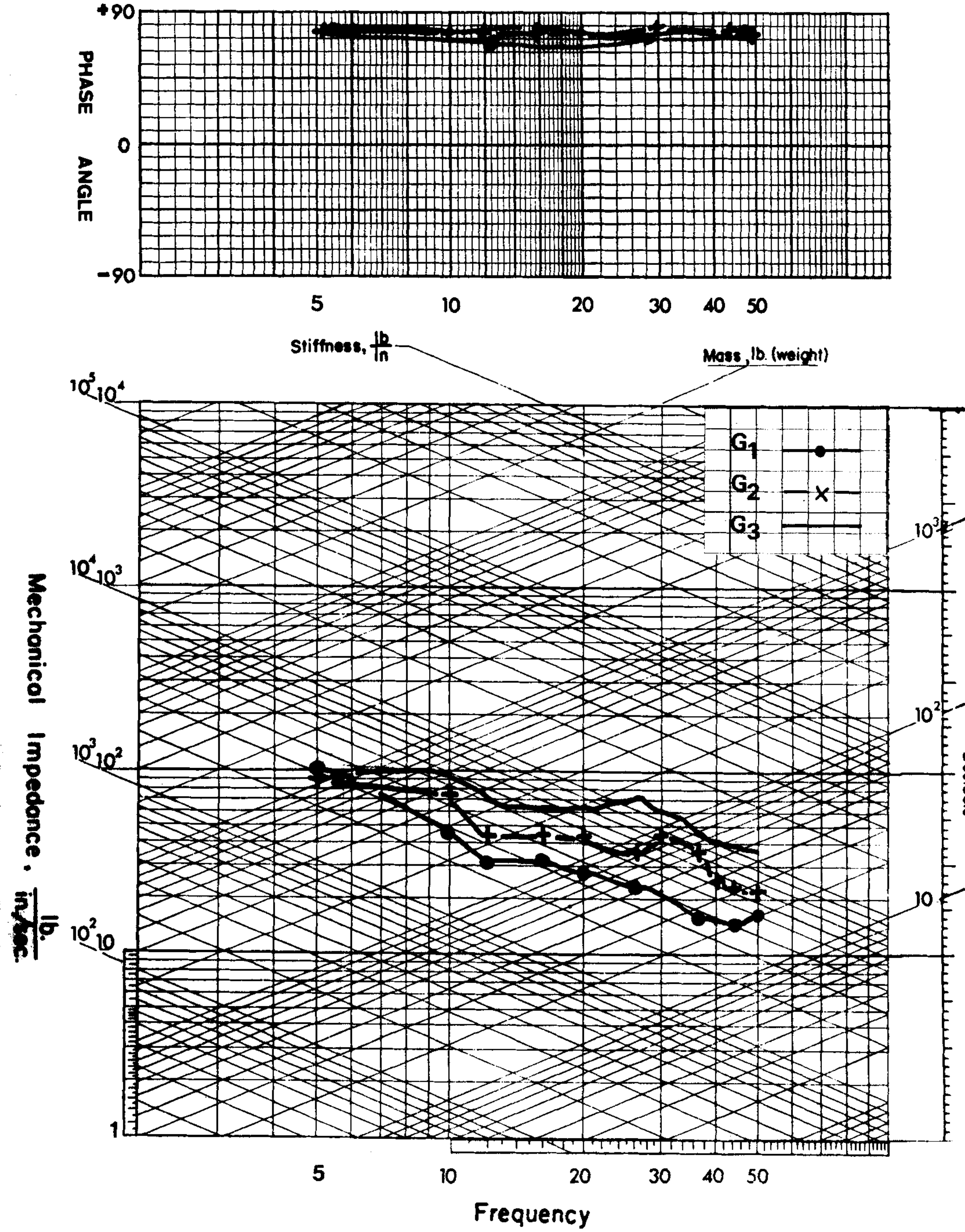


Figure 56. T_7—T_{12} / G_1—G_2—G_3 / P_3

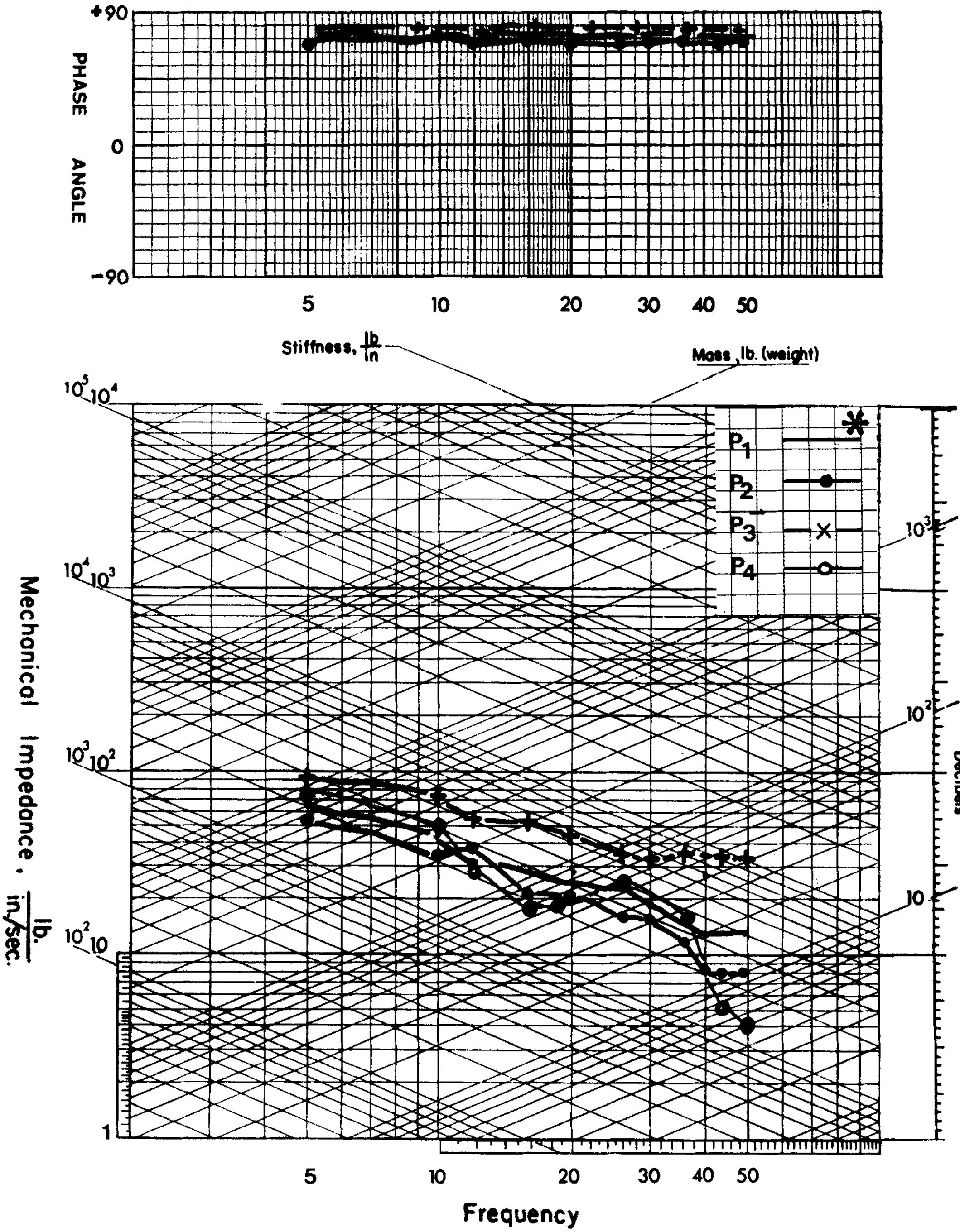


Figure 57. T7—T12 / P1—P2—P3—P4

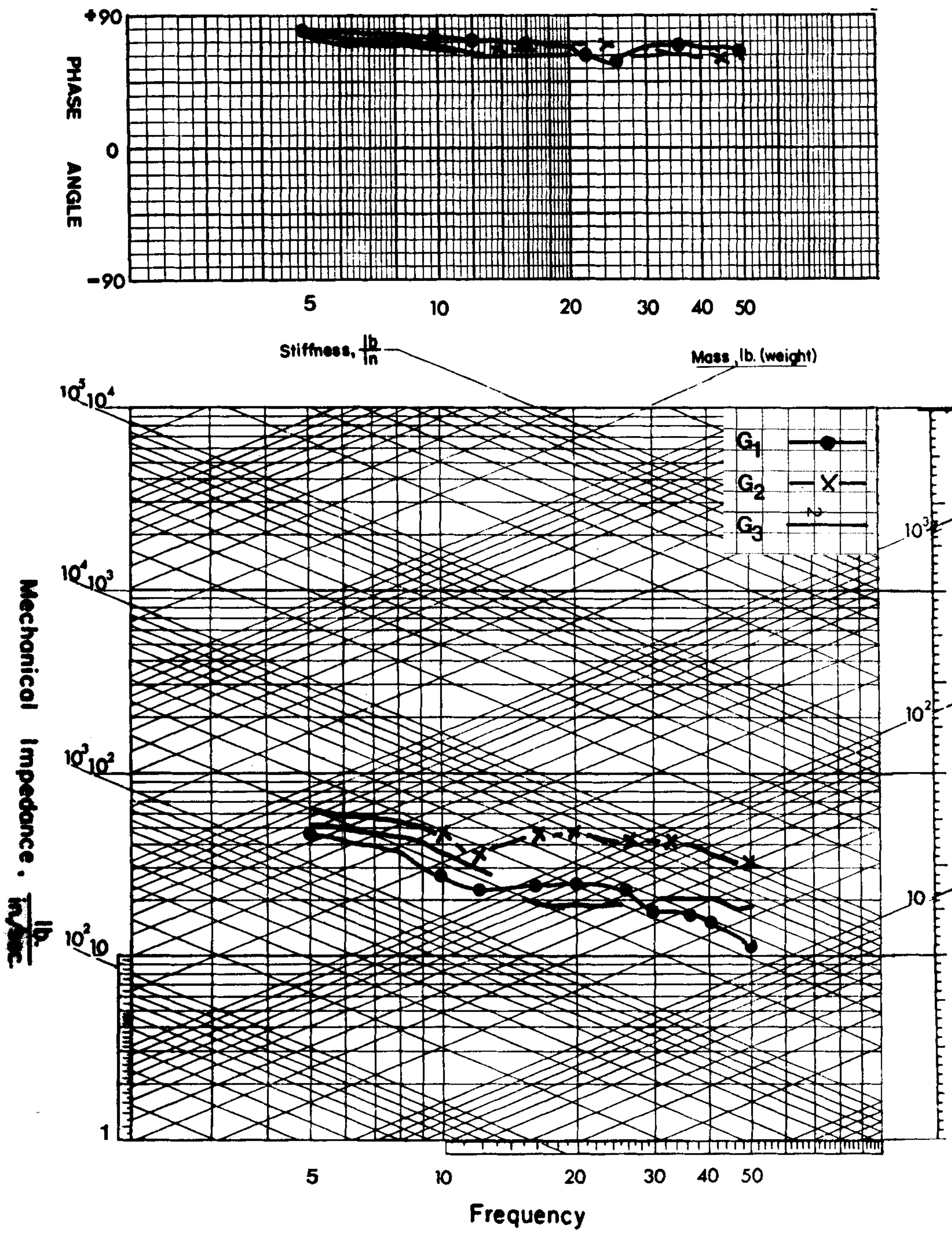


Figure 58. T_7—T_8—T_9 / G_1—G_2—G_3 / P_1

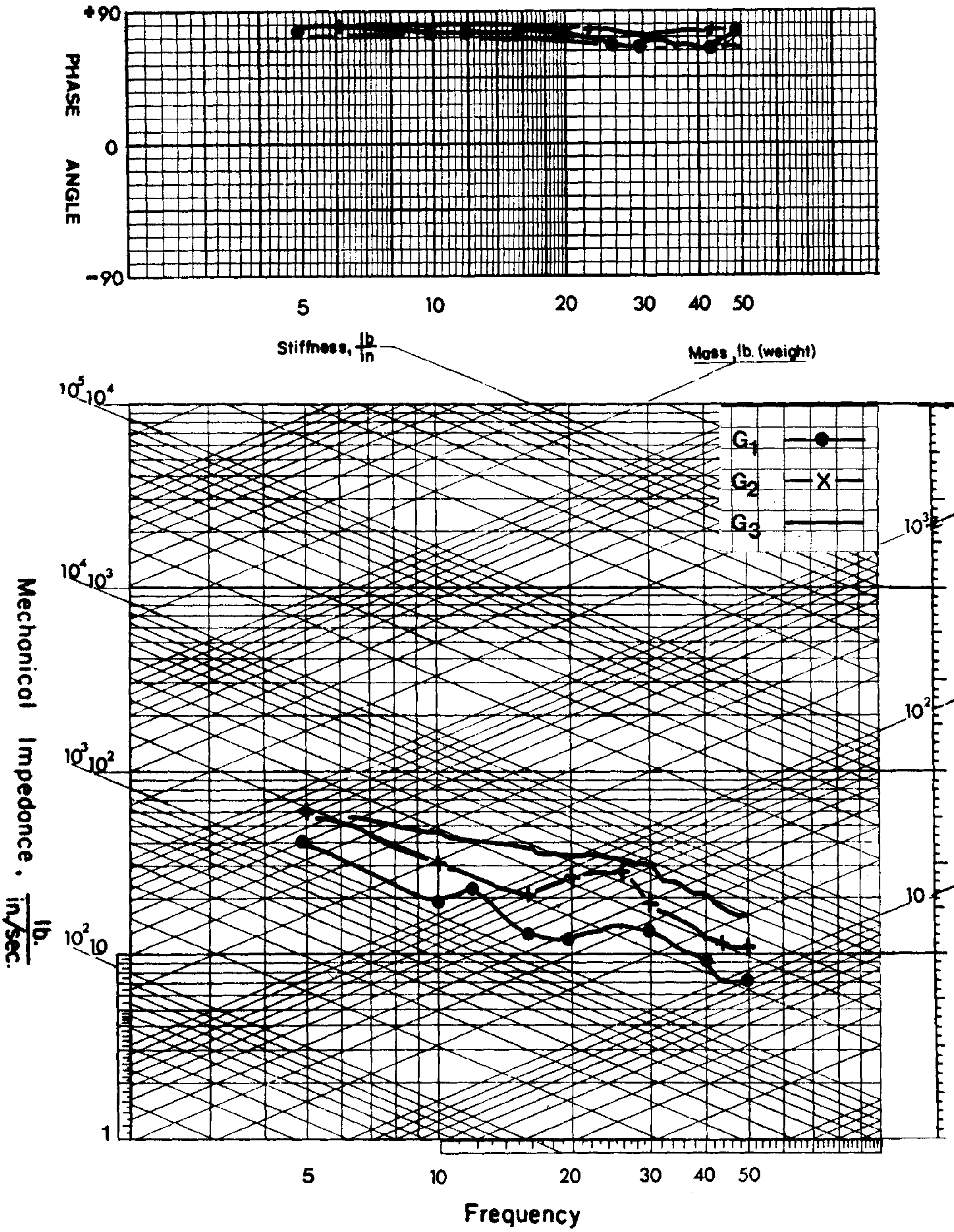
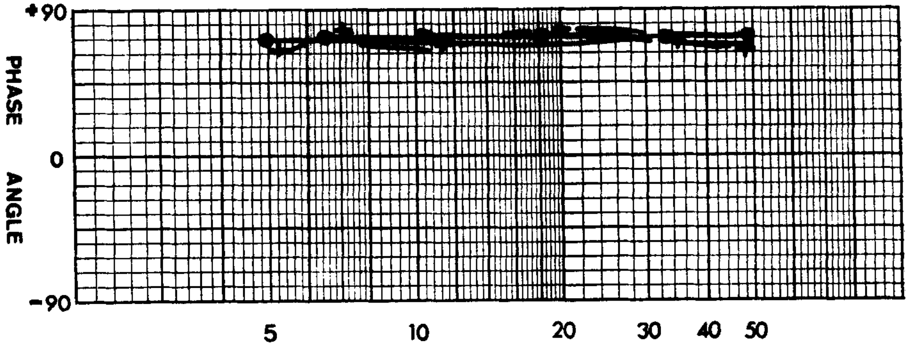


Figure 59. T_7—T_8—T_9 / G_1—G_2—G_3 / P_2



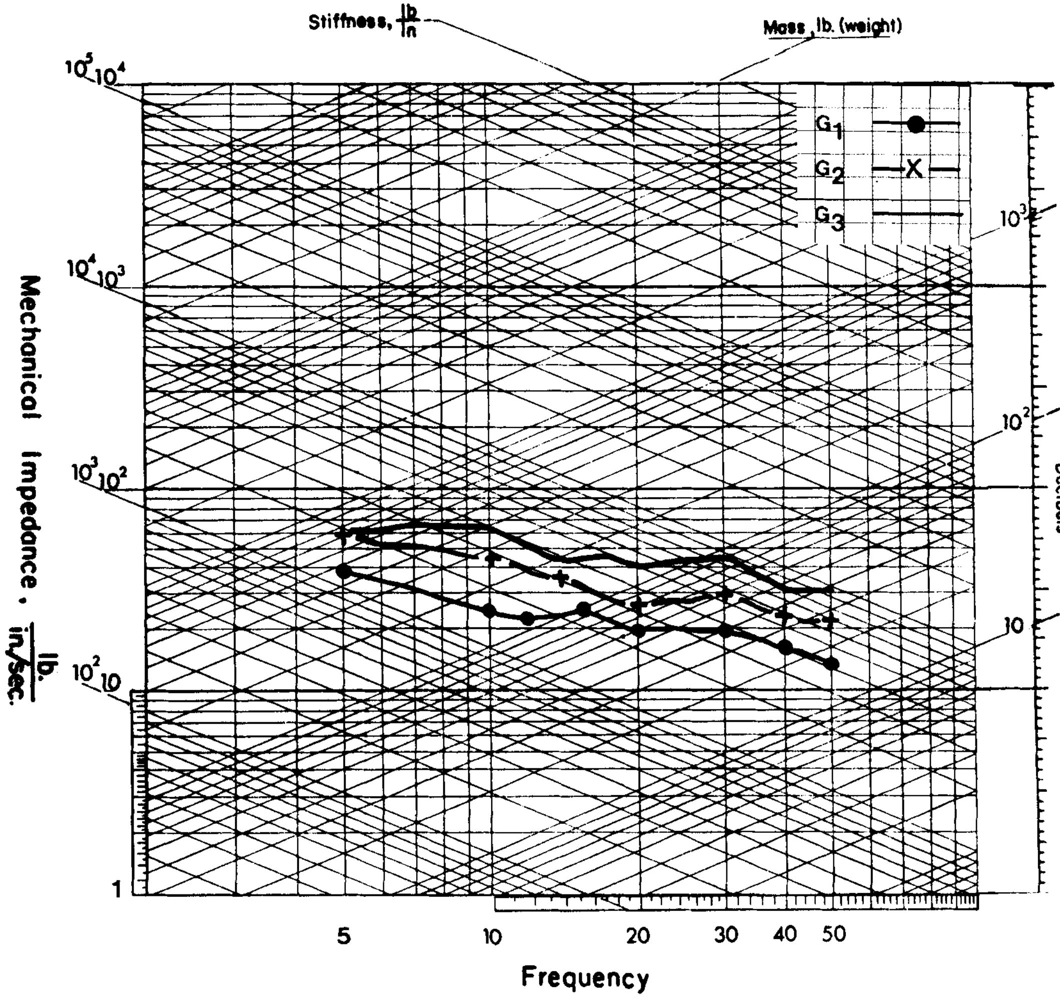


Figure 60. T_7—T_8—T_9 / G_1—G_2—G_3 / P_3

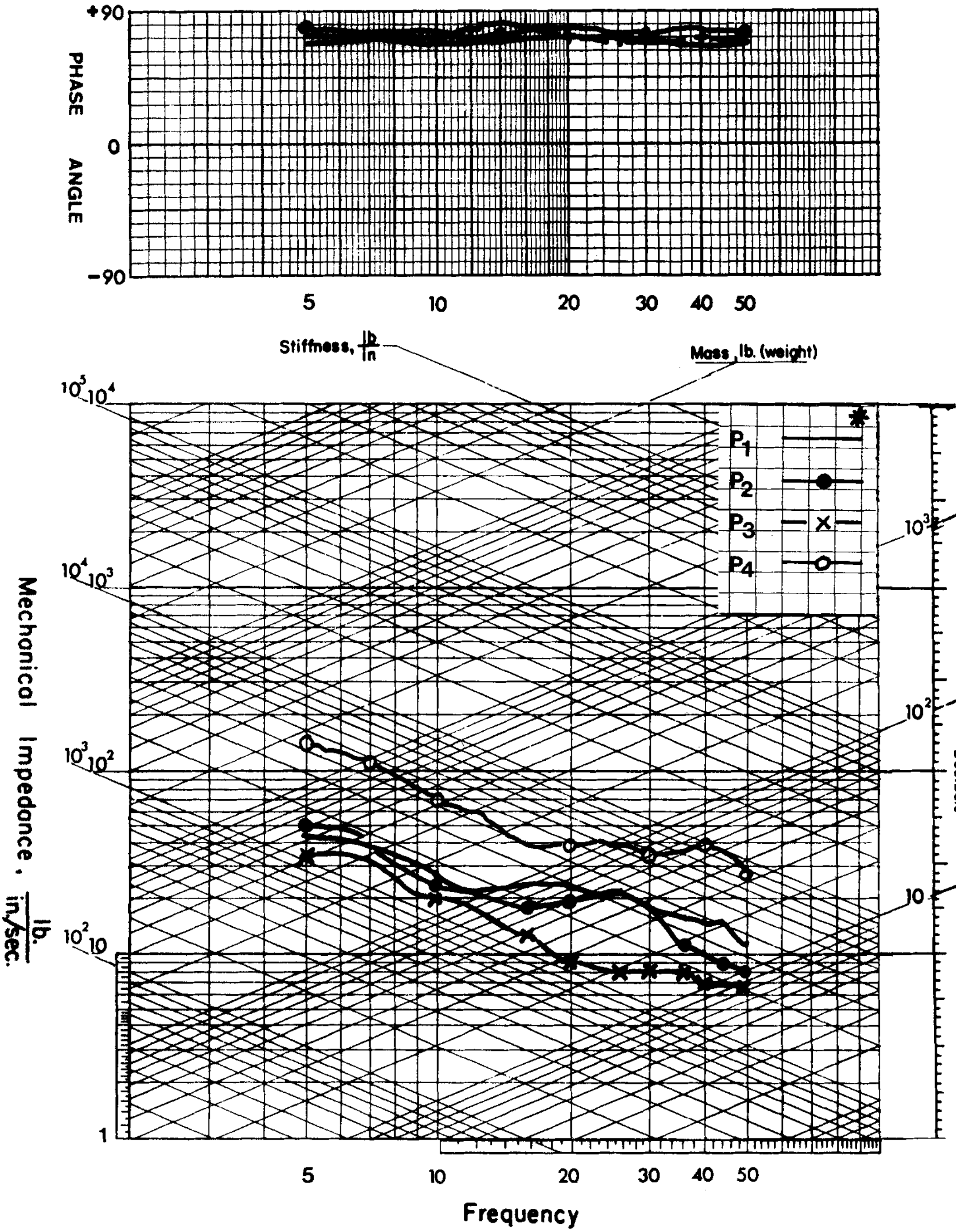


Figure 61. T_7—T_8—T_9 / P_1—P_2—P_3—P_4

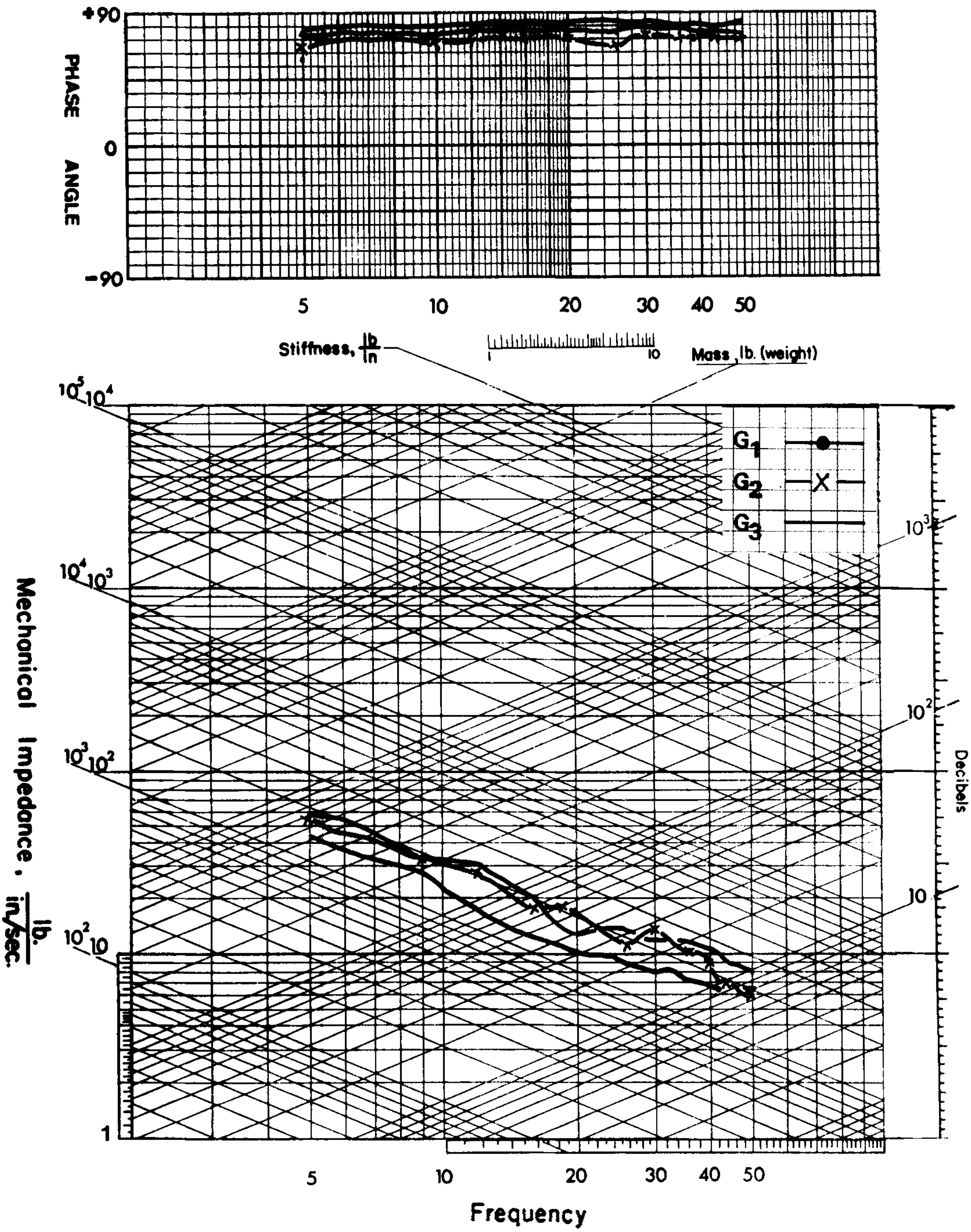


Figure 62. T_{10}—T_{11}—T_{12} / G_1—G_2—G_3 / P_1

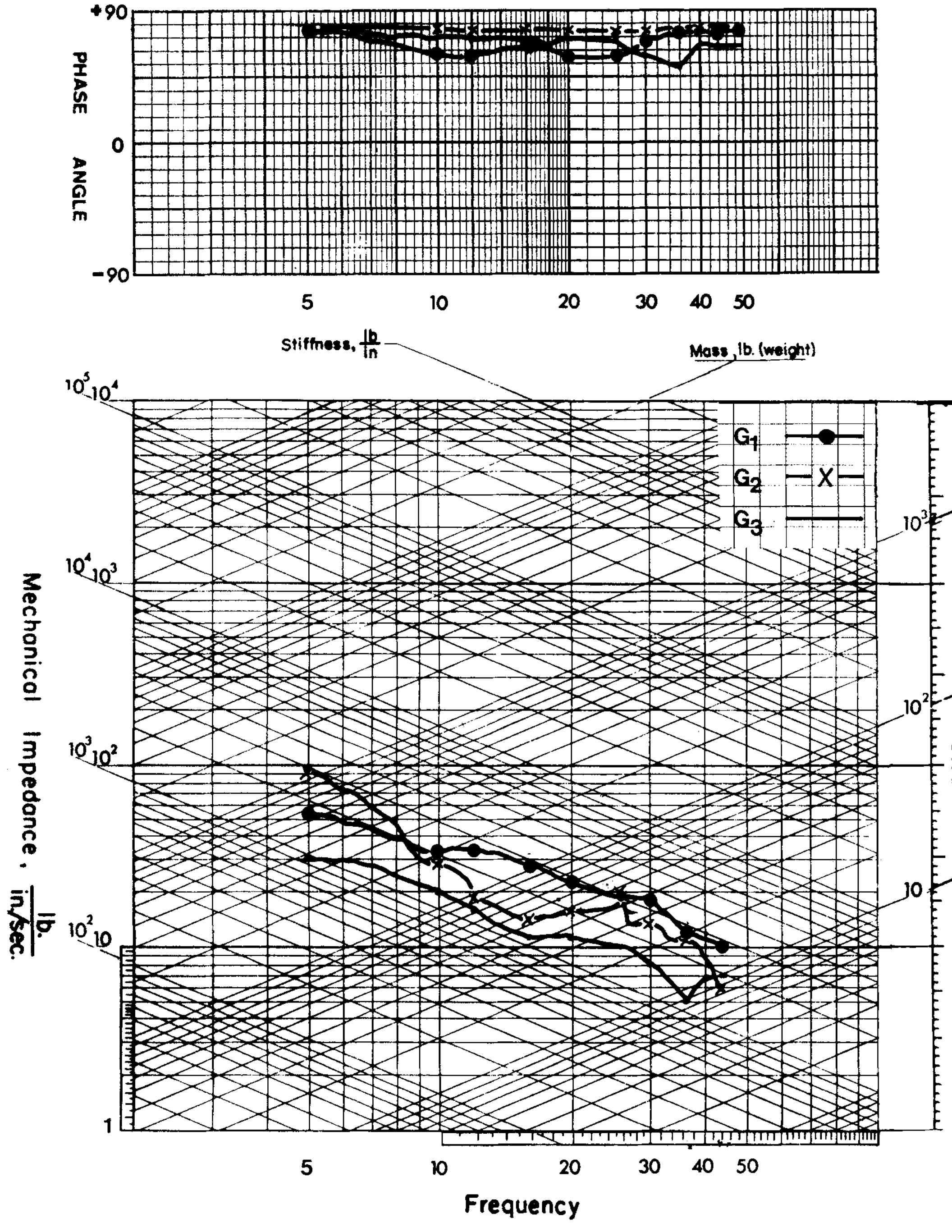


Figure 63. T_{10}—T_{11}—T_{12} / G_1—G_2—G_3 / P_2

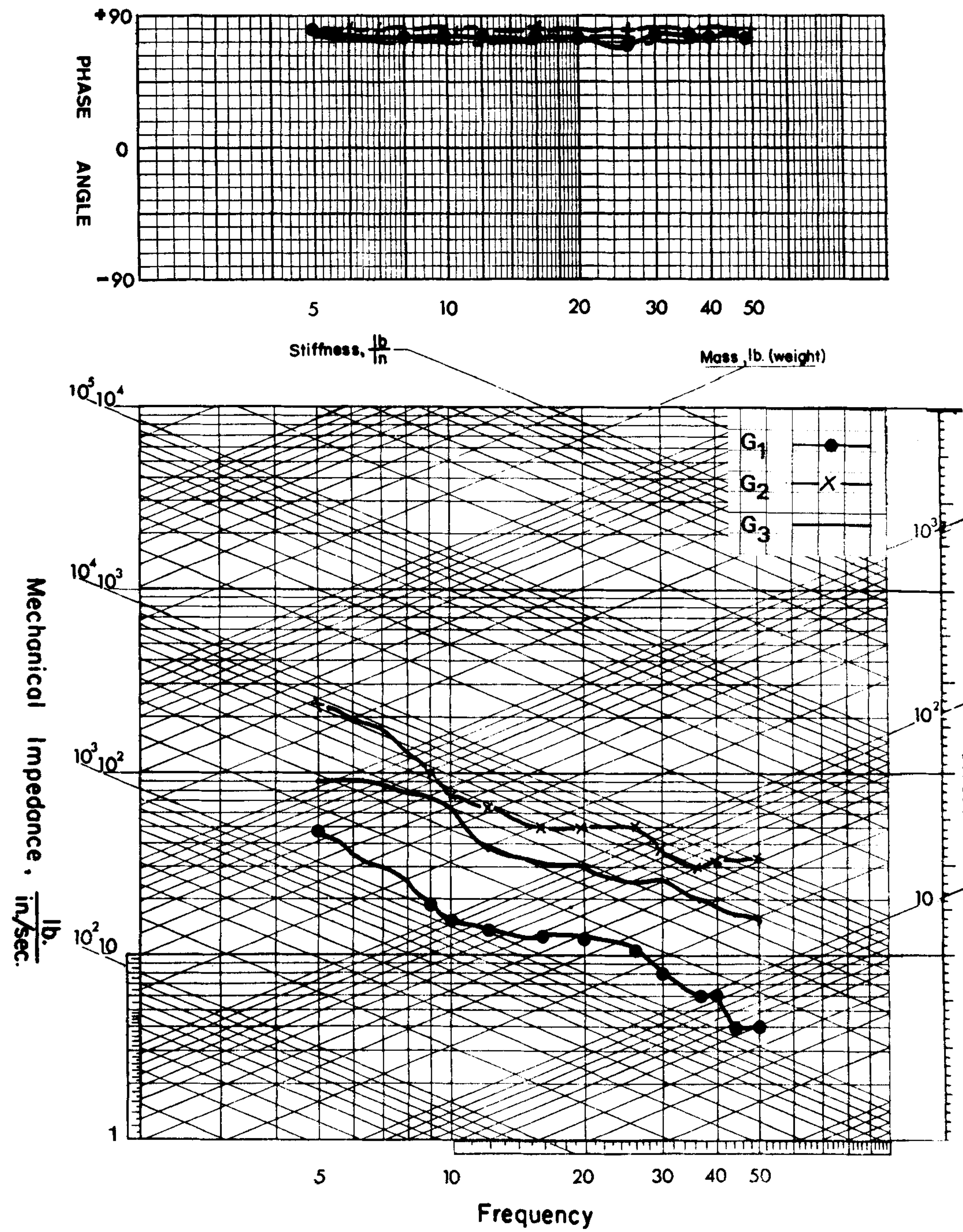


Figure 64. T_{10}—T_{11}—T_{12} / G_1—G_2—G_3 / P_3

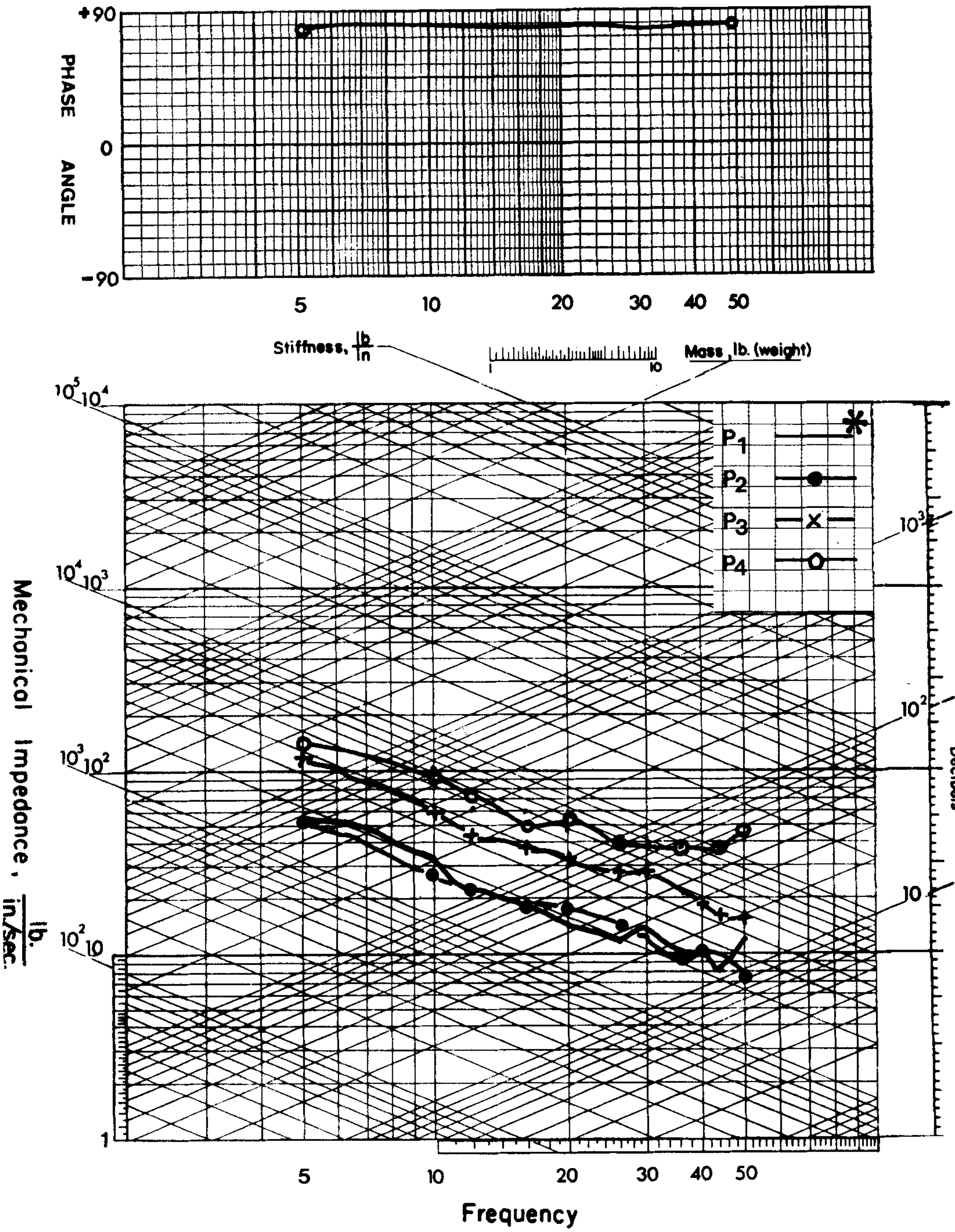


Figure 65. T_{10}—T_{11}—T_{12} / P_1—P_2—P_3—P_4

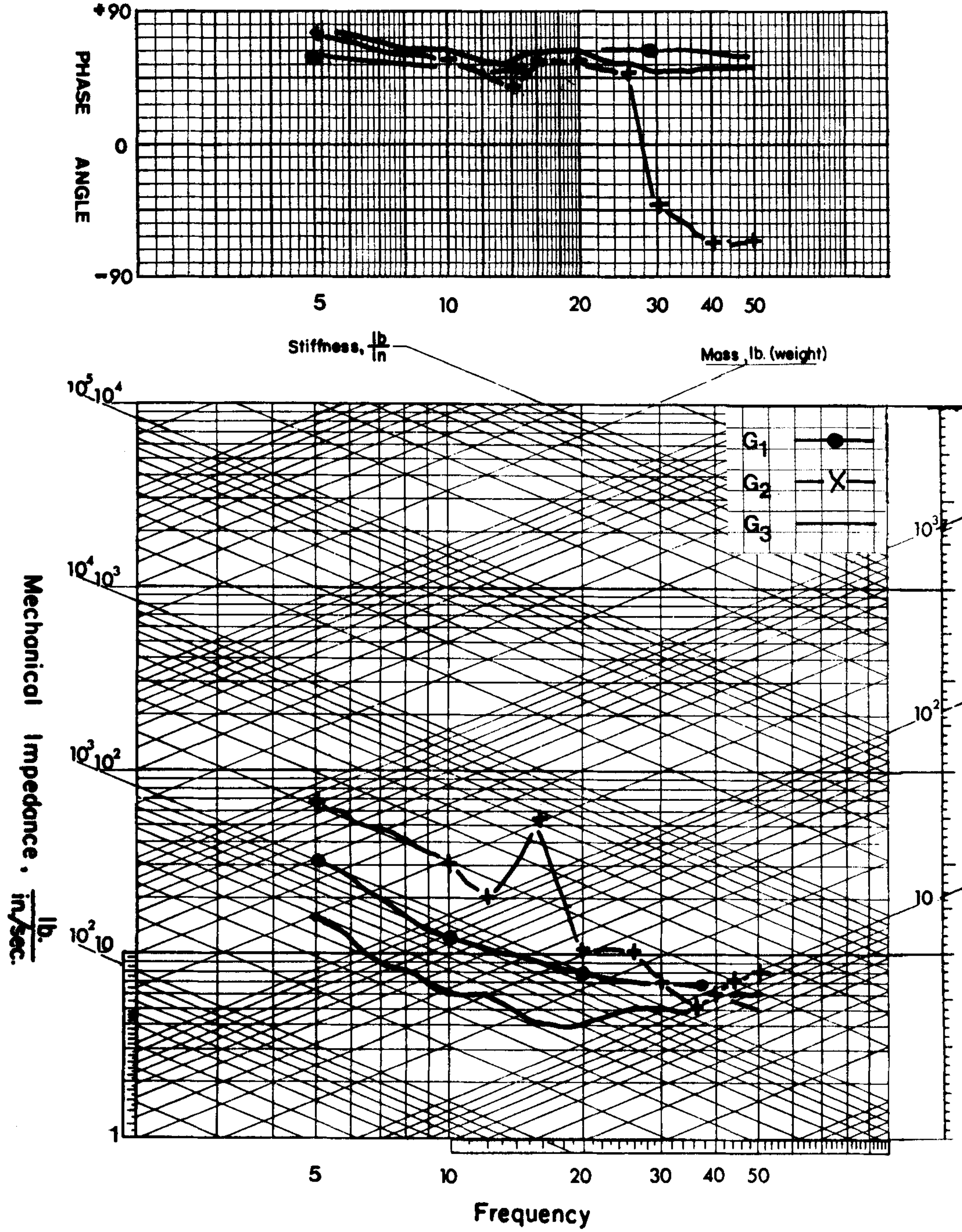


Figure 66. L_1—S / G_1—G_2—G_3 / P_1

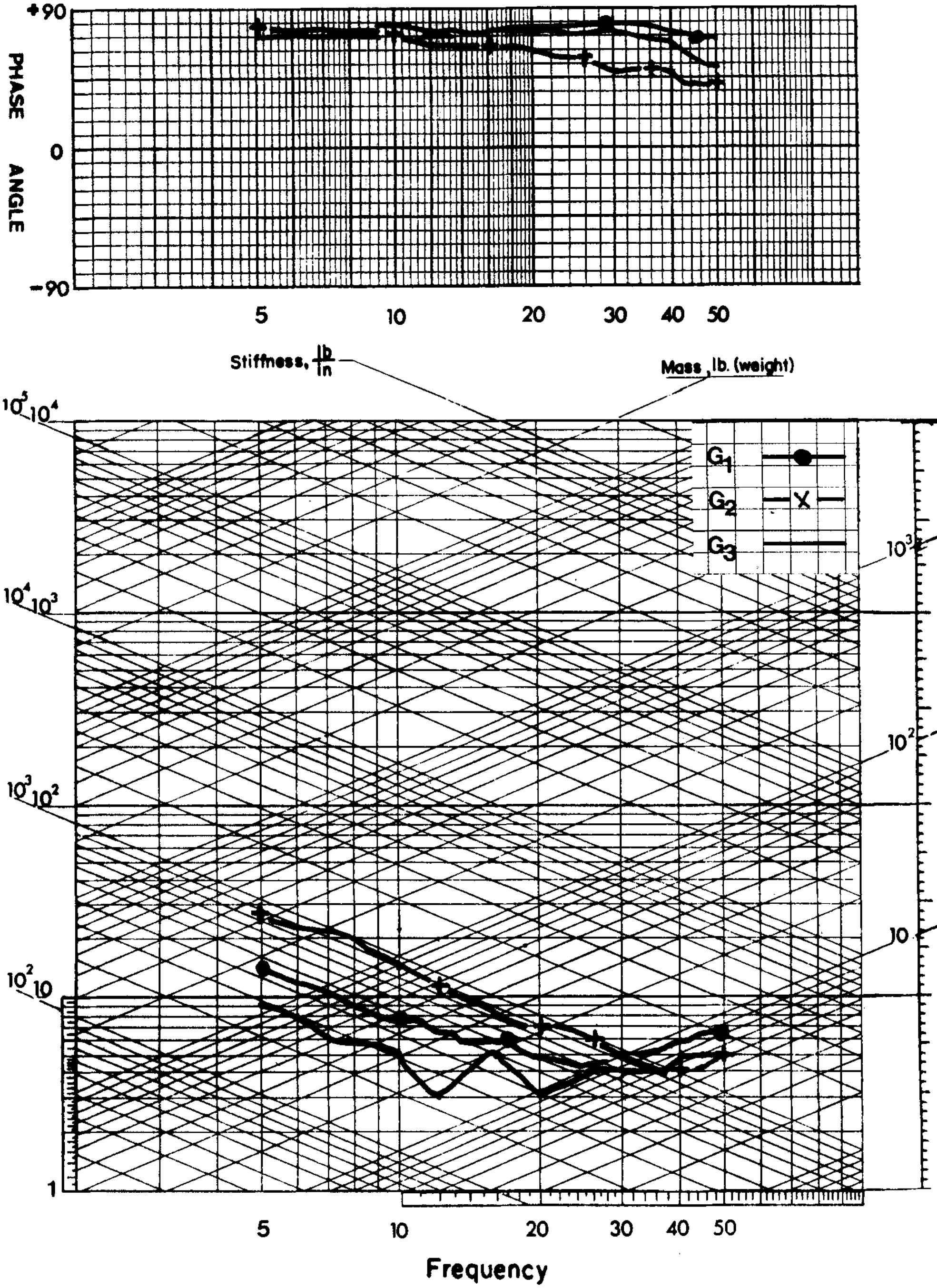


Figure 67. L_1—S / G_1—G_2—G_3 / P_2

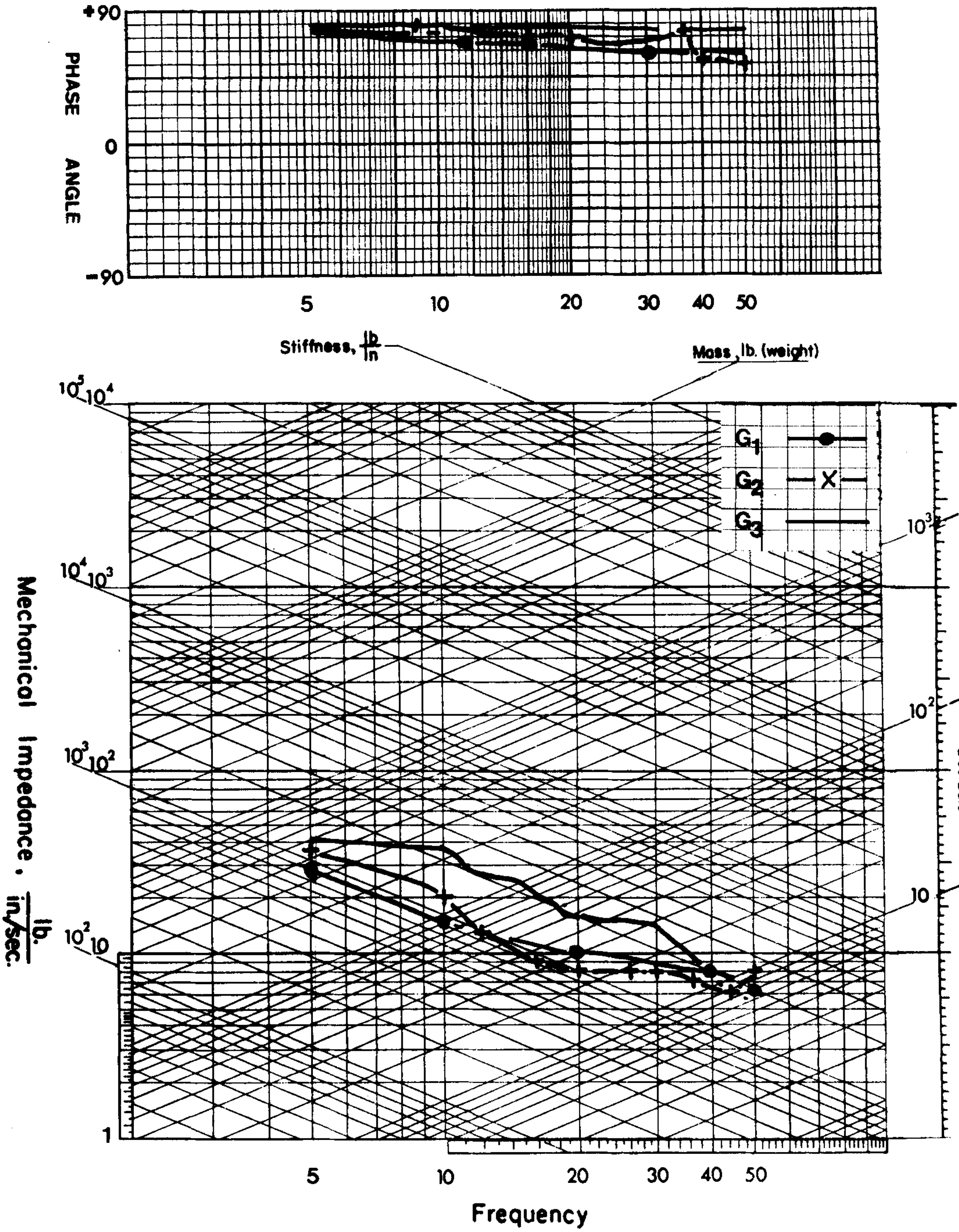


Figure 68. L_1—S / G_1—G_2—G_3 /P_3

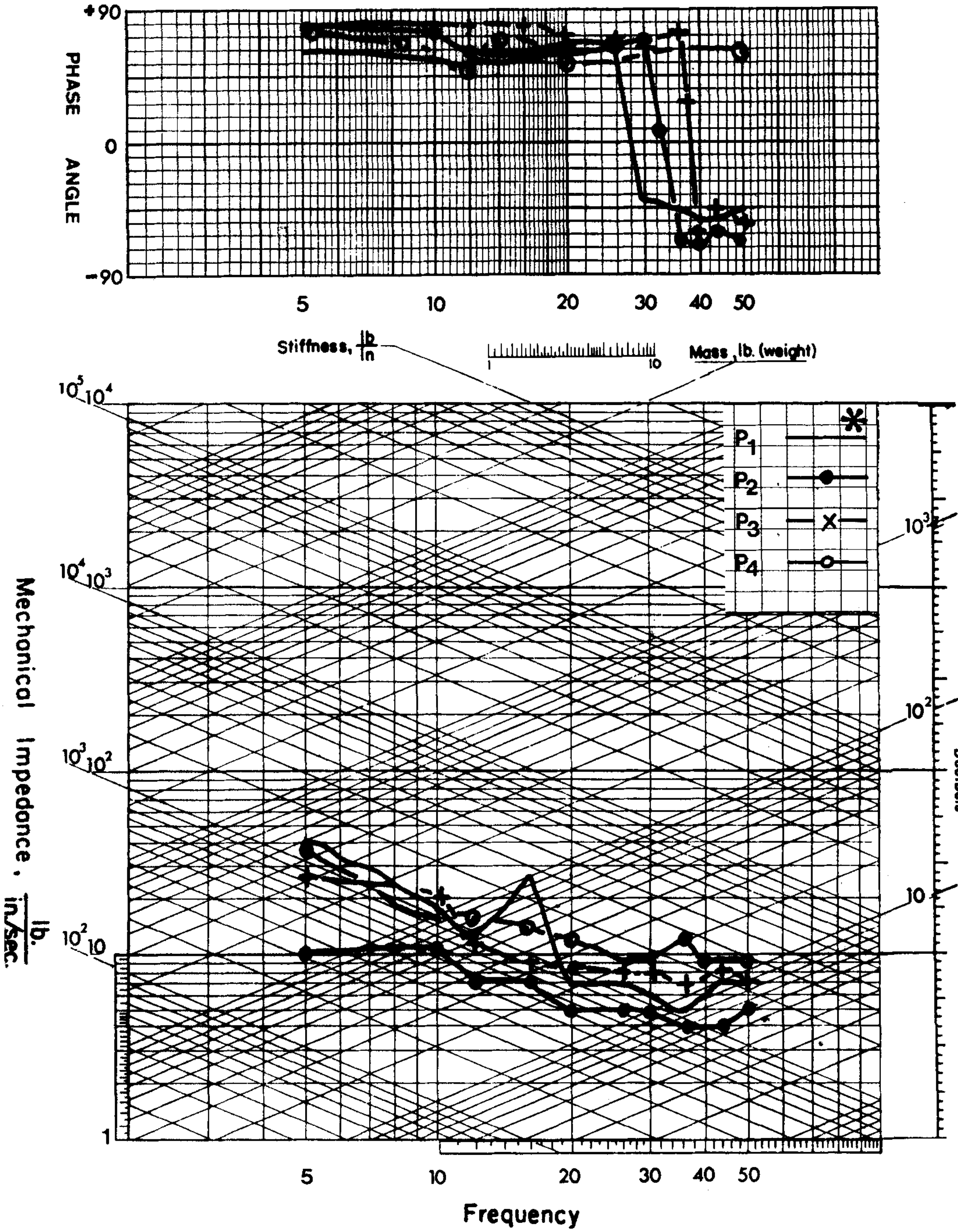


Figure 69. L_1—S / P_1—P_2—P_3—P_4

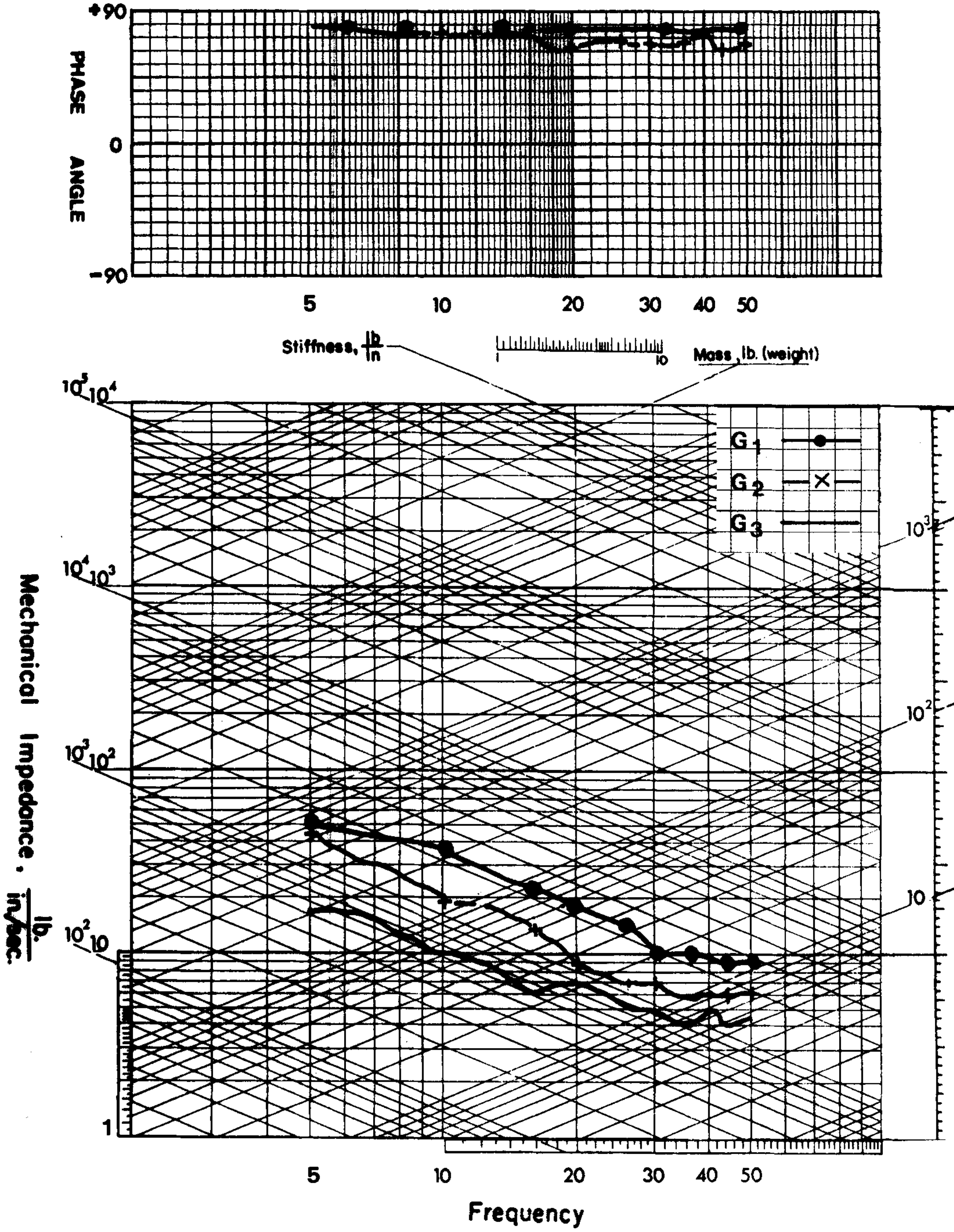


Figure 70. L_1—L_2—L_3 / G_1—G_2—G_3 / P_1

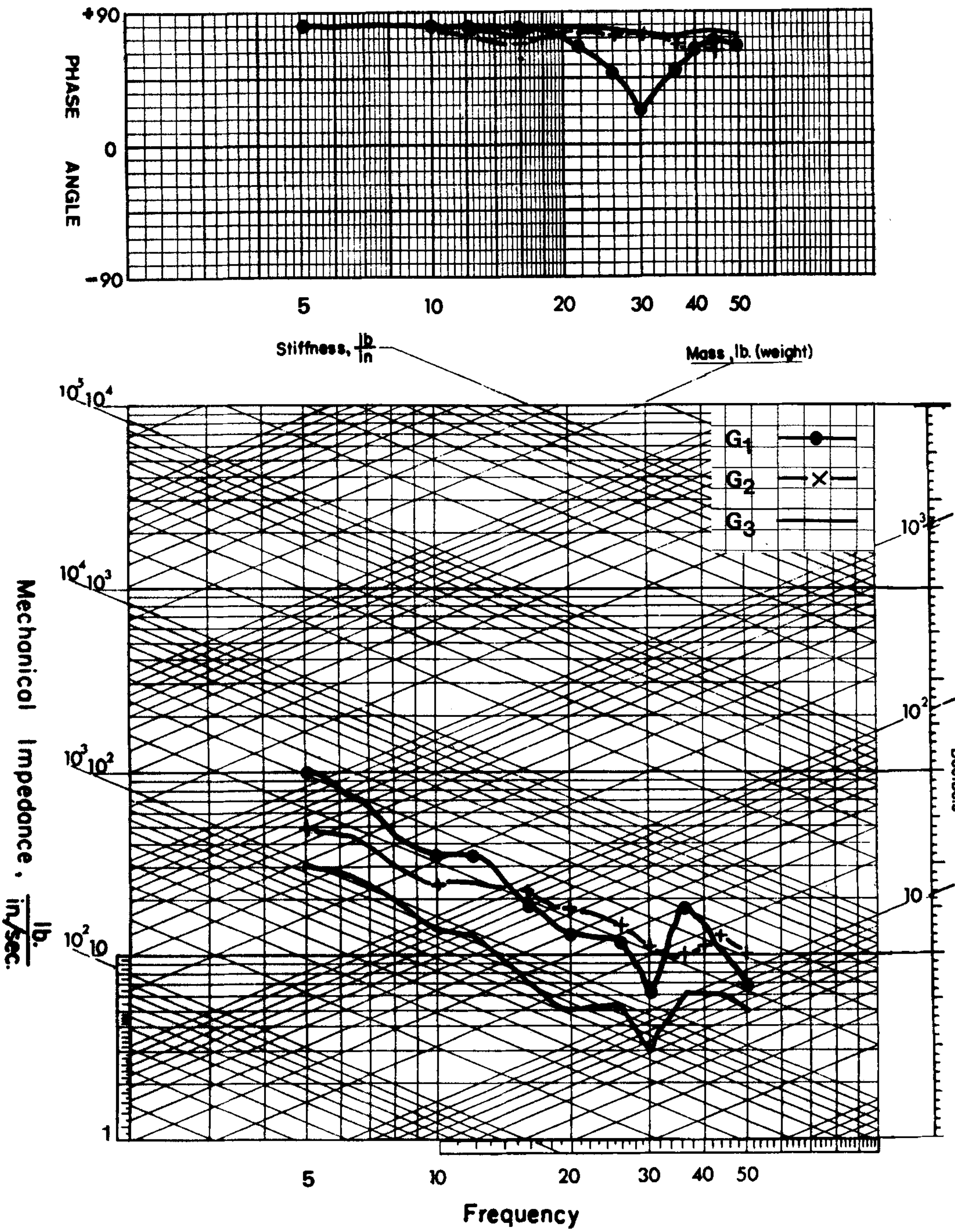


Figure 71. L_1—L_2—L_3 / G_1—G_2—G_3 / P_2

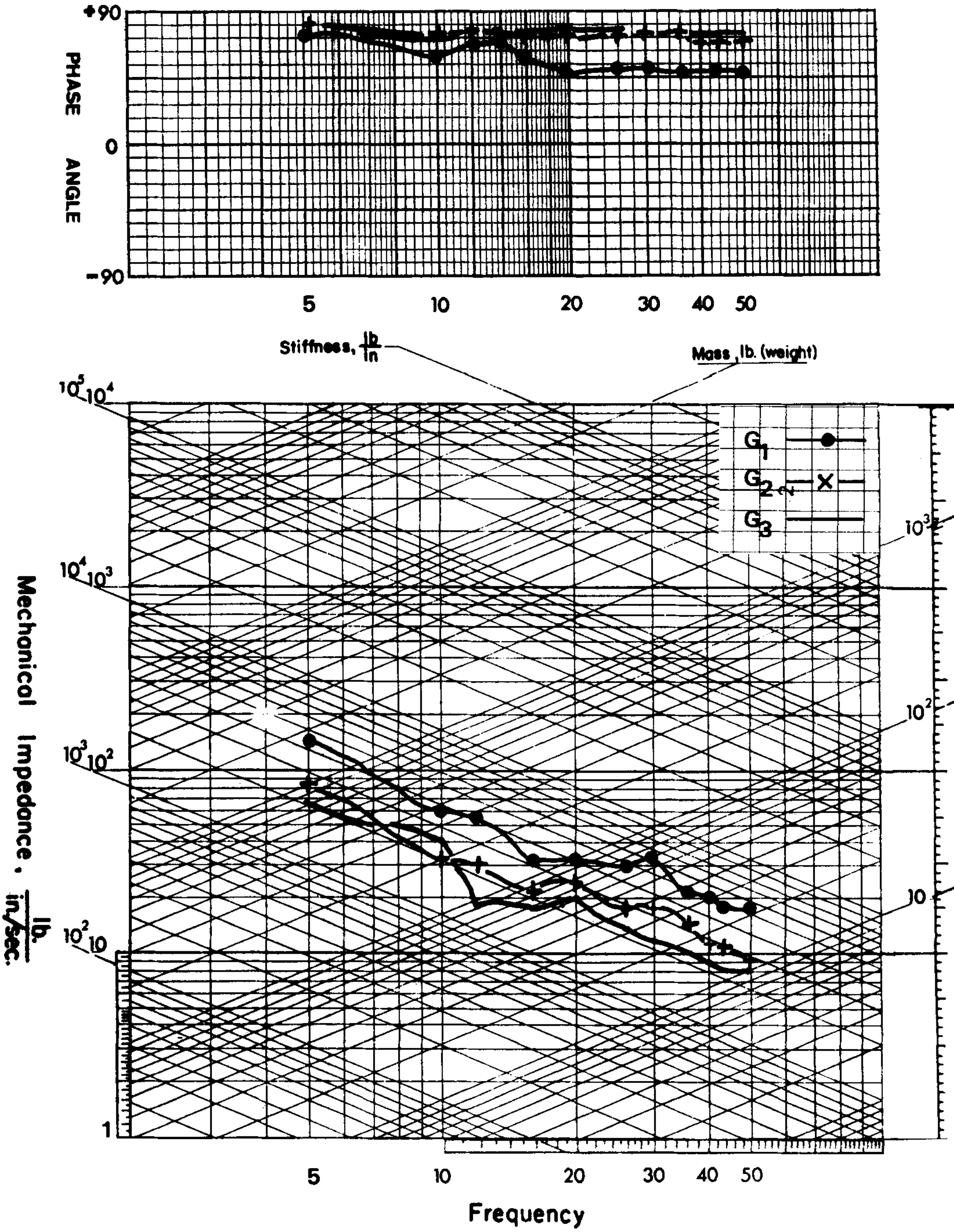


Figure 72. L_1—L_2—L_3 / G_1—G_2—G_3 / P_3

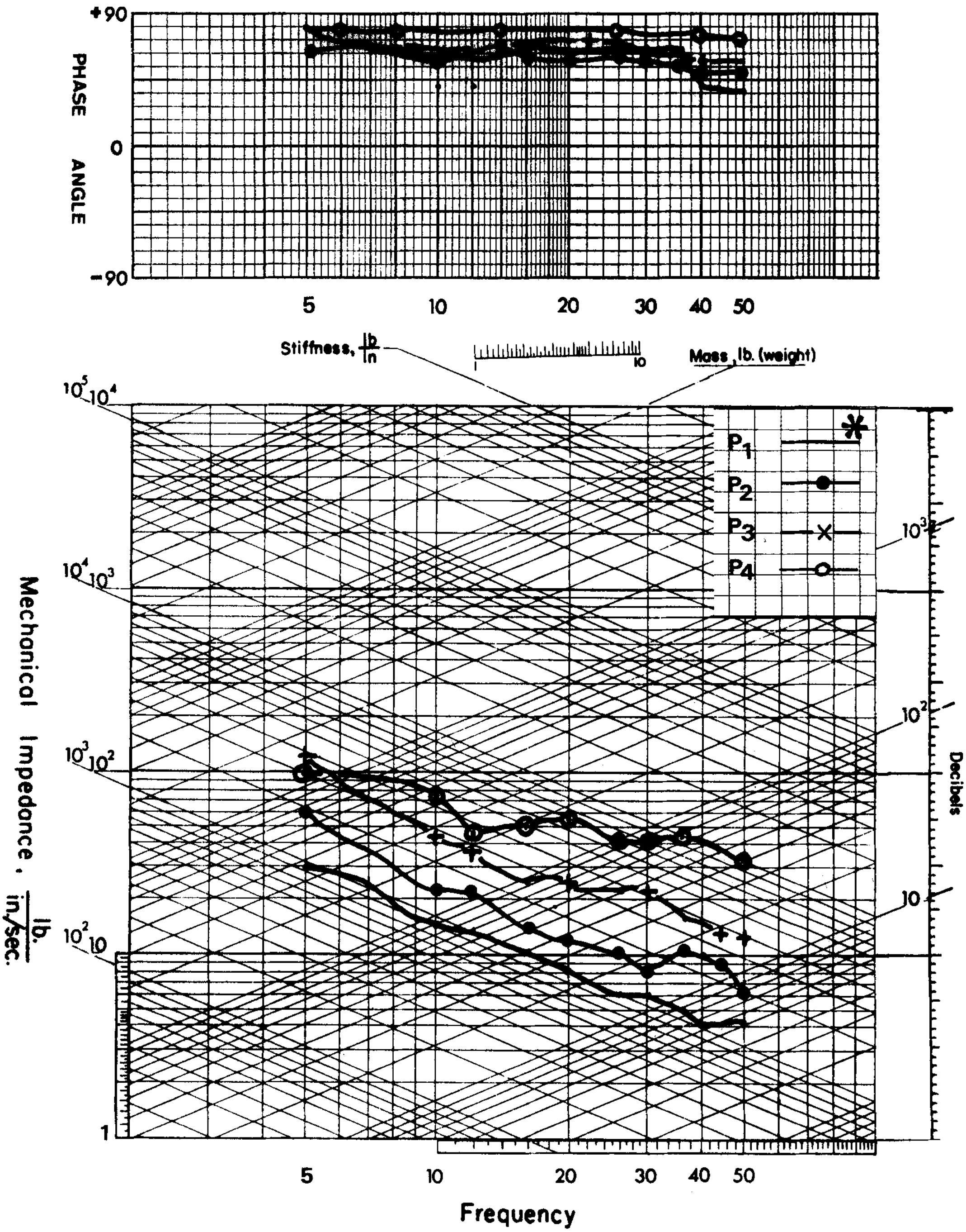


Figure 73. L_1—L_2—L_3 P_1—P_2—P_3—P_4

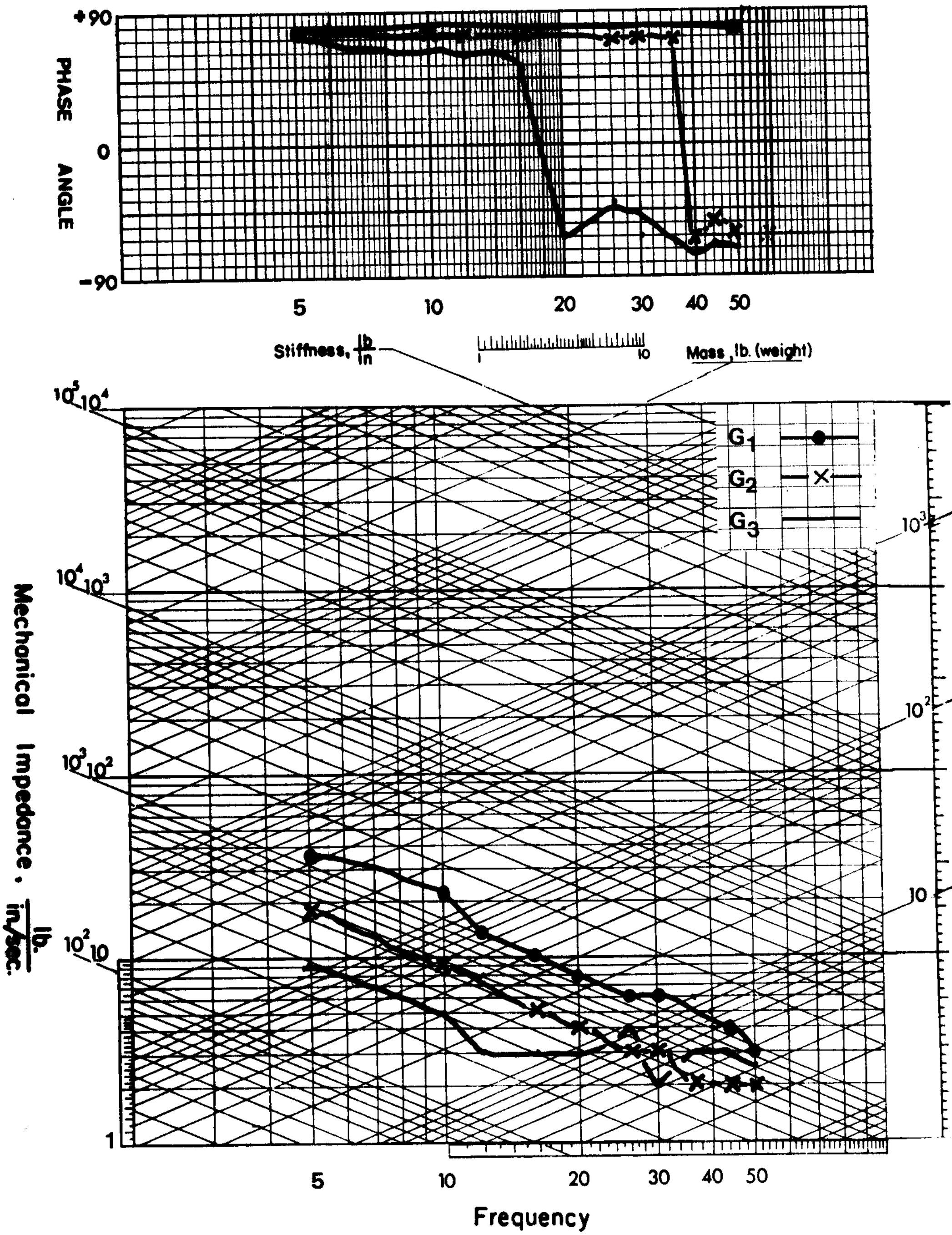


Figure 74. L_4—L_5—S / G_1—G_2—G_3 / P_1

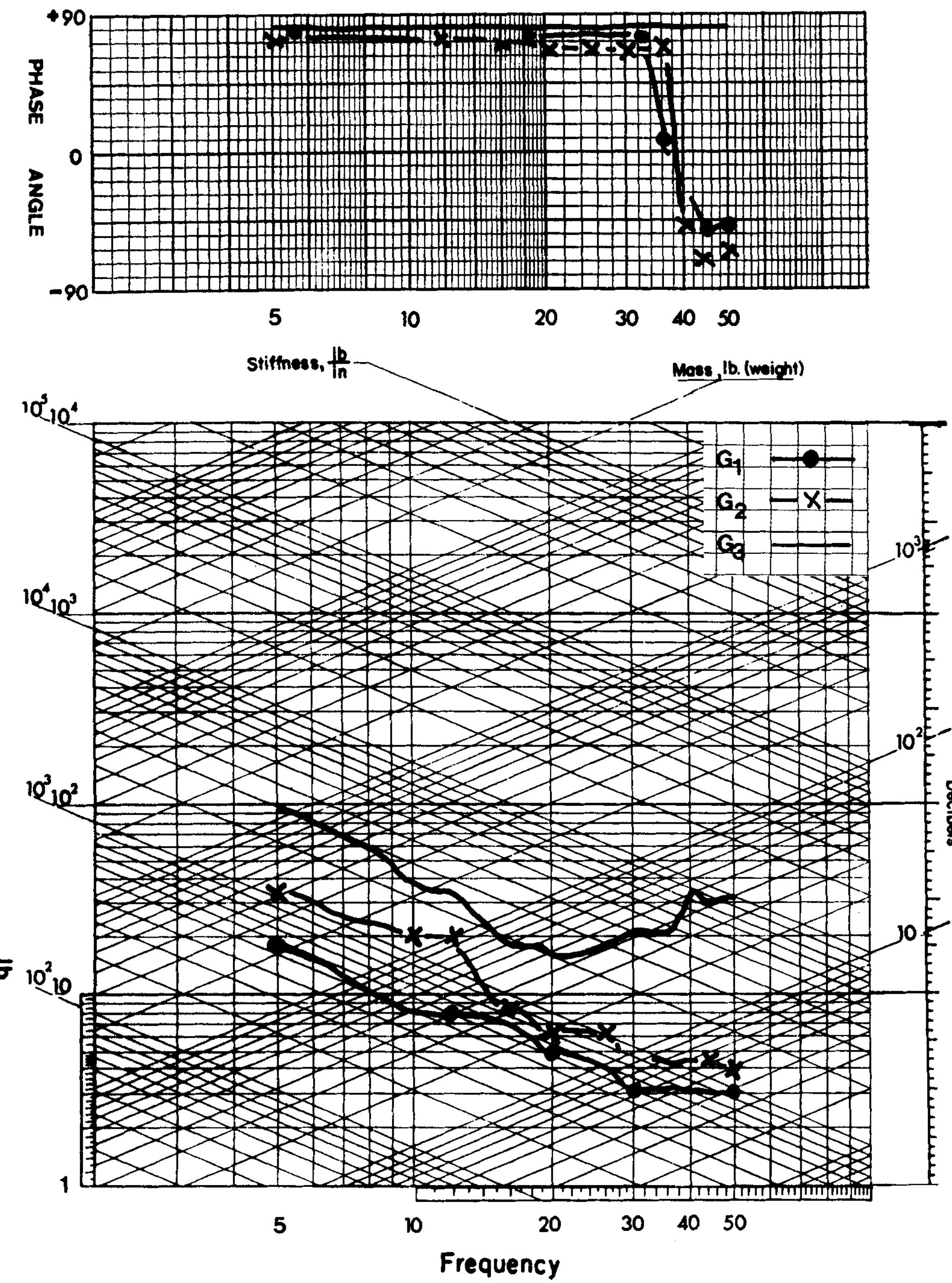


Figure 75. L_4—L_5—S / G_1—G_2—G_3 / P_2

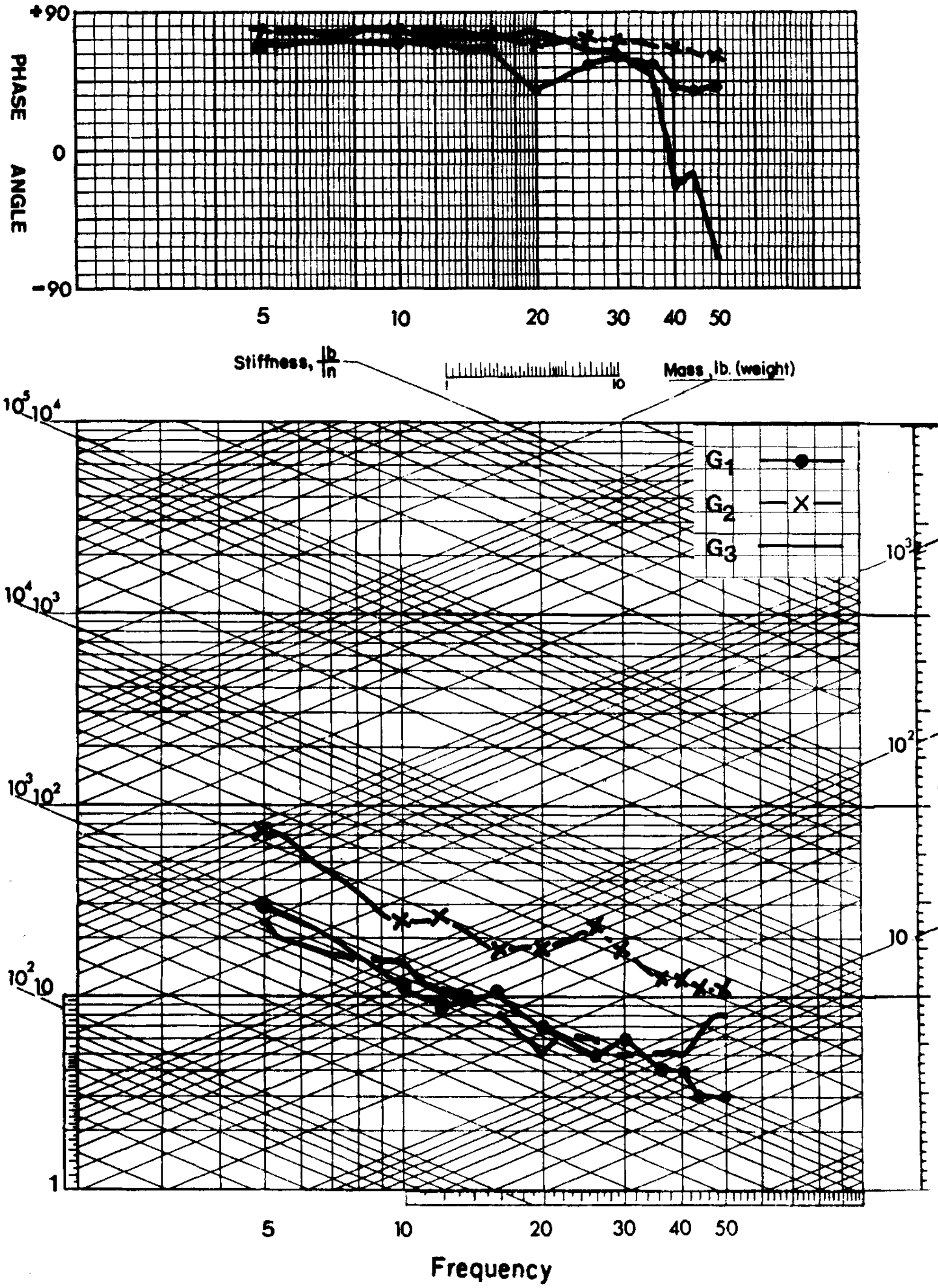


Figure 76. L_4—L_5—S / G_1—G_2—G_3 / P_3

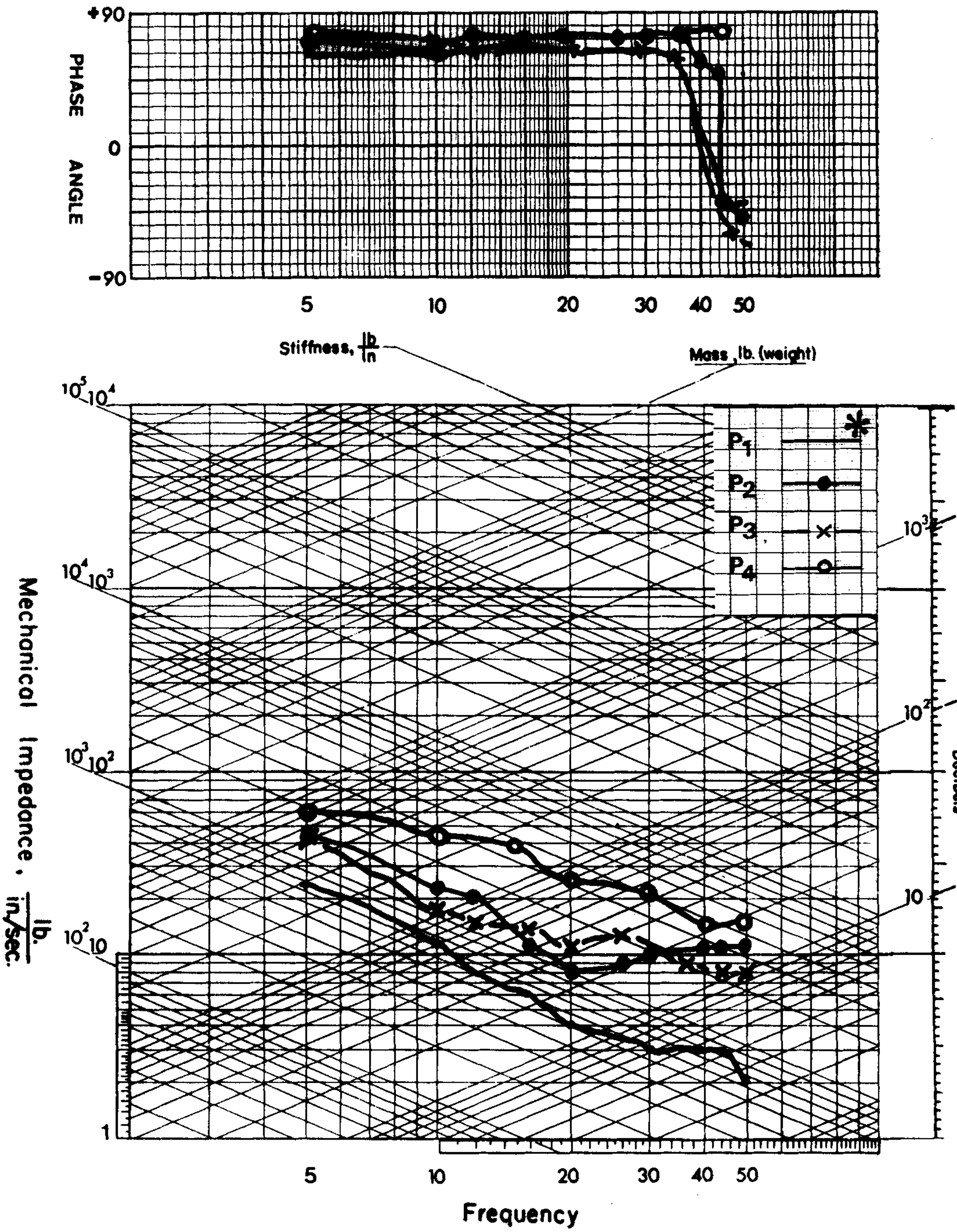


Figure 77. L_4—L_5—S / P_1—P_2—P_3—P_4

8. Summary

The present investigation deals with the application of a finite amplitude, sinusoidally varying longitudinal load, superimposed on a compression bias to the isolated segmented human vertebral column. To carry out this investigation a new experimental tool was designed, fabricated and made operational.

The method of analysis selected was mechanical driving point impedance in order to ascertain the complex frequency response function and its phase relationship with regard to various vertebral units. The experimental investigation was conducted in the frequency range from 5—50 Hz and was undertaken to characterize the behavioral nature of the various vertebral elements in the human spinal column under dynamic loading conditions and preload conditions. The overall results are based on a series of 90 cadaver subjects whose ages ranged from 17 to 90 years.

Preliminary repetitive impedance tests were carried out on single vertebral units and the following observations were made. When a single impedance test was conducted and its results contrasted to the second impedance test, the oscillograph tracings were noted to deviate. Comparative examination of oscillograph recordings between the first, second, third, and fourth and any subsequent tests showed further irregularities in the waveform patterns, but additionally intimated the behavior may equilibrate. The oscillograph tracings also indicated that an acceleration of transient actions could occur under dynamic loading, which appeared to be noticeably greater at certain frequencies of excitation and less at others.

With the highly limited supporting facts available, any decisions concerning the definition and manner of these tests were difficult. Therefore, it was decided that investigations ought to be conducted with special attention given to the following questions.

1. Does there take place a loss in axial height of a vertebral specimen as a function of constant compressive load, and if so, is its magnitude age-related and how long a period of time is necessary to achieve a new stability?

2. In what way does the response of the unconditioned vertebral specimen compare to the conditioned vertebral unit with and without its posterior arch?

3. How does the impedance response of any particular intervertebral disk vary as a function of constant peak to peak force over the length of the thoracic and lumbar column?

4. Do the creep magnitude and pattern deviate with respect to vertebral level, and if so, in what manner?

5. Does the impedance response vary as a function of the applied compression bias? If so, what is its pattern, and is it age-related when contrasted to adjoining vertebral units?

Five experiments were initiated to shed further light on these questions and identified respectively as,

Experiment A — The response of the vertebral unit to constant compressive loading

Experiment B — The dynamic response of the unconditioned and conditioned vertebral segment.

Experiment C — The impedance response of the unconditioned vertebral disk.

Experiment D — Creep response at the various vertebral levels.

Experiment E — The effect of the posterior arch on the anterior spinal column.

With regard to the first experiment, a humidified creep chamber was designed and made operational. It was discovered that the vertebral unit under constant load clearly exhibited creep tendencies. The magnitude and duration of the creep was found to be a function of the vertebral unit, its level and length, the age of the material and the external load. However, it was discovered by trial and error that a critical load magnitude existed prior to the initiation of creep which was also found to vary as a function of the vertebral level. All factors considered, the creep rheological mechanism dominated throughout all experiments.

In Experiment B, the nature of the creep was corroborated with the impedance response magnitude. The results primarily showed that following a period of conditioning, the mechanodynamics of the vertebral unit were altered when contrasted to its raw defrosted state. Furthermore, the conditioned vertebral unit could exhibit a hardening or softening impedance response with respect to the creep. With increasing vertebral unit deflection, the structural deformations increased and at each increment of deflection the impedance mode shifted. The answer to the second question was as follows: upon contrasting the raw and conditioned vertebral unit, a clear upward shift was recorded in the impedance mode in the former case, while its shape remained more or less constant for the latter case. When the posterior arches were removed the mode shape for the individual units deviated even more. These results, combined with those of the previous experiment, gave support to the idea that there must exist a normal domain of operation with regard to vibration amplitude and dynamic loading respective to the load carried at each intervertebral disk. Evidence was provided to substantiate the postulation that the thoracic-lumbar mortice could lock following a period of static conditioning. It was also brought to light, that if the superior and inferior articular facet joint planes of T_{11} were contrasted, a marked structural discontinuity may be found to exist and its dynamic response must be delineated in relationship to the mortice and the superior vertebral unit.

In Experiment C the relative impedance response of the intervertebral disk was investigated. It was learned that the disks at various vertebral levels showed individual differences in their physical traits and modes of response under constant load parameters and dynamic inputs.

In the following experiment (D) the creep conditioning chamber was modified because experiments A, B and C gave rise to the thought that perhaps an axial compressive load subjected to the isolated vertebral unit may also induce cross-coupling effects between a torsional and bending mode. The pattern and magnitude of deformation was felt to be primarily governed by hard tissue structures which could provide a physical stop, the configurational intricacy of the intervertebral disk (annulus fibrosus) and its physical properties. Since the physical nature of the latter element was found to vary as a function of spinal level, there was no reason to believe that form did not follow function and therefore, the actions of the individual vertebral bodies under similar loading conditions should also vary throughout the length of the column. Experimental observations using a modified creep chamber provided some insight into the nature and configurational rearrangements of the vertebral unit with regard to constant load application. It was learned that rotational creep could be greater than axial creep. Relative vertebral body rotations were found to occur above the level of the conditioned thoracic-lumbar mortice.

Utilizing the results of these experiments, a mode of action was theorized with regard to the early symmetry and the later asymmetry and decay of the column. The structural mechanisms and system dynamics to provide a rational fundamental basis for the assumption were delineated.

Experiment E was performed principally to delineate the effect of the posterior arches on the impedance function. Raw vertebral units were tested, their posterior arches were stripped and the anterior column was again impedance tested. In all cases the impedance response plot remained about the same shape but was found to shift to a lower range of impedance values. In addition, upon removal of the posterior arches a stiffening or softening action could be initiated, a factor dependent on the unit deflection to the applied constant load and specimen age. The role of the posterior arches was illuminated regarding their load-bearing, load-transmission peculiarities. The results seemed to connote that concurrently with increasing age and whenever interacting deflections were too great within or over the normal operational domain of a particular vertebral level, the articular facet joints engaged their mutual cartilaginous surfaces and reacted as an override snubber by interlocking to control and restrict the relative envelope of performance attributed to a particular vertebral level.

Limited answers were provided for the initial questions asked. Based on these findings, a discussion was put forth on why the vertebral column is a prestressed structure. The acceleration of creep under a compression bias and the implications of these findings with regard to the present experiment were discussed.

The final experiment was concerned with repetitively loading the vertebral unit at four different magnitudes of compression bias. This provided a family of impedance curves for the various vertebral levels, age groups and preload conditions. It was also found that the mean static bias caused a large change in the damping and enhanced vertebral unit stability and the principal mode of loading.

An experimental tool has been developed which provides further understanding and new insight into the dynamic structural interactions of the excised human vertebral column.

Although the data presented here are far from conclusive, the results to date have been extremely promising. It is difficult to draw upon any previous studies with regard to creep, relaxation, conditioning, prestressing, preloading, warping and dynamic testing of the human spinal column, for no similar tests are reported in any of the previous literature data or otherwise known to this writer.

9. References

Agate, I., Druett, H.: A study of portable vibrating tools in relation to clinical effects which they produce. *Brit. J. Indust. Med., 4:* 141, 1947.

Åkerblom, B.: Standing and sitting posture with special reference to the construction of Chairs. Stockholm, 1948.

Andrae, R.: Ueber Knorpelknötchen im Hinteren Ende des Wirbelbandscheiben im Bereich des Spinalkanals. *Beitr. z. Path. Anat. u. z. Allg. Path. 82:* 464, 1929.

Armstrong, J.: Lumbar disc lesions. E & S Livingstone Ltd, Edinburgh, 1958.

Arnold, J.: Morphology of bone in osteoporosis. Progress report AM 09379, PHS—NIH—IAMD., May, 1965.

Axelsson, S., Analysis of vibrations in power saws. *Royal College of Forestry, Stockholm, No. 59,* 1968.

Badgley, C. E.: The articular facets in relation to low-back pain and sciatic radiation. *J. Bone & Joint Surg., 23:* 481, 1941.

Barany, E.: A contribution to the physiology of bone conduction. *Acta Oto-Laryngolo., Suppl. No. 26,* 1938.

Bardeen, C.: The development of the thoracic vertebrae in man. *Amer.. J. of Anat.., 4:* 163, 1905.

Barr, J.: Sciatica caused by intervertebral-disc lesions. *J. Bone & Joint Surg., 19:* 323, 1937.

Bartelink, D.: The role of the abdominal pressure in relieving the pressure on the lumbar intervertebral discs. *J. Bone & Joint Surg., 39-B:* 718, 1957.

Beadle, O. A.: The intervertebral discs. Observations on their normal and morbid anatomy in relation to certain spinal deformities. Medical Research council. Special report series. London., No. 161, 1931.

Bell, G. H., Dunbar, O., Beck, J. S., Gibb, J.: Variations in strength of vertebrae with age and their relation to osteoporosis. *Calcified Tissue Research., 1:* 75, 1967.

Bick, E. M., Copel, J. W.: Longitudinal growth of the human vertebra. A contribution to human osteogeny. *J. Bone & Joint Surg., 32-A:* 803, Oct. 1950.

Bouche, R. R.: Instruments and methods for measuring mechanical impedance. TP 203, Endevco Corp., 801 Arroyo Parkway, Pasadena, California., 91109.

Bradford, F. K., Spurling, R. G.: The intervertebral disc. C. Thomas, Springfield, 1921.

Brailsford, J. F.: Deformities of the lumbosacral region of the spine. *Brit. J. Surg., 116:* 562, 1928—29.

Brailsford, J. F.: Radiology of the bones and joints. William Wood and Co., Baltimore. 1934.

Braus, H.: Anatomie des Menschen, Band I: Bewegungsapparat. Springer, Berlin, 1921.

Brown, L.: Conservative treatment of backache. *J. Bone & Joint Surg., 14:* 157, 1932.

Brown, T., Hansen, R. J., and Yorra, A. J.: Some mechanical tests on the lumbosacral spine with particular reference to the intervertebral discs. *J. Bone & Surg., 39-A(5):* 1135, 1957.

Cailliet, R.: Low back pain syndrome. F. A. Davis Company, Philadelphia, Pa, 1965.

Calve, J., Galland, M.: The intervertebral nucleus pulposus; its anatomy, its physiology, its pathology. *J. Bone & Joint Surg., 12:* 555, 1930.

Campbell, J., Jurist, J.: Mechanical Impedance of the femur: *J. of Biomech. 4:* 319, 1971.

Capener, N.: Sciatica. *Ann. Rheum. Diseases., 4:* 29, 1944.

Chalmers, J., Weaver, J.: Cancellous bone: Its strength and changes with aging and an evaluation of some methods for measuring its mineral content. I. Age changes in cancellous bone. *J. Bone & Joint Surg., 48-A:* 289, 1966.

Charnley, J.: The imbibition of fluid as a cause of herniation of the nucleus pulposus. *Lancet 1:* 124, 1952.

Clark, W., Lange, K., and Coermann, R.: Deformation of the human body due to unidirectional forced sinusoidal vibration. *Human Factors.,* Vol. 4, 1962.

Cleland, C.: On fibro-plates and intervertebral discs. *J. Anat. Physiol., 24:* 373, 1889.

Cloward, R., Buzaid, L.: Discography. *Am. J. Roentgenol., 68:* 552, 1952.

Coermann, R., Ziegenruecker, G., Wittwer, A., and von Gierke, H.: The passive dynamic mechanical properties of the human thorax-abdomen system and of the whole body system. *Aerospace Med., 31:* 1960.

Coermann, R.: The mechanical impedance of the human body in sitting and standing position at low frequencies. ASD-TR-61-492, Aeronautical Systems Division., WPAFB, Ohio, 1961.

Coermann, R.: Comparison of the dynamic characteristics of dummies, animals and man, *in* Impact Acceleration Stress Proceedings, Brooks AFB, Texas, NAS-NRC-977, Washinton D.C. 1961.

Corliss, E., Koidan, W.: Mechanical impedance of the forehead and mastoid. JASA., 27: 1955.

Coventry, M., Ghormley, R., and Kernohan, J.: The intervertebral disc. Part I. Its microscopic anatomy and pathology. Anatomy, development and physiology. *J. Bone & Joint Surg., 27:* 105, 1945.

Coventry, M., Ghormley, R., and Kernohan, J.: The intervertebral disc. Part II. Changes in the intervertebral disc concomitant with age. *J. Bone & Joint Surg., 27:* 233, 1945.

Coventry, M., Ghormley, R., and Kernohan, J.: The intervertebral disc. Part III. Pathological changes in the intervertebral disc. *J. Bone & Joint Surg., 27:* 460, 1945.

Crocker, J. F., Higgins, L. S.: Phase IV—Investigation of strength of isolated vertebrae. Final Tech. Report Contract No. NASw-1313, Technol. Inc., San Antonio, Texas, 1966.

Dadson, R., Robinson, D., and Greig, R.: The mechanical impedance of the human mastoid process. *British J. of Appl. Physics.,* Vol. 5, 1954.

Danforth, M., Wilson, P.: The anatomy of the lumbosacral region in relation to sciatic pain. *J. Bone & Joint Surg., 7:* 109, 1925.

Davis, P. R.: The thoraco-lumbar mortice joint. *J. Anat., 89:* 370, 1955.

Davis, P. R.: Studies on the functional anatomy of the human vertebral column. Thesis, London University, 1958.

Davis, P. R.: The medial inclination of the human thoracic intervertebral articular facets. *J. Anat., 93:* 68, 1959.

Decoulx, P., Rieunau, G.: Les fractures du rachis dorso-lombaire sans troubles nerveux. *Rev. Chir. Orthop., 44:* 254, 1958.

De Puky, P.: The physiological oscillations of the length of the body. *Acta Orthop. Scand., 6:* 338, 1935.

Dieckman, D.: A study of the influence of vibration on man. *Ergonomics, 1:* 345, 1958.

Donohue, W.: Pathology of the intervertebral disc. *Am. J. M. Sc., 198:* 419, 1939.

Eckert, C., Decker, A.: Pathological studies of intervertebral discs. *J. Bone & Joint Surg., 29:* 447, 1947.

Edwards, R. G., Lange, K. O.: A mechanical impedence investigation of human response to vibration. AMRL-TR-64-91, Aerospace Medical Research Labs., Wright-Patterson AFB, Ohio, 1964.
Epstein, B.: The spine — A radiological text and atlas. Lea & Febiger, Philadelphia, 1969.
Estes, W. L.: Causes and occurrence of functional scoliosis in college men. *J. A. M. A., 75:* 1411, 1920.
Evans, F. H., Lissner, H. R.: Biomechanical studies on the lumbar spine and pelvis. *J. Bone & Joint Surg.,* 41-A: 278, 1959.
Evans, G., Lissner, H.: Studies on the energy absorbing capacity of human lumbar intervertebral discs. 7th Stapp Car Crash Conference Proceedings., 386, 1965.
Farfan, H., and Sullivan, J.: The relation of facet orientation to intervertebral disc failure. *Canadian J. Surg., 10:* 179, 1967.
Farfan, H.: Effects of torsion on the intervertebral joints. *Canadian J. Surg., 12:* 336, 1969.
Farfan, H., Huberdeau, R. and Dubow, H.: Lumbar intervertebral disc degeneration. *J. Bone & Joint Surg., 54-A:* 492, 1972.
Farkas, A.: Physiological scoliosis. *J. Bone & Joint Surg., 39:* 607, 1941.
Farmer, H. L., Accessory articular processes in the lumbar spine. A review of 40 cases. *Amer. J. Roentgenol., 36:* 763, 1936.
Fick. R.: Handbuch der Anatomie und Mechanik der Gelenke, unter Berücksichtigung der bewegenden Muskeln. Erster Teil: Anatomie der Gelenke. In von Bardeleben's Handbuch der Anatomie des Menschen. Gustav Fischer, Jena, 1904.
Firestone, F.: The mobility method of computing the vibration of linear mechanical and acoustical systems, mechanical and electrical analogies. *J. Appl. Phys., 9:* 373, 1938.
Firestone, F.: The mobility and classical analyses. American Institute of Physics Handbook, McGraw-Hill, New York, 1957.
Franke, E.: Mechanical impedance measurements of the human body surface. AF Tech. Report No. 6469, AMRL-WPAFB, 1951.
Franke, E.: The mechanics of vibration in the human body. *Shock and Vibration Bull., Suppl.* No. 22, Sec. of Defense, Washington, D.C., 1955.
Franke, E.: The response of the human skull to mechanical vibrations. *JASA, 28:* 16, 1956.
Francis, C.: Variations in the articular facets of the cervical vertebrae. *Anat. Rec., 122:* 589, 1955.
French, A.: Vibrations and waves. Thomas Nelson and Sons, London, 1971.
Friedman, M., Fisher, F., and Van Demark, R.: Lumbosacral roentgenograms of one hundred soldiers. *Am. J. Roent., 55:* 292, 1946.
Frykholm, R.: Lower cervical vertebrae and intervertebral discs. Surgical anatomy and pathology. *Acta Chir. Scand., 101:* 345, 1951.
Furst, W.: The force to crush vertebrae. *Psychiat. Quart., 14:* 397, 1940.
Galante, J.: Tensile properties of the human lumbar annulus fibrosus. *Acta Orthop. Scand. Suppl.,* 100, 1967.
Galante, J., Rostocker, W., and Ray, R.: Physical properties of trabecular bone. *Cal. Tissue Res., 513:* 236, 1970.
Geertz, A.: Limits and special problems in the use of seat catapults. AAF Aero Medical Center ATI No. 56946, AMRL-WPAFB, 1946.
Ghormley, R., Kirklin, B.: The oblique view for demonstration of the articular facets in lumbosacral backache and sciatic pain. *Am. J. Roent., 31:* 173, 1934.
Gianturco, C.: Motion of lower lumbar vertebrae in normal individuals and in patients with low back pain. *Am. J. Roent., 52:* 261, 1944.

Göcke, C.: Das Verhalten spongiösen Knochens im Druck- und Schlagversuch. *Verh. Dtsh Orhop. Ges., 20:* 114, 1926.
Göcke, C.: Beitrage zur Druckfestigkeit des Spongiösen Knochens. *Beitr. Klin. Chir., 143:* 539, 1928.
Göcke, C.: Traumatische Wirbelumformung im Versuch. *Hefte Unfallheilk. H. 8:* 136, 1931.
Göcke, C.: Das Verhalten der Bandscheiben bei Wirbelverletzungen. *Arch. Orthop. Unfall-Chir., 31:* 42, 1932.
Goldman, D. R., von Gierke, H.: The effects of shock and vibration on man. Lecture and review series No. 60—3, Naval Medical Research Institute, Bethesda, Md., 1960.
Gordon, E. E.: Natural history of the intervertebral disk. *Archives of Physical Medicine and Rehabilitation., 42:* 750, 1961.
Güntz, E.: Die Erkrankungen der Zwischenwirbelgelenke. *Arch. Orthop. Unfall-Chir. 34:* 333, 1934.
Gurdjian, E., Hodgson, V., and Thomas, L.: Studies on mechanical impedance of the human skull: Preliminary report. *J. Biomechanics,* 3, 1970.
Hadley, L. A.: Subluxation of the apophyseal articulations with bony impingement as a cause of back pain. *Amer. J. Roentgenol. 33:* 209, 1935.
Hagelstam, L.: Retroposition of lumbar vertebrae. *Acta Chir. Scand.,* Suppl. 143, 1949.
Hallen, A.: The collagen and ground substance of human intervertebral disc at different ages. *Acta Chem. Scand., 16:* 705, 1962.
Hanraets, P.: The degenerative back. Elsevier Publishing Co., Amsterdam, 1959.
Happey, F., MacRae, T. and Naylor, A.: X-ray crystallographic investigation on the change with age in the structure of the human intervertebral disc. In Nature and structure of collagen. Ed. Randall, J. T., Butterworth's Scientific Publications, London, 1953.
Happey, F., Horton, W., MacRae, T., and Naylor, A.: The human intervertebral disc. *Nature, 175:* 1032, 1955.
Hardy, W. G., Lissner, H. R., Webster, J. E., and Gurdjian, E.: Repeated loading tests of the lumbar spine. *Surgical Forum, 9:* 690, 1958.
Harris, R., MacNab, I.: Structural changes in the lumbar intervertebral discs. Their relationship to low back pain and sciatica. *J. Bone & Joint Surg.,* 36-B: 304, 1954.
Hetenyi, M. (Editor): Handbook of experimental stress analysis. John Wiley and Sons Inc., London, 1950.
Hirsch, C.: Studies on the mechanism of low back pain. *Acta Orthop. Scand., 20:* 261, 1951.
Hirsch, C., Paulson, S., Sylvén, B., and Snellman, O.: Biophysical and Physiological investigation on cartilage and other mesenchymal tissues. IV. Characteristics of human nuclei pulposi during ageing. *Acta Orthop. Scand., 22:* 175, 1952.
Hirsch, C., Schajowicz, F.: Studies on structural changes in the lumbar annulus fibrosus. *Acta Orthop. Scand., 22:* 184, 1952.
Hirsch, C., Nachemson, A.: New observations on the mechanical behavior of lumbar discs. *Acta Orthop. Scand., 23:* 254, 1954.
Hirsch, C.: The reaction of intervertebral discs to compression forces. *J. Bone & Joint Surg., 37-A:* 1188, 1955.
Hirsch, C.: Efficiency of surgery in low-back discorders. Pathoanatomical, experimental, and clinical studies. *J. Bone & Joint surg., 47-A:* 991, 1965.
Hixson, E., Wittenborn, A.: The measurement of mechanical impedance and its applications. University of Texas, Defense Research Laboratory, 1957.
Hixson, E.: Mechanical impedance and mobility. Shock and Vibration Handbook, Mc Graw-Hill, New York, 1961.

Hodgson, V., Patrick, L.: Dynamic response for the human cadaver head compared to a simple mathematical model. The 11th Stapp Car Crash Conference, 1968.

Hodgson, V., Gurdjian, E., and Thomas, L.: The determination of response characteristics of the head with emphasis on mechanical impedance techniques. The 12th Stapp Car Crash Conference, 1970.

Horton, W.: Further observations on the elastic mechanism of the intervertebral disc. *J. Bone & Joint Surg., 40-B:* 552, 1958.

Humphrey, G.: A treatise on the human skeleton. London, 1858.

Ingelmark, B. E. and Ekholm, R.: Ueber die Kompressibilität der Intervertebral-scheiben. *Acta Soc. Med. Upsalien, N. S., 57:* 202, 1952.

Ingelmark, B. E.: De funktionellt anatomiska förhållandena i ryggraden med särskild hänsyn till dess småleder. Acta Univ. Gothoburg, 62, 1956.

Ingelmark, B. E.: Function of and pathological changes in the spinal joints. *Acta Anat.,* Suppl. 36, 1959.

Inman, V., Saunders, J. B.: Anatomic-physiological aspects of injuries to the intervertebral disc. *J. Bone & Joint Surg., 29:* 461, 1947.

Jefferson, G.: Discussion on spinal injuries. *Proc. R. Soc. Med., 21:* 625, 1927.

Joplin, R. J.: The intervertebral disc. Embryology, anatomy, physiology and pathology. *Surg., Gynec. & Obst., 61:* 591, 1935.

Kazarian, L., Boyd, D. and von Gierke, H.: The dynamic biomechanical nature of spinal fractures and articular facet derangement. U.S.A.F. AMRL-71-17, WPAFB, Ohio. U.S.A.

Keller, H. A., A clinical study of the mobility of the human spine, its extent and its clinical importance. *Arch. Surg., 8:* 627, 1924.

Keller, A.: Fundamentals for mechanical impedance analysis. TP No. M-2 Spectral Dynamics Corporation, San Diego, California.

Kellogg, L.: Impossibility of moving vertebrae. *J.A.M.A., 81:* 233, 1923.

Kelly, M.: Physical changes in the prolapsed disk. *Lancet, 2:* 584, 1958.

Keyes, D., Compere, E.: The normal and pathological physiology of the nucleus pulposus of the intervertebral disc. An anatomical, clinical and experimental study. *J. Bone & Joint Surg., 14:* 897, 1932.

King, V. P.: Elastic deformation characteristics of the spine. Proceedings of Workshop on Bioengineering Approaches to Problems of the Spine. NIH—Surgery Study Sections, 1970.

Kimberly, A. G.: Low back pain and sciatica. *Surg. Gynec. & Obstet., 65:* 195, 1937.

Kraft, L.: Facet synovial impingement. *Surg. Gynec. & Obst., 93:* 439, 1951.

Kuhns, J.: Developmental changes in the vertebral articular facets. *Radiology 25:* 498, 1935.

Lange, C.: Untersuchungen über Elasticitätsverhältnisse in den menschlichen Rückenwirbelin mit Bemerkungen über die Pathogenese der Deformitäten. *Ztschr. f. Orthop. Chir., 10:* 47, 1902.

Lewin, P.: The back and its disk syndromes. Lea & Febiger, Philadelphia, 1955.

MacConill, M. A.: The movements of bones and joints. 5. The significance of shape. *J. Bone & Joint Surg., 35-B:* 290, 1953.

Mac Nab, I.: Spondylolisthesis with an intact neural arch, the so called pseudospondylolisthesis. *J. Bone & Joint Surg., 32-B:* 325, 1950.

Magid, E., Coermann, R.: The reaction of the human body to extreme vibrations. *Proceedings of the Institute of Environmental Science,* National Meeting, 1960.

Manley, R.: The mechanical impedance of damped vibrating systems. *J. Roy. Aero. Soc.,* 1941.

Markolf, K., Steidel, R.: The dynamic characteristics of the human intervertebral joint. ASME. Paper No. 70-WA/BHF-6, 1970.

McGregor, H. M.: Impedance measurement techniques. *Environmental Qt.* 1967.

Meachim, G., Cornah, M.: Fine structure of juvenile human nucleus pulposus. *J. Anat.,* 107,2,337, 1970.

Meyer, V. H.: Statik und Mechanik des menschlichen Knochengerüstes. Leipzig, 1873.

Messerer, O.: Ueber Elasticität und Festigkeit der Menschlichen Knochen. J. G. Cotta, Stuttgart, 1880.

Miles, M.: Lateral vertebral dimensions and lateral spinal curvature. *Human Biology, 16:* 153, 1944.

Mitchell, G.: The significance of lumbosacral transitional vertebrae. *British J. Surg., 24:* 147, 1936.

Mixter, W., Barr, J.: Rupture of the intervertebral disc with involvement of the spinal canal. *New England J. Med., 211:* 210, 1934.

Moffatt, C. A., Howard, R. H.: The investigation of vertebral injury sustained during aircrew ejection. Technology, Inc., San Antonio, Texas; Quarterly Report No. 2, NASA Contract NAS 2-5062. 1970.

Molloy, C. T.: Use of four-pole parameters in vibration calculations. *JASA, 29:* 7, 1957.

Molloy, C. T.: Four-pole parameters in vibration analysis. ASME Colloquium on Mechanical Impedance Methods for Mechanical Vibrations. New York, 1958.

Mutch, W.: Shock, Vibration and Associated Environments. Bull. No. 30, Office of the Secretary of Defense, Research and Engineering, Washington D. C., 1962.

Nachemson, A.: Lumbar intradiscal pressure. *Acta Orthop. Scand. Suppl. 43,* 1960.

Nachemson, A.: The influence of spinal movements on the lumbar intradiscal pressure and on the tensile stresses in the annulus fibrosus. *Acta Orthop. Scand., 33:* 183, 1963.

Nachemson, A., Evans, J.: Some mechanical properties of the third human lumbar interlaminar ligament. *J. Biomechanics, 1:* 211, 1968.

Naylor, A., Smare, D.: Fluid content of the nucleus pulposus as a factor in the disc syndrome. Preliminary Report. *Brit. M. J., 2:* 975, 1951.

Naylor, A., Happey, F., and MacRae, T.: The collagenous changes in the intervertebral disc with age and their effect on its elasticity. An x-ray crystallographic study. *Brit. M. J., 2:* 570, 1954.

Naylor, A., Happey, F., and McRae, T.: Changes in the human intervertebral disc with age. A biophysical study. *J. Amer. Geriat. Soc., 3:* 964, 1955.

Naylor, A., Horton, W.: The hydrophilic properties of the nucleus pulposus of the intervertebral disc. *Rheumatism,* (April) 1955.

Naylor, A.: Changes in the human intervertebral disc with age. *Proc. Roy. Soc. Med., 51:* 573, 1958.

Naylor, A.: The biophysical and biochemical aspects of intervertebral disc herniation and degeneration. *Ann. Roy. Coll. Surg. Eng., 31:* 91, 1962.

Noyes, D., Clark, J. and Watson, C.: Mechanical input impedance of human teeth in vivo. *Med. & Biol. Engr., 6:* 487, 1968.

Novogrodsky, M.: Die Bewegungsmöglichkeit in der menschlichen Wirbelsäule. Bern 1911.

Odgers, P.: Lumbar and lumbosacral joints. *J. Anat., 67:* 301, 1933.

Oestreicher, H.: Field and impedance of an oscillating sphere in a viscoelastic medium with an application to biophysics. *JASA,* 23, 1951.

Olson, H.: Dynamic analogies. D. Van Nostrand, New York, 1943.

Oppenheimer, A.: Diseases of the apophyseal (intervertebral) articulations. *J. Bone & Joint Surg., 20:* 285, 1938.

Oppenheimer, A.: The apophyseal intervertebral articulations, roentgenologically considered. *Radiology, 30:* 724, 1938.
Oppenheimer, A.: Development, clinical manifestations and treatment of rheumatoid arthritis of the apophyseal intervertebral joints. *Amer. J. Roentgenol, 49:* 49, 1943.
Oppenheimer, A.: Pathology, clinical manifestation and treatment of lesions of the intervertebral disks. *New England J. Med., 230:* 95, 1944.
O'Reilly, A.: Abnormalities of the lower part of the back. *J.A.M.A., 89:* 1128, 1927.
Osgood, C.: Mechanical impedance of spacecraft structures. Prentice-Hall, New Jersey, 1965.
Patrick, L. M.: Caudo-cephalad static and dynamic injuries. Proc. of the Fifth Stapp Automotive Crash and Field Demonstration Conf., Univ. of Minn., 1962.
Perey, O.: Fracture of the vertebral end-plate in the lumbar spine. An experimental biomechanical investigation. *Acta Orthop. Scand. Suppl., 25:* 1957.
Petter, C. K.: Method of measuring the pressure of the intervertebral disc. *J. Bone & Joint Surg., 15:* 365, 1933.
Prives, M. G.: Influence of labor and sport upon skeleton structure in man. *Anat. Rec., 136:* 261, 1960.
Putti, V.: New conceptions in the pathogenesis of sciatic pain. *Lancet II,* 53, 1927.
Rabinovitch, R.: Diseases of the intervertebral disc. Charles C. Thomas, Springfield Illinois. 1961.
Rauber, A.: Elasticität und Festigkeit der Knochen. W. Engelmann, Leipzig. 1876.
Rockoff, D., Sweet, E., and Bleustein, J.: The relative contribution of trabecular and cortical bone to the strength of human lumbar vertebrae. *Calc. Tiss. Res., 3:* 163, 1969.
Rockwell, H., Evans, F. G. and Pheasant, H. C.: The comparative morphology of the vertebral spinal column. Its form as related to function. *J. Morph., 63:* 87, 1938.
Rodgers, S.: Mechanics of scoliosis. *Arch. Surg., 26:* 962, 1933.
Rolander, S.: Motion of the lumbar spine with special reference to the stabilizing effect of posterior fusion. An experimental study on autopsy specimens. *Acta Orthop. Scand. Suppl.* 90, 1966.
Rouviere, H.: Sur la texture des disques intervertebraux. *Comptes Rendus Soc. Biolog. 85:* 156, 1921.
Ruff, S.: Brief acceleration: Less than one second. German Aviation Medicine, World War II, 1:584, 1950. The Surgeon General: U. S. Air Force. Superintendent of Documents. Washington U. S.
Saunders, J., Inman, V.: Pathology of the intervertebral disk. *Arch of Surg., 40:* 389, 1940.
Schmorl, G.: Ueber die an den Wirbelbandscheiben vorkommenden Ausdehnungs- und Zerreizungsvorgänge und die dadurch an ihnen und der Wirbelspongiosa hervorgerufenen Veränderungen. *Verh. Dtsch. Path., 22:* 250, 1927.
Schmorl, G. Ueber bisher nur wenig beachtete Eigentümlichkeiten auswachsener und kindlicher Wirbel. *Arch. Clin. Chir., 150:* 420, 1928.
Schmorl, G.: Verkalkung der Bandscheiben der Wirbelsäule nebst Bemerkungen über das Verhalten der Bandscheiben bei infektiöser Spondylitis. *Fortschr. Röntgenstr.. 40:* 18, 1929.
Schmorl, G., Junghanns, H.: The human spine in health and disease. Grune & Stratton, New York, 1959.
Sedlin, E.: A rheologic model for cortical bone. A study of the physical properties of human femoral samples. *Acta Orthop. Scand. Suppl.* 83, 1965.
Sedlin, E., Hirsch, C.: Factors affecting the determination of the physical properties of femoral cortical bone. *Acta Orthop. Scand., 37:* 29, 1966.

Severin, E.: Degeneration of the intervertebral discs in the lumbar region. *Acta Chir. Scand., 89:* 378, 1943.

Shore, L.: On osteo-arthritis in the dorsal intervertebral joints. A study in morbid anatomy. *Brit. J. Surg., 22:* 833, 1935.

Soroka, W.: Analog methods of analysis. Chapter 29 in Shock and Vibration Handbook, (C. M. Harris and C. E. Crede, Eds.) McGraw-Hill, New York, 1961.

Southworth, J., Bersack, S.: Anomalies of lumbosacral vertebrae in five hundred and fifty individuals without symptoms referable to the low back *A. J. Roentgenol. 64:* 624, 1950.

Stalnaker, R. L., Fogle, J. L. and McElhaney, J. H.: Driving point impedance characteristics of the head. ASME, 70-BHF-14, 1970.

Steindler, A.: Kinesiology of the human body, under normal and pathological conditions. C. Thomas Publ., Springfield, Ill., 1955.

Strasser, H.: Lehrbuch der Muskel- und Gelenkmechanik, Bd 1—2. Springer, Berlin, 1908.

Sullivan, W., Miles, M.: The lumbar segment of the vertebral column. *Anat. Rec., 133:* 619, 1959.

Sylven, B.: On the biology of nucleus pulposus. *Acta Orthop. Scand., 20:* 275, 1950.

Taylor, R.: Anomalies of the lumbosacral articulations. *J.A.M.A., 113:* 463, 1939.

Tkaczuk, H.: Tensile properties of human lumbar longitudinal ligaments. *Acta Orthop. Scand. Suppl* 115, 1968.

Tkaczuk, H.: Experimental studies on longitudinal ligaments to provide explanation for their share in producing pains in the small of the back. Medical Academy in Warsaw. Warsaw, 1970.

Tylman, D., Ramotousky, W.: Study of strength of particular elements of the spine. *Chirurgia Narzadow Ruchu I Orthopedia Polska, 26:* 21, 1961.

Van Den Hooff, A.: Histological age changes in the annulus fibrosus of the human intervertebral disk. With a discussion of the problem of disk herniation. *Gerontologia, 9:* 136, 1964.

Viidik, A., Lewin, T.: Changes in tensile strength characteristics and histology of rabbit ligaments induced by different modes of postmortal storage. *Acta Orthop. Scand., 37:* 141, 1966.

Virgin, W.: Experimental investigation into the physical properties of the intervertebral disc. *J. Bone & Joint Surg., 33-B:* 607, 1951.

Virgin, W.: Anatomical and pathological aspects of the intervertebral disc. *Indian J. Surg., 20:* 113, 1958.

Vogt, H., Coermann, R., and Fust, H.: Mechanical impedance of the sitting human under sustained acceleration. *Aerospace Medicine, 39:* 7, 1968.

Von Bekesy, G.: The sensitivity of standing and sitting man to sinusoidal vibration. (Ueber die Empfindlichkeit des stehenden und sitzenden Menschen gegen sinusformige Erschutterungen.) *Akustische Zeit.,* 4, 1939.

Von Gierke, H.: Sound absorption at the surface of the body of man and animals. *JASA, 21:* 55, 1949.

Von Gierke, H.: Measurement of acoustic impedance and acoustic absorption coefficient of the surface of human body. AF-TR-6010, Wright-Patterson AFB, Ohio, 1950.

Von Gierke, H., Oestreicher, H., Franke, E., Parrack, H., and von Witter, W.: Physics of vibrations in living tissue. *J. Appl. Physiol., 4:* 1952.

Von Lackum, H.: The lumbosacral region. *J.A.M.A., 82:* 1109, 1924.

Vykukal, H.: Dynamic response of the human body to vibration when combined with various magnitudes of linear acceleration. *Aerospace Medicine,* 1968.

Weaver, J. K., Chalmers, J.: Cancellous bone: Its strength and changes with aging and an evaluation of some methods for measuring its mineral content. 1. Age changes in cancellous bone. *J. Bone & Joint Surg., 48-A:* 289, 1966.

White, A. A.: Analysis of the mechanics of the thoracic spine in man: An experimental study of autopsy specimens. *Acta Orthop. Scand., Suppl.* 127, 1969.

White, G. H., Jr., Lange, K. O. and Coermann, R.: The effects of simulated buffeting on the internal pressure of man. *Human Factors, 4:* 275, 1962.

White, G. H., Jr., Lange, K. O. and Coermann, R.: Human vibration research. Lippert, S., ed., Pergamon Press, Oxford, 1963.

Whitney, C.: Asymmetry of the vertebral articular processes and facets. *Am. J. Phys. Anthrop., 9:* 451, 1926.

Willis, T. A.: The lumbosacral vertebral column in man: Its stability of form and function. *Am. J. Anat., 32:* 95, 1923.

Willis, T. A.: An analysis of vertebral anomalies. *Am. J. Surg., 6:* 163, 1929.

Wittman, T., Phillips, N.: Human body nonlinearity and mechanical impedance analyses. *J. Biomechanics,* 2, 1969.

Wright, D. V.: Bureau of Ships letter to SSN Noise Advisory Committee Structural Impedance Panel, 1961.

Yamada, H.: Strength of biological materials. Williams and Wilkins Co., Baltimore Maryland, 1970.

Yorra, A. J.: The investigation of the structural behavior of the intervertebral disc. Thesis. Masachusetts Institute of Technology, 1956.

Ziegenruecker, G. H., Magid, E. B.: Short time human tolerance to sinusoidal vibration. Wright Air Development Center, Wright-Patterson Air Force Base, Ohio, WADC Technical Report 59—391, AD 227—341, 1959.

10. Appendix

SECTION 1

Units and Conversion Factors

TO CONVERT FROM	*TO*	*MULTIPLY BY*
FPS	CGS	
$\frac{lb}{IN}$	$\frac{KP}{CM}$	.1783
$\frac{lb}{IN^2}$	$\frac{KP}{CM^2}$	7.0308×10^{-2}

SECTION 2

The End Plate of the Intervertebral Disk

Figures A, B, C, and D show the morphology of a cervical end plate.

Figure A is a cephalic view of an end plate peeled away from a vertebral body. The shiny translucency of the structure is apparent. Its matrix is soft and pliable. This photograph was taken with the end plate air-dried.

Figure B is a caudal view of the same end plate as in Figure A. In this case the central "light" material are crystals which may assist to fix the end plate to the vertebral body centrum. The area of the nucleus pulposus appears to be smaller in pictures A & B.

Figure C is a cephalic view of an end plate removed from an older (compared to Figures A and B) subject. The relative position of the nucleus pulposus and annulus fibrosus is indicated. Its matrix and its central region are brittle.

Figure D is a caudal view of the end plate pictured in Figure C. The organized calcified ridges are well defined anteriorly, but less defined posteriolaterally. These may play a role in minimizing torsional vertebral motions in the axial mode of excitation by assisting to increase surface contact area.

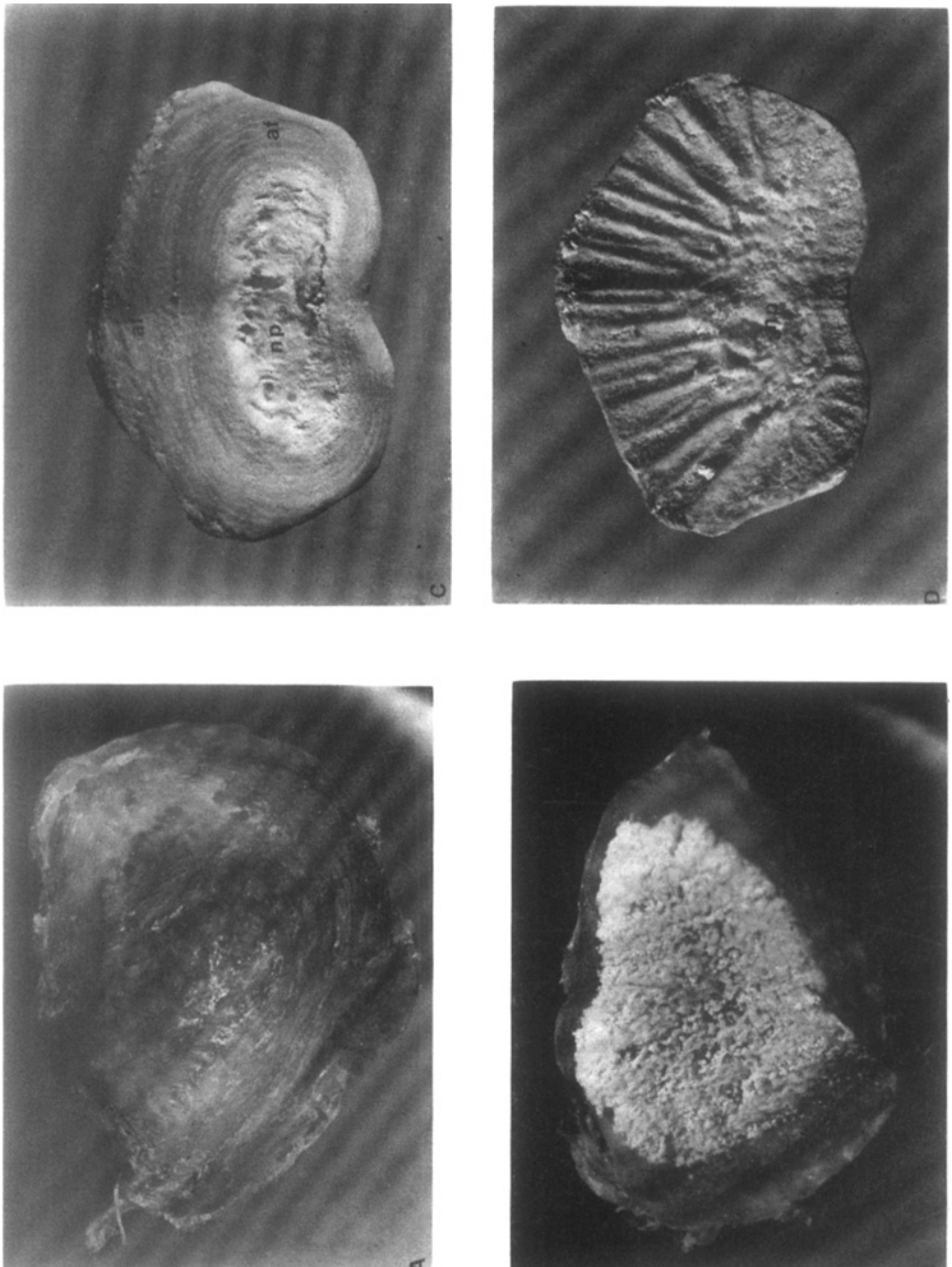


Figure 9 A.

SECTION 3

The Calibration Graph for the Strainsert Load Cell—least square solution

VOLTAGE OUTPUT VS LOAD

LOAD CELL CALIBRATION #4

S/N FLIU - 0224

LEAST SQUARE SOLUTION

n = 25	X LBS	Y VOLTS
1	0	0.009
2	50	0.37
3	100	0.78
4	150	1.07
5	200	1.42
6	300	2.13
7	400	2.83
8	500	3.53
9	600	4.22
10	700	4.92
11	800	5.61
12	900	6.32
13	1000	7.00
14	900	6.32
15	800	5.62
16	700	4.92
17	600	4.22
18	500	3.53
19	400	2.83
20	300	2.13
21	200	1.43
22	150	1.08
23	100	0.73
24	50	0.37
25	0	0.020
TOTAL	R_{cal} (324.5 K)	1.044

SLOPE = +.0069917957 VOLTS / LB

Y_{int} = +.0253729988 VOLTS

X_{int} = -3.628967420 LBS

Figure 2 A. The calibration graph for the Strainsert load cell.

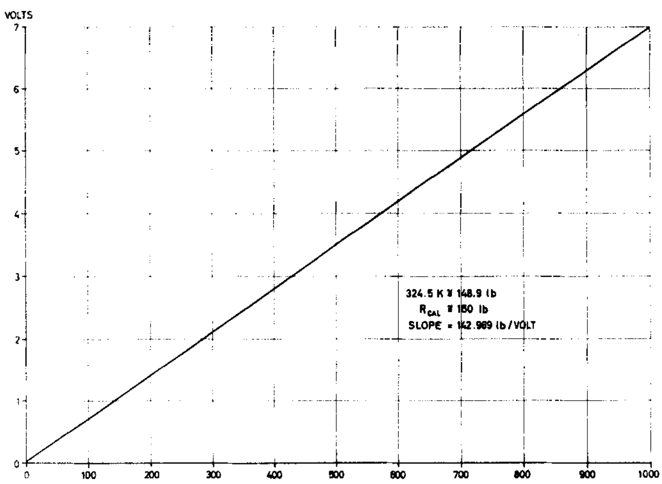


Figure 2 B. Voltage output vs load — calibration graph.

SECTION 4

Engineering Specifications for Impedance Tester

Maximum Full Scale Outputs

Force 150 lb peak = 10 volts peak
Velocity four selectable ranges
5 inches per second = 10 volts peak, G = 1
2.5 inches per second = 10 volts peak, G = 2
1.0 inches per second = 10 volts peak, G = 5
0.5 inches per second = 10 volts peak, G = 10

Real Power, four ranges determined by Velocity Gain Setting
750 lb- in/sec peak = 10 volts peak, G = 1
375 lb- in/sec peak = 10 volts peak, G = 2
150 lb- in/sec peak = 10 volts peak, G = 5
75 lb- in/sec peak = 10 volts peak, G = 10

Internal Calibration Signals

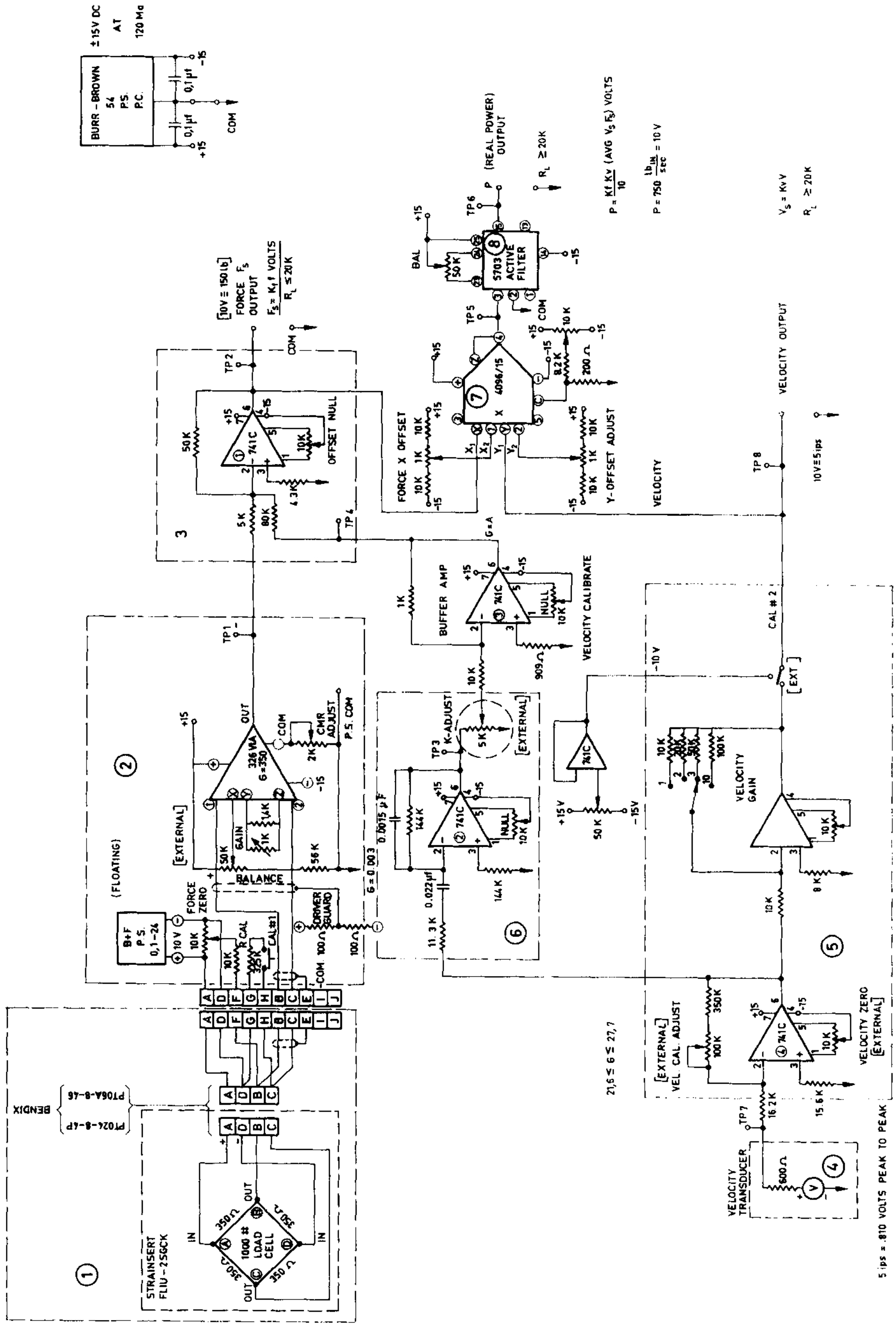
Force, 150 lb peak = 10 volts peak
Velocity, 5 ips peak = 10 volt peak
Real Power, 750 lb- in/sec = 10 volts peak

Mechanical

Maximum shaker output 150 lb peak force

Maximum shaker response with a constant full scale velocity of 5 ips and 20 lb load.

f Hz	velocity, $\dot{x}$ ips peak	acceleration, $\ddot{x}$ g's peak	displacement, x in-peak-to-peak
5	5 ips	0.41	0.32
10	5 ips	0.82	0.160
15	5 ips	1.23	0.106
20	5 ips	1.64	0.080
25	5 ips	2.05	0.064
30	5 ips	2.46	0.053
35	5 ips	2.87	0.045
40	5 ips	3.28	0.040
45	5 ips	3.00	0.030
50	5 ips	3.00	0.022



IMPEDANCE COMPUTER SCHEMATIC DIAGRAM

Test Frame Specifications

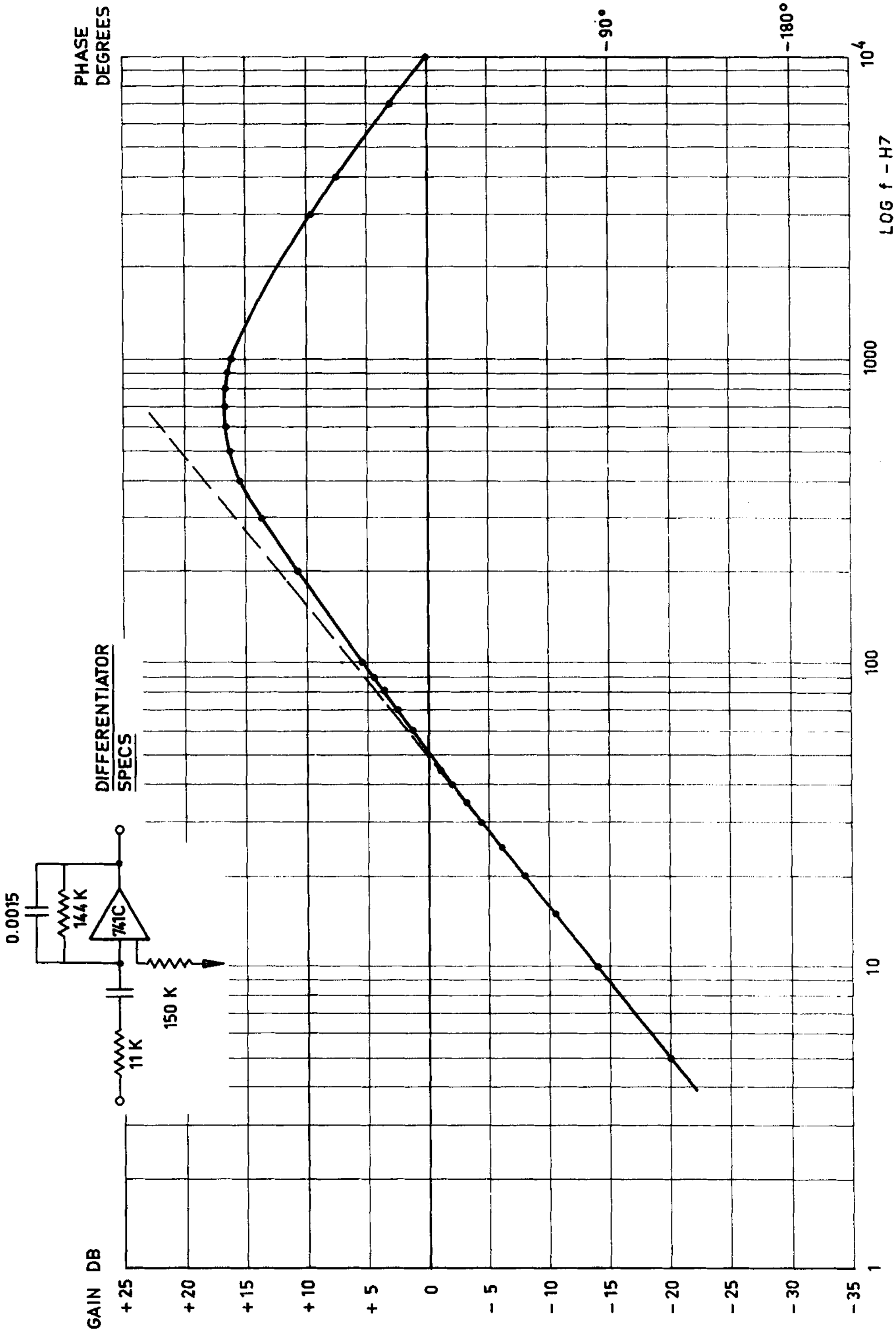
A. f_0 = 380 Hertz, axial direction

B. K frame = 9×10^5 lb/in upper position

C. Upper preload screw 1 inch diameter/14 threads/inch,
Advance = 0.0714 in/rev
= 0.0001 in/degree
= 0.009 in/quadrant
= 0.0015 in/15 deg. graduation

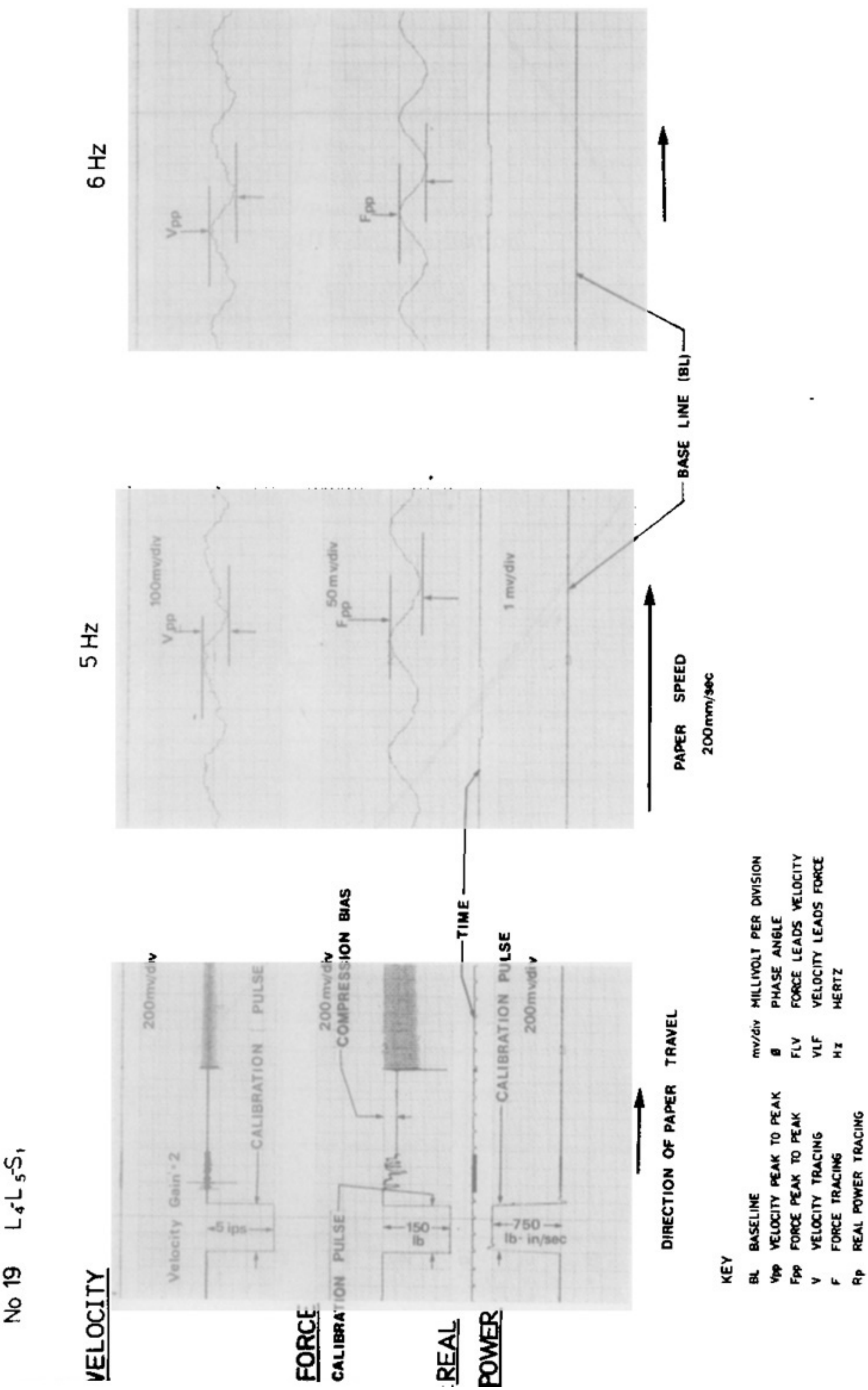
D. Lower pre-load system consists of a triangular plate supported by three springs in parallel. Adjustment screws are 1/4—28
Advance = 0.0359 in/rev.
= 0.0001 in/degree
= 0.009 in/quadrant
= 0.0015 in/15 deg. graduation.

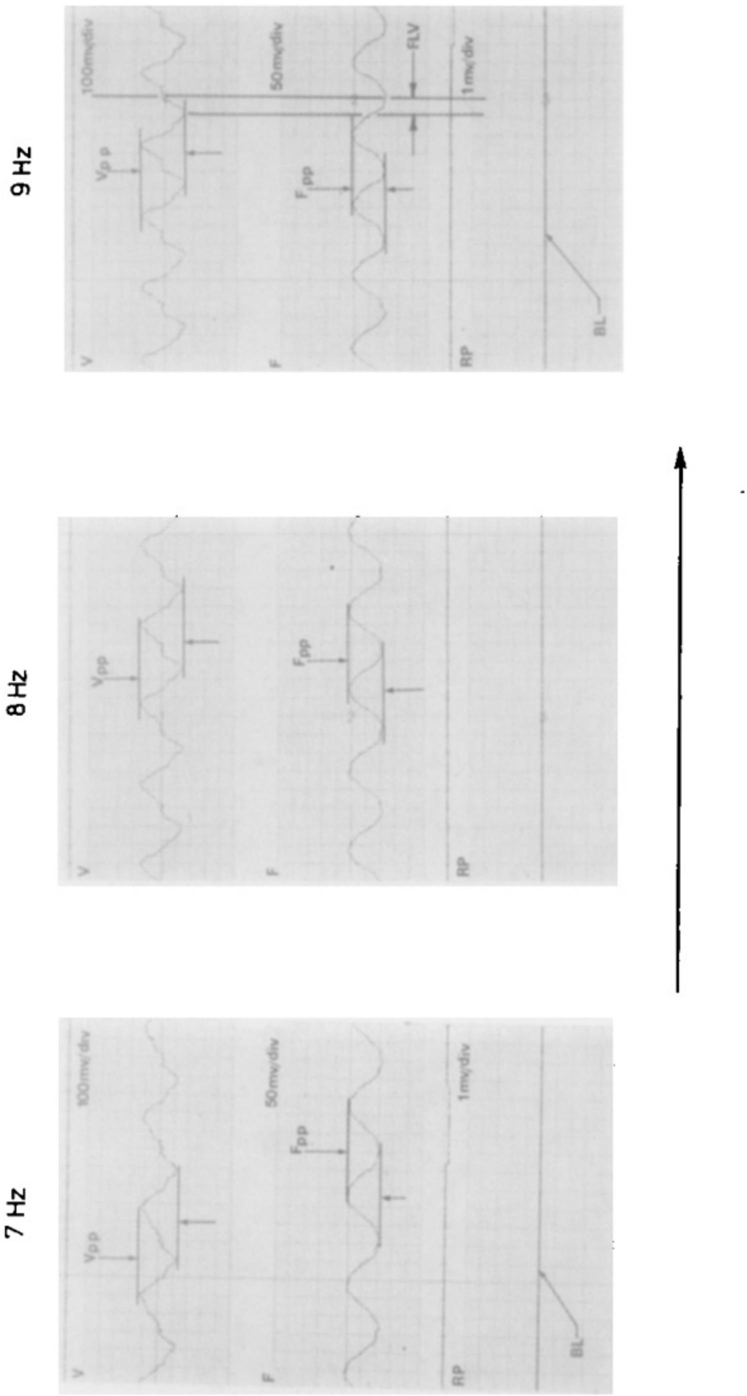
Graduations on dial scale for making screw adjustments are at 15 degree spacings.

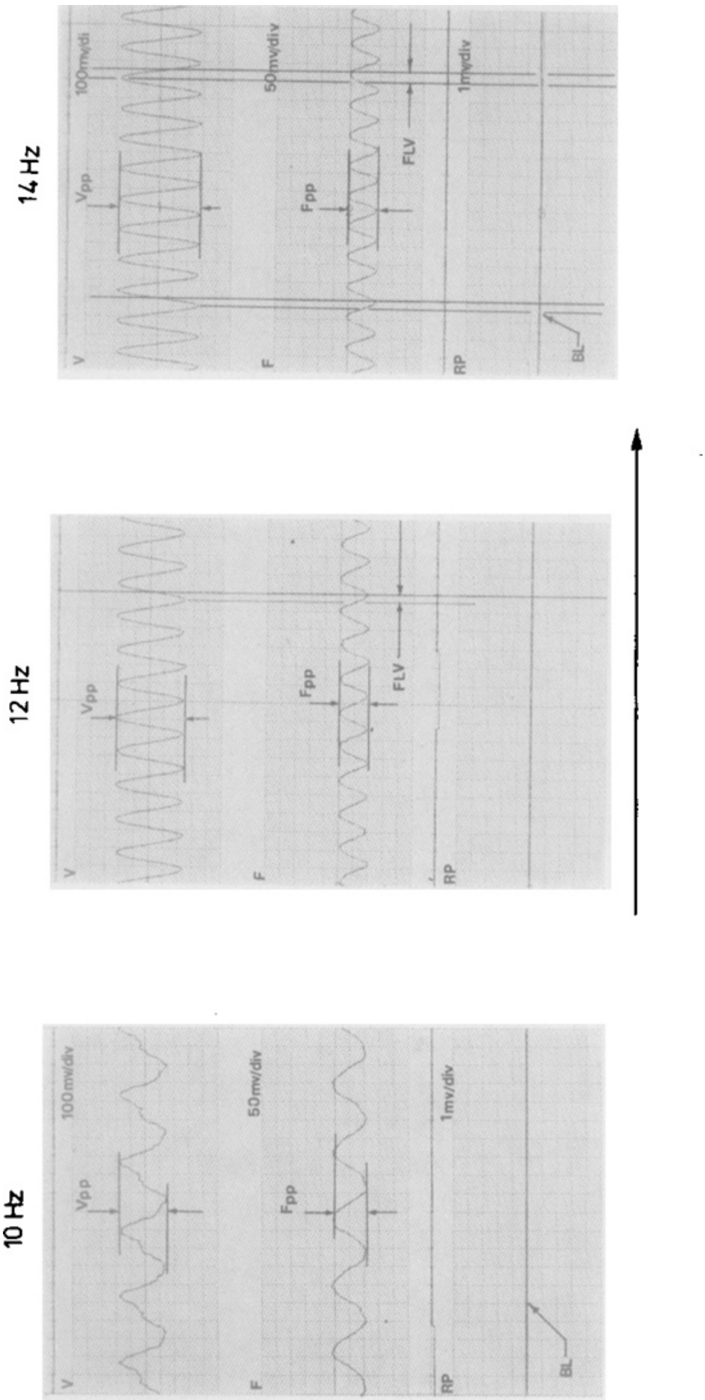


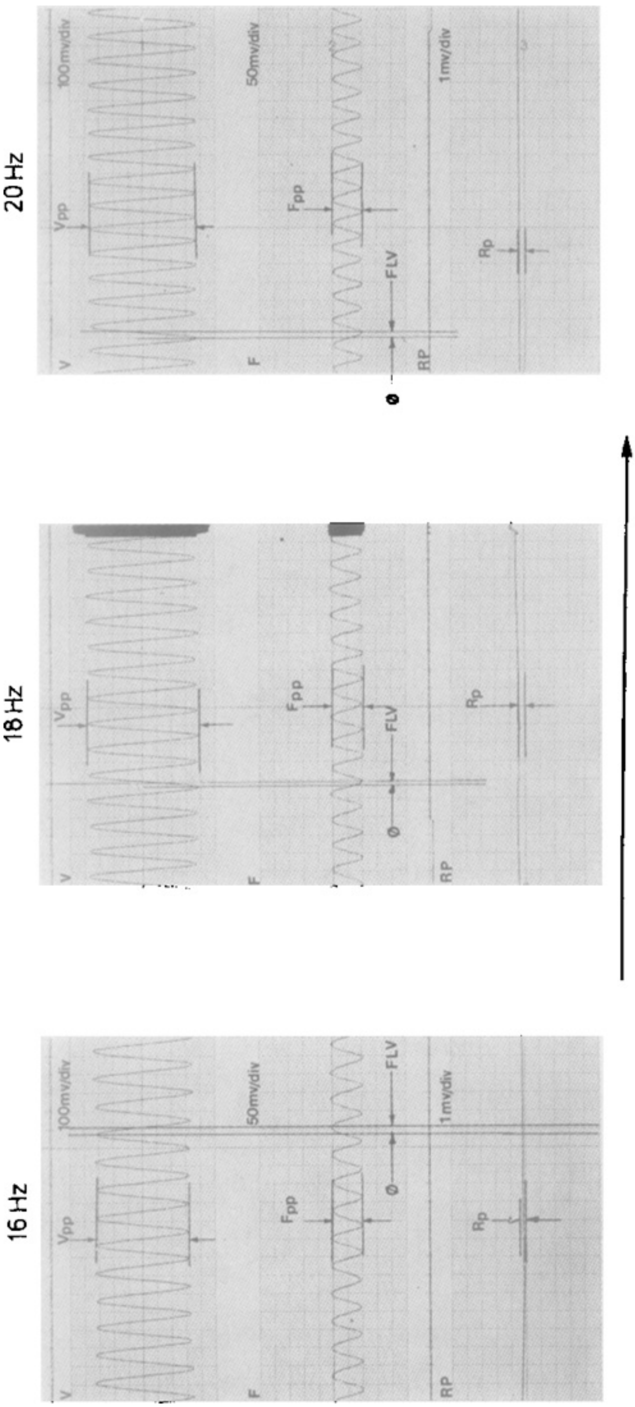
The Differentiator specifications.

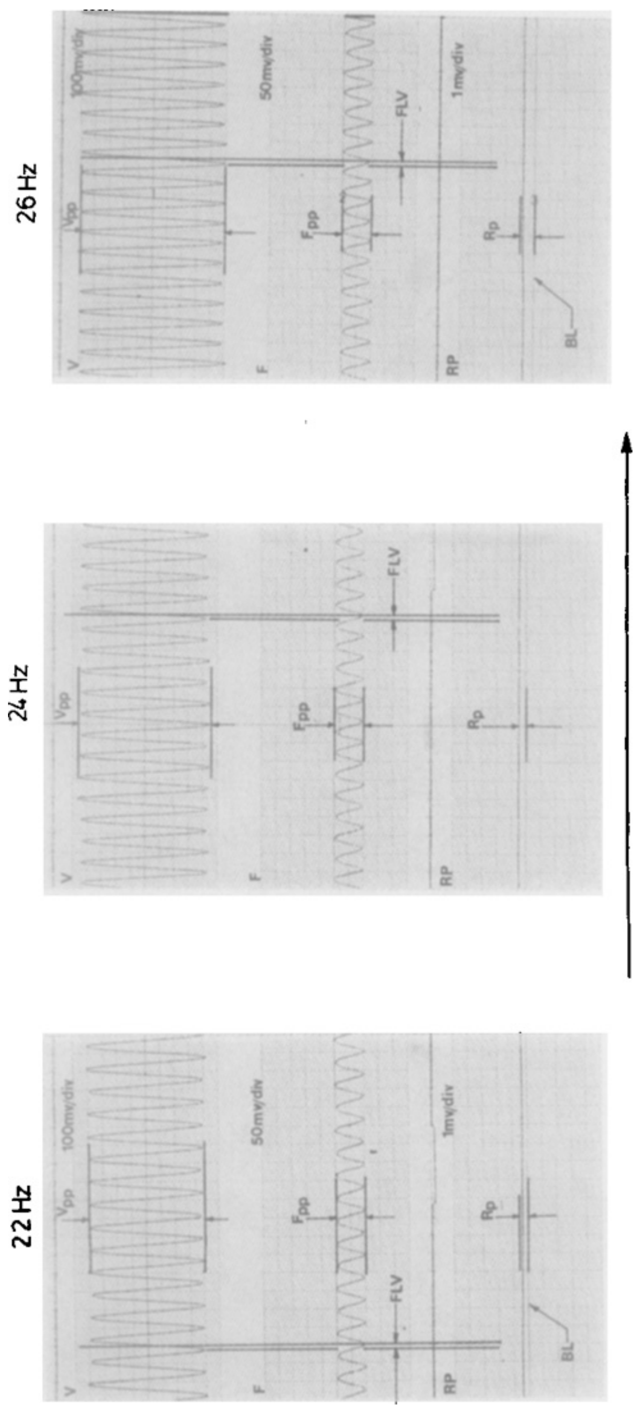
SECTION 5
Typical Waveforms from an Impedance Test

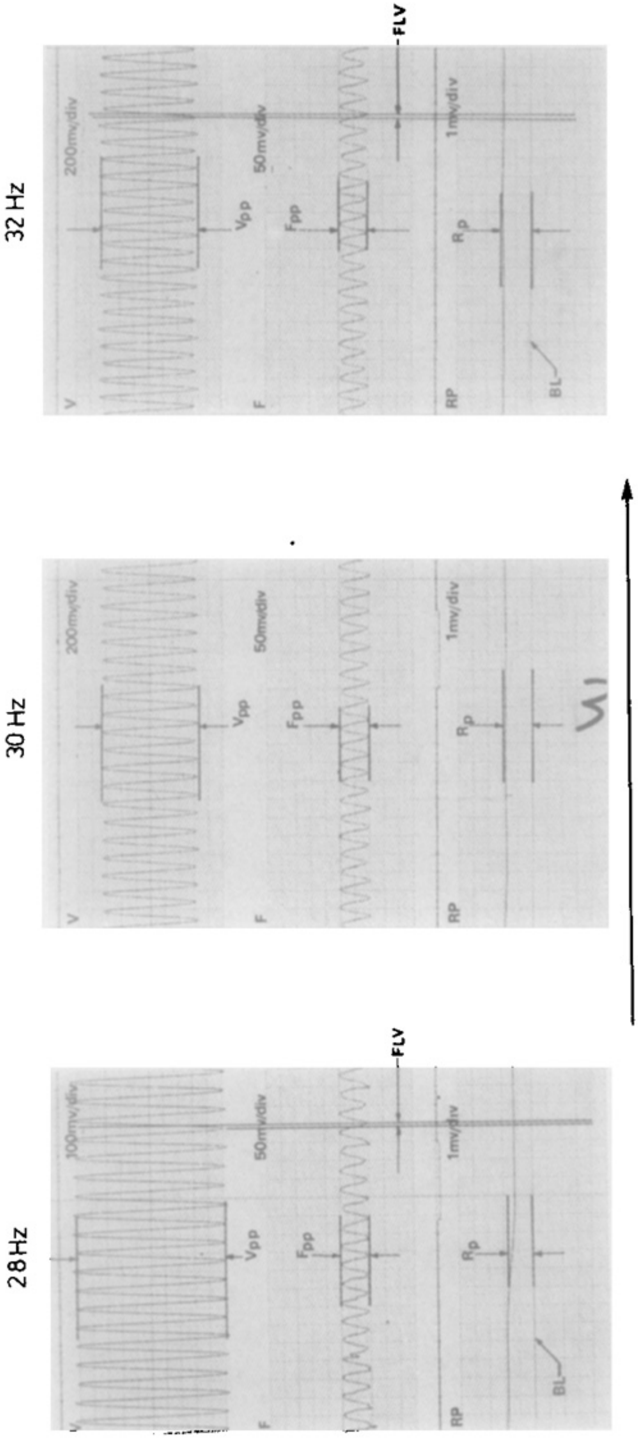


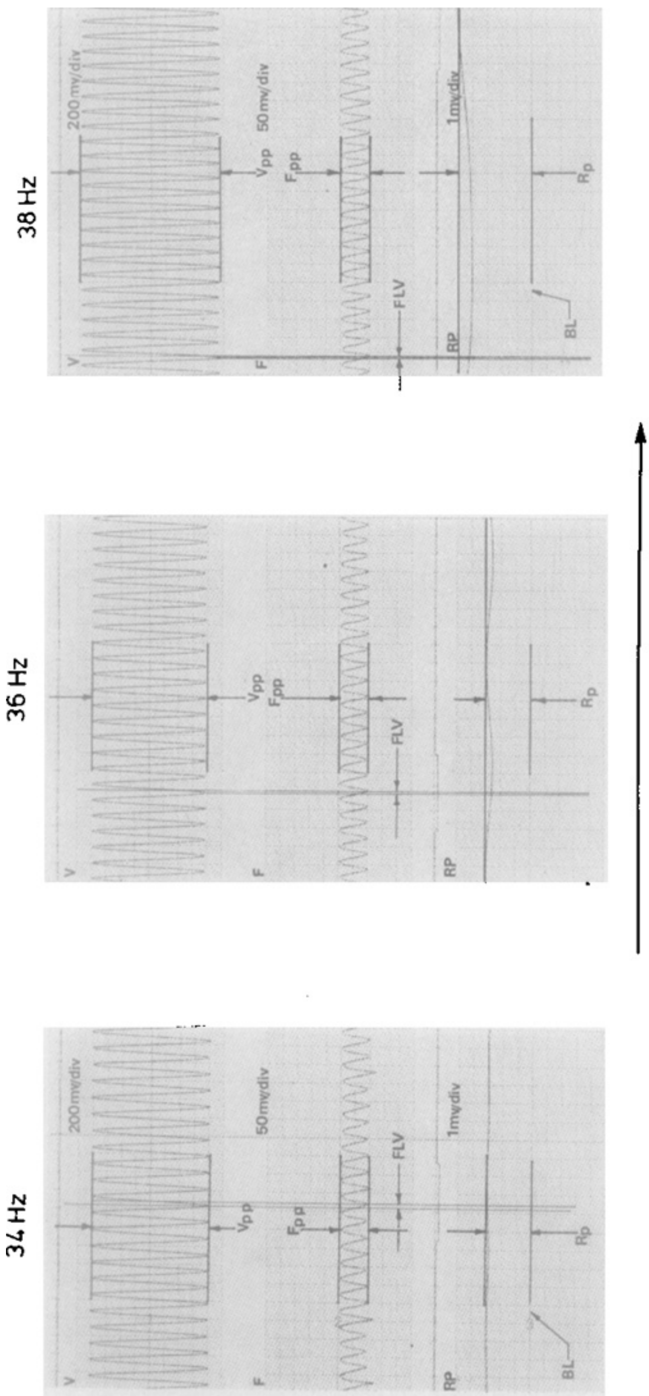

9 Hz
8 Hz
7 Hz
V
F
RP
100mv/div
50mv/div
1mv/div
Vpp
Fpp
FLV
BL


10 Hz
12 Hz
14 Hz
V
F
RP
V_{pp}
F_{pp}
FLV
BL
100mv/div
50mv/div
1mv/div

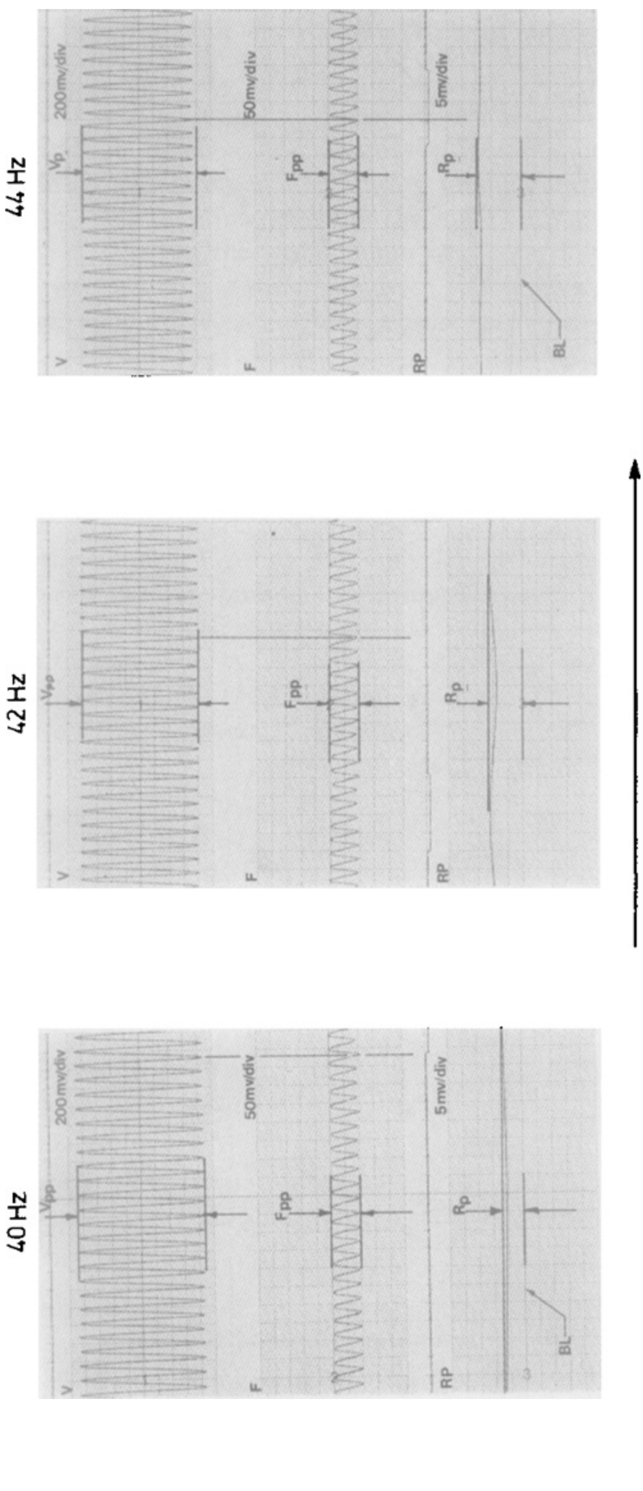

16 Hz
18 Hz
20 Hz
V
F
RP
100mv/div
50mv/div
1mv/div
Vpp
Fpp
Rp
FLV
Ø


26 Hz
24 Hz
22 Hz
V
F
RP
Vpp
Fpp
FLV
Rp
BL
100mv/div
50mv/div
1mv/div

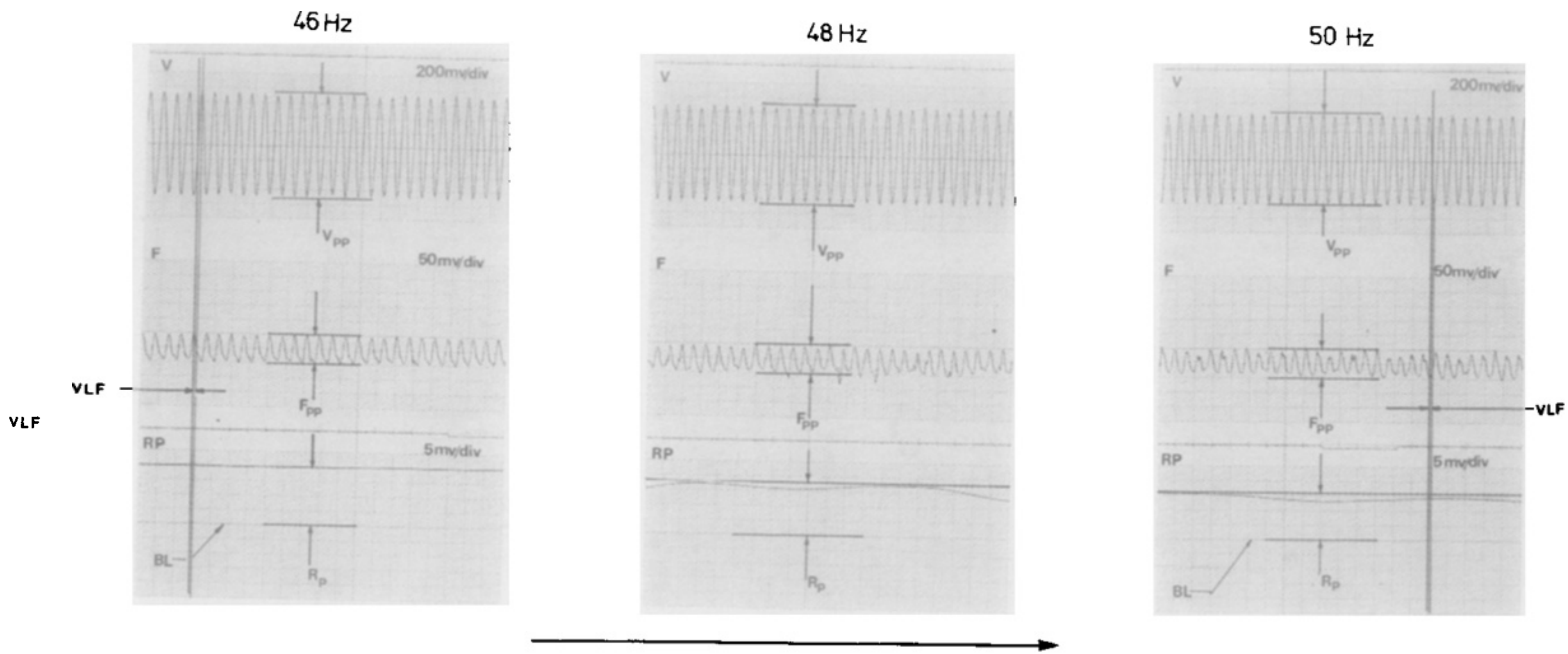

28 Hz
30 Hz
32 Hz
V
F
RP
100mv/div
200mv/div
50mv/div
1mv/div
Vpp
Fpp
Rp
FLV
BL



34 Hz
36 Hz
38 Hz
200mv/div
50mv/div
1mv/div
V
F
RP
V_{pp}
F_{pp}
FLV
R_p
BL



40 Hz
42 Hz
44 Hz
V
F
RP
200mv/div
50mv/div
5mv/div
Vpp
Vp
Fpp
Rp
BL


46 Hz
48 Hz
50 Hz
V
200mv/div
V_{pp}
F
50mv/div
F_{pp}
RP
5mv/div
R_p
BL
VLF
VLF

SECTION 6

Vertebral Unit Creep under Dynamic loading

Figure 1 A is an actual force waveform generated for a vertebral unit in the frequency range from 5 to 50 hertz. The waveform was obtained by utilizing a 2 channel Mark 220 Brush analog recorder in simultaneous operation with the 6 channel Mark 260 recorder. The tracing was obtained in the following manner: A set of leads were jumped from the force output of the impedance computer panel directly to the signal input terminals of the 220 recorder. Both recorders were set to the same sensitivity positions. Upon initiating the master calibration switch a rectangular wave pulse was written onto both force channels.

The paper speed and pen position control were continually adjusted on the 260 recorder throughout any particular test. However, the pen position control sensitivity and chart speed (at 1 mm/sec) were all locked into position and remained so throughout any sequence of the tests on the 220 recorder. In this manner the influence of creep with respect to the initially applied preload and baseline was examined. Figure 1 A is fully labelled and enables the reader to differentiate the various losses in compression bias as a function of frequency. For instance, the calibrated force value applied was 63.75 lbs (28.9 kgs), and identified as point A. At point B, the compression bias has been reduced to a value of 45.0 lbs (20.4 kgs). At the 50 hertz (point C) the preload is further reduced when compared to the point A.

VERTEBRAL UNIT CREEP UNDER CONSTANT LOAD AND AT EACH FREQUENCY INTERVAL