A DISTRIBUTED PARAMETER DYNAMIC MODEL OF THE HUMAN FOREARM

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ABSTRACT

The human hand-arm system has been modeled previously as a lumped parameter system. To provide better anatomical representation, a distributed parameter model has been developed. This model is composed of dual homogeneous Euler-Bernoulli beam elements representing the bones of the forearm; a simple discrete parameter system is used to represent the properties of the hand local to a vibrating handle. Validity of the model was verified by experiment.

INTRODUCTION

Mechanical impedance techniques have been used to investigate the vibration response of the human hand-arm system. A number of investigators have used the mechanical impedance approach to develop mathematical models to represent analytically the dynamic behavior of the human hand-arm system.

In most biological systems, however, the parameters that describe their behavior are not lumped in nature, but are distributed throughout the entire system. It is recognized that to represent continuous systems by lumped parameters is, at best, an approximation. The advantage of the distributed parameter model is that the physical structure of the system can be modeled as well as the extrinsic (overall) response, which may be evident from mechanical impedance studies. As well, the continuous system approach does not require the same degree of empiricism as does the lumped system approach.

Abrams[1] proposed a single-degree-of-freedom damped spring-mass system for each of three orthogonal directions of vibration. The frequency range of his study was from 70 to 1,670 Hz. Reynolds and Soedel[2] in a similar study proposed a lumped parameter six-degree-of-freedom model, two degrees of freedom being associated with each of three orthogonal directions. However, they found it convenient to separate the frequency range of the study (20 to 500 Hz) into two ranges with the division occurring at 73 Hz. This enabled a single-degree-of-freedom model to be fitted to the impedance data in each of the two smaller ranges. Mishoe[3] developed more complex, three-degree-of-freedom lumped parameter models for each orthogonal direction over the frequency range from 20 to 2,000 Hz.

DEVELOPMENT OF THE DISTRIBUTED PARAMETER MODEL

In this study, distributed mass and stiffness parameters were obtained by representing each long bone of the forearm, the radius and ulna, as a homogeneous flexural member. A distributed damping parameter was admitted to each beam by introducing the concept of complex stiffness to the beam stiffness parameter; it is the imaginary part of this quantity that contributes to damping whereas the real part is the static bending stiffness.

The soft tissue material of the forearm forms a viscoelastic field that surrounds the radius and ulna bones. Most of the damping along the length of the beam would be derived from this material; the damping contribution of the bone would be considerably less. By specifying the imaginary part of the complex beam stiffness as admitting viscoelastic damping, the viscoelastic field is eliminated, at least as far as its dissipative effects are concerned. The mass of the viscoelastic field is readily incorporated in the beam mass parameters.

It is assumed that the effective beam length is given by the distance between the elbow and the center of the gripped handle. The wrist is thus included as part of the beam. This may be partly justified by the fact that, when the hand is tightly gripping the handle, the wrist can become almost as stiff as the rest of the arm.

When modeling the mechanical impedance data collected for the system, it is not only necessary to model the forearm itself, but also the coupling between the driving point and the arm. Coupling to the arm is via the soft tissue material of the hand local to the driven handle of a vibratory power tool. This coupling was represented by a discrete parameter system:

a Kelvin viscoelastic model (parallel spring and damping elements) in series with a concentrated mass that represented the mass of tissue material of the hand being excited in a local mode. Independent coupling paths between the handle and each bone was assumed. Figure 1 shows the dual beam model of the forearm.

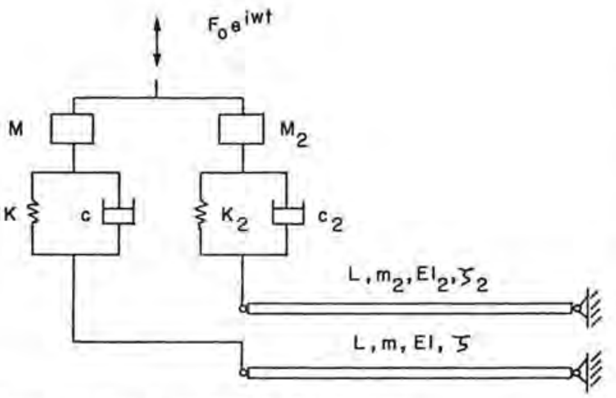

Figure 1. Beam model of the forearm.

The method of four pole parameters was used to derive the mechanical impedance expression for the model. This technique enables the overall dynamic characteristics of a composite system to be built up in terms of the characteristics of the individual components. The method applies equally well to lumped and continuous systems and to combinations of these systems. It is particularly useful when a continuous element is not directly attached to an inertial reference point. Briefly, each component of the composite system may be regarded as having an input force and velocity at an input junction and an output force and velocity at an output junction. Equations relating the input quantities and the output quantities may be obtained by solving the equations of motion of the element. A matrix of four pole parameters is obtained from these equations. Elements may be connected in series or parallel form by manipulation of these matrix quantities; further detailed discussion of this method is given by Wood.[4]

EXPERIMENTAL PROCEDURE

In vivo mechanical driving point impedance data were then collected for the forearm, which was dynamically isolated from the upper arm. Isolation was accomplished by constraining the arm to be simply supported at the elbow; each subject placed his elbow firmly (but comfortably) on a raised surface molded to fit the contour of his elbow. Care was taken to ensure that inplane rotation of the forearm was as free as possible.

Coupling of the vibration from the source to the hand was by means of a smooth tubular handle (1-inch diameter), which the subject gripped with a prescribed force. The handle was split and fitted with strain gauges, the output of which was monitored on a readout meter in the subject's view enabling him to maintain the prescribed grip force. All tests were conducted at a single grip force, thereby eliminating grip force as a variable in the system. The grip force corresponded to a firm grip on the handle of a typical power tool such as a chain saw.

Impedance data on the forearm were collected over a frequency range from 30 Hz to 1,000 Hz. The test configuration consisted of the arm lying midway between the pronated and supinated positions (the vibrating handle was positioned vertically). The International Standards Organization[5] has recommended an orthogonal coordinate system (having its origin in the head of the third metacarpal bone) to be used when reporting the direction of vibration. The direction of vibration corresponded closely to the X_h direction in the "handgrip" position. A diagram of the arm in test configuration is shown in Figure 2.

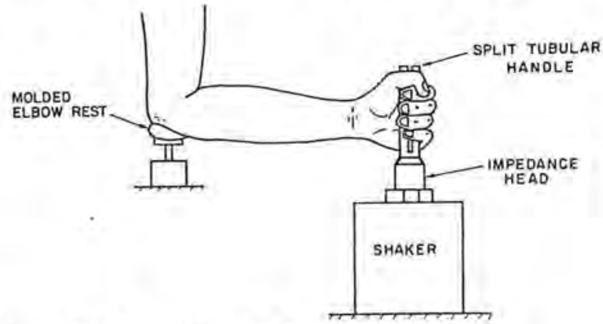

Figure 2. The forearm in test configuration.

DATA MEASUREMENT SYSTEM

Sinusoidal vibrations were produced at the handle by an electromagnetic vibrator. An impedance head was placed between the vibrator and the handle to measure force and acceleration inputs to the hand. A closed-loop feedback control system was used to maintain the acceleration level at 0.7 g peak-to-peak over the frequency range; a sweep oscillator was a part of the controller.

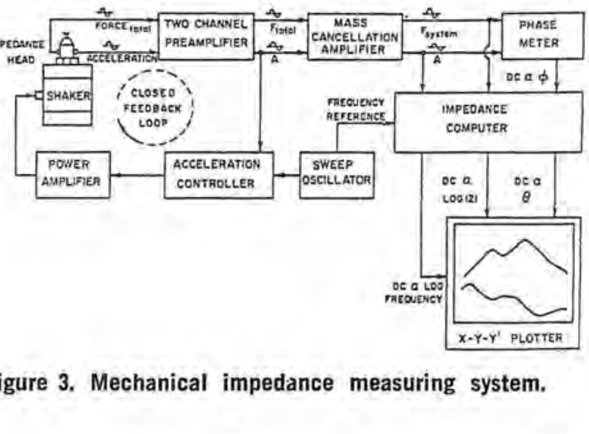

Figure 3. Mechanical impedance measuring system.

Mass cancellation of the force signal was necessary to eliminate the effect of the mass of the test fixtures above the force gauge, in this case the handle that coupled the vibrations to the hand. A phase meter was used to determine the phase relationship between the acceleration and corrected force signals. An impedance computer was then used to compute mechanical impedance, and the impedance magnitude and phase angle information were plotted directly as a function of frequency on an X-Y-Y′ recorder.

The experimental equipment is diagramed in Figure 3.

COMPARISON OF THEORETICAL AND EXPERIMENTAL RESULTS

To permit comparison with the experimental data, the frequency response of the model was obtained by using a digital computer to analyze the model equation. Measured arm length was used directly in the model equation and was fixed. An estimate of the mass parameter for each model was made by measuring the arm displacement and assuming an average arm density about 5% greater than that of water (soft tissue is composed largely of water). Estimates of the real part of the beam stiffness parameters were made by assuming each long bone to have a tubular cross-section 1/16 inch thick with an approximately ¼-inch diameter; a Young's modulus for bone of about 10^6 psi was assumed. Estimates of the parameters for the discrete system of the hand were obtained from Abrams[1] and Mishoe.[3] Note that the value of the discrete parameters depends not only on the properties of the hand itself but also on grip force and the type of handle being used to couple the hand to the vibration source. A constant grip force for the given handle was used in this study.

Starting with the initial estimates of the parameters, a trial and error method followed by a computerized optimization routine were used to obtain the best fit between the model response curves and the experimental data. In Figure 4, the model frequency response curve overlaying the experimental mechanical impedance data is shown. Best fit is obtained when the effective arm mass is about 10% below the estimated arm mass. This may be justified by considering the mass distribution of the arm with concentration of arm mass near the hinged elbow, where the translational velocities are smaller. The effective mass of the assumed uniform beam would be expected to be smaller than the estimated value when the total mass is evenly distributed along the length of the arm. Total arm mass is distributed between the two beams, and best fit is obtained when the distribution is unequal in approximately the ratio 4.5 to 1. This suggests that one bone is more effectively connected to the viscoelastic field (soft tissues) surrounding the forearm and the other bone is relatively free of the viscoelastic field. The beam with the most mass was the most heavily damped, which is to be expected if the damping of the beam is dependent

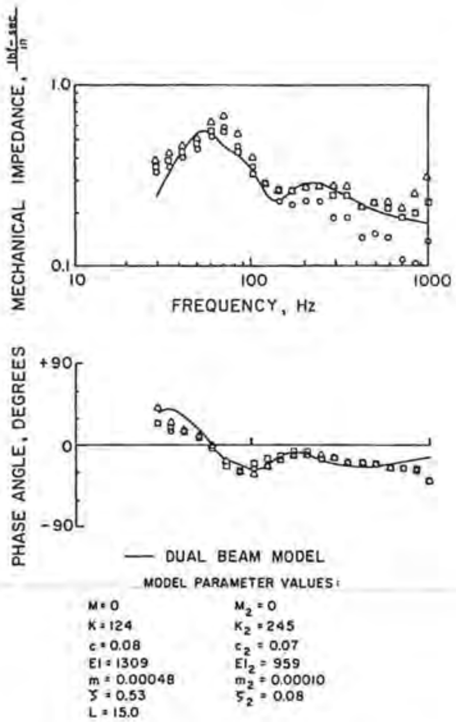

Figure 4. Model impedance curve compared with experimental data points.

mainly on the viscoelastic field surrounding the beam. The beam of larger mass was about 30% stiffer than the beam of the smaller mass. Coupling into the two bones differed significantly; although damping values of each path were similar, the spring coupling to the beam of smaller mass is about twice that coupling into the beam of larger mass.

It is evident from Figure 4 that above about 700 Hz, the model curve deviated from the experimental data. The most likely source of this behavior is the hand. Abrams,[6] in his study of vibration transmission along the arm of a human cadaver, concluded that high frequency vibration is absorbed largely near the point of exposure. His results indicated that low frequency vibrations are transmitted into the arm but that more than 90% of the vibration above 250 Hz never reaches the distal (wrist) end of the radius. As the high frequency response of the model is dominated by the discrete parameters representing the hand, it appears that the discrete model of the hand is deficient at the higher frequencies.

CONCLUSIONS

A distributed parameter dual beam model of the forearm has been developed that adequately represents the dynamic properties of the forearm from 30 Hz to about 700 Hz. In the model developed for the hand-forearm system, the parameters governing the behavior of the hand may be separated from the parameters governing the behavior of the forearm. The information yielded by this distributed parameter model appears more explanatory and is complementary to that yielded by a lumped parameter representation. It is evident, however, that an improved model of the hand is required for use at frequencies greater than about 700 Hz.

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QUESTIONS, ANSWERS, AND COMMENTARY

Comment (J. Guignard, Navy Aerospace Medical Research Laboratory): I'm particularly concerned about the question of what we mean when we talk about vertical and horizontal vibration relating to the hand. In a previous communication, we heard about a vertical vibration in this sense—that this is anatomically equivalent to the horizontal vibration discussed in the first communication of this session. In neither of your two communications were we informed of the origin and orientation of the biodynamic coordinate system to which the data are referenced. In this we have a standard system. We have no basis to transform data to anatomical receptor structures in the limb. We have no basis for making comparisons of data between labs, and I believe we have no basis for assessing the equivalence of data that are used for standards. So may I make a plea that we do try to agree to use equivalent coordinate systems between labs and in communications such as this.

Answer (C. Suggs): Your criticism is accepted. We'll try to follow the standards that have been set forth for the hand-arm vibration work, and thereby hope that our work will be more easily translatable into results from other laboratories and that better comparisons can be made. I think that's a very good comment and a very worthwhile goal.

Comment (H. Von Gierke, Aerospace Medical Research Laboratory): While we are making comments and, again, commenting on both of the last two communications, I would like to remark that to refer to the limit curves in the proposed ISO standard as damage risk criteria is probably not correct. As the ISO committee (TC108/SC4) chairman, I would say these are protective/preventive medicine criteria; they should really not be confused with damage/risk criteria wherein the reference is usually to one specific damage to a specific organ or particular site—criteria that usually give some percentage of probability of risk. As I said, these limit curves are protective criteria in the sense of good preventive medicine.

Question (R. Larsen, Outboard Marine Corporation): I know that in some of your curves you refer to the transmissibility decreasing as frequency increased. Then you talked about the power absorption increasing. Is this not similar to a vibration system —that as it goes into its isolation mode, the transmission decreases and the mass acts as inertia, as opposed to the input force, looking at the vector resolution at this point? Now, which mass are you talking about as being mass opposed into force? Mass 1, 2, or 3?

Answer: We know that as frequency increases, we have less transmissibility up the arm. Therefore, the masses actually participating in the vibration are decreased—that is, those masses remote from the point of input tend to stand still. There is also isolation going on at the connecting points, the springs between the lumped masses. This is where the isolation occurs.

Question (R. Larsen): Then these masses are opposed to the input force and absorbing the energy against the surface of the hand?

Answer: Yes. As the frequency increases, the mass that is active in the system is the one closer and closer to the point of application. At the higher frequencies, as I understand the system, it would have to be at the skin surface of the hand.

Question (R. Larsen): It is dissipating the energy and absorbing energy at that time?

Answer: Looks that way to me.

Comment (D. Reynolds, University of Texas): In my opinion, masses cannot absorb energy. The masses resist this motion. But the thing that appears to be absorbing the energy is the damping that is between the masses. In other words, when you put energy into the hand, it goes one of two ways. It either goes into storing potential kinetic energy or it goes into energy that is dissipating. Then, as I interpret it, as frequency increases, the level of storing kinetic energy in the hand system decreases as well, and the thing that increases is the dissipated energy.

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THE INTERNATIONAL OCCUPATIONAL HAND-ARM VIBRATION CONFERENCE

Sponsored by
National Institute for Occupational Safety and Health
Cincinnati, Ohio, U.S.A.
October 1975

Editors:
D. E. WASSERMAN
W. TAYLOR

Manuscript Editor:
M. G. CURRY

U.S. DEPARTMENT OF HEALTH, EDUCATION, AND WELFARE
Public Health Service
Center for Disease Control
National Institute for Occupational Safety and Health
Cincinnati, Ohio 45226
April 1977

For sale by the Superintendent of Documents, U.S. Government Printing Office, Washington, D.C. 20402

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DHEW (NIOSH) Publication No. 77-170