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UNDER SUSTAINED ACCELERATION

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Mechanical Impedance of Supine Humans Under Sustained Acceleration

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Measurements of the mechanical impedance of the supine human body were conducted to investigate the nonlinearities of the body system. A hydraulically driven shake table was installed on a centrifuge. Transmitted force and the acceleration of the platform, on which the subjects were lying, were recorded in the frequency range of 2Hz-20Hz. Sinusoidal acceleration amplitude was held constant at 0.5g. The impedance and phase results show that sustained acceleration up to +5G_x stiffens the human body with increasing G_x and shifts the resonance frequency from 6Hz under normal gravity to 8Hz under +2G_x and further up to 11Hz, 13Hz and 15 Hz under +3G_x, +4G_x, and +5G_x respec-

tively. The modulus of impedance grows with centrifuge acceleration. To explain these findings, a multi-degree-of-freedom model is proposed. Its model parameters were computed by way of an optimization procedure. The behavior of the model under sustained acceleration shows an increase of the effective masses near the driving point at the expense of the upper masses. The spring constants of the subsystems near the driving point increase with +G_x. The damping coefficients depend on mass and sustained acceleration. These results show that the human body behaves nonlinearly in an extreme dynamic environment. An expansion of the proposed model to greater complexity is necessary to explain its reactions to forces from other directions and to transients.

IN ORDER TO protect man from the forces of a high energy mechanical environment, his dynamic properties have to be known. As a first approach the human body may be considered as a complex system of masses, springs and dampers. These elements are interconnected and influence each other. If such a system is excited by vibration or impact, internal displacements will result. It has been shown that the relative displacement of the effective masses is the determining factor for the subjective tolerance in the low frequency range.[1]

The dynamic behavior of a simple mass-spring-damper system is determined by the equation of motion:

$$m\ddot{x} + c\dot{x} + kx = f(t) \qquad (1)$$

where:

m = mass
x = displacement
c = damping coefficient
k = spring constant

The solution of this equation is well known[3] but requires that the factors m, c and k are known. They cannot be measured on the human body directly and the assumption of a one-degree-of-freedom system to replace man is certainly a crude simplification. There is however a method which gives a good insight into the dynamic behavior of a complex system. Similar to the measurement of a complex electrical resistance, the mechanical impedance of the human body can be measured. It is defined as the complex ratio of transmitted force and vibrational velocity at the point of excitation:

$$\overline{Z} = \frac{F}{\dot{x}} \qquad (2)$$

where:

$\overline{Z}$ = complex impedance

F = force

$\dot{x}$ = velocity

The impedance and the phase relationship between the two values are frequency dependent.

A peak in the impedance plot indicates resonance. The phase between transmitted force and vibrational velocity is $+90°$ when the system behaves like a pure mass. If an undamped system goes through resonance the phase shifts through $0°$ to $-90°$. So the phase curve versus frequency shows the resonant frequencies, too. Furthermore effective masses, spring constants, and damping factors can be calculated from impedance. Also the relative displacement within the system at the natural frequency can be derived, which finally determines tolerance.

With this method, sitting,[2] standing,[1] and supine[3] humans have been tested. In the supine posture a fundamental resonance around 6Hz was detected. Smaller peaks of the impedance curve show up around 8 and 10Hz. The location of the latter were highly dependent on muscle tension.

Mathematical or mechanical models are widely used to explain the impedance results. Most authors assume a linear, simple, and passive system. Within the range of vibrational acceleration, tolerable for human subjects and occurring in an industrial environment, these assumptions may be satisfied. But for impacts, and certainly in the range of irreversible tissue damage, the presumption of a linear system is invalid. The impedance of a linear system does not change whether excited either by different levels of vibrational acceleration or while being prestressed by sustained acceleration or under transient conditions. Krause and Lange[4] measured a set of different impedance curves of supine pigs while shaking them with different levels of vibrational acceleration. In a previous study[5] we showed that the impedance of sitting humans changes under sustained acceleration on a centrifuge. The resonant frequencies shifted from the original value of 5Hz under normal gravity to 7 and 9Hz under $+2G_z$ and $+3G_z$ centrifuge acceleration. The modulus of impedance increased with G.

Similar results were reported from semi-supine sub-

jects[6] except for a decreasing damping factor in the latter experiment. Weis et al[7] published impedance curves obtained from impact tests and compared them with results from steady state conditions on a shake table. While steady state impedance decreases generally after the first resonance, the impedance measured under transient conditions increased in the higher frequency range. The authors explained these discrepancies by considering inherent nonlinearities in the human body such that the impedance is amplitude and/or velocity direction dependent.

The purpose of this investigation is to determine the nonlinear properties of the supine human body and to develop a simple model which explains its dynamic behavior under low frequency mechanical energy.

MATERIALS AND METHODS

An electrohydraulic shaketable was mounted on a centrifuge. Figure 1 gives an overall view of the machinery. The hydraulic actuator is suspended, freely swinging about a horizontal axis, from the centrifuge arm. The vector of the static acceleration is therefore always perpendicular to the x-axis of the subject. The subject is lying on a stiff platform which is suspended from the actuator by a rigid framework of welded tubing. For reasons of safety the subject was loosely restrained with lap belt and shoulder harness. The cylinder is controlled by an electrohydraulic valve which receives its sinusoidal input voltage from a function generator. The connection between the subject support and the welded framework is via six force cells. Their output is summed, filtered and fed into a recorder. In order to monitor sustained and vibrational acceleration, accelerometers are mounted directly underneath the platform. The radius of the centrifuge to the rotating axis of the shaketable was 5.5m and the distance between table platform and this axis 1.75m.

Ten young healthy male subjects volunteered for this test. Their height and weight are shown in Table I. They underwent the following procedure: First the centrifuge arm was rigidly connected to the floor by two supporting beams. Then the subject lay down, was loosely restrained, and the shaker was started. Measure-
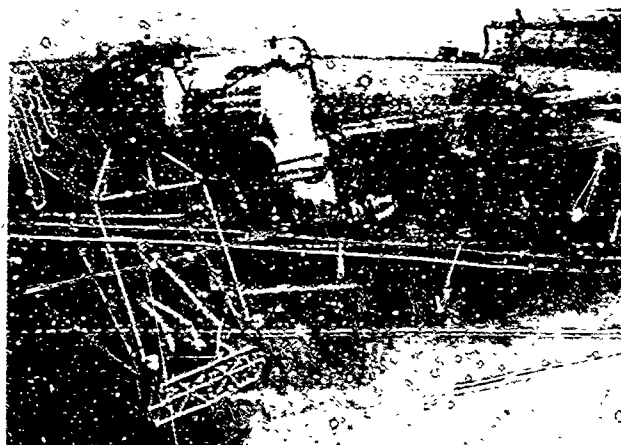

Figure 1. DFVLR Centrifuge with Shake Table and Subject in Position.

TABLE I. HEIGHT AND WEIGHT OF SUBJECTS

No.	Subject	Height (cm)	Weight (kg)
1	B.M.	173	71.00
2	D.V.	178	70.00
3	G.M.	174	80.05
4	G.T.	187	102.00
5	U.U.	183	91.00
6	M.E.	185	73.50
7	M.J.	175	68.00
8	W.D.	181	75.80
9	W.G.	178	71.00
10	W.F.	175	61.00

ments of transmitted force and vibrational acceleration were taken in the range from 2Hz to 20Hz. Acceleration was held constant at 0.5g amplitude. Both signals were electrically filtered. The 2-channel low-pass filter used had adjustable upper cutoff frequencies of 5Hz, 10Hz, 20Hz and 30Hz with a peaked rolloff of 40 dB/octave. The phase lag between the two filter channels never exceeded ±10°. Vibrational acceleration was monitored on a cathode ray oscillograph and was always effective for at least 15 sec before any recordings were made. Then, the same procedure was repeated with the centrifuge running at + 2G$_x$, + 3G$_x$, + 4G$_x$ and + 5G$_x$. A centrifuge run lasted about 10 min which was near the tolerance of most of the subjects. All of them encountered respiratory difficulties at +5G$_x$. They were never so severe that a run had to be terminated.

The force and vibrational acceleration recordings were evaluated by hand. In spite of the fact that the force curve was not always a pure sine wave, it was possible to draw the fundamental frequency with fair accuracy. However the accuracy for the force measurement was not greater than ±10%, and for the phase angle due to the filter error, not greater than ±10°. With respect to the changes due to the individual dynamic properties of the subjects this accuracy was sufficient to demonstrate the effect of sustained acceleration.

To obtain the force transmitted to the human body the acceleration force of the masses between the force cells and the subject had to be subtracted vectorially. The quotient of the resulting force and the velocity of the platform is the measured impedance.

RESULTS

The measured impedance values and phase angles were averaged over all 10 subjects and plotted against frequency. Figure 2 shows the results together with the $m\omega$ - line for a mean subject weight of 76.3 kg. There is one curve for each level of sustained acceleration. In the low frequency range the curves fall together and separate clearly above the first resonance. Impedance values grow with increasing centrifuge acceleration. Three relative maxima can be distinguished up to +3G$_x$ while at +4G$_x$ and +5G$_x$ only two peaks remain. With increasing sustained acceleration the first and second resonances shift to higher frequencies while a third peak remains stationary around 19Hz. The first resonance under normal gravity lies at the well known 6Hz
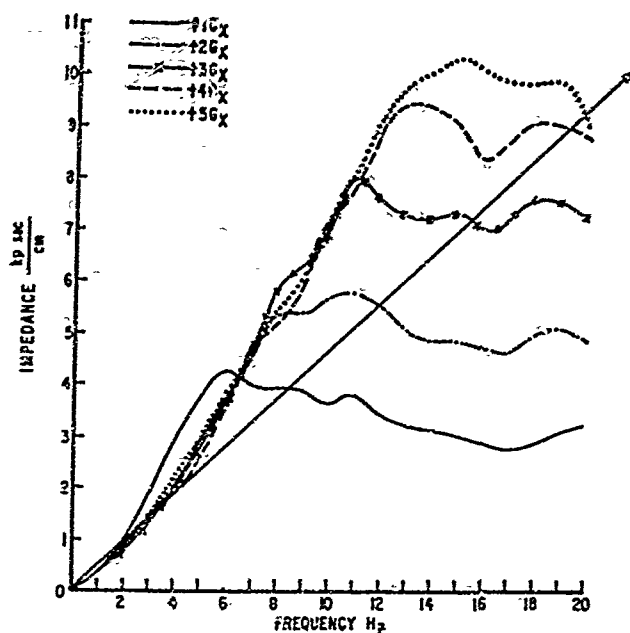

Figure 2. Mechanical Impedance for 1G$_x$ — 5G$_x$ Sustained g, Plotted vs. Frequency.
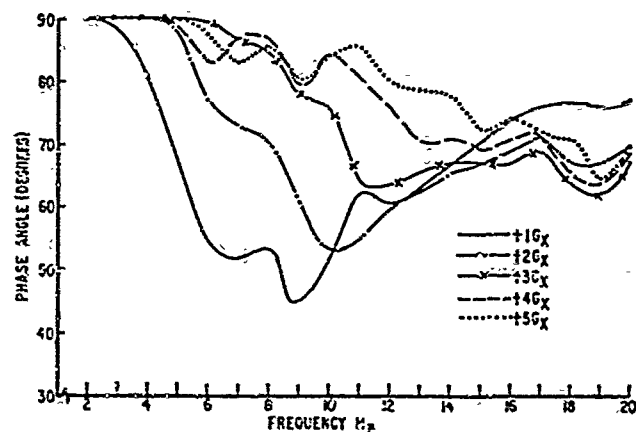

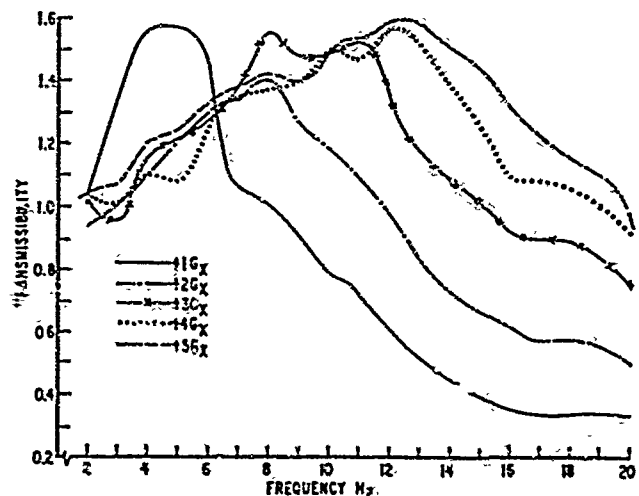Figure 3. Phase Angle between Transmitted Force and Velocity for Five Levels of Sustained Acceleration.

Figure 4. Transmissibility for Five Levels of Sustained Acceleration.

point. It shifts to 8Hz under $+2G_x$ and further on to 11Hz, 13Hz and 15Hz under $+3G_x$, $+4G_x$ and $+5G_x$ respectively.

The phase angle between transmitted force and vibration velocity is shown in Figure 3. It verifies the assertions of the impedance plots. The maximum phase lag is 45°, which is reached at 9Hz under normal gravity.

The ratio of the measured impedance to the mass impedance at the same frequency is called transmissibility. This factor represents the ratio between the force transmitted to a damped and sprung mass and the force which would be transmitted to the same mass without the spring and damper. If its value is greater than 1 the force transmission will be amplified by the spring and damper forces. On the other side, force transmission is diminished when transmissibility is smaller than 1. When it equals 1, springs and dampers have no influence on force transmission. In Figure 4 transmissibility is plotted versus frequency for five levels of sustained acceleration. Under normal gravity a value of 1.6 was calculated at the natural frequency of the human body. Transmissibility decreases slightly to 1.4 under $+2G_x$. For the remaining levels of centrifuge acceleration the maxima also shift to higher frequencies but never exceed the value of 1.6.

The impedance curves and the particular type of phase curves cannot be obtained from a one-degree-of-freedom system. The simplest model which could still reproduce the measured impedance and phase values with good accuracy is shown in Figure 5. Three mass-spring-damper systems are connected to an unsprung mass m_1. The values of the model parameters and their change under static acceleration were determined by an optimization procedure. For small displacement and velocity amplitudes, linear spring and damper characteristics were assumed. This assumption allowed the use of linear equations of motion for the model. The equations of motion were solved and the impedance equations were derived. The model parameters were changed in steps with an optimization procedure until a satisfactory correspondence was established between the experimental impedance and phase values and those calculated for the model. The calculations were done on a digital computer and were repeated for each level of sustained acceleration. Figure 6 shows a comparison between measured and calculated impedance for $+2G_x$ sustained acceleration.

The behavior of the model under the influence of static G shows a change in mass distribution (Figure 7). The three masses which are near to the driving point (m_1, m_{211} and m_{11}) increase, while the upper mass of the two-degree-of-freedom subsystem (m_{212}) becomes smaller. This is an indication for the fact that under sustained acceleration more and more of the supporting tissue of the human body acts like a pure mass. The total sum of masses for each level of centrifuge acceleration also increases, but the total gain stays within the 10% range and is therefore within the overall accuracy of the impedance measurements.
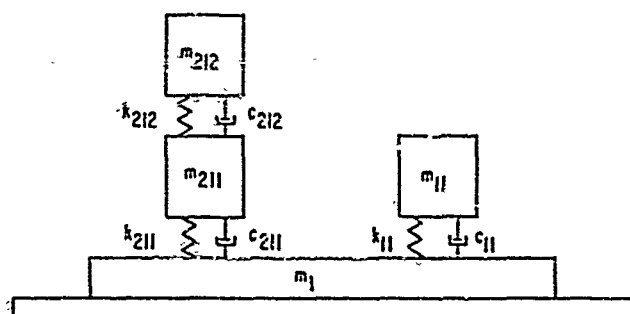

Figure 5. Model for the Supine Human Body.
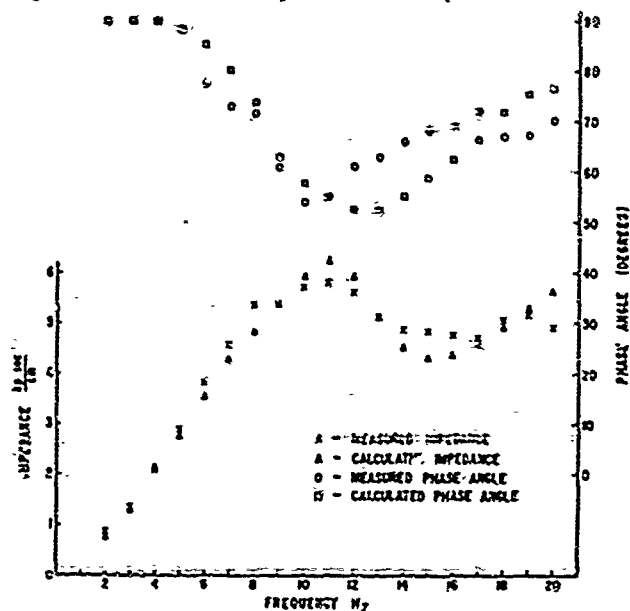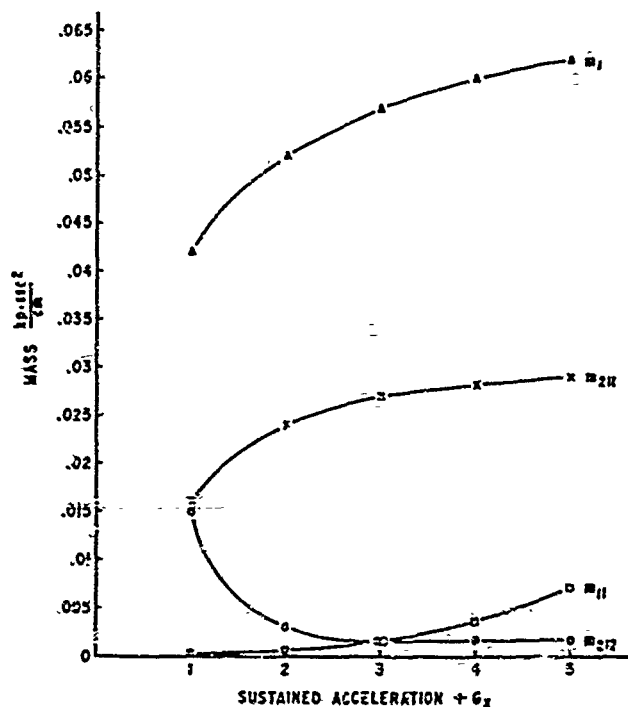

Figure 6. Comparison between Measured and Calculated Impedance Values and Phase Angles for $+2G_x$ Sustained Acceleration.

Figure 7. Changes in Mass Distribution of the Proposed Model under Five Levels of Sustained $+G_x$.

The measured impedance curves show a shift in resonances to higher frequencies. This should be represented by a similar behavior of the springs. While the characteristics of the lower springs (k_{11} and k_{211}) follow this prediction (Figure 8), the upper spring factor, k_{212}, is comparatively small and varies little. This can be explained by assuming nonlinear springs which increase their stiffness with displacement. The static deflection of springs which bear heavy and growing loads increases faster with static acceleration than the static deflection of those springs with a smaller static load. The lower springs of the model, k_{11} and k_{211}, thus reach the stiff sections of their stress-strain-curve faster than the upper springs.

Figure 9 shows the behavior of the three dampers c_{11}, c_{211}, and c_{212} under sustained acceleration. While c_{211} increases linearly with static G, c_{212} decreases between $+1G_z$ and $+2G_x$ and then stays constant around the value of $0.1 \frac{kp\ sec}{cm}$. Damper c_{11} reaches values up to $26 \frac{kp\ sec}{cm}$, which is about two orders of magnitude greater than the value for critical damping. It can therefore be considered as a rather stiff connection. Comparing damping coefficients with their proper masses, a similar behavior is obvious. This is in agreement with the fact that in biological tissue the damping capacity is distributed over the oscillating masses and is not concentrated in discrete elements.

DISCUSSION AND CONCLUSIONS

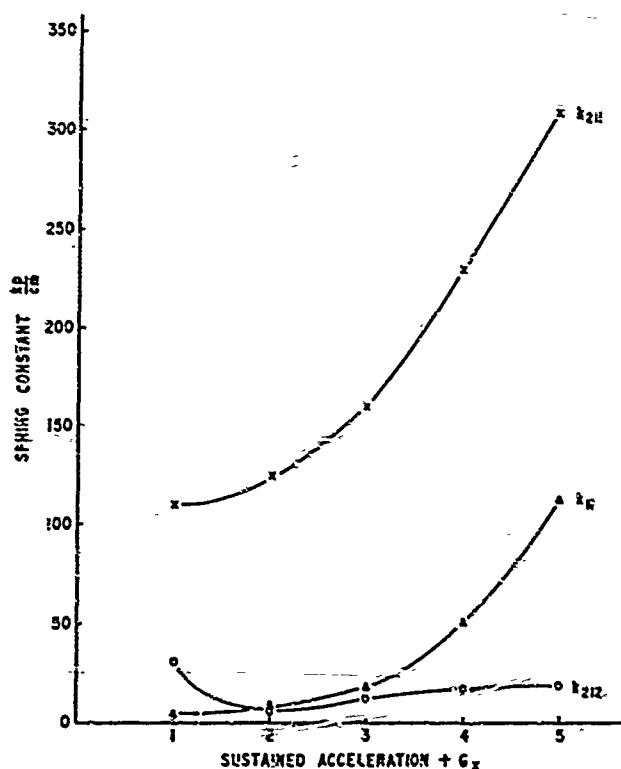There is only one study available which deals with the mechanical impedance of fully supine subjects, published by Edwards and Lange.[3] As they did measurements on two subjects only, a comparison between their findings and our data is not without difficulties. The frequency distribution of their impedance maxima compares well with our findings. Also the modulus of impedance at the first resonance as determined for a 77.8 kg subject with $3.76 \frac{kp\ sec}{cm}$ is in relatively good agreement with our value for a mean subject weight of 76.3 kg where we measured a first maximum of $4.3 \frac{kp\ sec}{cm}$. The main difference of the two studies lies in the behavior of the phase between transmitted force and velocity amplitude. Edwards and Lange's phase shift reaches a maximum of $-30°$ at 10Hz while our phase shift maximum is reached at 9Hz with a value of only $\div45°$. This difference is very important concerning the system parameters for designing a model. If the $-30°$ maximum were true, the supine human body could be replaced by a one-degree-of-freedom system with good accuracy.

In another series of shake table runs which are still unpublished and which were done on a different shake table, with different equipment and subjects and even in another part of the world we could verify our findings. Maybe Edwards' remark that his support structure had a resonance at approximately 18Hz explains this difference. He also looked for nonlinearities of the body system by using different levels of acceleration ampli-
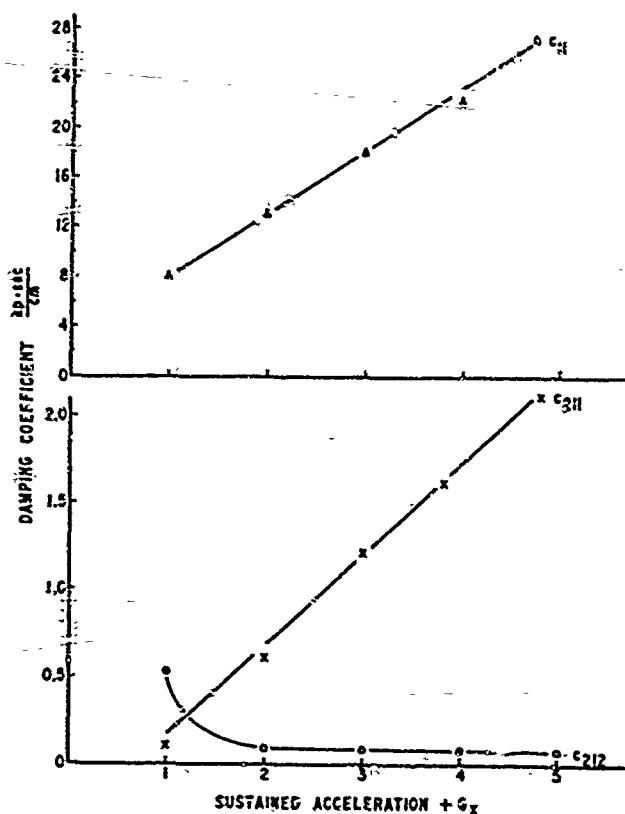
Figure 8. Behavior of the Model Spring Constants under Sustained $+G_x$.

Figure 9. Changes in Damping of the Proposed Model Due to Sustained $+G_x$.

tude. The resulting impedance values decreased with increasing vibrational acceleration, showing that transmitted force is not linearly related to the input amplitude.

In this study, impedance increases with sustained $+G_x$. This is mainly caused by two mechanisms: Firstly, the parameters of the proposed model show a simultaneous increase in the unsprung mass m_1 and in the effective masses near the driving point. This behavior of m_1 is supported by the tendency of the phase angle to deviate to a lesser extent from $+90°$ at higher levels of static acceleration. The unsprung mass is also the main contributor to impedance at frequencies above resonance. This may be explained by the following consideration: The impedance of the system in Figure 5 as a function of angular frequency ω is defined by the equation:

$$\bar{Z} = \left(\frac{1}{\omega^2} - \frac{\tilde{x}_{211}}{a}\right)(c_{211}\,\omega^2 - ik_{211}\,\omega) +$$

$$\left(\frac{1}{\omega^2} - \frac{\tilde{x}_{11}}{a}\right)(c_{11}\,\omega^2 - ik\omega) + i\omega m_1 \quad (3)$$

where: ω = angular frequency; $\bar{x}_{211}$, $\bar{x}_{11}$ = complex displacement amplitude of masses m_{211} and m_{11} respectively; a = acceleration amplitude of shake table; $\bar{Z}$ = complex whole body impedance; $i = \sqrt{-1}$.

The partial derivative of $\bar{Z}$ with respect to the unsprung mass m_1 is: $\frac{\delta \bar{Z}}{\delta m_1} = +i\omega.$ \quad (4)

Therefore the impedance increases with the unsprung mass m_1, particularly at great values of ω. Also it can be seen, that it is the imaginary component of the impedance vector which changes with m_1. This change occurs in the positive direction, thus shifting the phase of the impedance vector toward $+90°$.

Secondly, the contribution of elastic systems decreases above resonance. This is also shown by the transmissibility curves in Figure 4. Furthermore, the human body stiffens under centrifuge acceleration, as mentioned before, which results in higher resonant frequencies. This also leads to a greater modulus of impedance in this frequency range.

Until now there are no other data available which describe mechanical impedance of completely supine humans, measured under sustained $+G_x$ on a centrifuge. But there is Vykukal's work[6] on semisupine humans, done under the same conditions. He too found the modulus of impedance increasing with static $+G_x$ and a shift of resonances to higher frequencies. Obviously, the absolute values cannot be compared, because of the different body position. He also explains his findings by increased stiffness of the body but postulates an overall decrease in damping for an explanation. Our model shows that this is only true for those body parts which are located distant from the driving point. They diminish their damping together with their effective mass, while all other subsystems show an increase in effective mass, spring constant, and damping coefficient simultaneously.

We are aware of the fact, that the presented model is still a crude simplification of the human body. It gives no indication about the coupling of the parallel subsystems. Without these couplings the model represents only the response to forces in one direction. The number of degrees of freedom is too small to reproduce the measured responses in every detail. This shortcoming also results in the fact that the four masses and the three mass-spring-damper systems are in no way related to any body segments, organ groups, or organs of the live human body. Location and magnitude of internal stresses, which finally determine tolerance, can only be predicted if the model is expanded to a greater degree of complexity. Besides an increase in the number of degrees of freedom it is necessary to consider continuously nonlinear elements. This is also a prerequisite for the simulation of the response to transients.

In spite of this shortcoming, the following conclusions can be drawn from this study:

1) Supine acceleration stiffens the human body in the $+G_x$ direction and the spring factors of the lower systems increase.

2) The resonant frequency of the supine human body shifts to greater values with centrifuge acceleration.

3) The effective mass of body parts near the support increases, while the effective masses of distant parts diminish.

4) The damping coefficients depend on sustained G_x and mass distribution.

These findings clearly indicate that nonlinearities have to be considered when the response of the human body to extreme dynamic environments has to be evaluated.

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