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Human Responses to Vibration Therapy

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Abstract—**This paper outlines the progress that has been made in a study on human body dynamics during vibration therapy. At the Toronto Rehabilitation Institute, vibration therapy has been applied to spinal cord injured (SCI) patients in an effort to maintain the patients' bone density. Some clinical trials have not been successful, and thus a better understanding of human responses to vibration therapy is required to ascertain if and how it can be applied to maintain bone density in SCI subjects. Experiments with SCI and healthy subjects were conducted to determine the accelerations present in the lower extremities during vibration therapy. The results showed negligible differences between the responses of SCI and healthy subjects, but considerable differences between the responses of subjects with different body types. A mathematical model of a standing subject was also developed, and theoretical predictions using the model were found to match experimental data reasonably.**

Keywords—**Bone Density, Vibration Therapy**

I. Introduction

Doctors and rehabilitation therapists at the Lyndhurst Centre, Toronto Rehabilitation Institute (TRI) have conducted several clinical trials investigating the feasibility of using vibration therapy to regulate bone density in the lower extremities of spinal cord injured patients who have little or no motor control below the waist. TRI has an apparatus that secures the subject in a standing position with the feet bearing most of the subject's weight. Vibrations are then applied to the subject via the platform that they are standing on. The TRI apparatus vibrates the subject in a horizontal direction, unlike previous studies of this kind that applied vertical vibrations [1]. Because the apparatus vibrates the subject horizontally and constrains the body at certain points, the propagation of vibrations through the body may differ significantly from situations where the induced vibrations are vertical. The purpose of this study is to ascertain the degree to which differences in the type of therapy applied to a patient affect the parameters that are important in stimulating increases in bone density through vibration therapy, and to determine how different subjects will respond to vibration therapy.

A number of studies have been done to identify the most important parameters in bone remodeling through mechanical stimulation. Several studies conducted with animal subjects in the 1980's demonstrated that the strain magnitudes [2], strain rates [3], and strain distributions [4] present in bones have large effects on bone density. These parameters are functions of the forces present in the lower extremities.

To examine the forces in the legs during different types of vibration therapy, a lumped mass model has been developed using physiological data from literature. In the past, linear lumped mass models have been used to predict human responses to both uni-directional vertical vibrations [5] and multi-directional vibrations [6]. Experiments were also conducted with healthy and spinal cord injured subjects to determine the accelerations present in the lower extremities during different types of vibration therapy. The experimental results were analyzed to provide a better understanding of the mechanical properties of the human body during vibration therapy. These results were also used to calibrate the model in order to produce more accurate theoretical predictions.

II. Modeling Equations

The rigid body model was developed to predict the forces acting on the femur and tibia bones during vibration therapy. Modeling equations were developed based on treating the body as a system of four lumped masses connected by linear translational and linear rotational springs and dampers connected in parallel. The model is shown in Figure 1.

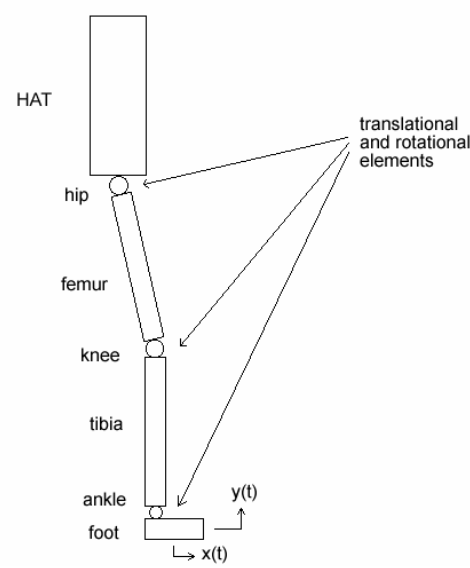

Fig. 1. Model for Rigid-Body Analysis

The foot, tibia, and femur are each represented by lumped masses. The head, arms, and trunk (HAT) are all lumped together as a single mass.

Using this model, the kinetic energy of a segment T_a, the potential energy of a segment P_a, the spring energy stored in a joint U_a, and the energy dissipated in a joint R_a can be expressed as

$$T_a = \frac{1}{2}(m_a \bar{v}_a^2 + J_{a,cg}\omega_a^2) \qquad (1)$$

$$P_a = m_a q y_{a,cg} \qquad (2)$$

$$U_a = \frac{1}{2}\Big(k_{x,a}(x_{1,a} - x_{2,b})^2 + k_{y,a}(y_{1,a} - y_{2,b})^2 + k_{\theta,a}(\theta_a - \theta_b)^2\Big) \qquad (3)$$

$$R_a = \frac{1}{2}\Big(c_{x,a}(\dot{x}_{1,a} - \dot{x}_{2,b})^2 + k_{y,a}(\dot{y}_{1,a} - \dot{y}_{2,b})^2 + k_{\theta,a}(\dot{\theta}_a - \dot{\theta}_b)^2\Big) \qquad (4)$$

where the stiffness and damping co-efficients (all k and c values) are constants from literature, m_a and $J_{a,cg}$ are the mass and mass moment of inertia of a about the centre of gravity, and $\bar{v}_a$ and ω_a are the planar and angular velocities of a. The points $x_{1,a}$, $y_{1,a}$, $x_{2,b}$, and $y_{2,b}$ are the displacements of the bottom and top of the segments a and b respectively from their quiet standing positions. The $y_{a,cg}$ term is the displacement of the centre of gravity of the segment a from the quiet standing position. The energy terms (1) to (3) can be combined to establish the Lagrangian of the segment, which is the difference between the kinetic and potential energies, as follows:

$$L_a = T_a - U_a - P_a \qquad (5)$$

The R_a term is known as the Rayleigh Energy Dissipation Function. The L_a and R_a terms for each segment can be summed together, and the resulting terms L and R are the Lagrangian and the Rayleigh Energy Dissipation Function for the entire system.

Using basic trigonometry and the small angle approximations $\sin\theta = \theta$ and $\tan\theta = \theta$, the Lagrangian for the system can be expressed in terms of the x and y displacements of the center of mass of each segment relative to the quiet standing position, and the angle of rotation of each segment relative to the quiet standing position. The following form of Lagrange's equation can then be used to obtain global equations of motion

$$\frac{d}{dt}\left(\frac{\partial l}{\partial \dot{\mathbf{q}}}\right) - \frac{\partial L}{\partial \mathbf{q}} = -\frac{\partial R}{\partial \dot{\mathbf{q}}} + \mathbf{Q} \qquad (6)$$

where $\mathbf{q}$ is a vector containing the state variables of the system, and $\mathbf{Q}$ is a vector of forcing functions acting on the system. The equations resulting from (6) were simulated in state-space form in Matlab's Simulink environment.

The constants in the equations of motion were all obtained from accepted literature. All of the mass (m) and inertia (J) constants were calculated based on subject size and basic anthropometry [7]. The vertical spring and damping coefficients were obtained from previous work in human vibration response analysis [8]. Since no experimentally measured values for horizontal spring and damping coefficients in the lower extremities could be found, these values were inferred based on the vertical spring and damping values, and previous work done involving vertical and horizontal spring and damping coefficients linking slender body segments [8,6]. The ankle and knee rotational spring and damping coefficients were taken from studies on passive moments in the lower extremities [9,10]. The rotational spring coefficient at the hip was taken from a study on passive hip moments [11], and the hip rotational damping coefficient was calculated based on the spring coefficient and a damping ratio of 0.2, which is typical for rotation about the joints in the lower extremities [9, 10].

III. Experimentation

The experiment was conducted with two healthy male subjects, and two spinal cord injured (SCI) male subjects. The subjects were able to stand independently, were roughly 180 cm tall, and weighed between 61 kg and 93 kg. The subjects were subjected to vibrations using two vibration platforms, a device manufactured by Exogen and a device manufactured by the Bloorview MacMillan Children's Centre specifically for TRI. The Exogen device was used to apply the vertical vibrations at a fundamental frequency of 30 Hz. The device was designed to have a peak-to-peak displacement of 0.1 mm. The TRI device was used to apply the horizontal vibrations at a variety of frequencies. The device was designed to have a peak-to-peak displacement of 3.0 mm. Accelerations were measured in x- y- and z-directions using 12 Atech Instruments 3041A4 accelerometers. The accelerometers were powered by an Atech Instruments 12 channel current source power unit. The data acquisition card was connected to a PC, and data was recorded at a sampling frequency of 1000 Hz. The constraining device, which ensured safety and stability of SCI subjects during the previous clinical trial, held the subjects in place near or at the hip, depending on specific adjustments and the height of the subject. The device is made of an aluminum frame, a wood platform and a foam brace.

The following procedure was performed with each subject for each different type of therapy separately.

1. The accelerometers were mounted on three sides of four square mounting blocks.
2. The mounting blocks were attached to the subject's shank, upper thigh, and hip with string and two-sided tape.
3. While the subject was being assisted for balance, the vibrating device was turned on. The assistance was withdrawn once the subject was comfortable on the vibrating device.
4. Once the system had reached steady-state (this happened within seconds), the accelerations of the base plate, shank, thigh, and hip were recorded via the accelerometers for approximately 60 seconds.

RMS acceleration analysis and frequency domain analysis using the Fast Fourier Transform (FFT) were done on each set of data in Matlab. Foot to tibia and foot to femur RMS acceleration ratios were also calculated for each subject

IV. Results

The experimental results showed behavior that was generally as expected. The magnitude of the RMS acceleration was typically shown to decrease from segment to segment moving from the foot to the upper body, and the high frequency content of the acceleration signals was strongly attenuated by successive body segments in all of the subjects.

When SCI and healthy subject data were compared, the results generally showed negligible differences in the mechanical properties of these two subject groups. RMS acceleration ratios varied much more between subjects with different body types than between healthy and SCI subjects. All of the subjects displayed similar frequency responses, regardless of body type or health status.

Almost completely uniformly, the data showed an *increase* in the foot to tibia and foot to femur acceleration ratio values when the subject was constrained, for both healthy subjects and for SCI subject No. 2. The most notable exceptions to this trend were the measured hip accelerations. These accelerations were shown to decrease when the constraining device was applied in most cases. In contrast to the healthy subjects and SCI subject No. 2, SCI subject No. 1 showed uniform decreases in acceleration ratio values during constrained therapy.

The FFT data showed that the frequency content of the input and output signals was very similar. Although some frequencies that were present in the input signal were not present at the same magnitude in the output signals, there were few, if any, frequencies present in the output signals that were *not* present in the input signal.

When simulated with the recorded input signals, the model gave results that were consistent with expected behavior, and reasonably consistent with the experimental data. Completely uniformly, the model did not remove enough high frequency content from the input signal and did not attenuate the input signal enough to match the experimental data exactly. The damping coefficients of the model were modified in an attempt to reduce high frequency content in the output signals, and in some cases this significantly improved the model predictions. Figure 2 shows typical experimental results overlaid on top of results from a model simulation.

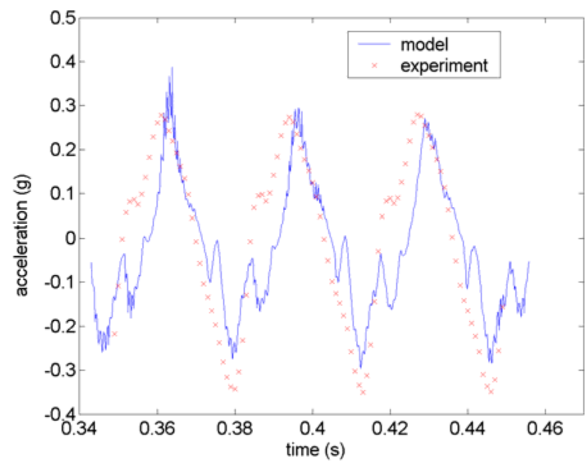

Fig. 2. Acceleration of femur, healthy subject.

The results also showed that the z-direction accelerations (horizontal accelerations causing side-to-side motion of the body) present in the system were significant during all types of vibration therapy. In some cases, z-direction accelerations were larger than x- or y-direction accelerations. Acceleration ratios in all directions were on the same order of magnitude for all subjects. Accelerations in all directions were smaller during therapy with the Exogen device.

V. Conclusion

The experimental results demonstrate that although subject health status was a negligible factor in human responses to vibration therapy (i.e. negligible differences between able-bodied subjects and spinal cord injured subjects), subject body type was a significant factor. The results also show that responses to therapy with and without the constraining device and responses to therapy with the two different devices were quite different.

The FFT analysis of the acceleration signals indicates that the human body can be represented as a linear mechanical system during vibration therapy, because the frequency content of the input and output signals was similar. RMS analysis of the foot to tibia and foot to femur acceleration ratios in each direction suggest that an uncoupled model would be adequate to predict the response of the human body to vibration therapy. Uncoupled models have been shown to reasonably match experimental data when predicting human responses to vibration [6]. Development of a three-dimensional uncoupled model with flexible tibia and femur segments is being conducted to produce preliminary estimates of the strain fields present in the tibia and femur during vibration therapy.

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