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A STUDY OF TIBIAL VIBRATIONS DURING THE TIBIAL RESECTION PROCEDURE OF TOTAL KNEE ARTHROPLASTY

by

ANGELA L. MORRIS

A THESIS SUBMITTED IN PARTIAL FULFILLMENT OF THE REQUIREMENTS FOR THE DEGREE MASTER OF SCIENCE

APPROVED, THESIS COMMITTEE:

[Signatures]

P. Spanos, Professor
Mechanical Engineering

J.E. Akin, Professor
Mechanical Engineering

T.P. Harrigan, Adjunct Professor
Mechanical Engineering

Houston, Texas

May, 1996
Abstract

A Study of Tibial Vibrations During The Tibial Resection Procedure of Total Knee Arthroplasty

by

Angela L. Morris

Vibrations of the tibia induced during the tibial resection procedure of Total Knee Arthroplasty are studied. Experimental modal analysis, performed on cadaver specimens, shows that natural frequencies of the in-situ tibia occur at 250, 350, and 650 Hz. A finite element beam model of the tibia, which consists of 3-D elements with varying material and geometrical properties, is used. The model indicates that the first bending mode of the in-situ tibia may occur below 100 Hz. The cadaver specimens are then cut with an oscillating saw to observe the effect of the oscillating saw on the vibrational characteristics. When the tibial resection is performed with the oscillating saw, it is shown that the oscillating saw excites the resonant frequencies of the in-situ tibia.
Acknowledgments

I would like to thank Dr. Timothy Harrigan and the University of Texas Health Science Center Department of Orthopaedic Surgery for providing me with the opportunity to perform this study. Thanks to Dr. Pol Spanos and Dr. J.E. Akin for their guidance and support throughout this study. I would also like to thank the UTHSC morgue for consistently supplying the cadaver specimens. Additionally, thanks to Kyle Wilson and Zimmer for providing the use of the oscillating saw and for assisting with the tibial resection. I would also like to express my gratitude to the National Science Foundation for funding me throughout this project.

Finally, I would like to thank my wonderful husband, Mark, for his selfless love, support, and friendship. You make my life complete.
# Table of Contents

<table>
<thead>
<tr>
<th>Chapter</th>
<th>Title</th>
<th>Page</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>Introduction and Background</td>
<td>1</td>
</tr>
<tr>
<td>1.1</td>
<td>Introduction</td>
<td>1</td>
</tr>
<tr>
<td>1.2</td>
<td>Background</td>
<td>1</td>
</tr>
<tr>
<td>1.2.1</td>
<td>Total Knee Arthroplasty</td>
<td>1</td>
</tr>
<tr>
<td>1.2.2</td>
<td>Methods of Fixation</td>
<td>2</td>
</tr>
<tr>
<td>1.2.3</td>
<td>Tibial Resection Surgical Procedure</td>
<td>3</td>
</tr>
<tr>
<td>1.2.4</td>
<td>Importance of Accuracy</td>
<td>5</td>
</tr>
<tr>
<td>1.3</td>
<td>Purpose of this Study</td>
<td>5</td>
</tr>
<tr>
<td>2</td>
<td>Previous Related Work</td>
<td>7</td>
</tr>
<tr>
<td>3</td>
<td>Methods</td>
<td>9</td>
</tr>
<tr>
<td>3.1</td>
<td>Modeling the Tibia</td>
<td>9</td>
</tr>
<tr>
<td>3.1.1</td>
<td>Previous Related Work</td>
<td>9</td>
</tr>
<tr>
<td>3.1.2</td>
<td>Simple Tibia Model</td>
<td>11</td>
</tr>
<tr>
<td>3.1.3</td>
<td>Beam Theory</td>
<td>14</td>
</tr>
<tr>
<td>3.2</td>
<td>Experimental Modal Analysis</td>
<td>16</td>
</tr>
<tr>
<td>3.2.1</td>
<td>Test Specimens</td>
<td>16</td>
</tr>
<tr>
<td>3.2.2</td>
<td>Instrumentation</td>
<td>17</td>
</tr>
<tr>
<td>3.2.3</td>
<td>In-Situ Impact Testing</td>
<td>17</td>
</tr>
</tbody>
</table>
Table of Contents (cont'd)

<table>
<thead>
<tr>
<th>Section</th>
<th>Page</th>
</tr>
</thead>
<tbody>
<tr>
<td>3.2.4 Bone Cutting</td>
<td>17</td>
</tr>
<tr>
<td>3.2.5 Data Processing</td>
<td>18</td>
</tr>
<tr>
<td>Chapter 4 - Results</td>
<td>22</td>
</tr>
<tr>
<td>4.1 Impact Testing</td>
<td>22</td>
</tr>
<tr>
<td>4.2 Beam Model</td>
<td>27</td>
</tr>
<tr>
<td>4.3 Bone Cutting</td>
<td>30</td>
</tr>
<tr>
<td>Chapter 5 - Discussion</td>
<td>33</td>
</tr>
<tr>
<td>5.1 Impact Results</td>
<td>33</td>
</tr>
<tr>
<td>5.2 Beam Model</td>
<td>35</td>
</tr>
<tr>
<td>5.3 Bone Cutting</td>
<td>38</td>
</tr>
<tr>
<td>Chapter 6 - Concluding Remarks</td>
<td>40</td>
</tr>
<tr>
<td>References</td>
<td>41</td>
</tr>
<tr>
<td>Appendix A: Accelerometer and Impact Hammer Specifications</td>
<td>44</td>
</tr>
<tr>
<td>Accelerometer and Impact Hammer Specifications</td>
<td>45</td>
</tr>
<tr>
<td>Appendix B: X-Accelerance, Impact Data Set 2</td>
<td>46</td>
</tr>
<tr>
<td>Y-Accelerance, Impact Data Set 2</td>
<td>47</td>
</tr>
<tr>
<td>Appendix C: X,Y,Z Accelerance, Sawing Data Set 2</td>
<td>48</td>
</tr>
</tbody>
</table>
List of Tables

Table 3-1. Variations in bone material properties................................. 13
Table 4-1. Natural frequencies of composite simple beam model.................. 28
Table 5-1. Natural frequencies of refined tibia model................................ 35
Table 5-2. Calculated natural frequencies of tibia after material properties changes... 38
List of Figures

<table>
<thead>
<tr>
<th>Figure</th>
<th>Page</th>
</tr>
</thead>
<tbody>
<tr>
<td>Figure 1-1. Smooth-stem TKA component</td>
<td>3</td>
</tr>
<tr>
<td>Figure 1-2. Porous-coated TKA prosthesis component</td>
<td>3</td>
</tr>
<tr>
<td>Figure 1-3. Extramedullary and intramedullary guides aligned on tibia</td>
<td>4</td>
</tr>
<tr>
<td>Figure 1-4. An oscillating saw</td>
<td>4</td>
</tr>
<tr>
<td>Figure 2-1. Frequency response function of a femur</td>
<td>7</td>
</tr>
<tr>
<td>Figure 3-1. Geometrically-correct finite element model of the tibia</td>
<td>10</td>
</tr>
<tr>
<td>Figure 3-2. Material distribution along the tibial axis</td>
<td>12</td>
</tr>
<tr>
<td>Figure 3-3. Circular cross-section of beam element</td>
<td>12</td>
</tr>
<tr>
<td>Figure 3-4 A simple beam in transverse vibration</td>
<td>15</td>
</tr>
<tr>
<td>Figure 3-5. Block diagram of the experimental setup</td>
<td>18</td>
</tr>
<tr>
<td>Figure 3-6. The FRF of a typical 3 DOF system</td>
<td>19</td>
</tr>
<tr>
<td>Figure 3-7. Decay record of a damped system</td>
<td>20</td>
</tr>
<tr>
<td>Figure 4-1. Raw impact data</td>
<td>23</td>
</tr>
<tr>
<td>Figure 4-2. Z-Accelerance during impact testing, showing noise</td>
<td>23</td>
</tr>
<tr>
<td>Figure 4-3. X-Accelerance and X-Inertance FRF</td>
<td>24</td>
</tr>
<tr>
<td>Figure 4-4. Y-Accelerance and Y-Inertance FRF</td>
<td>25</td>
</tr>
<tr>
<td>Figure 4-5. X-Accelerance plots at five impact points</td>
<td>26</td>
</tr>
<tr>
<td>Figure 4-6. Y-Accelerance at five impact points</td>
<td>27</td>
</tr>
<tr>
<td>Figure 4-7. Composite beam model of the tibia</td>
<td>28</td>
</tr>
<tr>
<td>Figure 4-8. First bending mode of model in X-Y plane</td>
<td>29</td>
</tr>
<tr>
<td>Figure 4-9. First bending mode of model in X-Z plane</td>
<td>29</td>
</tr>
</tbody>
</table>
List of Figures (cont'd)

<table>
<thead>
<tr>
<th>Figure</th>
<th>Page</th>
</tr>
</thead>
<tbody>
<tr>
<td>Figure 4-10. Raw data during cutting</td>
<td>30</td>
</tr>
<tr>
<td>Figure 4-11. X-Accelerance of tibia during cutting</td>
<td>31</td>
</tr>
<tr>
<td>Figure 4-12. Y-Accelerance of tibia during cutting</td>
<td>31</td>
</tr>
<tr>
<td>Figure 4-13. Z-Accelerance of tibia during cutting</td>
<td>32</td>
</tr>
<tr>
<td>Figure 5-1. FFT of impact hammer impulse</td>
<td>36</td>
</tr>
<tr>
<td>Figure 5-2. X-Accelerance of Tibia During Impact</td>
<td>37</td>
</tr>
<tr>
<td>Figure 5-3. Y-Accelerance of Tibia During Impact</td>
<td>37</td>
</tr>
</tbody>
</table>
Chapter 1
Introduction and Background

1.1 Introduction

In un cemented total knee arthroplasty (TKA) it is essential that the bone and implant achieve very close contact so that the bone tissue can grow into the porous coated surface of the implant. This bone ingrowth creates the adhesion that is desired. For the tibial component of TKA, the tibial surface is created by sawing off the top 8-10 mm of the tibia with an oscillating saw. If this surface is not flat enough, the desired bone ingrowth may not occur. It has been shown that some tibial surfaces that are created by cutting with an oscillating saw are not flat enough to allow this bone ingrowth (Toksvig-Larsen et. al, 1994). There is no definitive reason for this non-flat surface; some factors may be the inherent tool chatter from the inexact cutting instrumentation and the vibrations that are induced in the tibia by the oscillating saw. The latter possibility can be addressed by performing modal analysis on a tibia in situ and then examining the vibrational response of the tibia when the cutting is performed.

1.2 Background
1.2.1 Total Knee Arthroplasty

Total Knee Arthroplasty (TKA) is a surgical procedure in which the articulating surfaces of the distal femur, patella, and proximal tibia are replaced by artificial materials. The procedure is generally performed on patients whose bones have deteriorated and who experience pain, stiffness, and deformity in their knee (Laskin, 1991). There are three basic cutting procedures performed during a TKA operation: tibial resection, femoral resection, and patellar resection. During these resections the bones are cut using complex instruments and cutting guides so that the implant components'
alignment and fit are precise. The tibial resection is the procedure that is examined in this study.

1.2.2 Methods of fixation

There is a continuing debate on the best method of implant fixation. There are two basic kinds of implants: cemented and noncemented.

TKA components were initially fixed to the bone with bone cement, which is composed of poly-methylmethacrylate (PMMA). Fixation was achieved by the irregular configuration of the bony surface and penetration of the cement into the microstructure of cancellous bone (Laskin, 1991). The components used in cemented arthroplasty have smooth stems and can be seen in Figure 1-1. Problems have arisen with the use of PMMA such as fragmentation, thermal necrosis of adjacent bone, and immunological defects in leukocytes (decreased leukotaxis and phagocytotic rates) (Laskin, 1991). There is also concern about the long-term performance of PMMA. Under cyclic loading and upon curing, cracks can occur at the bone-cement-implant interface system and can propagate. Failure at the bone-cement or cement-implant interface can occur, which leads to relative motion between bone and cement or implant. This movement causes bone resorption, fibrous tissue formation, and generation of particulate debris (Goldberg, 1991).

Because of the problems associated with PMMA and cemented implants, research on noncemented implants began. The noncemented implants have porous metallic coatings, such as sintered beads or fiber mesh, that are applied to solid metal substrates. These porous coatings comprise a system that provides the required mechanical properties and appropriate surface for the ingrowth of the bone tissue (Haddad et. al, 1987). A porous-coated implant is shown in Figure 1-2. Uncemented TKA is less forgiving and more
exacting than cemented TKA. But, it is also less troublesome because one avoids the steps of bone preparation for cementing, cement clearing, and curing (Krackow, 1990). Despite their potential for better long-term results given the elimination of failure modes associated with the PMMA, their long-term success is not yet proven (Haddad et. al, 1987). Bony in growth into the porous-coated implants depends on the design of the prosthesis, the accuracy of the prosthesis fit, and the accuracy of limb alignment.

![Figure 1-1. Smooth-stem TKA component (Niwa et. al, 1988).](image1)

![Figure 1-2. Porous-coated TKA prosthesis component (Niwa et. al, 1988).](image2)

1.2.3 Tibial Resection Surgical Procedure

The tibial resection is a procedure in which the proximal tibia is cut with an oscillating saw essentially perpendicular to its anatomical-mechanical axis. This cut creates a
surface on which a noncemented tibial component sits. The cut is made by first aligning an extramedullary or intramedullary guide with the tibial shaft, as shown in Figure 1-3. Once the alignment rod has been properly placed, the tibial resection guide, either a block or a slotted guide, is affixed to it and lowered to a reference cutting mark. Once the cutting guide is in place, the tibia is cut with an oscillating saw. An oscillating saw is shown in Figure 1-4. Some surgeons "finish off" the proximal tibia cut with a rotary planer. But there is a risk that too much peripheral bone could be removed by the common leaning side to side during the planing process (Krackow, 1990).

Figure 1-3. Extramedullary and intramedullary guides aligned on tibia (Inman, 1991).

Figure 1-4. An oscillating saw (Stryker, Inc.).
1.2.4 Importance of Accuracy

The success of cementless TKA is highly dependent on the appropriate contact between the machined bone of the proximal tibia and the prosthesis. Gaps at the interface between the porous surface and host bone decrease the amount of bone ingrowth and the strength of fixation (Goldberg, 1991). One study revealed that new bone will grow up to and within the porous structure of an implant at an ideal rate and degree of maturity and mineralization if the gap widths are 0.5 mm or less (Sandborn et. al, 1988). It has been shown that the current method of the tibial resection can create gaps between the implant and bone that would hinder this bone ingrowth. In one study (Toksvig-Larsen et. al, 1994), tibial bone cuts were prepared, using an oscillating saw, for noncemented tibial prostheses. The flatness of the cuts was quantified using sterilized dental imprint material and measured by Zeiss Universal Centres 850. A clinically "flat surface" was found to be uneven. The distance between the uppermost and lowermost points was 1.71 mm. The standard deviation of the measuring points was 0.26 mm (0.16 - 0.38 mm). Therefore, gaps were created between the bone and prosthesis interface that were large enough to prevent bone ingrowth when using noncemented fixation. A possible explanation for the non-flat surface could lie in the ability of the surgeon to accurately cut the bone so as to achieve a flat surface. This fact could be attributed to the vibrations induced in the tibia during the cutting process.

1.3 Purpose of This Study

As mentioned earlier, the success of cementless TKA is highly dependent on the appropriate contact between the prosthesis and the machined bone. This study concentrates on the procedure of the tibial resection and the possibility that the oscillating saw induces vibrations in the tibia by exciting its natural frequencies. A finite element model of the tibia is created, and the model is verified by performing modal analysis.
experiments on a tibia *in situ* Then the response of the tibia is examined during the resection cutting process.
Chapter 2
Previous Related Work

This work is similar to a study (Haste, 1994) that focused on the specific concern of the vibration of the femur during the reaming procedure which is done to prepare the femoral canal for implant insertion in a total hip arthroplasty. The dynamic behavior of the femur, its bending in the anterior-posterior (A-P) and medial-lateral (M-L) directions and rotation about the anatomical axis, was studied through the measurement of accelerometer response to recorded impulses and surgical reaming. The experiments were performed on cadaveric human legs. Figure 2-1 shows the frequency response function from a femur during an impulse test. The peaks in the graph indicate a natural frequency (resonant frequency).

Figure 2-1. Frequency response function of a femur (Haste, 1994).
Since the accelerometers measured acceleration, the displacement components were obtained by dividing the acceleration frequency components by $\omega^2$ ($\omega$ is the radian frequency). The frequency data showed that frequencies above 1 kHz corresponded to displacements of less than 10 microns, which is insignificant. Therefore, the first one or two natural frequencies (<1 kHz) was of most interest.
Chapter 3
Methods

3.1 Modeling the Tibia

3.1.1 Previous Related Work

Mathematical models of the human tibia have been proposed by several authors (Cornellissen et. al, 1986; Christensen et. al, 1986; Collier et. al, 1987) to study its mechanical resonances. The objectives of most of these models were to provide quantitative measures of the state of fracture healing. These models generally consisted of two elastic Euler-Bernoulli beams connected by a region of lower stiffness material that represented the fracture site. Finite element models of long bones have been used by other authors (Hobatho, 1991; Collier, 1982; Hight, 1980; Thomsen, 1990) to study the effects of geometrical properties, mass distribution, pre-twist, shear deformation, rotary inertia, and boundary conditions on the resonant frequencies.

Thomsen (1990) first modeled the tibia as a straight, twisted, non-uniform Timoshenko beam, made up of three different materials: compact bone, cancellous bone, and bone marrow. The partial differential equations describing the model were discretized by a finite element procedure, using the 2-node, 12 degree-of-freedom, constant cross-section Timoshenko beam element. Each beam element was described by 32 parameters characterizing material and geometry. The material parameters were: Young's modulus, shear modulus, and density of the three materials. The geometrical parameters were element length, axial element position, cross-sectional area, principal moments of inertia, rotation of principal axes, Timoshenko shear coefficients, and a torsional stiffness factor. The calculated first seven natural frequencies were within six percent of the measured natural frequencies via modal analysis of an excised bone. Thomsen also developed a more simple model, based on uniform beam theory, that could
be used to calculate the few lowest vibrations modes accurately. This simplified approach would be to model the tibia as a uniform Timoshenko beam, using cross-sectional properties at tibial midshaft (Thomsen, 1990). To accurately reproduce the first two vibration modes and frequencies, shear deformation, rotational inertia, axis twist, and the stiffness of cancellous bone could be neglected. The influence of the cancellous bone was simulated by lumped point masses at the extremities.

Hobatho et. al (1991) developed a geometrically accurate, three-dimensional finite element model of the tibia. The geometry was constructed by digitizing CT scans to develop contours. Adjoining surfaces were created between contours to define the outer surfaces of the tibia. The geometry was meshed automatically and manually by using hex elements. The material properties considered were: Young's moduli of cortical and cancellous bone, and mass densities of cortical and cancellous bone. The complete FE model can be seen in Figure 3-1.

![Figure 3-1. Geometrically-correct finite element model of the tibia (Hobatho et. al, 1991).](image-url)
When the calculated natural frequencies were compared to measured data from modal analysis of a dry excised tibia, the finite element model had predicted frequencies with a mean percent relative error of three percent.

### 3.1.2 Simple Tibia Model

For the first part of this study, a simple finite element model is constructed. A three-dimensional beam element is used. Important material parameters are: Young's moduli, Poisson's ratios, and mass densities of cortical and cancellous bone.

#### Geometrical Properties

When using 3-D beam elements, the important geometrical properties are the cross-sectional shape and measurements, from which the moment of inertia is calculated, and the cross-sectional area. The data from which the geometrical properties are derived come from the literature (Thomsen, 1990). Figure 3-2 shows the material distribution along the tibial axis. From this plot, cross-sectional area and transverse moments of inertia for cortical and cancellous bone are derived. Bone marrow contributes no stiffness, only mass. It is felt that this mass is insignificant compared to the cortical and cancellous masses, so the effect of bone marrow is neglected.

If a circular cross-section is assumed, as pictured in Figure 3-3, radii can be calculated from the data in the plot of Figure 3-2 by:

\[
\begin{align*}
    r_{\text{conc}} &= \sqrt{\frac{A_{\text{conc}}}{\pi}} \\
    r_{\text{cort}} &= \sqrt{\frac{A_{\text{cort}} + r_{\text{conc}}^2}{\pi}}
\end{align*}
\]

(3.1)  
(3.2)
Figure 3-2. Bone material distribution along the length of the tibia (Thomsen, 1990).

Figure 3-3. Circular cross-section of beam element, showing cancellous and cortical bone.

Once $r_{\text{canc}}$ and $r_{\text{cort}}$ are calculated, the moments of inertia ($I_{11}$, $I_{22}$) can be calculated using the following formulas:

- cancellous bone: $I_{11} = I_{22} = \frac{\pi r_{\text{canc}}^4}{4}$ \hspace{1cm} (3.3)
- cortical bone: $I_{11} = I_{22} = \frac{\pi (r_{\text{cort}}^4 - r_{\text{canc}}^4)}{4}$ \hspace{1cm} (3.4)
Material Properties

Young's moduli of compact and cancellous bone vary throughout the literature. Table 3.1 shows these variations.

For this study, the following values are used: $E_{\text{cortical}} = 23 \text{ GPa}$

$E_{\text{cancellous}} = 1 \text{ GPa}$

$\rho_{\text{cortical}} = 1700 \text{ kg m}^{-3}$

$\rho_{\text{cancellous}} = 600 \text{ kg m}^{-3}$

<table>
<thead>
<tr>
<th></th>
<th></th>
<th></th>
</tr>
</thead>
<tbody>
<tr>
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</tr>
<tr>
<td>$E_{\text{cancellous}}$ (GPa)</td>
<td>0.12</td>
<td>0.6 - 2.2</td>
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<tr>
<td>$\rho_{\text{cortical}}$ (kg m$^{-3}$)</td>
<td>2254</td>
<td>1600 - 1800</td>
</tr>
<tr>
<td>$\rho_{\text{cancellous}}$ (kg m$^{-3}$)</td>
<td>597</td>
<td>200 - 1100</td>
</tr>
</tbody>
</table>

Table 3.1. Variations in bone material properties

Poisson's ratio is taken equal to 0.3. The material is assumed to be linear, elastic, and isotropic.

Boundary Conditions

In a study conducted by Van der Perre et. al (1983) the in vivo vibration modes of a human tibia were identified by modal analysis. One of the tests of the experiment consisted of evaluating the in vivo medium upon the vibration behavior of the tibia. A systematic dissection was made on an above-the-knee amputated underleg, and a complete modal analysis was made under the following conditions:

1. intact specimen

2. skin removed from medial face of tibia
3. all muscles removed, joint capsules intact
4. foot and ankle joint removed
5. knee joint removed
6. free tibia

It was found that the difference in natural frequency between the *in vivo* tibia and the free tibia was almost completely accounted for by the presence of the muscles. After removal of the muscles, the joints seemed to play no significant role. It was concluded that the shape of the bending modes and the effects of the muscles and joints on the natural frequencies indicated free-free boundary conditions. Therefore, free-free boundary conditions are used for this model.

*Model Construction*

The model is built using ANSYS, a finite element analysis software. The length of the model is 34 cm, the average length of an adult human tibia. Eighteen nodes span the length of the model. First, seventeen 3-D beam elements with cancellous bone material and geometrical properties are entered. Then, on top of these elements, using the same nodes, seventeen additional 3-D beam elements with cortical bone material and geometrical properties are added. Modal analysis is performed with ANSYS, implementing subspace iteration.

*3.1.3 Beam Theory*

A classical Euler-Bernoulli beam has plane cross-sections, initially normal to the beam's axis, that remain plane, normal to the beam axis, and undistorted. The internal virtual work rate is associated with the axial strain and torsional shear only. A simple beam in transverse vibration is shown in Figure 3-4. For an Euler-Bernoulli beam bending laterally, the governing equation is
\[
\frac{\partial^2}{\partial x^2} [EI(x) \frac{\partial^2 y(x,t)}{\partial x^2}] = -\rho A(x) \frac{\partial^2 y(x,t)}{\partial t^2}
\]
where EI is the flexural stiffness, and A(x) is the cross-sectional area.

\[c = \sqrt{\frac{EI}{\rho A}}\]

Figure 3-4. A simple beam in transverse vibration (Inman, 1994).

A more simplified form of equation 3.5 with EI(x) taken as constant is
\[
\frac{\partial^2 y(x,t)}{\partial t^2} + c^2 \frac{\partial^4 y(x,t)}{\partial x^4} = 0, \quad c = \sqrt{\frac{EI}{\rho A}}.
\]

Since equation 3.6 contains four spatial derivatives, it requires four boundary conditions and two initial conditions (in time). A separation-of-variables solution is assumed of the form \(y(x,t) = X(x)T(t)\). This is substituted into equation 3.6 to derive
\[
c^2 \frac{X''''(x)}{X(x)} = -\frac{T(t)}{T(t)} = \omega^2, \quad \left( X'''' = \frac{d^4 X}{dx^4}, \ddot{T} = \frac{d^2 T}{dt^2} \right).
\]

This leads to temporal and spatial equations:
\[
\ddot{T}(t) + \omega^2 T(t) = 0 \quad (3.8)
\]
\[
X''''(x) - \left( \frac{\omega}{c} \right)^2 X(x) = 0 \quad (3.9)
\]

By defining \( \beta^4 = \frac{\omega^2}{c^2} = \frac{\rho A \omega^2}{EI} \), the spatial equation becomes:
\[
X(x) = a_1 \sin \beta x + a_2 \cos \beta x + a_3 \sinh \beta x + a_4 \cosh \beta x \quad (3.10)
\]
The values for $\beta$ and the four constants of integration, $a_1$, $a_2$, $a_3$, and $a_4$, are determined from the four boundary conditions and the two initial conditions. The values of $\omega^2$ are the eigenvalues of the system, and the mode shapes, $X_n(x)$, are the eigenfunctions. These equations have been solved for beams with various boundary conditions (Inman, 1994). The first four modes of a free-free beam can be determined by relying on the following equations:

$$\beta_n L = 0 \text{ (rigid-body mode)}$$
$$4.730$$
$$7.853$$
$$10.9956$$

and:

$$\omega_n = \beta_n^2 \sqrt{\frac{EI}{\rho A}}$$

The Euler-Bernoulli beam model ignores the effects of shear deformation and rotary inertia. It is safe to ignore the shear deformation as long as the $h_x$ and $h_y$ illustrated in Figure 3-4 are small compared with the length of the beam. In the case of modeling the tibia, this condition is satisfied.

In the finite element model, the structure is made of many beam finite elements. In order to determine the structure's eigenvalues and eigenfunctions, the mass and stiffness matrices must be calculated.

3.2 Experimental Modal Analysis

3.2.1 Test Specimens

Impact testing was performed on a fresh-frozen cadaveric leg. The leg was supported at 45 degrees of flexure to simulate operating conditions. The knee was exposed by a resident surgeon as it would be in a TKA operation. The tissue overlying the anterior surface of the tibia was removed down to the medial calcaneus.
3.2.2 Instrumentation

A Wilcoxon triaxial accelerometer (frequency response 2-4000 Hz ± 10%) was attached to the tibia by exposing the bone and drilling and tapping a small hole into a firm part of the bone approximately 6 cm from the lateral plateau. A PCB instrumented impulse hammer was used to perform the impact at six points along the shaft of the tibia. Specifications for the accelerometer and impact hammer can be seen in Appendix A. The impact points were spaced every 4 cm starting 4 cm below the accelerometer. Data was filtered via a set of active filters with a cutoff frequency of 10 kHz, and was sampled at a rate of 20 kHz by an Analogic data acquisition board mounted in an 80486 PC running Labtech Notebook data acquisition software. A block diagram of the instrumentation set-up can be seen in Figure 3-5.

3.2.3 In-Situ Impact Testing

Excitation of the tibia was performed by the impact hammer at several points along the length of the bone. The bone's vibrational acceleration was recorded by the attached accelerometer. The excitation values of the accelerations measured by the force transducer in the hammer and the response measured by the accelerometer on the tibia were stored in the PC after passing through the anti-aliasing filters and data acquisition card. Multiple tests at each impact point were performed to be certain that the data were recorded. Due to memory limitations, each run could only last 0.4 seconds at the 20 kHz sampling rate. The impact and response were complete in 0.1 sec, so that the entire response was captured in the 0.4 sec interval.

3.2.4 Bone Cutting

The tibia was cut with an oscillating saw (Zimmer, VersiPower Plus Oscillating Saw) to determine if the saw excited the tibia's natural frequencies. The fresh-frozen cadaver
leg was supported at 45 degrees of flexure, and the knee was exposed as during the impact testing. The accelerometer was attached to the bone 6.5 cm below the tibial plateau, and an external cutting guide (Zimmer, NexGen Complete Knee Solution extramedullary tibial alignment guide) was attached. The cutting was performed at 10 mm below the tibial plateau. Acceleration data were collected as before. To examine the vibrational response of the bone to the sawing, the accelerances in all three directions were calculated.

![Block Diagram](image)

**Figure 3-5.** Block diagram of the experimental setup.

### 3.2.5 Data Processing

**Acceleration, Frequency Response Function**

After the accelerometer time histories were stored by the PC, they were transferred to a UNIX workstation for analysis in MATLAB. The signals were normalized to remove small voltage offsets. Both excitation and response time-based signals were transformed into frequency spectra via a Fast Fourier Transform (FFT) in MATLAB, i.e. they were resolved into their frequency spectra $F(\omega)$ and $A(\omega)$. The spectrum $A(\omega)$ is known as the accelerance. The complex frequency response function (transfer function):

$$H(\omega) = A(\omega) / F(\omega)$$ (3.13)
was calculated, which contained amplitude as well as phase information, and from which natural frequencies were derived. This transfer function is also called the inertance frequency response function (FRF). It is important to mention that the transfer function is defined for a linear structure independent of the excitation force. The assumption that the tibia is a linear structure is made here since only very small deformations are excited in the bone.

*Modal Data Extraction*

Once the frequency response function (FRF) and accelerance were measured, the desired information was the natural frequencies and damping ratios associated with each resonant peak. A typical ideal FRF for a three degree of freedom system is shown in Figure 3-6. Each peak in the graph represents a natural frequency of the system and can be confirmed by examining the value of the phase angle at each frequency, which should be $\pm 90$ degrees. A peak in the accelerance plot also represents a natural frequency of the system.

![Figure 3-6. The FRF of a typical 3 DOF system (Inman, 1994).](image)

The damping ratio, $\zeta$, of the system can be determined from the raw data by using the logarithmic decrement method. In the logarithmic decrement method, the amplitude of
motion, \( U_p \), at the beginning of a cycle and the amplitude, \( U_Q \), at the end of the cycle are measured.

![Diagram of decay record of a damped system](image)

**Figure 3-7.** Decay record of a damped system (Craig, 1981).

The response of a damped system can be written

\[
u(t) = U e^{-\zeta \omega_n t} \cos(\omega_d t - \alpha) .
\]

(3.14)

At the end of a period (one cycle), the value of \( \cos(\omega_d t - \alpha) \) returns to the value it had at the beginning of the cycle. Therefore, from equation 3-6, we derive the expression

\[
\frac{U_p}{U_Q} = e^{\zeta \omega_n T_d} .
\]

(3.15)

The logarithmic decrement is defined by the equation

\[
\delta = \ln \left( \frac{U_p}{U_Q} \right) = \zeta \omega_n T_d .
\]

(3.16)

The damped natural period is

\[
T_d = \frac{2\pi}{\omega_d} = \frac{2\pi}{\omega_n \sqrt{1 - \zeta^2}} .
\]

(3.17)

Therefore

\[
\delta = \zeta \omega_n T_d = \frac{2\pi \zeta}{\sqrt{1 - \zeta^2}} .
\]

(3.18)
For small damping approximations ($\zeta < 0.2$), $\delta = 2\pi \zeta$. Finally, the damping ratio can be obtained from the equation

$$\zeta = \left( \frac{1}{2\pi} \right) \ln \left( \frac{U_p}{U_q} \right).$$

(3.19)
Chapter 4
Results

4.1 Impact Testing

Figure 4-1 shows a sample data set collected during an impact test. The large peak is the force data, and the smaller peaks are the acceleration responses. The damping ratio is calculated from the raw data using the logarithmic decrement method (Craig, 1981). Using the first two highest acceleration peaks, $U_p = 0.8, U_Q = 0.625$. Using equation 3-19, $\zeta$ is calculated to be 0.04. Since $\zeta$ is such a small number, it is apparent that the system does not have much damping. That is,

$$\omega_d = \omega_n \sqrt{1 - \zeta^2} = \omega_n \sqrt{1 - (0.04)^2} = \omega_n$$

(4.1)

Thus, the damped natural frequencies are equal to the undamped natural frequencies. During the test, the accelerometer response in the z-direction was quite noisy, as can be seen in of Figure 4-2, the plot of the z-accelerance during impact. Figure 4-3 and 4-4 show a plot of the X and Y acceleration vs. time and the X and Y inerterance FRF vs. time. The x plots share the same major peaks below 1000 Hz at 250 and 650 Hz, and the y plots share the same peak at 350 Hz. The accelerance plots contain much less noise, making it a much cleaner plot. Therefore the accelerance plots were chosen to display the natural frequencies of the tibia.

Figure 4-5 shows the X accelerance plots from impact data taken at five of the impact points. Figure 4-6 shows the Y-accelerance plots from the same impact data. The major peaks below 1000 Hz that occur consistently within these plots are at 250, 350, and 650 Hz. Data was taken twice at each impact point. The X and Y accelerance plots for the other set of data can be seen in Appendix B.
Figure 4-1. Raw impact data.

Figure 4-2. Z-Accelerance during impact testing, showing noise.
Figure 4-3. X-Accelerance and X-Inertance FRF.
Figure 4-4. Y-Accelerance and Y-Inertance FRF.
Figure 4-5. X-Accelerance plots at five impact points - 6 cm is the proximal end, 30 cm is the distal end.
Figure 4-6. Y-Accelerance at five impact points - 6 cm is the proximal end, 30 cm is the distal end.

4.2 Beam Model

The tibia beam model was constructed using ANSYS, as mentioned earlier. The model was 34 cm long with a total of 34 3-D beam elements spanning 18 nodes. The model is shown in Figure 4-7.
Because the model has free-free boundary conditions, it has six rigid body modes, or modes with natural frequencies of zero. The next three natural frequencies can be seen in Table 4-1.

<table>
<thead>
<tr>
<th>Mode</th>
<th>Model Frequency (Hz)</th>
<th>Experimental Frequency (Hz)</th>
</tr>
</thead>
<tbody>
<tr>
<td>7</td>
<td>48.27</td>
<td>250</td>
</tr>
<tr>
<td>8</td>
<td>48.27</td>
<td>350</td>
</tr>
<tr>
<td>9</td>
<td>153.06</td>
<td>650</td>
</tr>
</tbody>
</table>

Table 4-1. Natural Frequencies of Composite Simple Beam Model

Mode 7 (48.27 Hz) is the model's first bending mode in the X-Y plane, as pictured in Figure 4-8. Mode 8 is the model's first bending mode in the X-Z plane as shown in Figure 4-9. It makes sense that modes 7 and 8 have the same frequency because the cross-section of the model is circular, thus symmetric.
Figure 4-8. First bending mode of model in X-Y plane.

Figure 4-9. First bending mode of model in X-Z plane.
4.3 Bone Cutting

Figure 4-10 is a plot of the raw data taken when the tibia was cut with the oscillating saw. Figure 4-11, 4-12, and 4-13 are the x, y, and z acceleration of the tibia during the cutting. It is important to note that in all three plots, the sharp spikes are evenly spaced every 220 - 230 Hz, corresponding to 13,200 - 13,800 rpm. Looking at the peaks below 1000 Hz, the greatest response of the tibia occurs at the peaks located at 250, 400, and 650 Hz. Again, two sets of data were taken during the cutting process. The X, Y, and Z acceleration plots for the second data set can be seen in Appendix C.
Figure 4-11. X-Accelerance of Tibia During Cutting.

Figure 4-12. Y-Accelerance of Tibia During Cutting.
Figure 4-13. Z-Accelerance of Tibia During Cutting.
Chapter 5
Discussion

5.1 Impact Results

The major peaks in an accelerance plot indicate a maximum response of a system to an input. The location of those major peaks indicates a natural frequency of the system. From Figure 4-5 and 4-6, the major peaks occur at 250, 350, and 650 Hz, indicating three natural frequencies of the tibia below 1000 Hz. Because the damping ratio of the system is so small (0.04), damping is not considered here. Thus, the system can be modeled with the equation:

\[ M\ddot{x} + Kx = 0 \]  

(5.1)

The undamped solutions are harmonic, in the form:

\[ x = x_n e^{i\omega t} \]  

(5.2)

Equation 5-1 then becomes:

\[ (-\omega^2 M + K)x_n = 0 \]  

(5.3)

The first two natural frequencies of the tibia have been shown to be the first two bending modes about the transverse axes (Thomsen, 1990; Van der Perre et. al, 1983). In one study, Van der Perre identified the in-vivo vibration modes of human tibia by modal analysis. He showed that the first two bending modes of the tibia occur at 264 Hz and 336 Hz with the skin intact. When the outer skin layer was removed, as in the case here, the first two natural frequencies occurred at 267 Hz and 341 Hz, which are 6.8 and 2.8 percent, respectively, from the natural frequencies determined in this study. In Thomsen's study, the third mode of vibration that was identified was a torsional mode. He examined the structural vibrations of an excised human tibia. Therefore, there was no effect of added mass from the skin and muscle tissue. The third mode that he identified had a natural frequency of 1113 Hz, which is quite a bit higher than the third mode identified in
this study. This makes sense because in this study, the tibia is in situ, surrounded with skin and muscle tissue. The skin and muscle tissue add mass to the system, decreasing its natural frequencies from its excised state. This decrease in frequency due to added mass is confirmed by the study by Van der Perre (1983) where the first two natural frequencies were found for a dry excised tibia. These first two frequencies were around 500 and 644 Hz. The presence of bone marrow was simulated by injecting a grease into the medullary canal of the dry tibia, and measurements were taken again. The first two natural frequencies then decreased to 357 and 465 Hz.

If a beam in transverse vibration is free at one end, the deflection and slope at that end are unrestricted, but the bending moment and shear force vanish

\[
\text{bending moment} = EI \frac{\partial^2 \omega}{\partial x^2}
\]

\[
\text{shear force} = \frac{\partial}{\partial x} \left( EI \frac{\partial^2 \omega}{\partial x^2} \right)
\]

Therefore, the solution of equation 3.2 can be solved by applying the four boundary conditions and the initial conditions for deflection and velocity. This solution for a free-free beam is

\[
X_n(x) = \cos \beta_n x + \sin \beta_n x + \cosh \beta_n x + \sinh \beta_n x
\]  

(5.4)

where

\[
\beta_n l = 0
\]

\[
\omega_n = \beta_n^2 \sqrt{EI / \rho A}
\]

4.73

7.85

11.0

From these equations, it can be shown that the second bending frequency of the beam in transverse bending is 2.75 times greater than the first bending frequency. From our experimental data, the first three natural frequencies of the in-situ tibia were found to be 250, 350, and 650 Hz. The mode at 650 Hz is approximately 2.6 times greater than the mode at 250 Hz, indicating that it may be the second bending mode in that same plane.
5.2 Beam Model

For the model that is developed here, the first bending modes in the X-Y and X-Z planes are equal due to the fact that a circular cross-section is assumed. It has been shown in several studies (Thomsen 1990, Van der Perre et. al 1983, Hight 1980) that these two modes actually differ by at least 100 Hz due to the transverse moments of inertia in these planes having different values. Therefore, the assumption of a circular cross-section for this model is a simplification, but it is felt that the choice is sufficient to verify the experimental results to some degree of confidence.

The natural frequency values that were found for this model were quite low compared to the experimental range of frequencies. To be more confident in the model, the number of elements was doubled, and the analysis was performed again. The results are shown in Table 5-1.

<table>
<thead>
<tr>
<th>Mode</th>
<th>Model Frequency (Hz)</th>
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<tbody>
<tr>
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<td>48.74</td>
</tr>
<tr>
<td>8</td>
<td>48.74</td>
</tr>
<tr>
<td>9</td>
<td>154.77</td>
</tr>
<tr>
<td>10</td>
<td>154.77</td>
</tr>
</tbody>
</table>

Table 5-1. Natural frequencies of refined tibia model

These results indicate that the first bending modes of the model in the X-Y and X-Z planes are 48.74 Hz, and the second bending modes are at 154.77 Hz. This very low first bending mode frequency introduces the possibility that we did not identify the first natural frequency in our experimental work. Perhaps 250 Hz and 350 Hz are the second bending modes of the *in-situ* tibia.
There are some possible reasons for not seeing these first bending modes in our data analysis. One reason may be that the resolution ($\Delta \omega$) of the FFT is not fine enough (Ewins, 1984). The resolution of the FFT is defined as

$$\Delta \omega = \frac{\omega_s}{N} \quad (5.5)$$

where

$$\omega_s = \text{sampling (digitizing) rate}$$

$$N = \text{number of discrete values}$$

Another explanation may be that the impulse hammer did not excite the low frequencies. Additionally, perhaps the displacements of the low frequencies were quickly damped out.

In this study, the resolution of the FFT was: $\Delta \omega = \frac{\omega_s}{N} = 20 \text{ kHz} / 4096 = 4.8 \text{ Hz}$, which is low enough to show a frequency of 50 Hz. To examine if the impact hammer excited the low frequencies, an FFT of the impact hammer was calculated and is shown in Figure 5-1. As indicated, the low frequencies are being excited.

![FFT of Impact Hammer Impulse](image)

**Figure 5-1.** FFT of impact hammer impulse.
If we look at some X and Y acceleration plots, shown in Figures 5-2 and 5-3, we can see that there are some small peaks below 100 Hz, indicating the possible low frequency for which we are looking.

Figure 5-2. X-Accelerance of tibia during impact

Figure 5-3. Y-Accelerance of tibia during impact
It should be noted that the frequencies found for this model could vary ±20% due to the various values for the material properties of cortical and cancellous bone that have been found. For example, if the density of cortical bone is changed to 1600 kg/m³, Young’s modulus of cancellous bone is changed to 2.2 GPa, and density of cancellous bone changed to 200 kg/m³, the calculated frequencies are:

<table>
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<tr>
<th>Mode</th>
<th>Frequency (Hz)</th>
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<tr>
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<tr>
<td>8</td>
<td>60.79</td>
</tr>
<tr>
<td>9</td>
<td>182.05</td>
</tr>
<tr>
<td>10</td>
<td>182.05</td>
</tr>
</tbody>
</table>

*Table 5-2. Calculated natural frequencies of tibia after material properties changes*

In order for the second bending mode to be closer to the 250-350 Hz range, some slightly unrealistic material parameters may have to be used.

5.3 Bone Cutting

As seen in Figures 4-11 through 4-13, the greatest acceleration peaks below 1000 Hz that occur during cutting are at ~250, 400 and 650 Hz. These peaks are located near or at the tibia’s natural frequencies. Because the greatest response of the tibia does occur near its own natural frequencies, it appears that the saw excited the tibia's natural frequencies.

In the plot of the acceleration in the Y-direction during cutting (Figure 4-12), we see some large peaks in the range of 1000-2000 Hz. The Y-axis of the accelerometer was aligned with the anatomical axis of the tibia. These peaks could be the result of a couple of situations: They may be due to the vibrations of the saw blade itself, or they may be from the interaction of the blade and the external cutting guide.
It is interesting to note that the evenly spaced peaks that occur in the acceleration plots (Figure 4-11 - 4-13) are 220-230 Hz apart. The 220-230 Hz intervals correspond to 13,200 - 13,800 rpm. The oscillating saw that was used has a maximum operating speed of 14,000 rpm. Therefore, the fundamental operating frequencies of the oscillating saw are responsible for exciting the natural frequencies of the tibia, and therefore possibly causing some significant vibrations.
Chapter 6
Concluding Remarks

This study examined whether the oscillating saw, which performs the cutting process of a tibial resection, excites the natural frequencies of the in situ tibia. From the experimental data, it can be concluded that during the tibial resection, the oscillating saw excites some natural frequencies of the in-situ tibia. The frequencies that the saw excites are near 250, 350, and 650 Hz.

Based on the experimental modal analysis and the finite element beam model, it was shown that the first bending modes of the in-situ tibia may be below 100 Hz. The experimental values that were found at 250 and 350 Hz may be the second bending modes. With a realistic change of material properties, the second bending modes of the beam model were still ~30% below the 250-350 Hz range. Therefore, the simplified beam model may lead to the use of some slightly unrealistic material properties. Perhaps a more complex three-dimensional model, created with the use of CT scans, might be more accurate.

Clearly, further studies regarding this problem are warranted. They may lead to the design of a dynamic vibration absorber, attached to either the bone or the saw, that is tuned to cancel out certain vibrations. This cancellation of vibrations may result in a smoother tibial surface and a longer-lasting implant.
References


Appendix A: Accelerometer and Impact Hammer Specifications

Model 733
Triaxial Accelerometer

Features:
- Small size, lightweight
- Easy to mount
- Shear mode design
- Ground isolated - eliminates ground loops
- Mechanical & thermal isolation

Applications:
- Machinery monitoring
- Helicopters
- Laboratory research
- Predictive maintenance
- General purpose

SPECIFICATIONS

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<tr>
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<td></td>
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<td>Acceleration Range</td>
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<td>Spectral</td>
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<tr>
<td>Shock Limit</td>
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<table>
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<tr>
<td>Case Material</td>
<td>Hardened aluminum</td>
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<tr>
<td>Mounting</td>
<td>10-32 socket screw</td>
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<td>Output Connector</td>
<td>Molex Miniature DR-8-1 (R&amp;B)</td>
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<td>Cabling, Max Length Connector</td>
<td>J92, 9 ft. 4 conductor shielded</td>
</tr>
<tr>
<td>Recommended Cable</td>
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</tbody>
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NOTES: *Signal distortion can occur when measuring high vibration levels, especially with long cables. A 24-VDC powering source is recommended for minimizing oscillation.*

ACCESSORIES AVAILABLE: Cable assembly, power supplies, amplifiers, signal conditioners.

TYPICAL TEMPERATURE RESPONSE

TYPICAL FREQUENCY RESPONSE

Due to continued research and product development, Wilkerson Research reserves the right to amend this specification without notice.
Appendix A: Accelerometer and Impact Hammer Specifications

HAMMER BEHAVIOR: functional transfer vs frequency

Striking an object with an instrumented hammer excites it with a nearly constant force and a broad frequency range. The frequency range is determined by the hammer structure, which can be varied with different tips and extenders.

GK291C01 (0.2 lb; 0.5-inch dia aluminum head), tests light to medium structures at medium and high frequencies. Kits include miniature Models 399A and A353B17 Accelerometers.

GK291C01, C02, C04 (0.3 lb; 0.5-inch dia steel head), tests medium structures such as car frames, engines and machine parts at low and medium frequencies. Kits include two general purpose accelerometers. Kit Model GK291C01 and C02 include Models 353B03 and A353B17; GK291C04 include Models 353B03 & 353B33. Optional kit GKL291D13 includes hammer, 4-channel power unit and triaxial accelerometer Model 335A11.

GK291C05 (1.0 lb; 1.0-inch dia steel head), tests medium to heavy structures such as machinery at low and medium frequencies. Kit includes general purpose accelerometers Models 353B03 and 353B33.

GK291C20 (3 lb; 2-inch dia aluminum head), tests medium and heavy structures such as machine tools and tanks at low and medium frequencies.

GK291C50 (12 lb; 3-inch dia steel head), tests very heavy structures like buildings, locomotives, ships and manufacturing at low and very low frequencies. Kit includes high sensitivity 393C and 353B33 accelerometers.

GK291C80 (0.005 lb; 0.25-inch dia steel head), tests very light structures such as compressor blades, sheet metal parts and printed circuit boards at low, medium, high and very high frequencies. Kit includes miniature A353B17 and 399A accelerometers.

Specify 'modal-tuned' for the most precise impact measurement.

The hammers are also available separately as Model Numbers 353B01 through 353B20, C50 and C80. All "GK" Kits have power units with gain x1, x10, x100 and BNC connectors.
Appendix B: X-Accelerance, Impact Data Set 2
Appendix B: Y-Accelerance, Impact Data Set 2
Appendix C: X,Y,Z Accelerance, Sawing Data Set 2