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Technical Brief

Evaluation of Filler Materials Used for Uniform Load Distribution at Boundaries During Structural Biomechanical Testing of Whole Vertebrae

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This study was designed to compare the compressive mechanical properties of filler materials, Wood's metal, dental stone, and polymethylmethacrylate (PMMA), which are widely used for performing structural testing of whole vertebrae. The effect of strain rate and specimen size on the mechanical properties of the filler materials was examined using standardized specimens and mechanical testing. Because Wood's metal can be reused after remelting, the effect of remelting on the mechanical properties was tested by comparing them before and after remelting. Finite element (FE) models were built to simulate the effect of filler material size and properties on the stiffness of vertebral body construct in compression. Modulus, yield strain, and yield strength were not different between batches (melt-remelt) of Wood's metal. Strain rate had no effect on the modulus of Wood's metal, however, Young's modulus decreased with increasing strain rate in dental stone whereas increased in PMMA. Both Wood's metal and dental stone were significantly stiffer than PMMA (12.7 ± 1.8 GPa*,* 10.4 ± 3.4 GPa*, and* 2.9 ± 0.4 GPa*, respectively). PMMA had greater yield strength than Wood's metal (*62.9 ± 8.7 MPa *and* 26.2 ± 2.6 MPa*). All materials exhibited size-dependent modulus values. The FE results indicated that filler materials, if not accounted for, could cause more than 9% variation in vertebral body stiffness. We conclude that Wood's metal is a superior moldable bonding material for biomechanical testing of whole bones, especially whole vertebrae, compared to the other candidate materials.* [DOI: 10.1115/1.2133770]

Keywords: spine biomechanics, vertebra, mechanical property, mechanical testing, finite element modeling, polymethylmethacrylate, Wood's alloy, dental cement, viscoelasticity

Introduction

Mechanical testing, especially in combination with computational analysis, is a powerful approach for understanding the relationship between biomechanics and pathology of bones. For structural testing of whole bones, it is often necessary to embed a part of the specimen in some moldable material for stable fixing and uniform load transfer from the test machine to the specimen. Embedding is required to prevent premature and unrealistic failure that does not represent the in vivo condition. For accurate experimental and computational analyses of bone structures, however, it is necessary both to use adequate embedding materials during testing and also to account for their presence in computational models.

The choice of appropriate embedding or "potting" materials for structural testing of bones has been a subject of long discussions within the biomechanics community (see, for example, the archives of the Biomechanics and Movement Science Listserver since 1992)[2]. The materials used are usually chosen from among the three choices; polymethylmethacrylate (PMMA), dental stone (high strength plaster) or low-temperature melting point alloys. However, no comparative study of the mechanical behavior of these materials has been undertaken, thus the choice is largely a matter of personal preference and experience.

It is especially important that load transmission is facilitated through a layer of filler material that needs to be conformed to the uneven surface of vertebral endplates during the mechanical testing of whole vertebral bodies (Fig. 1). The use of PMMA is common for this purpose [1,2]. However, the low modulus and highly time-dependent properties of PMMA [3] may pose difficulties in interpreting the test results and in computational analysis of bone constructs. Low-temperature melting point alloys are alternative fillers for biomechanical testing as they can be easily molded into shape, removed from the bone and reused. Indeed, Wood's metal

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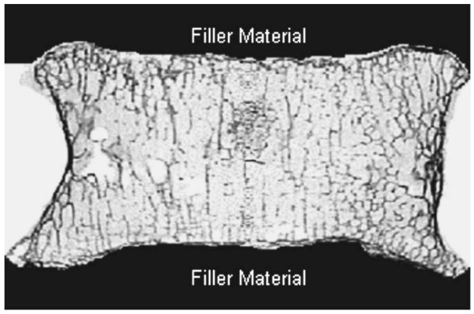
[2]http://isb.ri.ccf.org/biomch-l/archives



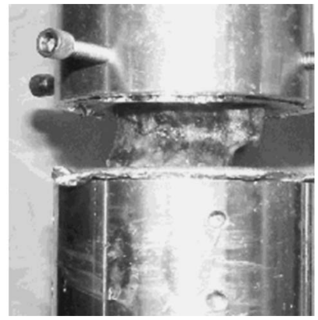


Fig. 1 Loading surfaces of whole bones such as human vertebrae are very nonuniform and need to be covered with a layer of suitable material for uniform load transmission during the mechanical testing (left). An actual test fixture with the specimen molded into Wood's metal is shown (right). Micro-CT-based FE simulations were based on this configuration.

(a low-temperature melting point alloy composed of bismuth, lead, tin, and cadmium) has been successfully utilized in the mechanical testing of human vertebrae [4].

More information on the mechanical behavior of these filler materials would enhance the experimental and computational efforts for characterization of biomechanical constructs as well as provide a basis for comparing results from different methodologies. For this purpose, we examined the effect of strain rate and specimen size on the mechanical properties of Wood's metal and compared them to those of dental stone and PMMA. In addition, as Wood's metal can be reused after remelting, the effect of remelting on the mechanical properties was tested by comparing them before and after remelting. Finite element (FE) models of vertebral body-filler constructs were utilized to examine the magnitude of potential effects of these filler materials on stiffness of vertebral bodies.

Methods

Wood's metal (typical chemical purities: Sn=12.5%, Pb =25.0%, Bi=50%, and Cd=12.5%; melting temperature=158°F, Reade Advanced Materials, Reno, NV) was melted and poured into cylindrical borosilicate disposable cell culture tubes with a nominal outer diameter of 10 mm. This consistently resulted in Wood's metal specimens with a diameter of 8.52 mm (±0.03 mm). After the metal solidified and cooled (24 h after molding), the glass tubes were broken. Wood's metal rods were machined into 17, 10, or 5 mm long cylindrical specimens in a lathe. A second batch of specimens was prepared similarly (10 and 5 mm only) by remelting the Wood's metal, in order to examine the effect of recycling the material between tests. A total of 20 specimens (4 of each size/batch) were prepared.

Dental stone (Coecal™, GC Lab Technologies, Alsip, IL) specimens were mixed according to the manufacturer's instructions and 17, 10, and 5 mm long (8.5 mm diameter) specimens (4 of each) were prepared following a procedure similar to that of Wood's metal.

PMMA samples were prepared using hand mixing techniques at room temperature. 80 grams of Surgical Palacos® R Radiopaque bone cement (Biomet, Warsaw, IN) was mixed with 40 ml of liquid methylmethacrylate monomer. After mixing, the cement preparation was placed into a 50 ml syringe and injected into a series of Eppendorf tubes which were used as molds. Similar to those of Wood's metal and dental stone, 17, 10, and 5 mm long (8.8 mm diameter) specimens (4 of each) were cut out of molds.

In order to examine the effect of strain rate on the modulus, all specimens were nondestructively tested in compression between lubricated platens in a servohydraulic materials testing machine (Instron 8051) at strain rates of 0.01 and 0.001 s^{-1} up to 0.2% strain for Wood's metal and dental stone. Because the stress-strain curves of PMMA specimens exhibited a more pronounced toe-in region during initial tests, the nondestructive strain level was increased to 3% strain for PMMA specimens. Afterwards all specimens were loaded beyond yield at a strain rate of 0.01 s^{-1} in order to determine yield strength.

Displacements were read with an extensometer attached to the upper and lower compression plates. Modulus and yield stress were calculated from the stress-strain curves [5]. ANOVA was used for statistical analysis. The type and details of ANOVA are given next to the pertinent result in the next section. A statistics software package was used for the analysis (Sigma Stat, SPSS Inc., Chicago, IL).

In order to estimate the potential effect of filler materials on vertebral body stiffness during a mechanical test and during a computational analysis, voxel-based linear large-scale FE models were constructed from the microcomputed tomography (micro-CT) image (voxel size=119 μm) of a typical human vertebral body (L3). The details of the model are as published elsewhere [6], except that the solver was modified to handle nonhomogeneous moduli [7]. The filler material was added to the image using a custom-written code (Fig. 1). The minimum amount of filler material that would be enough to flatten the irregular surfaces of vertebral end plates was used in the simulation. A tissue modulus of 5 GPa was assigned to bone whereas modulus values for filler materials were assigned based on experimental values determined in this study. In order to compare with nonrigid filler cases, a rigid boundary at the surface of end plates was also simulated by assigning a high modulus value to the filler material (200 GPa). A Poisson's ratio of 0.3 was assumed for all materials. A uniaxial compressive displacement corresponding to a 0.005 strain in the vertebral body was applied to all models. Stiffness of the vertebral construct was computed by dividing the sum of reaction forces (total force) by the total displacement. The total displacement in the vertebral body was computed as the difference in displacements between the upper and the lower end plates of the vertebra. The total force was divided by the estimated total displacement in the vertebral body to compute the stiffness of the vertebral body. Models were rerun after increasing the thickness of filler-material layers by five voxels at each end (about 0.6 mm additional thickness/end). The effect of each filler material on vertebral body stiffness was examined by comparing the percent differences in the stiffness of vertebral body between the rigid and nonrigid filler models. The total experimental error in stiffness caused by the assumption of rigidity for fillers was estimated as the percent differences in stiffness between vertebra-filler constructs with rigid fillers and those with nonrigid fillers.

Results

The stress/strain curves of PMMA and Wood's metal were smooth and showed a ductile behavior with a long period of pos-

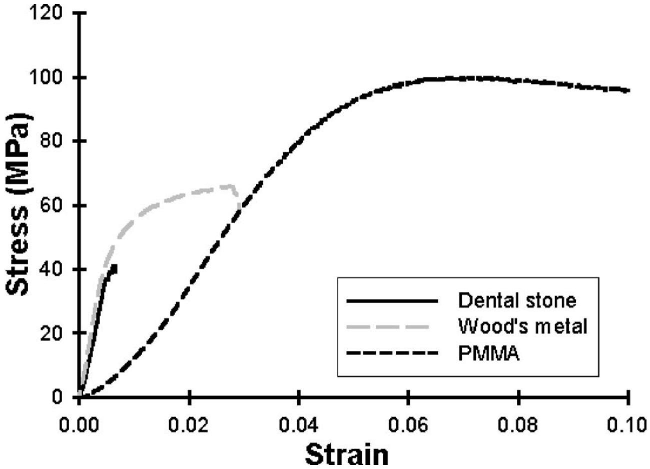


Fig. 2 Typical stress-strain plots from compressive testing of 10 mm long Wood's metal, dental stone, and PMMA specimens at 0.01 s^{-1}.

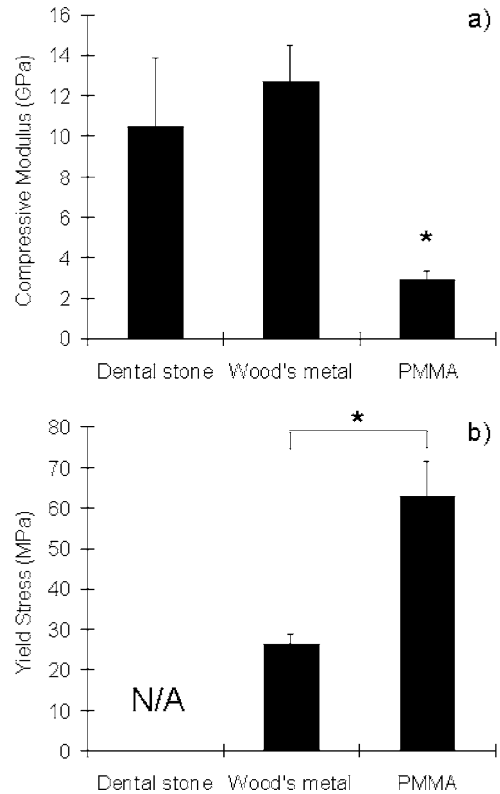


Fig. 3 (*a*) Mean compressive modulus of dental stone, Wood's metal, and PMMA. * indicates the significantly lower modulus of PMMA. (*b*) Mean yield stress of dental stone, Wood's metal, and PMMA. Because failure of dental stone was brittle, failure stress is plotted instead. * indicates the significant difference between Wood's metal, and PMMA. Failure stress of dental stone was not included in the statistical analysis. Error bars represent standard deviation.

tyield (Fig. 2). The stress/strain curves of dental stone specimens exhibited a brittle behavior with a zig-zag shaped postyield region. Failure was not consistent with some specimens breaking during nondestructive test attempts. Therefore, modulus only was examined for dental stone.

No significant difference in modulus, yield strain and yield strength ($p>0.4$, $p>0.4$ and $p>0.05$, respectively, two-way ANOVA) was found between batches (melt-remelt) of Wood's metal for short or long specimens. Wood's metal specimens from batch 1 and batch 2 were pooled thereafter.

Two-way repeated measures ANOVA (RMANOVA) followed by Bonferroni's test revealed that size had a significant effect on modulus with longer specimens being stiffer than shorter ones ($p=0.008$). Strain rate had no effect on the modulus of Wood's metal ($p=0.476$). Longer specimens were stiffer for dental stone and PMMA as well ($p<0.001$ and $p<0.03$, respectively). Unlike for Wood's metal, however, modulus decreased with increasing strain rate in dental stone whereas increased in PMMA ($p<0.05$, two-way RMANOVA for both). There was no interaction effect between size and strain rate on modulus for all filler materials ($p>0.35$, two-way RMANOVA).

Both Wood's metal and dental stone were significantly stiffer ($p<0.001$, Tukey's test) than PMMA (12.7 ± 1.8 GPa, 10.4 ± 3.4 GPa, and 2.9 ± 0.4 GPa, respectively) whereas the difference between the two was not significant ($p>0.32$) (Fig. 3). On the other hand, PMMA had a greater yield strength than Wood's metal (62.9 ± 8.7 MPa and 26.2 ± 2.6 MPa, respectively, $p<0.001$) and than the failure stress of dental stone which didn't exhibit a consistent yield or damage behavior (16.1 ± 4.1 MPa, $p<0.001$).

There were up to 4.5% difference in FE-estimated stiffness of the vertebral body between the rigid and nonrigid filler models (Fig. 4(*a*)). Within the same type of filler, the FE-estimated vertebral body stiffness decreased as the amount of filler material increased, as would be expected. However, the difference was small ranging from 0.12% to 0.42% for an additional layer of about 0.6 mm at both ends. On the other hand, the differences between the vertebral body stiffness and the construct stiffness (representing an experimental case where the filler is completely ignored due to the assumption of rigidity) could be as high as 9.1% (Fig. 4(*b*)).

Discussion

This study was concerned with the evaluation of materials used for in vitro testing of whole bones, specifically vertebrae. It should be noted that the common practice of filling the irregular surfaces on the superior and inferior ends of vertebrae with moldable stiff materials to accomplish a uniform load transfer during in vitro tests creates boundary conditions that are different than those in vivo. Despite large differences between in vitro and in vivo conditions, the use of stiff fillers in mechanical testing of whole bones is usually an acceptable approach when the biomechanical integrity of the bone tissue is of main concern.

None of the materials examined had modulus values large enough to be ignored when compared to that of hard tissue in cortical or cancellous bone, including that of vertebral shell (Fig. 3) [6,8–14]. Thus, it is necessary to account for the material properties of the filling layer in finite element stress analyses of mechanical tests where the interface is close to the region of interest in the analysis.

All of the filler materials exhibited size-dependent modulus values indicating that the actual thickness of the filling layer may be important in calculations. Size dependence of material properties is a commonly observed phenomenon and can be caused by many factors including flaws and impurities in the material, stress-state and boundary conditions, and characteristic size of the particles/features in the material [3,12,15–18]. The lower stiffness observed for shorter specimens in our experiment is more consistent with the effect of a defect and/or boundary conditions being more pronounced in a smaller volume. However, the exact cause of size dependence was not investigated for the materials used in this study.

The moduli of the dental stone and PMMA were rate-dependent consistent with previous reports [3]. Accounting for the time-

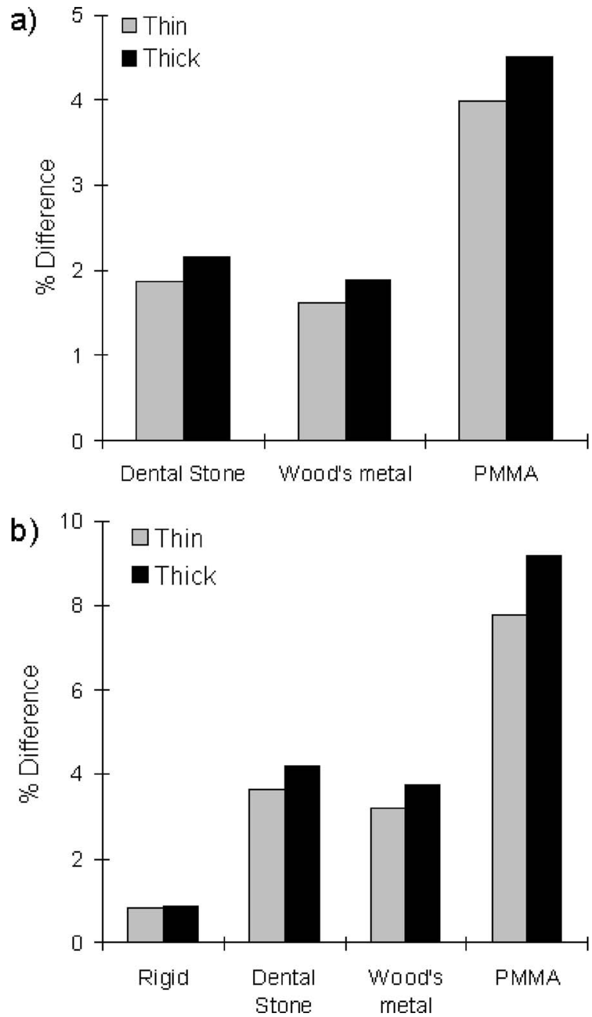


Fig. 4 (*a*) Percent difference in the FE-estimated stiffness of a vertebral body between simulations with rigid (modulus: 200 GPa) and nonrigid fillers (Dental Stone, Wood's metal, and PMMA). Simulations with minimal filler material (enough to flatten the end surfaces) are shown as "thin" and those additional layers of filler material (about 0.6 mm at both ends) are shown as "thick." (*b*) Percent difference between the FE-estimated stiffness of the vertebral body as calculated from the rigid-filler model and that of the construct with rigid and nonrigid fillers.

dependent behavior of the filling may introduce additional complexity in the mechanical behavior of the bone/filler construct.

Low melting point alloys, in general, are considered to exhibit temperature and time dependent behavior [19]. However, a great majority of work on the rate dependence of these alloys has been focused on the strain rate sensitivity of the superplastic behavior [20–23]. A recent article reported increased compressive yield stress with increasing strain rate for a lead-bismuth alloy [24]. A temperature effect was noted on the stiffness of the same alloy but the effect of strain rate on stiffness was not clear. These results would collectively suggest that Wood's metal stiffness is rate-dependent, however, it was not found to be the case in our study. It should be noted that although the rate dependent alloys have some components in common with Wood's metal, the presence or absence of additional components may have a significant effect on the mechanical behavior of the alloy, perhaps much more than the effect of different concentrations of the same component [25,26]. It should also be stressed that strain rates outside the range of values used in this study may affect the stiffness of Wood's metal. It is noteworthy, however, that strain rates used in this study are those commonly used in biomechanical testing of bone tissue and in the order of physiologically encountered values [27–29]. Our finding that the modulus of Wood's metal within these strain rates is not rate-dependent suggests that errors in experiments and models originating from the time-dependent behavior can be reduced compared to those caused by PMMA or dental stone by using Wood's metal.

The finding that there are no differences in Wood's metal modulus values between batches is consistent with previous results from other alloys of lead, tin, and bismuth that elastic stiffness is relatively insensitive to the alloy composition, i.e., percentages of tin and bismuth [25]. The addition of cadmium (10%) to these alloys increases the stiffness considerably but further changes in composition do not cause changes in stiffness [26]. Yield strength, on the other hand, might be expected to be affected by melt/remelt procedures since hardness is more sensitive to the alloy composition, although statistical significance has not been reported in the previous results [25,26]. Our finding that there are no differences in Wood's metal modulus values between batches means that remelting and reusing Wood's metal does not introduce errors into the mechanical testing of vertebral bodies. It might be advisable, however, to check the mechanical properties of the alloy over a period of intensive use since accumulation of contaminants from tissue may eventually cause noticeable changes in the properties of Wood's metal.

The present simulations demonstrated that the stiffness of a vertebral body is affected by the presence of the filler material and that the effect is larger when a thicker layer of filler is used (Fig. 4(*a*)). The effect of filler thickness on vertebral body stiffness was lower for stiffer filler materials although this difference in size dependence within the same type of filler was negligible for the typical vertebra. It appears that the effect of filler stiffness on vertebral body stiffness is relatively small, especially for higher-stiffness fillers such as Wood's metal. These are true boundary condition effects of the filler on vertebral stiffness and should not be confused with the dependence of the filler material modulus on specimen length observed during materials testing. These effects, though might be limited to the vicinity of end plates, are indicative of changes in stress patterns. Away from the boundaries, these effects are likely minimal according to St. Venant's principle. Nevertheless, there is room for improvement of computational models by inclusion of filler properties in simulations. On the other hand, the differences between the vertebral body stiffness and the construct stiffness (representing a mechanical test where the filler is completely ignored due to the assumption of rigidity) could be as high as 9.1% (Fig. 4(*b*)). It should be noted that these are the lower bounds of filler effects as the amount of filler simulated in the models was minimal. In experimental applications, it is expected that a larger amount of filler would be used which can potentially cause errors that are larger than estimated in this study.

Overall, the results show that, among the filler materials tested, Wood's metal is the most desirable material for compression testing of vertebral bodies. However, while Wood's metal might be a desirable filler material for vertebral compression testing, it may not be advisable for the testing of denser bones (e.g., long bone diaphyses) due to its low yield stress unless the locations affected by the applied loads are sufficiently away from locations of potting.

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