NEW MINIMALLY INVASIVE TECHNIQUES IN APPROACHING OF OSTEOPOROTIC FRACTURES IN SPINE NEUROSURGERY

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Abstract - In this article we proposed to describe two new minimally invasive surgery at the vertebral column level, from technical and medical point of view, vertebroplasty and kyphoplasty. These techniques have a major contribution in the treatment of spinal deformities. Osteoporosis is a major health problem that can cause a variety of spinal deformities, and this include those who can be treated with great success through vertebroplasty and kyphoplasty techniques.

Osteoporosis is a major health problem, particularly in women. Of the more than 100 million women affected worldwide, a significant number have painful vertebral compression fractures. Two new image-guided minimally invasive procedures (vertebroplasty and kyphoplasty), which use the internal application of bone cement, give remarkable results in ameliorating the pain in 80% to 90% of painful osteoporotic vertebral compression fractures. In addition, kyphoplasty provides an opportunity to reduce unwanted kyphosis before stabilization, accordingly to the Journal of Women’s Imaging 2000;2:117-124.

Keywords – osteoporotic fractures, neurosurgery

In this article we proposed to describe two new minimally invasive surgery at the vertebral column level, from technical and medical point of view, vertebroplasty and kyphoplasty. These techniques have a major contribution in the treatment of spinal deformities. Osteoporosis is a major health problem that can cause a variety of spinal deformities, and this include those who can be treated with great success through vertebroplasty and kyphoplasty techniques.

As historical terms, in the 1970s, bone cement replacement had became the standard of care for giant cell tumors and other benign bone tumors. This was an open surgical procedure during which benign tumors were removed and the cavity was then filled with cement. In the mid-1980s, Dr. Herve Deramond, a French interventional neuroradiologist, performed the first vertebroplasty to manage a painful spinal hemangioma. Based on the findings from open surgical cementations of benign bone tumors, he performed a bone cement (polymethylmethacrylate) injection into the vertebral body and succeeded in ameliorating the pain. This procedure became known as vertebroplasty. He expanded the treatment to include patients with painful osteoporotic fractures, with excellent results. The first vertebroplasties were reported in the United States in 1995. Most vertebroplasties in the United States are performed for painful osteoporotic compression fractures. However, some are performed for vertebral compressions from metastatic disease (particularly osteolytic metastases), multiple myeloma, and painful hemangiomas.
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**Figure: Environmental Scanning Electron Microscope**

Polymethyl methacrylate (PMMA) or poly(methyl 2-methylpropanoate) is the synthetic polymer of methyl methacrylate. This thermoplastic and transparent plastic is sold by the tradenames Plexiglas, Limacryl, R-Cast, Perspex, Plazcryl, Acrylex, Acrylite, Acryplast, Altuglas, Polycast and Lucite and is commonly called acrylic glass or simply acrylic. The material was developed in 1928 in various laboratories and was brought to market in 1933 by Rohm and Haas Company.

The material is often used as an alternative to glass. Differences in the properties of the two materials include:

**PMMA:**
- is less dense; its density can range from 1150-1190 kg/m³. This is less than half the density of glass which ranges 2400 to 2800 kg/m³;
- has a higher impact strength than glass and does not shatter but instead breaks into large dull pieces;
- is softer and more easily scratched than glass. This can be overcome with scratch-resistant coatings;
- is typically processed at a lower temperature than glass, just 240-250 °C;
- transmits more light (up to 93% of visible light) than conventional glass; however, there exist special optical grades of glass that can transmit up to 98% of light.

Unlike glass, PMMA does not filter ultraviolet (UV) light. PMMA transmits UV light down to 300 nm. Some manufacturers[3] add a coating to PMMA sheets to make them absorb UV light. PMMA molecules have great UV stability compared to polycarbonate.

**Tabel 1: Vertebroplasty: Indications**

<table>
<thead>
<tr>
<th>Painful vertebral body compression from:</th>
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<tbody>
<tr>
<td>1. Osteoporotic vertebral compression fractures</td>
</tr>
<tr>
<td>2. Metastatic disease</td>
</tr>
<tr>
<td>3. Multiple myeloma</td>
</tr>
<tr>
<td>4. Painful vertebral hemangioma</td>
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**Medical technologies and implants**

PMMA has a good degree of compatibility with human tissue, and can be used for replacement intraocular lenses in the eye when the original lens has been removed in the treatment of cataracts. Hard contact lenses are frequently made of this material. Soft contact lenses are often made of a related polymer, where acrylate monomers containing one or more hydroxyl groups make them hydrophilic.

In orthopaedics, PMMA bone cement is used to affix implants and to remodel lost bone. It is supplied as a powder with liquid methyl methacrylate (MMA). When mixed these yield a dough-like cement that gradually hardens. Surgeons can judge the curing of the PMMA bone cement by pressing their thumb on it. Although PMMA is biologically compatible, MMA is considered to be an irritant and a possible carcinogen. PMMA has also been linked to cardiopulmonary events in the operating room due to hypotension. Bone cement acts like a grout and not so much like a glue in arthroplasty. Although sticky, it primarily fills the spaces between the prosthesis and the bone preventing motion. It has a Young's modulus between cancellous bone and cortical bone. Thus it is a load sharing entity in the body not causing bone resorption.

Dentures are often made of PMMA, and can be colour-matched to the patient's teeth. In cosmetic surgery, tiny PMMA microspheres suspended in some biological fluid are injected under the skin to reduce wrinkles or scars permanently.

**Technique notes on vertebroplasty:**

Careful physical examination should be correlated with imaging studies. Informed consent should be obtained.

Large-bore needles are placed via a transpedicular approach, but alternative approaches could include single posterolateral approach in larger lumbar vertebra or costovertebral approach in the upper thoracic areas when the pedicles are not sufficiently large to accommodate the large-bore needles.

Percutaneous direct vertebral venography is performed by most operators, with real-time filming usually on the lateral view. Nonionic, intrathecal appropriate contrast agent should be used. Usually
3-4 cc iodine contrast per injection is adequate. If this contrast does not wash out before bone cement injection, then injection of a small amount of saline is usually adequate to wash this out and clear it from view.

If there is exuberant contrast flow into epidural veins or into inferior vena cava, the bone cement can be allowed more time to harden before injection, or gelfoam embolization might be considered using large pledget gelfoam (3-4 mm pledget size) placed in an iodine contrast slurry. Approximately 1 cc of this mixture is injected, and the epidural venogram can be repeated in approximately 10 minutes to determine whether this is adequately reducing the venous flow.

Mixing the bone cement: various brands of bone cement are available. Six grams of USP sterile barium (Bryan Corp, Woburn, MA) is added for opacification. Mix vigorously and rapidly, so that a chemical reaction takes place to cause the material to become somewhat liquid. The mixture should be stirred until it becomes a thick toothpaste-like consistency. Transfer this into a 10 cc syringe, then fill smaller 1 cc luer lock syringes. The 1 cc syringes will be used to inject the bone cement through the 11- or 13-gauge needles under high pressure. It is helpful to time the mixing and filling sequence. Usually, mixing takes 1 minute, then loading the 1 cc syringes and preparing to inject takes approximately 3-5 minutes. However, this may vary depending on the temperature of the cement (to gain a longer working time the packages of cement kit may be refrigerated before use).

Bone cement application: when the bone cement has reached a viscous doughy state, injection of the bone cement should be performed slowly under direct fluoroscopic visualization. The injection should proceed by filling the needle and then slowly pushing the bone cement aut into the vertebra. One should fill from the anterior third in a retrograde manner posteriorly. As the anterior aspect of the vertebra is filled, then the needle can be withdrawn somewhat to fill the more posterior aspects. If the bone cement becomes very difficult to inject through the needle, the syringe should be removed promptly, and the stylet of the needle can be used as a ramrod to push the bone cement aut through the shaft of the needle. More bone cement then can be injected through the needle. Complete filling of the vertebral body may be ideal. However, the pain relief after injection of less than 2.5 cc polymethacrylate into a vertebral body, particularly when the anterior two thirds of the vertebra are filled. If there is unwanted embolization of bone cement into either the epidural veins or into the inferior vena cava, the injection should be halted immediately and momentarily. This should prevent further leakage of bone cement into unwanted areas. After perhaps 30-40 seconds, the injection can be resumed. In some cases, one might have to come back into the same entry hole with a replacement needle if the bone cement in the needle hardens such that the bone cement cannot be evacuated from the needle shaft. Once the injection procedure is complete, the needle should be removed promptly to prevent cementing the needle in the bone.

The patient is not moved from the prone position on the x-ray table until the remaining bone cement in the mixing bowl is hardened.

Patient follow-up: keep the patient resting supine under observation for approximately 4 hours after the procedure in the recovery area before discharge.
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In the early 1990s, a modification of vertebroplasty was designed and became known as kyphoplasty. In this procedure, a bone balloon is used in an effort to restore vertebral body height to correct the spinal deformities before cementation. The first kyphoplasty procedure was performed in 1998 by Mark A. Reiley, MD, Orthopedic Surgeon, Berkeley Orthopedic Medical Group, Berkeley, California.

The instruments used in this technique are;
- IBT – Inflatable Bone Tamp;
- BFD – Bone-Filler Device;
- Large-bore needles;
- Guide pin;
- Blunt dissector;
- Cannula;
- 3.3 mm hand drill.

In the following is a short presentation of the instruments used, all this correlated with a brief description of the method, to better understanding this minimally invasive surgery technique.

The instruments

a) The Inflatable Bone Tamp

b) The instruments for bone acces
Balloon Kyphoplasty Procedural Techniques

Careful physical examination should be correlated with imaging studies. Informed consent should be obtained.

Large bore needles usually are placed via transpedicular approach, but alternative approaches could include single posterolateral approach in larger lumbar vertebra or extrapedicular approach in the upper thoracic areas when the pedicles are not sufficiently large to accommodate the large-bore needles or are too lateral for required medial position medial position of subsequent tools.

Guide pin is inserted, followed by the removal of large-bore needle. Then the blunt dissector is placed over guide pin and the cannula over the dissector, stopping at the vertebral body wall for creation of a working channel into the bone.

Remove the blunt dissector. Use a 3.3-mm hand drill to create a channel through the vertebral body (aiming medial and inferior if procedure is transpedicular or extrapedicular), ending near the anterior cortex.

Position the inflatable bone tamp prepared with radiocontrast inflation medium, so that the marker bands are evenly spaced between the anterior and posterior ends of the drill channel. Inflate slowly, monitoring x-ray images, inflation volume, and pressure. This process continues until the vertebral body height is completely reduced, when the balloon reaches an endplate or cortical wall, when the balloon reaches its maximum volume, or when there is no further pressure decay. Note final balloon volume for each balloon used.

Cement preparation: polymethylmethacrylate
(40 cc powder and 10 cc liquid monomer) is mixed with 6 gm barium and antibiotics and loaded into a bone-filler device via syringe. Cement injection is begun 3-5 minutes after mixing. The appropriate injection time depends more on the consistency of the cement than on an actual time. The cement should be thick enough to not drip from a test spatula in the mixing bowl. It should be moldable with minimal stickiness.

Cement placement: under x-ray image control, position the bone-filler device into the cavity and almost touch the anterior wall. Using continuous monitoring, push the thick cement into the cavity using a stylet. Bring nozzle back into center of cavity and continue adding cement around the tip of the nozzle. Be aware of the volume of cement (each bone-filler device holds 1.5cc), watch carefully as you exceed the cavity volume. It is appropriate to fill the cavity and see some interdigitation with no cement leaks. Confirm appropriate placement with line fluoroscopy and final radiographs.

Remove the cannula and bone cement-filler device, and steristrip or suture wounds with a small number of stitches if required.

Do not move the patient until the remaining bone cement in the mixing bowl is solid. Keep the patient supine for 4 hours to help prevent hematomas, especially for lumbar spine kyphoplasty. The patient should be observed for 4 hours after the procedure, although some operators may wish to hospitalize their patients overnight, particularly if the patient is very frail.

From an engineer's point of view, the design of numerical models (using a discretization with finite elements) for vertebral column bodies, is proved to be a very useful instrument. The study of mechanical stresses, statics and dynamics equation and strains which appear in the bone mass can reproduce the physiological activity under the form of external mechanical stresses.

\[
\frac{d\rho(x,t)}{dt} = B \sum_{i=1}^{n} f_i(x) \cdot \frac{U_i}{\rho_i} - k
\]

for \(0 < \rho \leq \rho_{cb}\)

where:
\(n\) - the numbers of finite elements;
\(U_i\) – density of the strain energy, estimated to be concentrated in the center of the finite element;
\(\rho_i\) – density of the bone tissue of the finite element;
\(k\) – the reference value of the mechanical signal;
\(B\) - remodelling constant of the bone.

This model of remodelling require a spatial function which physically simulate the influence of sensorial cells of the bone tissue from vicinity. That can be analytically expressed as:

\[
f_i(x) = e^{-\frac{d_i(x)}{D}}
\]

where:
\(d_i(x)\) – represent the signal transmitted from the „i” cell, situated on „x” position;
\(x\) – the cell position;
\(D\) – the range of cells which are influenced.

The Young’s modulus of an finite element can be expressed as:

\[
E = C \cdot \gamma
\]

where \(C\) and \(\gamma\) are constants.

The (1) equation can be rewrite as an iterative equation:

\[
\frac{\Delta \rho(x,t)}{\Delta t} = B \sum_{i=1}^{n} f_i(x) \cdot \frac{U_i}{\rho_i} - k
\]

for \(0 < \rho \leq \rho_{cb}\)

\[
\rho(x,t + \Delta t) = \rho(x,t) + \Delta \rho(x,t)
\]

The process of remodelling is convergent if one of the following condition is true:

1. An reference value of the stimulus is reached, which represent the density of strain energy and the density of the bone tissue ratio;
2. Is reached te value of the density of the cortical bone, \(\rho = \rho_{cb}\);
3. Is achieved an complete resorbtion of the
bone tissue, reaching the value of the bone tissue $\rho = 0.01\, \text{g/cm}^3$.

Fulfilling of one of the conditions mentioned above means reaching the point of equilibrium for the modeling process.

The iterative process of modeling continues until these conditions are fulfilled for all the finite elements.

The geometric model of the bone can be created through reconstruction of an set of CT scans, sequentially achieved on antero-posterior direction, as can be seen in fig 2.

At the bone-cement interface, due to low absorption of the cement, can be performed an analysis of the manner of its volume changing (fig 3). An low volume and irregular borders will cause partial volume effects which influence the values measured through standard measuring procedures.

Fig 2: 3D segmentation process using CT method. Can be used automated calculus algorithms for various parameters of the bone

Fig 3 Reconstruction of two intervertebral cavities filled with cement

The design of the model with finite elements can be achieved by two ways:

1. **In vivo** – using CT scans;
2. **Using an sensor**

The creation of an finite element model of the vertebral body is based on CT scans. The images obtained are analysed through an semiautomat algorithm. The various density of the bone are represented by various intensity of colour (see fig 4).

The full specter is divided in 200 discrete intensities, every intensities having attributed specific material properties, after a pre-callibration.

Fig 4 Level of intensity for various materials

The last stage consist in the generation of nodes an base elements for CT scans, using HEXAR programme (Cray Research inc., Minneapolis, USA). This specific programme has the possibility of digitization a structure of random type with hexaedric elements. This particular digitization represent an elevated degree of precision and avoid irregularities specific to tetraedric digitization. From the amount of calculation point of view, the method used in HEXAR programme is about time faster versus those who use tetraedric elements for digitization.

3D automatic generation:

A precise 3D finite element model is an important factor in understanding and monitoring the changes in the state of vertebral body

Nowadays, could be build an preprocessor and an solver on the ANSYS programme capacities. Thus, uniform geometries and dimensions can be obtained, through minimization of dimensional variation and his orientation. The experience that has been made shown that through using of semiautomats algorithms the load of the calculus system was reduced with 42%.

Fig 5 An example of automatic 2D digitization of an trabecular bone

At the execution of the tridimensional model has been calculated a large amount of points in 2D, the distance between planes measuring 1mm (fig 5). These points were merged on their luminosity basis. Between trabecular and cortical bone it is a great difference.

The geometry are saved in the ANSYS
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Programme slide by slide and then they are computed through using interpolation algorithms.

**Fig 6: 3D model of an digitized vertebral body bone**

To obtain an 3D model, the intensity curves was merged on the basis of existing polylines, generating 3D surfaces (fig 6).

**Fig 7: Solid model of an undigitized lumbar vertebral body**

Material properties:

On the basis of virtual models can be made real models which can be used to validate on experimental basis teh method of calculation. Can be made experiments on a solid structure. To produce such models can be used CNC machines. The real models obtained through this procedure will allow the investigation on mechanical basis for the various degree of mechanical properties (related to osteoporosis) and various geometries. Beside, in this case can be developed, design adn test new materials and evaluates new hypotheses.

An important element consist in the ability to evaluate the risk of fractures, an for achieving this, it is essential to understand the effects of macroscopic and microscopic strains and stresses. This can be obtained through microstructural analysis and histological data that can be used at the construction of the models. In this way can be assessed local state of load and stress and their influences to local morphology at bone macro scale. The Young’s modulus of an cortical bone is usually 17.4 GPa on longitudinal direction.

The computation models use data from the tabel below. To see what is the effect of the osteoporosis, the haversian porosity, lacunar and the percentage of osteonal bone can be varied. It is observed that the Young’s modulus vary significantly as haversian porosity vary. Lacunar porosity is reduced with 9% in the case of osteonal bone.

**Material properties and information microstructure utilized**

<table>
<thead>
<tr>
<th>Material</th>
<th>Dimension</th>
<th>Strength modulus transversal/longitudinal</th>
</tr>
</thead>
<tbody>
<tr>
<td>Osteon bone</td>
<td>200-300μm</td>
<td>14.36GPa, 21.33Gpa</td>
</tr>
<tr>
<td>Interstice bone</td>
<td>10-40% of total</td>
<td>16.03GPa, 23.03Gpa</td>
</tr>
<tr>
<td>Cement</td>
<td>1.5μm</td>
<td>unknown ≤6Gpa</td>
</tr>
</tbody>
</table>
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<table>
<thead>
<tr>
<th>Harvesian canals</th>
<th>Less than 50μm diameter</th>
<th>negligible</th>
</tr>
</thead>
<tbody>
<tr>
<td>Lacunar bone</td>
<td>Elliptical pits, beam 3-11μm</td>
<td>negligible</td>
</tr>
</tbody>
</table>

Fig 9 Dependence of E1 modulus, function of fiber volume and of haversian porosity

Fig 10 Dependence of E3 modulus, function of fiber volume and of haversian porosity

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