INTRODUCTION

Bone is a living, growing tissue. The architecture of the inner porous (trabecular) bone that gives bone its strength is continually changing as a result of mechanical adaptation to the external loads. The aging process can be mathematically modeled to simulate the process of adaptive bone remodeling. The idea proposed in this research is to use realistic mechanical loading to simulate the structural change in trabecular bone over time. Realistic mechanical loading data was used to calculate the forces on the joints. These calculated mechanical forces were then used in a remodeling algorithm to predict the future geometry of the trabecular bone.

METHODS

The long-term goal of this work is to use real-world motion data as the driving factor for the bone remodeling algorithm, outlined in Figure 1. Virtual human modeling and simulation software (LifeMOD, LifeModeler, Inc.) is used to calculate the forces at various joint locations or at specific marker location. The force data is then used in the remodeling algorithm to alter the geometry which follows the mechanical adaptation principal. The modified geometry is then imported back in to LifeMOD’s virtual human model to calculate the new set of force data. The highlighted sections in Figure 1 indicate the focus of this work.

In this experiment, μCT scans of the 4th lumbar vertebrae with a resolution of 18μm X 18μm X 36μm were used. Subsequent calculations were performed in MATLAB (The MathWorks). These images were thresholded with at 22.4% of the maximal grayscale value to separate bone from the marrow [1]. A 3D geometry was constructed from these images using an isosurface algorithm. A set of parameters that characterize the bone structure were calculated. Comparing these parameters with the values published based on clinical studies validates one aspect of the simulation.

Osteocyte cell locations were randomly distributed within the 3D reconstruction [3]. Strain Energy Density (S.E.D) rate triggers the remodeling process [2]. The S.E.D. ($R_i(t)$) at an osteocyte location is proportional to the product of strain and stress tensors at that location (Eqn 1). The effective mechanical stimulus on the surface of the bone at a location is the sum of the effect of all the osteocytes within its vicinity whose intensity reduces exponentially (function $f_i(x)$) with distances ($d_i(x)$ and $D$) (Eqn 2) [3].

$$R_i(t) = \frac{1}{2} \sigma_i \varepsilon_i$$

$\sigma_i = \text{Strain tensor}$

$\varepsilon_i = \text{Stress tensor}$

$$P(x,t) = \sum_{i=1}^{N} f_i(x) \mu_i R_i(t)$$

$f_i(x) = e^{-\left(d_i(x)/D\right)}$

$\mu_i = \text{Mechanosensitivity of osteocyte i}$
The final change \((\text{dm/dt})\) in relative density at location on the surface is proportional (by \(\tau\)) to the effective mechanical stimulus (\(P\)), the bone formation threshold (\(k_{th}\)) and the relative amount of bone resorbed per day (\(r_{oc}\)) \([2]\). This was modeled using the equations as described in (Eqn 3).

\[
\frac{dm}{dt} = \begin{cases} \tau P(x,t) - k_{th} & \text{for } P(x,t) > k_{th} \\ -r_{oc} & \text{for } P(x,t) \leq k_{th} \end{cases}
\] (3)

**RESULTS AND DISCUSSION**

An example of image processing on one slice is shown (Fig. 2). The set of images are stacked and stitched using MATLAB to produce a 3D geometry.

![Figure 2](image)

**Figure 2**: (a) Original CT (b) ROI (c) Binary image

An important intermediate calculation is that of the mid-axis of the structure and the orientation of each trabeculae with respect to the horizontal plane (Fig. 3). The parameters were calculated on several small sections about 3mm\(^3\) selected randomly from the vertebrae and compared (Table 1) with previously-reported values \([4]\). Loads calculated from LifeMOD were used to mathematically simulate bone remodeling on these sections. An example result of the remodeling algorithm is shown (Fig. 4).

![Figure 3](image)

**Figure 3**: (a) Mid-axis overlapped on the structure and (b) Orientation.

Future work will parallelize the remodeling algorithm to gain ability to run the simulation on whole vertebrae.

**Table 1: Calculated parameter and the clinical data.**

<table>
<thead>
<tr>
<th>Parameters</th>
<th>Samples</th>
<th>Clinical Study</th>
</tr>
</thead>
<tbody>
<tr>
<td>rTb.Th (mm)</td>
<td>0.092 – 0.120</td>
<td>0.139 (0.028)</td>
</tr>
<tr>
<td>pTb.Th (mm)</td>
<td>0.103 – 0.150</td>
<td>0.139 (0.028)</td>
</tr>
<tr>
<td>Tb.Sp (mm)</td>
<td>0.070 – 0.130</td>
<td>0.854 (0.143)</td>
</tr>
<tr>
<td>Tb.N (mm(^{-1}))</td>
<td>2.788 – 4.990</td>
<td>1.161 (0.181)</td>
</tr>
<tr>
<td>BV/TV (%)</td>
<td>12.2 – 33.6</td>
<td>8.7 (0.033)</td>
</tr>
<tr>
<td>BS/BV (mm(^{-1}))</td>
<td>9.71 – 30.29</td>
<td>21.17 (3.59)</td>
</tr>
</tbody>
</table>

![Figure 4](image)

**Figure 4**: Example Result: (a) Original geometry (b) After 14 iterations of remodeling.

**CONCLUSIONS**

We have implemented the algorithm which uses real-world motion data to drive a remodeling algorithm in vertebral bone. Clinically, these results could be used to adapt the lifestyle to sustain stronger bones for longer period of time. Future work collecting more diverse motion data will give insight to activity-related bone remodeling.

**REFERENCES**