Evaluation of Heat Transfer on Bone Cemented Hip Replacement

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Abstract
Evaluation of observable response of heat of bone cement in hip replacement was studied. In an exothermic reaction, bone cementing in a polymerization reaction between the liquid and the powder monomer, provides the bonding mechanism between the prosthesis stem and the femur cavity. This is of concern to biomechanics engineers on the mechanism of heat transfer between the femur bone, the cement and the prosthesis stem. The ANSYS software together with Autodesk software was used to model the scenario, steady state thermal structural analysis was used to simulate it. From the observation, the PMMA polymer (used as bone cement) temperature from the exothermic reaction raised the temperature in the assembly thereby creating a flow of heat amounting to 5.11x10⁻⁷W/m² in which only 2.83x10⁻⁷ W/m²reached to the femur bone as others has been absorbed by the femur bone and prosthesis stem.0.59kJ/kg.K and 1.297kJ/kg. Kare the values of specific heat capacities of femur bone and PMMA respectively while the values of young modulus for femur bone and PMMA are 18.79GPa and 28.78GPa respectively. This result shows how possible it is to determine these properties from the studies of simulation.

Keywords-Bone cement, Bonding Femur bone, PMMA, Prosthesis stem, Simulation, Specific heat capacity, Temperature.

1. Introduction
The increasing number of arthritis cases that cause pain and loss of movement among adult is becoming over alarming. These cases which are seen more in aged women are said to be caused by lack of calcium that aids in strengthening the bone and bone joints, due to bone weakening occurs more as a result of child birth, hence, need for recommendations on the use of bone cementing [1]. Bone cementing over the years has been very successfully used in providing aid for artificial joints such as shoulder and elbow joints, knee joints, hip joints for more than half a century [2]. Artificial joints like the hip joints are usually anchored by bone cementing which is meant to fill the free space that would exist between the bone and the prosthesis. Hence it plays a vital part in creating an elastic limit or zone due to thermal loads agitated by external effects such as walking, running, sitting and even body weights [3].

But also, to the usage of bone cementing, there exist some external problematic factors that leads to adverse effects of bone cementing. This can be seen as hip damages due to fatigue and thermal loadings internally in the joints. With different loadings acting on the joint such as body weight, walking, running, the internal friction increases thereby increasing the internal temperature of the cemented bone. This leads to an obstruct expansion of the cemented bone above the elastic zone and hence causes aseptic loosening [4-5].
This challenge is due to incomplete heat transfer in the bone cement prosthesis system. Also it is believed that cement and thermal mechanical properties and chemical injuries to the bone tissue in the bone tissue in the interface of bone cement are main two factors that affect transfer of heat process in the system. Lewis, [6]; Akanksha et al, [7]; Chaodi et al, [8] looked descriptively on the pre-cooling and pre-heating procedures effect on cement polymerization and cemented hip replacements thermal osteonecrosis. Chi-Chung et al, [9] looked at frictional heating analysis of total hip prosthesis in a two dimensional finite element model. In their work, the frictional heating resulting during enunciation of entire hip prosthesis was been proven to occur using wear test hip simulators. Péreza et al, [10] worked on bone cement polymerization computation modelling were the residual stresses and temperature were significantly analysed. They stressed that the major two concerns linked to bone cement usage are the residual stresses generation and probably thermal necrosis of bone surrounding. Fenton et al, [11] stated that cement fragmentation and foreign body reaction to wear debris was one of the major drawbacks of bone cement in joint replacement, resulting in loosening of prosthetic and periprostheticosteolysis. Charnley after experimenting with various materials while working at Manchester University, settledeventuallyon PMMA-a viscous dough which he formed through mixing the liquid monomer and the powder [12]. This has drawn the attention of several researchers in this field to carry out quality research for their country to achieved a sustainable economic development [13-14]. There is a higher rate incidence of thigh pain with proximally fitted stems. Irrespective of the concerns with regards to the thigh pain and the proximal stress shielding associated femoral stems coated exclusively, good long-term results have been recorded [15]. Recently, there has been an interest in the calcium phosphate cements developments. These cements when they harden they form one or more calcium and phosphate containing compounds and ultimately transforms into apatites in vivo. Hence, these cements are theoretically capable of bonding chemically to bone and in some cases bony resorption and substitution by osteoconduction unlike polymethylmethacrylate [16].

The increase in the cases of arthritis due to loss of calcium mostly in women during child bearing which causes hip damages, there is need for replacement of hip with bone cement which serves as a glue. But heat generated by the cement causes adverse effect of fatigue and thermal loading. The need to determine heat transfer, stress, thermal deformation and strain energy at different temperature values come to play for safe measures and their data recorded will be used for reference purposes. This study also evaluates the thermal behaviour of bone cement in a hip replacement joint.

2. Materials and Methods
2.1 Design of the Hip, Femur and the Ball and Socket Joint

**Design of the Model:** The design of the model was done with inventor professional. At first, the femur bone was modelled which spanned the length of an average adult having a length of 200mm in it, the cavity was created with a cut-extrude operation. This cavity is to permit the insertion of the prosthesis stem which will be used for the thermal simulation. The gap that will be created will be used for filling in the bone cement and thus the thermal behaviour will be characterized.

**Geometry Modelling:** The modelling was done basically with the inventor software. This is due to the flexibility in changing of shapes and volumes which turn into models that depicts your desired aim. In the human anatomy, the femur stands as the largest bone in the body. It is also one of the two strongest bones in the body. The average adult male has a femur of 48 cm in length and
2.34 cm in diameter and can withstand about 30 times the weight of an adult. It is a substantial part of the hip joint and part of the knee joint. These have to be replicated in the design so as to achieve almost similar characteristics of the femur in the real world. The shaft of femur is cylindrical with a rough line on its posterior surface. The revolve tool is the most prominent tool which will be used in the design of the femur. The cavity was created for the insertion of the prosthesis stem using the extrude tool in conjunction with a spline cut. This is because of the un-uniformity of the prosthesis stem as it is to be inserted in the femur cavity.

**Materials Properties Modelling:** The material property which includes density, thermal conductivity and heat capacity were assign for simulation. It is sure to simulate what is obtainable in real life. The cement, the femur bone (socket), prosthesis stem and the space for air void will be accommodated in the material selection.

**Finite Element Formulation:** The heat equation just like as stated earlier are the models that governs the behavioural characteristics of the process. Since it will involve several elemental regions, the numerically based models act based for the boundary conditions to act. This numerical formulation is set to take its foundation of the finite element methods and it’s imperative that we insinuate it in the modelling process. To solve this numerically by the finite element method, an isotropic body with temperature-dependent heat transfer was considered.

The following form is a basic equation of heat transfer;

\[-(\frac{\partial qx}{\partial x} + \frac{\partial qy}{\partial y} + \frac{\partial qz}{\partial z}) + Q = \rho c \frac{\partial T}{\partial t}\]  

(1)

Where; qx, qy and qz are heat flow components through the unit area; the inner heat-generation rate per unit volume is \(Q = Q(x,y,z,t)\), material density is \(\rho\), heat capacity is \(c\), temperature is \(T\) and time is \(t\).

The components of heat flow according to Fourier’s law can be expressed as follows [17]

\[qx = -k \frac{\partial T}{\partial x}\]
\[qy = -k \frac{\partial T}{\partial y}\]
\[qz = -k \frac{\partial T}{\partial z}\]

(2)

Where the thermal conductivity coefficient is \(k\). Substituting Eq. (2) into Eq. (1) gives the heat transfer equation as shown in Eq.(3);

\[\frac{\partial}{\partial x}(k \frac{\partial T}{\partial x}) + \frac{\partial}{\partial y}(k \frac{\partial T}{\partial y}) + \frac{\partial}{\partial z}(k \frac{\partial T}{\partial z}) + Q = \rho c \frac{\partial T}{\partial t}\]

(3)

In assumption, the following types can be the boundary conditions: Specified heat flow; \(qxnx+qymn+qznz = qs\) on \(S_2\), Specified temperature; \(Ts = T1(x,y,z,t)\) on \(S_1\), Radiation; \(qxnx+qymn+qznz = \sigma e T4s\) \(-\alpha qr\) on \(S_i\), Convection boundary conditions; \(qxnx+qymn+qznz = h(Ts-T)\) on \(S_3\). Were the convection coefficient being \(h\), convective exchange temperature is \(Te\), unknown surface temperature is \(Ts\), surface emission coefficient is \(e\), surface absorption coefficient is \(\alpha\), Stefan–Boltzmann constant is \(\sigma\), and incident radiant heat flow per unit surface area is \(qr\).

It is necessary to specify an initial temperature field for a body at the time \(t = 0\) for transient problems.

**Numerical Model:** Residual stress and heat generation. Transfer of heat in the polymerizing PMMA mixture is govern by the form of unsteady heat conduction equation [18-20]
\[ C_p \frac{dT}{dt} = \nabla \cdot (k \nabla T) + S \]  
(4)

Where temperature is \( T(\text{oC}) \), time is \( t(\text{s}) \), specific heat capacity is \( C(\text{J/kg\text{oC}}) \), thermal conductivity is \( K(\text{W/mm\text{oC}}^{-1}) \), density is \( \rho(\text{kg/mm}^3) \), and rate of heat generation per unit volume is \( S(\text{W/mm}^3) \).

The polymerization fraction of therapeutic mass of cementing was given by (Perez et al, 2009);

\[ P = \left( \int_0^t \dot{Q} dt \right) / Q_{tot} \cong \frac{1}{Q_{tot}} \sum_{i=1}^n S_i \Delta t_i \]  
(5)

Where \( \dot{Q}_{tot} \) is a function of temperature, its value is constant and assumed to be 0.121J/mm³. The rate of heat generation as a function of polymerization fraction and temperature is given

\[ R(T) = a(T(T/100) + a_2(T/100)^2 + a_3(T/100)^3 + a_4(T/100)^4 + a_5(T/100)^5) \]  
(6)

From equations (4) and (5), cement polymerization is;

\[ P = (1/Q_{tot})R(T)(P - P^2) \]  
(7)

**Finite Element Discretization of Heat Transfer Equations:** Interpolation of temperatures are used in shape functions \( N_i \) inside a finite element of \( T = [N]/[T] \). Where \([T]\) is \((T_1, T_2, \ldots)\), \([N]\) is \((N_1, N_2, \ldots)\).

Temperature interpolation equation differentiation gives interpolation relation for temperature gradients as following;

\[ \frac{\partial T}{\partial x} = \frac{\partial N_1}{\partial x} + \frac{\partial N_2}{\partial x} \{T\} = [B] \{T\} \]  
(8)

Where matrix of shape functions is \([N]\), matrix for temperature interpolation gradient is \([B]\), and vector of temperatures at nodes is \([T]\).

Galerkin method was used in rewriting Eq. (1).

Thus; \( \int_N \left( \frac{\partial q_x}{\partial x} + \frac{\partial q_y}{\partial y} + \frac{\partial q_z}{\partial z} \right) N_i \; dv \)  
(9)

Divergence theorem was applied to the first-three terms as shown in Eq. (10);

\[ \int_{\partial N} \rho C \frac{dT}{dt} N_i \; dv - \int_N \left[ \frac{\partial N_i}{\partial x} \frac{\partial N_j}{\partial y} \frac{\partial N_k}{\partial z} \right] \{q\} q_i \; dv = \int_N \dot{Q} N_i \; dv - \int_N \{n\}^T \{n\}^T N_i \; ds \]  
(10)

where \( \{q\} = (q_x, q_y, q_z), \{n\} = (n_x, n_y, n_z) \), and outer normal to the surface of the body is \( \{n\} \).

introduction of boundary conditions into Eq. (10), the discretized Eq. (11) is;

\[ \int_{\partial N} \rho C \frac{dT}{dt} N_i \; dv - \int_N \left[ \frac{\partial N_i}{\partial x} \frac{\partial N_j}{\partial y} \frac{\partial N_k}{\partial z} \right] \{q\} q_i \; dv = \int_N \dot{Q} N_i \; dv - \int_N \{n\}^T \{n\}^T N_i \; ds + \int_2 q_s N_i ds - \int_3 h(T - T_e) N_i ds - \int_4 (\sigma e T^3 - a q_T) N_i ds \]  
(11)

**Simulation Effects:** With the basic information attained and boundary conditions specified, the simulations effects are the results to perform from the thermal analysis, after a series of iterative numerical solutions and considerations with the physics orientated functions and equations been placed at the node of the elements formed. The corresponding effects are as follows: stresses (Von Mises), strains deformations along the x, y and z axis), safety factors, heat flux and total heat flux.

**VonMises Thermal Stress:** As the thermal inputs are being initiated on the cement, there will be a transmission of energy which will result into stresses in to prosthetic stems and the body parts.
It comes in a von Mises stress nature and it essential we determine it magnitude. Von Mises stress is widely used by designers to check whether their design will withstand a given load condition. According to distortion energy failure theory;

\[
J_2 = \frac{1}{6}[(\sigma_2 - \sigma_3)^2 + (\sigma_1 - \sigma_2)^2 + (\sigma_3 - \sigma_1)^2]
\]

(12)

**Thermal Deformations:** The length of the body increases as the temperature falls, the stability distance between neighbouring particles decreases. The spatial transformation model can give a mathematical insight of what takes place in the x, y and Z axis;

\[
dl = L_0 \alpha (t_1 - t_0)
\]

(13)  
Where the initial length of object is \(L_0\), the linear expansion coefficient is \(\alpha\), the change in object length \(\Delta l\), the initial temperature is \(t_0\), and the final temperature is \(t_1\).

**Geometrical Association for Parametric Relationship:** The prosthesis stem which is inserted into the femur bone replicates parametrically for a detailed simulation. Hence, it is essential dimensionally design each component of the hip replacement for determining the heat transfer characteristics. The prosthesis stem spans 155.5 mm long, with a ball of 36mm in diameter (18mm radius). The stem exhibits 3 diametric segments. The top most stem is 20.96 mm, the second segment is 17.79 mm and the bottom segment of the stem is 14.31 mm. This is because of the non-uniformity of the femur hip and for the easy insertion and removal of the prosthesis stem.

![Fig. 1: The prosthesis stem](image1.png) ![Fig. 2: The Femur Bone](image2.png)

For the femur design, the design was also accomplished using the Autodesk Inventor software. The top section bone was designed with a diameter of 30.91 mm while the bottom section with 21 mm. a cavity bored using the cut extrude function was made having a diametric hole of the top section of 18.81 mm while the bottom section of the cavity with 20 mm. the gap between the femur and the cavity gives the allowance for the bone cementing operation. The bone spans with a length of about 200mm which was considered from the design standard from most literatures.

**2.2 Experimentation**

**Material selections:** PMMA used in this study as the bone cement are the liquid and the powder. The densities of the powder and liquid as reported by Bergmanna et al, [21] is about 1100kg/m³,
the final PMMA mixture of the material properties are considered in order to simplify the model. Baliga et al., [22] proposed the mass density of bone cementing mixture to be constant \((1.1 \times 10^{-6} \text{ kg/mm}^3)\). According to Perez et al, [18], the bone cement specific heat \(c\) is either temperature dependent \(c = 1.25 \times 10^3 + 6.5 T_J / \text{kg oC}\) or constant altering between 1450 and 2000J/kg oC. Proposed a constant value for the thermal conductivity, \(k\) to be 0.0002W/mm°C. Depending on the modulus of elasticity \(E\) which is time-dependent and the polymerization process involved, bone cement material properties changes. In experimental studies for fully solidified cement, the modulus of elasticity has a wide range of 1583–4120MPa [23], [18]. An average value of 2400MPa was assumed in the present model. In the numerical model proposed for the fully solidified cement and its polymerization fraction, the modulus of elasticity of the PMMA mixture as a function of the modulus of elasticity. The prosthesis stem however, is materialized based on the metal on metal formation. Both the ball and socket are made of titanium, stainless steel, cobalt, chromium or combination of these. For this study, titanium is used. The physical properties, mechanical and thermal characteristics of PMMA considered in this study was reported by [24-25], while the properties of titanium and human femur bone considered was reported by [26].

The only experiment done was in the mixing of the cement to determine the appropriate temperature for the simulation studies. PMMA bone cement preparation involves mixing the solid component, powder, with the liquid component. The ratio is usually 2g of powder to 1ml of liquid [27]; however, this composition can vary depending on the cement type used. PMMA, polymer, benzoyl peroxide, plus the initiator are the solid part while methyl methacrylate, pure monomer, plus the activator are the liquid part. During the polymerization process, monomer MMA is converted into PMMA, which involves an exothermic reaction. Immediately after mixing, a thermocouple was placed on the bone cement to determine the temperature. The mixing of the bone cement performed at the ambient temperature of 25oC was repeated seven times within the temperature range 43oC to 47oC at time frame of 160s. The values of maximum temperature and time for each experiment were recorded on the table 3.1. This temperature range is within the reported range of values by Perez et al, [18]. During polymerisation, the maximum temperature attained ranged from 60oC to 70oC [28-29] Bone cement peak temperature of 60oC and 70oC was reported by Homsy et al, [30].

| SN | Max. Temperature (°C) | Time (s) |
|----|----------------------|----------|
| 1  | 45.0                 | 105      |
| 2  | 47.0                 | 160      |
| 3  | 44.4                 | 100      |
| 4  | 46.5                 | 158      |
| 5  | 47.0                 | 155      |
| 6  | 46.7                 | 159      |
| 7  | 47.0                 | 157      |

Since physical experiment using human femur was not performed, the results of the temperature values in table (1) were used for simulation studies only.

3. Result and Discussion

3.1 Model Analysis

At this phase, the finite element analysis (FEA) was performed on the femur bone to simulate the thermal behaviour as the bone cement was visualized placed alongside the prosthesis stem of the femur cavity. The ANSYS software was deployed alongside with the Autodesk Inventor through
computer aided Engineering to efficiently and effectively simulate the whole process. After the importation of the assembled model, the physical and mechanical properties of the materials, PMMA, Titanium, prosthesis stem and the bone structure, as defined [24-26] were used.

3.2 Model Preparation
The geometry was modelled using the ANSYS software. The co-ordinates were set to locate the axis and origin of the geometry. Since the geometry consists of no movable parts, the contacts areas were set as fixed shown in Fig. 3.

![Assembled Geometry for Mesh Preparation](image1)

![Meshing of the Assembled Model](image2)

Fig.3: Assembled Geometry for Mesh Preparation  Fig. 4: Meshing of the Assembled Model

After this, the model was discretized into finite elements. The discretization reduced the model into smaller elements with nodes. For a discrete setting, the meshing process took the tetrahedral elemental structure. This element accommodated the processing of 3D models in which the x, y and z axes were considered in the setting of the boundary conditions and the acknowledgment of their effects. Fig. 4 is the meshed geometry showing the nodes and elements of the assembled bone. The total elements and nodes recorded were 20545 and 127247 respectively. A finite element model of the bone cemented joint was produced using the Computer Aided Engineering (CAE) package, ANSYSv15.0. The precision and accuracy of the model depends upon its element size or number of nodes and time step size. The increase in number of nodes not only increases the accuracy of the model, but also increases the processing time of the model. An optimum solution could be reached by increasing node density near the region of high temperature gradient, which is in the vicinity of weld line, and decreasing node density near the region of low temperature gradient, which is away from the weld line.

3.3 Boundary Conditions
The boundary conditions were set with the three different materials assembled. The one basic input is the temperature at which the bone cement gets to the termination stage of the polymer process; it is therefore imperative that we determine the effect of that input. The conductive and convective factors are put in place from the prosthesis stem and bone cement respectively. The outputs needed are the heat flux and the structural effects occurring around the assembly. Hence, a thermostatic analysis was carried out with ambient temperature of 25°C. Since the initiation and termination of the polymerization process of the PMMA polymer is between 40°C and 47°C, extreme boundary condition will occur when the initiation process reaches a peak temperature of 50°C. Here, the heat
flux will create several effects thermally and structurally. Bioactive bone cement which consists of CaO-SiO₂-P₂O₅-MgO-CaF₂ has been clinically tested and that gives maximum temperature of 60°C during polymerisation [31-32]. The maximum surface temperature for bioactive bone cement and PMMA cement composite are 35°C and 60°C respectively.

3.4. Simulation Results
The simulation carried out shows a maximum temperature of 57°C at the bone cement. The experimental values of temperature obtained in this work can be seen to be within ranges of other temperatures recorded. With the ambient temperature taken at 25°C for all simulations using ANSYS software, the total surface area was found to be 2.95x10⁻⁶ m².

Fig. 5: 40 Degrees PMMA Thermal Effect   Fig. 6: Heat flux as a function of temperature

From the coding of the simulation effect in Fig. 5, it is observed that femur bone received a heat flux of about 1.249x10⁻⁷ W/m² which emanated from the PMMA bone cement and the heat flux reaching the prosthesis stem was 2.810x10⁻⁷ W/m². Although the heat flux may do little or nothing to the femur or prosthesis stem, it should be noted that the heat is also transferred to the blood vessels which might have some effects generally. The heat flux can be seen in both cases to increase with an increasing temperature. It ranged from 1.25x10⁻⁷ W/m² at 40°C to 2.83x10⁻⁷ W/m² at 50°C for the femur bone, and from 2.81x10⁻⁷ W/m² at 40°C to 5.11x10⁻⁷ W/m² at 50°C for PMMA. The slopes of the graphs Fig. 6, for both the PMMA and the femur bone gave approximately the specific heat capacities of the two materials respectively: 

\[ C_p = \frac{d\theta}{dT} \]

These are given for both the PMMA and Femur bone as: 

\[ C_p = 1.297 \text{kJ/kg.K} \] and \[ C_p = 0.59 \text{kJ/kg.K} \] respectively. According to Landgraf et al [1] the specific heat capacity of the PMMA was 1.2kJ/kg.K. This differs from the result of this work by about 7.6%. [30] recorded the specific heat capacity of the bone cement (PMMA) as 1.6kJ/kg.K which is about 18% different from the value reported in this work. For the femur bone (cortical bone), [30] found the specific heat capacity to be 0.46kJ/kg.K which is about 22% different from the result of this work. These percentage differences may be attributable to minor errors in the simulation studies.

The data obtained were fitted to polynomial of the form: 

\[ C_p = \alpha + \beta T + \gamma T^2 + \lambda T^3 \]

to give the temperature dependence of the specific heat capacity of the femur bone as:

\[ C_p = 1 \times 10^{-6} - 9 \times 10^{-8} T + 5 \times 10^{-10} T^2 + 4 \times 10^{-12} T^3 \]

with \( R^2 = 0.9774 \). For the bone cement the temperature dependence of the specific heat capacity was found to be:

\[ C_p = 3 \times 10^{-7} - 2 \times 10^{-8} T + 5 \times 10^{-10} T^2 + 4 \times 10^{-12} T^3 \]

with \( R^2 = 0.9926 \). The deformation experienced was found to be very
small. At the bottom of the assembly, the value was about $4.29 \times 10^{-5}$m. The ends of the assembly experienced the most deformational effect. The centre received the least deformation effect of about $8.65 \times 10^{-7}$m.

Thermal-structural analysis was carried out. The equivalent Von Mises stress, Von Mises strain and strain energy were determined parametrically within the ranges of loads from the boundary conditions [29-30]. At $40^\circ$C, the structural effects were observed.

From Fig. 7, in terms of the equivalent stress, the femur bone received lower stress of about 2.573MPa from the lower to the upper part of the femur bone while the bone cement and the prosthesis stem received a maximum of 12.39MPa. These stresses are far below the yield strength or ultimate strength of the femur or prosthesis stem but still have some impact on the general assembly. In Fig. 8, it was observed that the Von Mises stress increases as the temperature increases for the PMMA cement but the femur shows a very slight increase of an equivalent stress due to the thermal load. A slight increase was observed from the initialization stage but the propagation and termination showed very minimal increase in the Von Mises stress of the femur bone. From Fig. 8, the Von Mises stress of the PMMA is larger than that of the femur bone because the cement is the source of heat been generated. Also due to the density and nature of PMMA compared to the femur bone, the cement will experience more stress than that of bone. The Von Mises as well as the deformation experienced on the femur bone is far less than that of the Stem/PMMA.
In Fig. 9, the strain also which describes the ratio of the elongation to the original length in the x, y and Z coordinates shows a strain of 0.000143 m at the femur bone and about 0.000431 m at the bone cement. This is also the case in the strain energy as well see Fig. 12. Fig. 10, show that the strain increases as the temperature increases for the PMMA cement but the femur also shows a very slight increase of an equivalent strain just like that of the Von Mises Stress due to the thermal load. The equivalent strain occurring as a result of the temperature effect is seen to be quite related at a coefficient of determination of about 73.25% for the femur bone but about 100% for the PMMA.

The young modulus for each of PMMA cement and the Femur was determined from the slope of stress-strain curves of Fig.11 as 3.44 GPa and 17.06 GPa, respectively. The Young Modulus for PMMA of 3.44 GPa is larger than the range of values reported in MIT property data base (2017). The differences in the values of Young’s Modulus may be due to the fact that the simulation in this work used different cement properties than those used at MIT; they considered pure PMMA while this work considered a combined assembly of the PMMA and the titanium prosthesis stem. The incorporation of titanium of course would make the prosthesis stronger, and the fact that the result of this work is far larger is expected.
The Young’s Modulus for femur bone of 17.06GPa is comparable with literature values. Rho et al., [31], while using both ultrasonic and micro-tensile measurements found the average trabecular Young’s Modulus to be 14.8±1.4GPa and 10.4±3.5GPa for bone cement giving an overall average value of 12.6GPa. This result is smaller than that of this study by about 26%. They found the average Young’s Modulus of micro-specimens of cortical bone measured ultrasonically and mechanically as 20.7±1.9GPa and 18.6±3.5GPa giving an overall average of 19.65GPa. This is higher than that of this work by about 13%. The result of this work appears to correlate more with that of micro-specimens of cortical bone determined by the ultrasonic and mechanical tests.

The strain energy is the energy required to strain a material by causing deformations against its original length. It could be observed that the strain energies of both the femur bone and the PMMA bone cement increased significantly with an increase in temperature Fig.13. The strain energy per degree rise in temperature is 0.3193x10^{-5}J/C for PMMA and 0.1185x10^{-5}J/C for the femur. This shows that the cement is more sensitive to temperature rise than the bone.

### 3.5 Response Surface

The response surface plots show how the temperature, the stress (Von Misses) and the strain are connected and related. The von misses stress and strain are attained as a function of the temperature change.

From Fig. 14a, an optimal value at 45°C produces a correlated relationship between the von misses and the strain. But at maximum temperature, the effects can be seen with a maximum stress and
strain. Fig. 14b follows a similar pattern of the PMMA but shows lower amplitude. The tip of the vertices shows higher correlation between the three factors. This is attributed at the highest and lowest temperature values unlike the centre where the correlation reduced.

4 Conclusion
The heat transfer effect which is instigated by the temperature loads from the polymeric reaction of the PMMA cement successfully created some heat transfer effects on the femur bone and the prosthesis stem. The external factors like ambient temperature showed negligible effects on the heat transfer rate. The temperature range of 40°C to 47°C is within the reported range. The value of heat capacities of the femur bone and PMMA cement are 5.9x10⁻¹KJ/kg.K and 1.297KJ/kg.K respectively. The value of young modulus of femur and PMMA cement are 17.06GPa and 3.440GPa respectively which are comparable with literature values. Thermal stresses, deformations, heat fluxes, strain and strain energy were studied between the temperatures of 40°C and 46°C. The temperature emanated from the exothermic reaction of the PMMA polymer was observed to raise the temperature in the assembly thereby creating a heat flux of about 5.11x10⁻⁷W/m² in which only about 2.83W/m² got to the femur bone as other has been absorbed by the prosthesis stem and the femur bone at which the blood vessels would have gained.

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